Bib:- 535531

THE BIOMECHANICAL BASIS OF SPINAL MANUAL THERAPY

Thesis and computer disc presented for the degree of

Doctor of Philosophy

by

Raymond Y.W. Lee, M.Phil.

Bioengineering Unit

University of Strathclyde

Glasgow

August 1995

Abstract

Mobilisation and traction are commonly employed manual therapy techniques in the management of low back pain. The present work examined the biomechanical effects produced by these procedures.

Lateral radiographs were taken of normal living subjects who were subjected to posteroanterior mobilisation loads. The motion segments were found to extend, except the L5/S1 segment which showed a less consistent pattern. The upper segments also translated posteriorly and the lower ones anteriorly. In a separate cadaveric motion segment study, the movements produced by mobilisation loads were examined after sequential dissection of the anatomical structures. The disc was found to be the principal structure that resisted the loads.

The intervertebral movements produced by posteroanterior mobilisation are small in magnitude. It is unlikely that the therapists can palpate the movements reliably. The results suggest that the technique is a passive test of the spine in three-point bending. Posteroanterior stiffness is likely to be affected by disc lesions and scarring of soft tissues.

The movements produced by traction were examined in cadaveric lumbosacral spines. An attempt was made to reproduce the in vivo loading conditions as accurately as possible. The motion segments were found to flex and translate anteriorly. These were accompanied by increases in the foraminal sizes. Most of these mechanical changes occurred after the application of flexion moment which simulated the Fowler's position. The time-dependent effects of traction were then studied by subjecting the specimens to repeated cycles of traction loads. It was revealed that most of the mechanical effects of traction were lost within 15 minutes after treatment.

The results suggest that traction may enlarge a pathologically narrowed foramen and reduce a posterior disc bulge and. It may also have the potential effect of stimulating the mechanoreceptors of the posterior elements, producing a relief of pain. These therapeutic effects may persist after treatment, although the spine recovers mechanically within a short period of time.

The experimental results have added distinctly to the body of knowledge on the scientific basis of manual therapy, but it is felt that further research in this area is still necessary.

Acknowledgements

First and foremost I must thank Professor John Evans (Jockey Club Rehabilitation Centre, Hong Kong Polytechnic University) for his continuous support, encouragement and guidance. I want to express my good fortune in having had him as my supervisor, in having benefited from the gifts of his uncommon insight, skills and dedication. To Professor John Paul (Bioengineering Unit, University of Strathclyde) I also owe a great debt for his constructive advice and for allowing me to undertake my research study at his Unit.

I would like to express my gratitude to a number of people who, in various ways, contributed to the study. Professor Nikolai Bogduk (Department of Anatomy, Newcastle University, Australia) who always asked critical questions that made me think deeper and provided valuable suggestions on the work. Dr. John Shaw Dunn (Department of Anatomy, Glasgow University, UK) for his excellent tutoring on spinal anatomy and dissection. Dr. Mike Adams (Departments of Anatomy and Orthopaedic Surgery, University of Bristol, UK) for showing me his mechanical testing techniques. Professor Lance Twomey (Curtin University of Technology, Australia) for the interesting and fruitful discussions on the work.

I would like to thank the staff members of the Bioengineering Unit and the Jockey Club Rehabilitation Engineering Centre for their technical support and help in constructing the experimental apparatus, and in particular, Mr. Dave Smith, for his willingness to always help and for collecting the cadaveric specimens. I also wish to acknowledge the assistance provided by the morticians at the Glasgow Royal Infirmary, and staff of the Radiography Department, Duchess of Kent Hospital, Hong Kong, in which the radiographic study was undertaken.

I am indebted to the Hong Kong British Council for the award of the Rehabilitation Engineering Fellowship which allowed me to pursue the first year of my study in Glasgow, and for sponsoring my study travels in the later years. I am also grateful to the Staff Development Committee of the Polytechnic University for allowing leaves of absence to conduct the experimental studies in Glasgow.

Finally, to my wife Helen and my family, I would like to express my love and appreciation for their understanding and continued moral support.

Raymond Lee

Content

■ *Chapter 1*

Introduction

Introduction

Low back pain is one of the most common disorders seen today. Previous studies showed that 60-80% of people suffered from symptoms related to the low back at some time in their lives (Frymoyer et al, 1983; Svensson and Andersson, 1982). Generally accepted risk factors for back pain include manual material handling (frequent lifting while twisting), static postures, driving of motor vehicles, exposure to whole-body vibration and cigarette smoking (Frymoyer et al, 1983; Svensson and Andersson, 1983; Pope et al, 1985; Kelsey et al, 1992).

The social impact of back pain is considerable. It was found that impairments of the back frequently caused activity limitation among persons under 45 years of age (Kelsey et al, 1992). Activity impairment has created serious social handicap, leading to sickness absence, loss of employment and participation in social and community activities (Office of Home Economics, 1985).

Back pain generates a substantial economic burden. The Office of Home Economics of United Kingdom (1985) reported the costs associated with back pain during 1982-1983. The medical costs were estimated to be £156 million. Payment of benefits from the social security fund as a result of certified incapacity for work due to back pain was about £193 million. However, a more serious economic consequence of incapacity for work is the loss of potential industrial output. It was estimated that back pain had deprived the United Kingdom of output worth £1,018 million in 1982-93. Furthermore, there are other less obvious costs associated with back pain which are difficult to quantify, for instance, personal expenditure on alternative medicine or home remedies, wages paid to replacement workers, and wages paid between the time of injury and when compensation payment begins.

Clearly, with the continuing increase in the social and economic burdens, strategies have to be implemented to prevent back pain and improve the management of patients. Incidence of back pain should be reduced by pre-employment screening, training and the application of ergonomic principles to task design to reduce the workers' exposure to the risk factors. Medical and health professionals should have a clear and effective management plan, and all effort should be made to reduce the duration of the disability so as to minimise the undesirable social and economic consequences.

Management of low back pain is a clinical challenge. It is presented with both diagnosis and treatment problems. In most cases, a specific lesion responsible for the low back pain cannot be determined (Nachemson, 1976; Jayson, 1984). In addition, a multitude of treatment approaches are available ranging between simple bed rest and major surgery, and the efficacy of most of these interventions is open to question. Low back pain is usually a self-limiting condition, and it was shown that most patients recovered within 6 weeks irrespective of the type of treatment given (Jayson et al, 1981; Frymoyer, 1988). The apparent favourable response to clinical treatment may simply result from the natural history rather than the treatment itself. There is also a significant possibility of the placebo effect.

Manual therapy is widely used by physiotherapists, osteopaths and chiropractors in the treatment of low back pain. Its common use suggests some degree of success in its application. However, like many other treatment methods, objective data evaluating its efficacy are few. There are several hypotheses relating to the mechanisms of action, but most of these "explanations" comprise observations and opinions which lack substantiating evidence, and few investigations demonstrate scientific rigour. Farfan (1980) commented that what had been said about the mechanical effects of manipulative therapy belongs in the realm of armchair reasoning, or at best, an educated guess.

Posteroanterior mobilisation and traction are among the most frequently employed treatment methods in spinal manual therapy. Their indications and techniques of applications are unclear, and vary between different therapists, clinical disciplines and authorities (Cyriax, 1978; Maitland, 1986; Grieve, 1988).

Posteroanterior mobilisation generally involves the application of oscillatory posteroanterior forces over the spinous process of a given vertebra while the subject is in prone-lying (Maitland, 1986; Grieve, 1988). The forces may be applied by the thumbs or the pisiform bone of the therapist with different magnitude, frequencies and inclinations depending on the preference of the therapist. The technique may be used for clinical assessment as well as treatment. It is believed that the therapist can "feel" the intervertebral movements produced and thereby identify the level of lesion, but this is highly questionable.

Maitland (1986) pointed out that traction is not different from mobilisation. It also involves the application of oscillatory forces, and should be regarded as a form of mobilisation applied along the longitudinal axis of the spine. Traction may be applied manually by pulling the legs (Cyriax, 1978). Nowadays, it is more often delivered by motorised machines with preset periods of "hold" and "rest" (Maitland, 1986; Grieve, 1988). Different positionings of patients have been advocated by different authors, but most commonly, the patient is positioned supine with the legs flexed and supported on a stool (the Fowler's position). A split traction table is usually used to reduce the friction encountered during traction. The upper trunk is fixed with a harness onto the upper half of the split table, and traction force is applied through another harness to the pelvis which is resting on the lower half of the table. There is no consensus on the magnitude of traction force that should be applied and it may be as much as half of the body weight.

Interest in manual therapy appears to continue to grow among clinicians and educators. However, the techniques have passed into popular acceptance from authorities or pioneers of manual therapy without much critical dialogue, scientific data and outcome research. The poor understanding of the scientific basis of manual therapy techniques has led to the lack of consensus on their indications and how the techniques should be applied. It also explains why the efficacy of the techniques cannot be successfully demonstrated.

Manual therapy has been heavily criticised in the scientific community and in medical practice. There is no doubt that a strong need exists to conduct research on manual therapy. The biomechanical effects produced by mobilisation and traction are still not fully understood. This has prompted the present work to examine the deformations of the spine produced by the techniques. By critically evaluating the hypotheses underlying the techniques on the basis of the experimental evidence obtained, it is intended to formulate a theoretical basis for practice and thus improve the management of back pain.

■ *PART I*

REVIEW OF RELATED LITERATURE

Chapter 2

Mechanical Characteristics of the Elements of the Lumbar Spine

---

Mechanical Characteristics of the Elements of the Lumbar Spine

2.1 INTRODUCTION

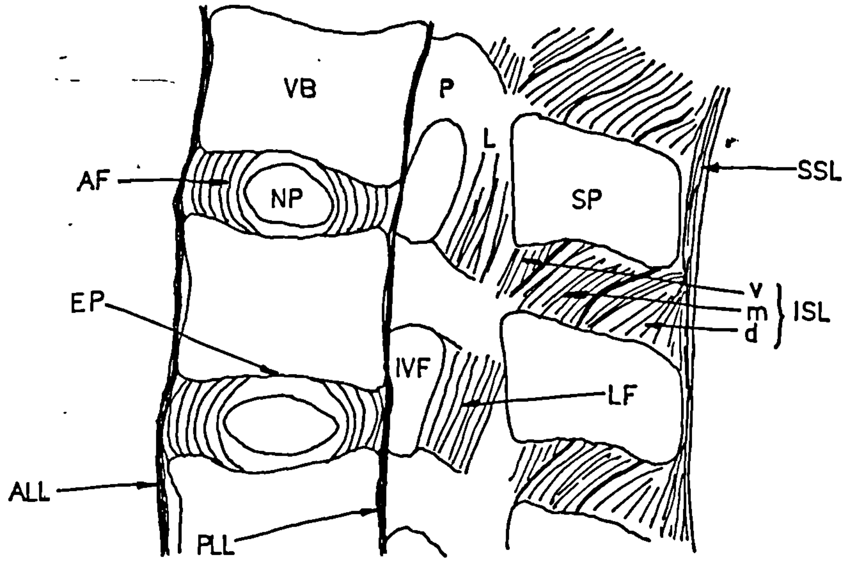
A critical review of the biomechanics of the lumbar spine was undertaken. This is essential in order to understand the mechanical mechanisms underlying the manual therapy procedures employed to treat low back pain. The literature review is presented under three separate headings: mechanical characteristics of the spinal elements, movements of the lumbar spine and previous studies of manual therapy.

This chapter reviews the mechanical characteristics of the various elements of the lumbar spine. It provides the readers with background information and will help understand the materials presented in subsequent chapters. The review also includes those aspects of structural anatomy that are pertinent to the discussion, but avoids reproducing the basic data which can be found in many anatomical textbooks (Romanes, 1981; Anderson, 1985; Williams et al, 1989; Bogduk and Twomey, 1991).

In the next chapter, previous studies on the movements of the lumbar spine are reviewed. These works laid the foundation of the present work which would examine the intervertebral movements produced by mobilisation and traction. The contentious issues in the literature are highlighted and the deficiencies in experimental techniques discussed.

Finally, in order to appreciate the need for a thorough investigation of the biomechanical basis of mobilisation and traction, both in the bioengineering and clinical sense, a review of previous studies on the mechanisms and clinical efficacy of these therapeutic procedures is presented in chapter 4.

(a)



(b)

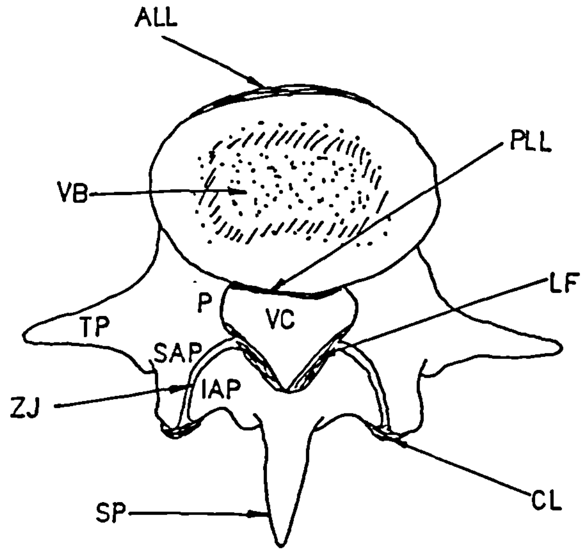


Figure 2.1
Structures of the osteoligamentous spine. (a) A median sagittal section and (b) A horizontal section. (After Bogduk and Twomey, 1991) SSL - supraspinous ligament. ISL - interspinous ligament. v,m,d - the ventral, middle and dorsal parts of the interspinous ligament. SP- spinous process. LF - ligamentum flavum. P - pedicle. L - lamina. IVF - intervertebral foramen. VB - vertebral body. NP - nucleus pulposus. AF - annulus fibrosus. EP - vertebral end plate. ALL - anterior longitudinal ligament. PLL - posterior longitudinal ligament. CL - capsular ligament. ZJ - zygapophyseal joint. SAP - superior articular process. IAP - inferior articular process. TP - transverse process

2.2 THE LUMBAR VERTEBRA

A vertebra consists of an anterior block of bone, the body, and a posterior bony ring, known as the vertebral arch, from which arise spinous, transverse and articular processes (figure 2.1). The vertebral arch consists of pedicles and laminae, and together with the posterior surface of the body, form a vertebral canal, which transmits the spinal cord that the spine protects.

The vertebral body is a shell of cortical bone surrounding a cancellous cavity with a weight-bearing function. Its compressive strength was found to decrease with ageing (Perky, 1957). Under 40 years of age, 55% of the compressive load was borne by the cancellous core, and this share of load was found to be smaller in older specimens (Rockoff et al, 1969).

2.3 THE INTERVERTEBRAL DISC

The intervertebral disc consists of two basic components - a central nucleus pulposus, surrounded by a peripheral annulus fibrosus. The top and bottom aspects of the disc are covered by layers of cartilage which are known as vertebral end-plates.

The nucleus pulposus consists of a randomly oriented network of fine collagen fibrils enmeshed in a proteoglycan-water gel. The water content of the nucleus ranges from 70% to 90%, and tends to decrease with degeneration or ageing (Gower and Pedrini, 1969).

The annulus fibrosus consists of collagen fibres arranged in between 10 and 20 concentric lamellae (Bogduk and Twomey, 1991). The orientation of all the fibres in a given lamella is the same and measures about 65-70^{o} from the caudocranial axis of the spine (Horton, 1958; Hickey and Hukins, 1980). However, the direction of this inclination alternates in successive lamellae.

While the annulus lacks a blood supply, nerve fibres have been identified in its superficial layers (Bogduk et al, 1981). The sources of the nerve endings are the sinuvertebral nerves, and branches of the lumbar ventral rami and the grey rami communicantes of the sympathetic trunk. These nerve endings have been ascribed a nociceptive function (Bogduk et al, 1981; Bogduk, 1983; Bogduk and Twomey, 1991).

2.3.1 Degenerative Changes of the Disc

Structural changes in the intervertebral disc with ageing or degeneration are well documented in the literature (Gower and Pedrini, 1969; Pritzker, 1977; Vernon-Roberts and Pirie, 1977; Bogduk and Twomey, 1991). With degeneration, the nucleus pulposus dries out and becomes more fibrous. The distinction between the nucleus and the annulus becomes less apparent. The collagen lamellae of the annulus become increasingly fibrillated, and cracks and cavities may develop that may enlarge to become clefts and overt fissures.

Earlier authors (Nachemson, 1960; Rolander, 1966; Galante, 1967) evaluated the degree of degeneration of intervertebral disc accordingly to the above-mentioned macroscopic changes. An integer scale of 0 to 3 was employed:

- grade 0: macroscopically normal with shiny gelatinous nucleus;
- grade 1: a somewhat more fibrous nucleus;
- grade 2: clear deterioration of the nucleus with yellowish discolouration, boundary between annulus and nucleus not distinct;
- grade 3: marked ruptures and sequestra in the nucleus or the annulus.

The present work would use this widely adopted method in the evaluation of disc degeneration.

It is generally believed that disc degeneration is an inevitable accompaniment of old age. A large-scale post-mortem study of 204 cadavers has refuted this notion (Twomey and Taylor, 1987). It was found that although the incidence of disc degeneration increased in old age, 72% of the elderly discs examined did not show evidence of degeneration.

2.3.2 Mechanical Properties of the Disc

The compression test has been the most popular test for the study of the intervertebral disc, probably because of its weight bearing function. A thorough review of the compressive characteristics of the disc had been provided by many authors (Farfan, 1973; Pearcy, 1979; White and Panjabi, 1990; Bogduk and Twomey, 1991). It was generally concluded that the disc was a strong structure. It did not fail under compression, even at very high loads and when incisions were made in the posterolateral part of the annulus. The first structure that failed in a vertebra-disc-vertebra construct was the vertebra due to fracture of the end-plates (Brown et al, 1957).

Much less attention has been paid to the tensile and shear properties of the disc. However, tension and shear loads are not uncommon clinically. They are often induced during the application of manual therapy techniques. For instance, clinical traction therapy produced flexion of the spine (Colachis and Strohm, 1969; Twomey, 1985) which would induce tensile stresses in the posterior annular fibres. Furthermore, during posteroanterior mobilisation, the spine was subjected to shear (Lee, 1990; Lee and Evans, 1994). The motion segments also extended producing tensile stresses in the anterior annular fibres.

It was reported that the disc was more flexible in tension than in compression (Markolf, 1972). This was due to the hydrostatic pressure generated within the nucleus during compression. For tension applied along the longitudinal axis of the spine, the anterior and posterior regions of the disc were stronger than the lateral region, and the central region, consisting of the nucleus pulposus, was the weakest (Brown and associates, 1957). The tensile strength of the disc varied to a great extent with the direction of force application (Galante, 1967). It was stronger when tension was applied along the fibre direction than when it was along the anteroposterior direction.

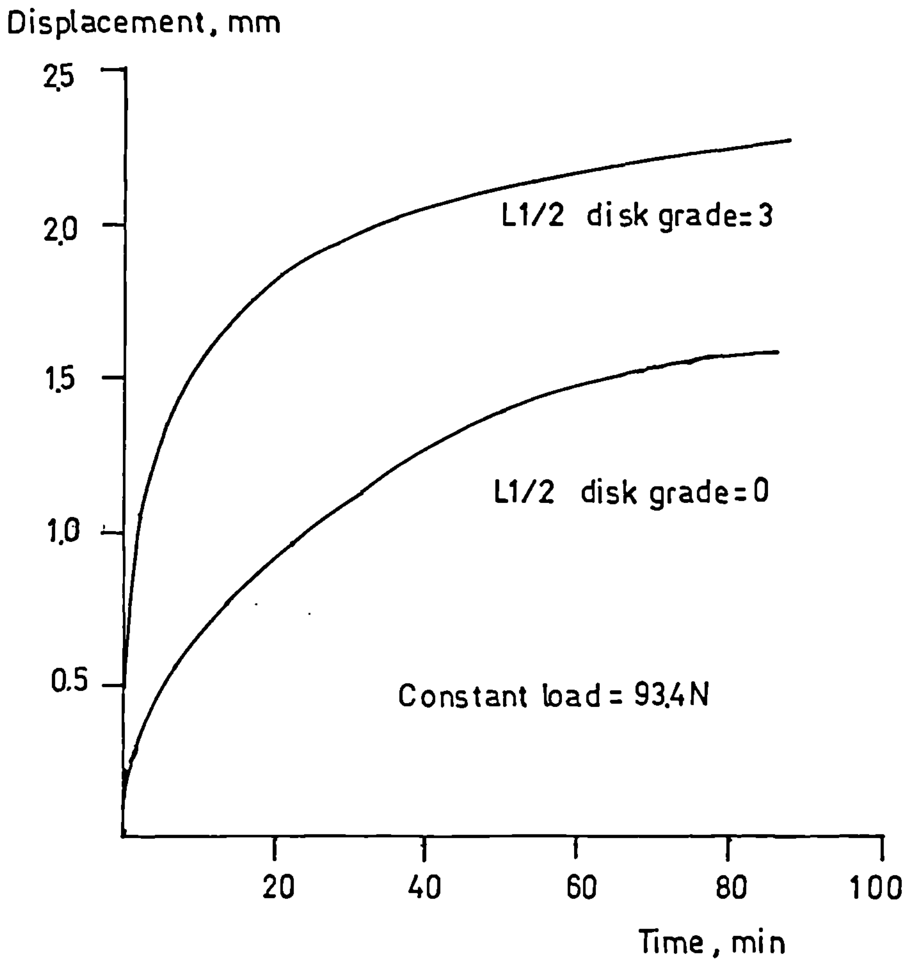


Figure 2.2
Creep curves of lumbar intervertebral discs of different grades of degeneration under constant compression. (After Kazarian, 1975).

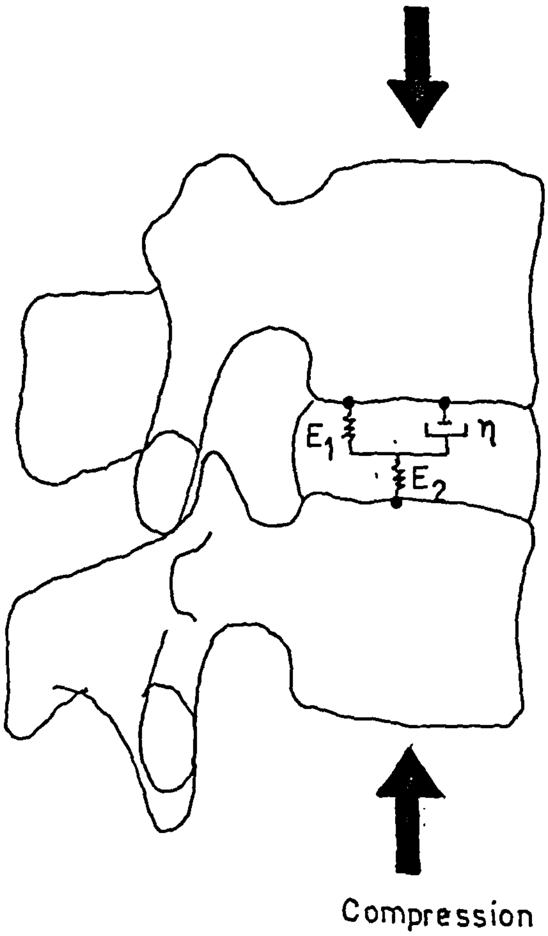


Figure 2.3
Three-parameter solid model of intervertebral disc. A Kelvin solid composed of a Hookean body (E_1) and a dashpot (η) is connected in series with a Hookean body (E_2). (After Keller et al, 1987).

Markolf (1972) performed anteroposterior and lateral shear tests of discs, using specimens which composed of three vertebrae and the two intervening discs with the posterior structures removed. The shear stiffness was similar in both the anteroposterior and lateral directions. The deformations of the discs observed in shear were small compared those in bending and torsion.

2.3.3 Viscoelastic Properties of the Disc

The intervertebral disc is a viscoelastic or time-dependent material. It had been found to exhibit creep and stress relaxation (Hirsch and Nachemson, 1954, Kazarian, 1975; Burns and Kaleps, 1980; Burns et al, 1984; Keller et al, 1987). The stiffness of the disc was also influenced by the rate of loading (Farfan, 1973).

Figure 2.2 shows the creep curves of isolated intervertebral discs under constant compressive loading (Kazarian, 1975). They generally possess the following characteristics: an initial rapid deflection immediately following the application of the load, a stage of creep at a decelerating rate, and a stage of creep at an approximate constant deflection, which in the extreme case approaches zero creep rate.

The creep characteristics were found to be dependent on the degree of degeneration of the specimen (Kazarian, 1975). Degenerated discs creeped faster and equilibrated in a shorter time and deformed as much and usually more than discs with no signs of degeneration (grade 0) (figure 2.2).

Rheological laws provide a mathematical description of the creep behaviour of the motion segments. Previous authors attempted to model the experimental creep data by using a three-parameter solid (Burns and Kaleps, 1980; Burns et al, 1984; Keller et al, 1987). This mechanical model consists of a Kelvin solid (which is a parallel combination of spring and dashpot) connected in series with a spring, as shown in figure 2.3, where E_1 and E_2 are the elastic moduli, and η is the viscosity coefficient. It was shown that there was good correlation between the values predicted by the model and those obtained experimentally, with the average error less than 1% (Keller et al, 1987).

The three parameter solid model explains the creep behaviour of the motion segment physically. With a sudden application of force, the elastic element (E_2) enables the joint to undergo instantaneous deformation directly proportional to the applied load, whereas the viscous element (η) produces infinite resistance and remains fixed. When the load is maintained, the viscous element offers no resistance initially and the elastic element (E_1) is deformed. However, as this happens, the viscous element will offer resistance decreasing the rate of creep deformation. Equilibrium is reached when the forces developed in the two elastic elements are equal.

2.3.4 Intradiscal Pressure

Nachemson (1960) pioneered the measurement of intradiscal pressures. The technique was adopted in a number of investigations (Nachemson and Morris, 1964; Nachemson and Elfstrom, 1970; Andersson et al, 1974 and 1983; Nachemson, 1975; McNally and Adams, 1992). It generally involved the insertion of a especially constructed hollow needle into the nucleus pulposus. Pressure variations in the nucleus were transmitted to a pressure transducer in the tip of the needle.

Posture and spinal movements had significant effects on intradiscal pressure (Nachemson and Morris, 1964; Nachemson and Elfstrom, 1970; Nachemson, 1975 and 1992). It was found that the pressure was highest in the sitting position, followed by standing, and then side lying, and lowest in supine lying. The pressures observed during forward and lateral bending movements, coughing, walking and jumping were higher than that in upright standing at ease.

In supine lying, it was demonstrated that intradiscal pressure was considerably less when the Fowler position (with the legs supported so that the hips and knees are flexed) was adopted (Nachemson, 1992). This explains why spinal traction therapy is often delivered with the patient in such position (Hinterbuchner, 1985), and why most patients with back pain are most comfortable with their hips and knees flexed (White and Panjabi, 1990).

The effects of traction therapy on intradiscal pressures had been examined by Andersson et al (1983). A traction force of 550N was passively applied over a period of 30 seconds. No significant change in intradiscal pressure was reported and negative pressures were never observed. Such findings contradicted the theory proposed by Cyriax (1978) that negative pressure developed during traction might suck back a disc protrusion. Andersson et al (1983) also observed that during active traction (that is, subject applied the traction force himself by pulling with the arms), intradiscal pressure was found to increase considerably. This was probably because it was accompanied by strong contractions of trunk muscles.

2.3.5 The Mechanism of Disc Prolapse

Adams and Hutton (1985) conducted cyclic loading tests of motion segments to study the mechanism of disc prolapse. Prior to testing, the nucleus pulposus of each disc was stained with a small quantity of blue dye and radiopaque solution. This enabled any disc prolapse to be monitored by direct observation and discogram. The specimens were subjected to eccentric compressive loading which were offset anteriorly and laterally from the disc centre so that one of the posterolateral corners of the disc was more highly stressed than the other. The peak load was increased at regular intervals during the 5 hour testing period.

It was found that specimens which had signs of annular ruptures before testing were not shown to be particularly susceptible to prolapse. Most of the specimens failed by fractures of the endplates and gradual disc prolapse was observed in only 6 out of the 49 specimens tested. In these cases with prolapse, blue stained nuclear pulp first appeared rather like a bruise, just under the surface of the annulus. It then worked its way to the surface and oozed out. These discs were generally young and

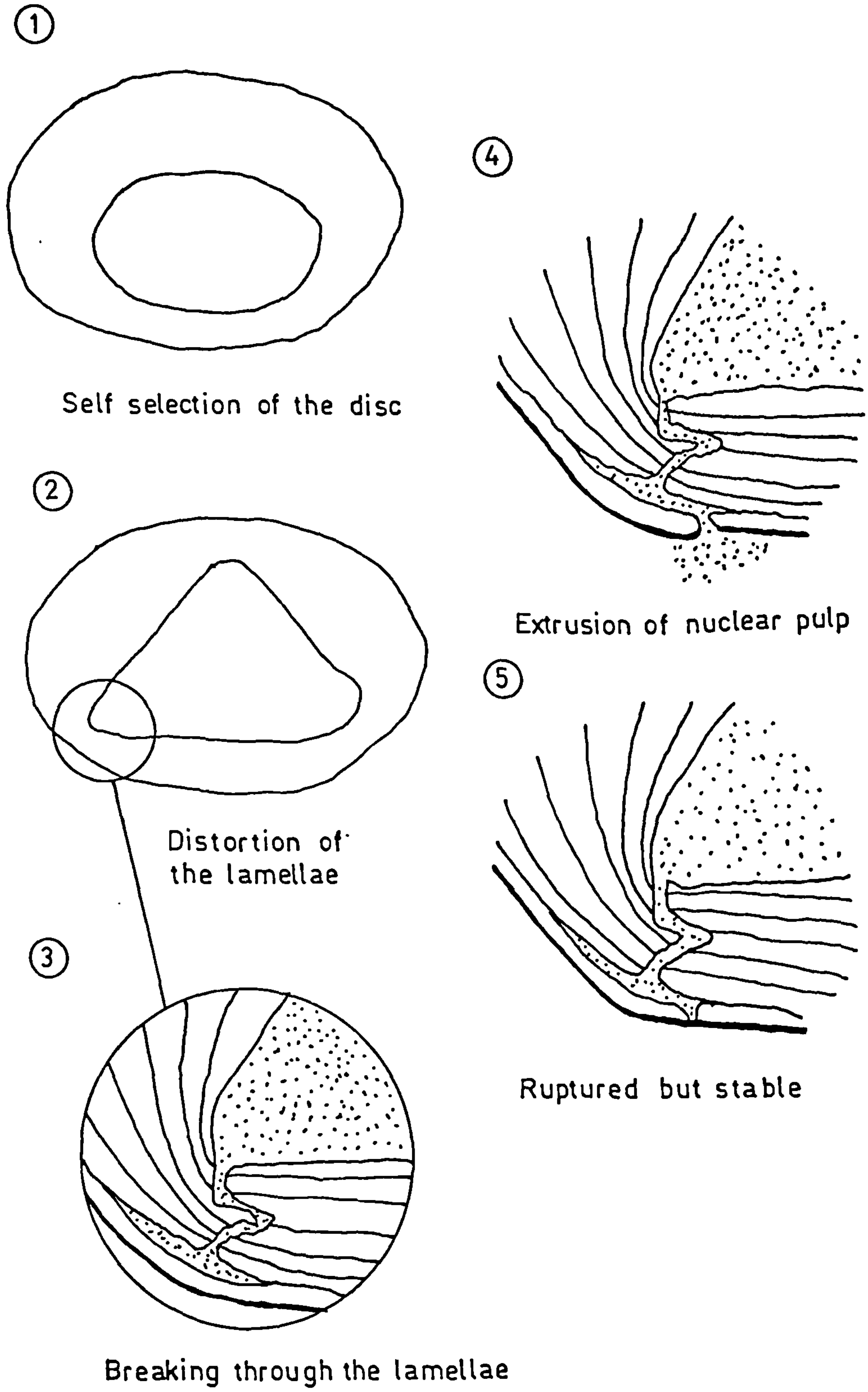


Figure 2.4
The five stages of intervertebral disc prolapse. (After Adams and Hutton, 1985)

discograms showed they had particularly thin posterior annulus and complete posterolateral fissures were formed during the testing.

Based on the above findings, Adams and Hutton (1985) suggested that there are several stages of disc prolapse (figure 2.4). Firstly, young discs with thin posterior annulus and a soft pulpy nucleus are most likely to develop prolapse. With repeated loadings, the annular lamellae are distorted and become tightly packed together. Nuclear pulp gradually breaks through the lamellae, creating a narrow and often tortuous channel, and finally extrude from the disc. As it was found that discs with signs of rupture did not get prolapse more easily than the normal ones, the authors suggested that there is a "self-sealing mechanism" and the disc becomes stable after rupture.

Bogduk and Twomey (1991) documented that symptoms may be produced well before there is a prolapse or rupture of the outer annulus which is pain sensitive. When sufficient amount of annular fibres has been ruptured, excessive strains will be developed in the outer annulus and the disc will become symptomatic. There may also be diffusion of inflammatory chemicals into the outer annulus producing a pain response.

The relationship between the degree of annular disruption and back pain was borne out by the study of Vanharanta et al (1987). Annular disruption was assessed by injecting radiopaque dye into the intervertebral discs of back pain patients and then taking computerised tomography scans of these discs. The degree of disruption observed was found to be correlated with the pain produced by the injection. It was demonstrated that when disruption had extended into the middle third of the annulus, more than 65% of the discs showed exact or similar reproduction of the patients' actual clinical pain.

An understanding of the mechanism of disc prolapse will help appreciate the mechanisms through which manual therapy techniques may reduce a disc lesion. The gradual process of disc prolapse may be reversed or at least stopped in the early stage by developing sufficient tension in the outer intact annular fibres and thereby preventing migration of nuclear materials. The present study would examine whether traction techniques would be able to elicit this mechanism. It should be noted that reversal of the process is dependent on the integrity of the outer annular fibres. Hence, at the later stage when these fibres have been broken and disc prolapse has occurred, the above mechanism will be unlikely to take place.

2.4 THE SPINAL LIGAMENTS

The longitudinal ligaments line the anterior and posterior surfaces of the vertebral bodies and the intervertebral discs of the whole vertebral column (figure 2.1). The ligaments of the posterior elements are the ligamentum flava, the interspinous ligaments, the supraspinous ligaments and the intertransverse ligaments (figure 2.1). The longitudinal ligaments and the inter- and supra-spinous ligaments are well-innervated (Bogduk and Twomey, 1991), and nerve endings were also identified in the outermost layer of the ligamentum flavum (Hirsch et al, 1963). When there is an abnormal increase in motion segment compliance, these tissues may be excessively stretched during the application of manual therapy techniques, eliciting a pain response.

There are some misunderstandings of the structural anatomy of spinal ligaments in the literature. Many authors have mistakenly pictured the fibres of the interspinous ligament running in the posterocaudual direction (Hamilton et al 1976; Goel et al, 1985; Williams et al 1989). Its fibres actually cross the interspinous space in a posterocranial direction and the ligament has three identifiable parts (figure 2.1) (Heylings, 1978; Bogduk and Twomey, 1991). The ventral part was a posterior extension of the ligamentum flavum; the middle part formed the main component of the ligament; and the dorsal part merges with the fibres of the supraspinous ligaments. Heylings (1978) showed that the ligament was bilateral anteriorly, and there was a

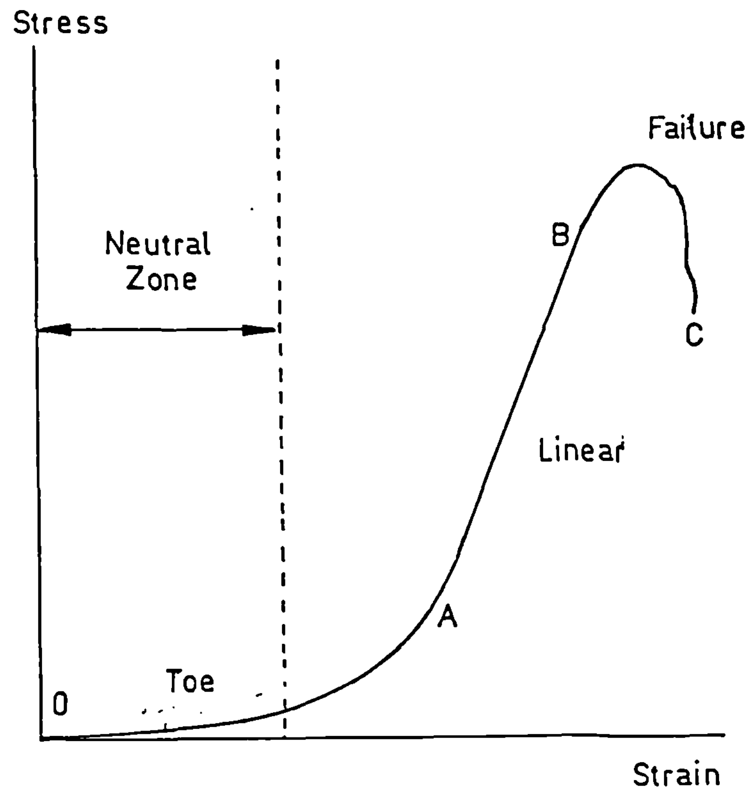


Figure 2.5
The stress-strain curve of a collagenous tissue. (After Nachemson and Evans, 1968; Tkaczuk, 1968; Waters and Morris, 1973; Chazal et al, 1985)

slit-like midline cavity, usually filled with fat. No midline cavity was identified posteriorly.

In addition, the intertransverse ligament is a thin membranous structure, and should not be considered as a true ligament (Bogduk and Twomey, 1991). Unlike other ligaments, it lacks a distinct border and its collagen fibres are not densely packed and regularly orientated. It is also stated that the ligament is weak and has no mechanical significance as bond of union (White and Panjabi, 1990).

2.4.1 Tensile Properties of Spinal Ligaments

The tensile properties of spinal ligaments had been extensively investigated by previous authors (Nachemson and Evans, 1968; Tkaczuk, 1968; Waters and Morris, 1973; Chazal et al, 1985). Their stress-strain curves were generally similar to that of a collagenous structure (figure 2.5). The curve has a nonlinear characteristic and can be conveniently divided into three regions. The initial toe region (from O to A) is concave in the direction of the stress axis. In general, the elongation reflected in this region was believed either to be the result of the wavy pattern of the relaxed collagen fibres which became straighter as more stress was imposed, or perhaps to be caused mainly by interfibrillar sliding and shear of the interfibrillar gel (ground substance) (Viidik, 1968; Barbenel et al, 1973; Tkaczuk, 1968). The curve becomes less concave under the influence of increasing load, and when finally the wavy pattern disappears and the collagen fibres have assumed an orientation in the line of the stress, a straight line is achieved which is the second linear region (from A to B) of the curve. In the third region (from B to C), progressive failure of the collagen fibres takes place and rupture occurs at point C.

White and Panjabi (1990) described the initial portion of the stress-strain curve as the "neutral zone". The ligament could be readily deformed within the zone with minimum application of external loads. Outside this neutral zone, increasing higher loads are required to produce ligament deformation. They also commented that the regions OA and AB constituted the physiological range of motion and the region BC is the traumatic range.

Tkaczuk (1968) examined the tensile characteristics of the anterior and posterior longitudinal ligaments. The strengths of the two ligaments were similar for samples of the same sizes and smaller in older specimens. Shrinkage of the ligaments was observed when they were removed from the spine, indicating that they were in a state of prestress in vivo (2.0-2.5N in the anterior ligament and 2.2-3.4N in the posterior ligament).

Nachemson and Evans (1968) demonstrated that the resting tension in ligamentum flavum (ranging from 4.2-17.6N) was much higher than those of the longitudinal ligaments. These values were found to decrease linearly with age, but was also dependent on the condition of the disc. It was suggested that this resting force would prestress the disc, at least in young individuals, to create an intradiscal pressure of 68.7kPa in the upright position.

Mechanical tests of ligamentum flavum in tension showed that it possessed elastic properties over much of its deformation range (Nachemson and Evans, 1968). This behaviour was probably due to the high content of elastic fibres in the tissues, the proportion of elastic to collagen fibres being 2 : 1. During extension of the spine, the high elasticity of the ligament, together with its pre-tension, minimise the chances of any protrusion of the tissue into the spinal canal. Stress relaxation tests of the ligament indicated that its time-dependent properties were negligible at very low stresses but increase in magnitude with increase in stress.

Water and Morris (1973) conducted tensile tests on samples of interspinous ligaments from scoliotic subjects. It was shown that the mechanical properties of the ligament were similar in patients with idiopathic scoliosis and those with scoliosis of known origin. The tissue was relatively inextensible when compared with ligamentum flavum.

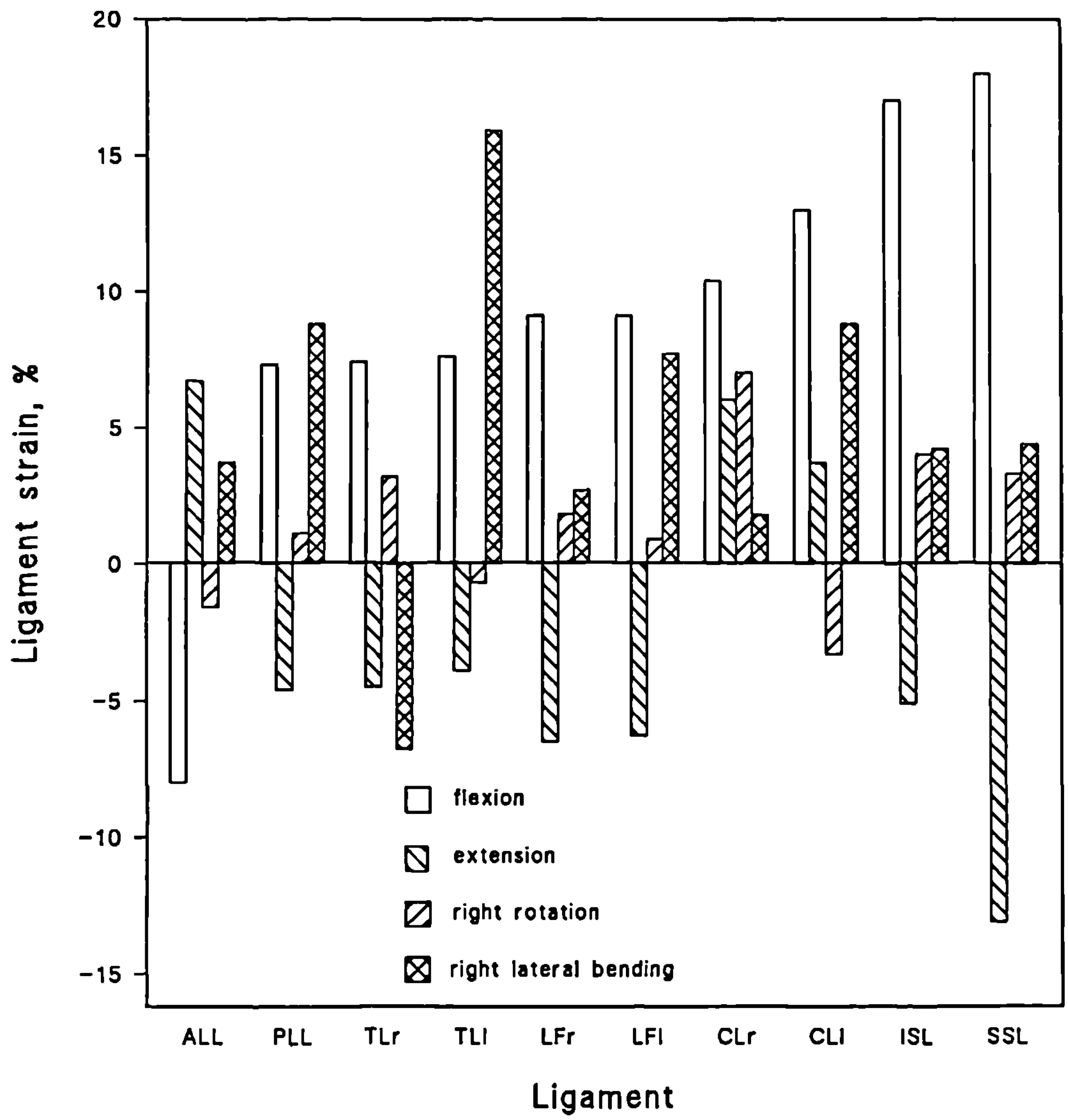


Figure 2.6
Physiological strains in the ligaments of the lumbar motion segments during physiological motions. (After Panjabi et al, 1982)
ALL - anterior longitudinal ligament. PLL - posterior longitudinal ligament. TL- transverse ligament. LF - ligamentum flavum. CL - capsular ligament. ISL - interspinous ligament. SSL - supraspinous ligament. l and r denotes left and right respectively.

2.4.2 In Situ Mechanical Behaviour of Spinal Ligaments

Biomechanical studies of isolated ligament specimens discussed in the previous section provide excellent information about the material properties of the individual ligaments, but in none of them was the ligament tested in situ.

Panjabi et al (1982) carried out an in vitro studies on the in situ behaviour of the spinal ligaments. Changes in the coordinates of the attachment points of the ligaments during movements of the motion segments were computed using a mathematical model. These allowed the ligamentous strains or the changes in ligament length to be determined.

The results of the experiment are summarised in figure 2.6. In flexion, all ligaments, except the anterior longitudinal ligament, were found to be stretched. Highest strains were produced in the supraspinous and interspinous ligaments. In extension, moderate strain was found in the anterior longitudinal ligament. Axial rotation produced maximum strains in the capsular ligament of the same side and the intertransverse ligament of the opposite side carried the highest strain during lateral bending. It was interesting to note negative values of ligament strains in their results. For example, ligamentum flavum had a mean strain of -6.4% during extension. This implied that the ligament had a tendency to bulge during the motion and such tendency was prevented by the presence of resting strain or pretension in the ligament.

Radiographic data provide useful information on the in vivo mechanical behaviour of spinal ligaments. Pearcy and Tibrewal (1984) measured the changes in interspinous distances on lateral radiographs of the lumbar spines which were taken in the upright, fully flexed and extended positions. In flexion, there were very large increases in these distances at all spinal levels (mean percentage changes ranged from 76% at L1/2 to 369% at L3/4). However, tensile tests on interspinous ligament specimens indicated that the average elastic limit of the ligament was only 28%. This implied that the interspinous ligament was lax in the upright position and really functioned only in the extremes of flexion. It was suggested that although the neutral

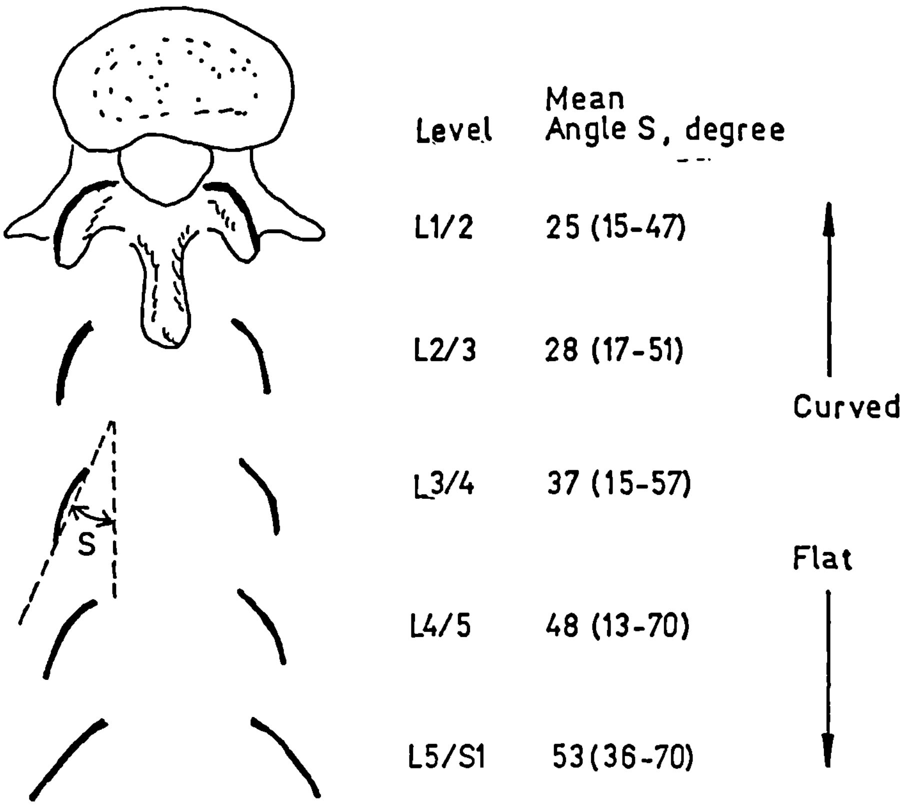


Figure 2.7
The shape and orientation of the superior articular facets of the lumbar zygapophyseal joints in the transverse plane. Angle S represents the inclination of the average facet plane with the anteroposterior axis of the vertebra. (After Taylor and Twomey, 1986; White and Panjabi, 1990)

zone of the load-deformation curves of motion segments was primarily due to the initial low stiffness characteristics of the soft tissue elements (Panjabi et al, 1982; White and Panjabi, 1990), ligament slackness also contributed to the presence of such a zone.

Pearcy and Tibrewal (1984) also reported that in extension, the interspinous distances decreased from 3.5-9.5mm to 0-4.0mm. The bifid characteristic of the interspinous ligament (Heylings, 1978) probably allowed it to buckle laterally on both sides, leaving space for the spinous process to approach one other. The idea of ligament buckling was also substantiated by the in vitro data of Panjabi et al (1982) who reported negative strain values of the ligament in extension.

2.5 THE ZYGAPOPHYSEAL JOINTS

The zygapophyseal joint is clinically important as it is a potential source of low back pain. Mooney and Robertson (1976) demonstrated that pain in the back and leg could be produced experimentally by injecting hypertonic saline into the lumbar zygapophyseal joints of normal individuals and patients with chronic back pain and sciatica. The joint is innervated by the medial branches of the posterior rami from at least two spinal levels (Bogduk and Twomey, 1991). King et al (1990) demonstrated that the fibrous capsule, primarily its inner layers and the area close to the inferior articular recess, was well-innervated by nerve fibres with diameters commonly in the range of 0.5-5.0µm which might conduct nociceptive and proprioceptive sensations.

In the transverse plane, the articular facets may be flat or planar, or may be curved to varying extents (Taylor and Twomey, 1986; Bogduk and Twomey, 1991). The upper lumbar and midlumbar joints are more consistently curved, while the lower joints are often truly planar (figure 2.7).

Variations in the orientations of the lumbar zygapophyseal joints were observed among different individuals and at different segmental levels (figure 2.7) (Taylor and Twomey, 1986; Bogduk and Twomey, 1991; White and Panjabi, 1991;

Panjabi et al, 1993). The angle between the superior articular facet surface and the anteroposterior axis was found to increase from 25° at L1 to 53° at L5. This suggested that facet orientation changed from a medial to a more backward inclination. However, the angle between the facet surface and the transverse plane did not show segmental variation and was approximately 80° (Panjabi et al, 1993), indicating that the facets were also vertical.

It is suggested that the variation in facet orientation has significant ramification on the biomechanics of these joints (Bogduk and Twomey, 1991). The medial orientation of the superior articular facets allows them to resist rotation of the upper vertebra, thereby protecting the intervertebral disc from torsional injuries (Adams and Hutton, 1983). On the other hand, the backward orientation of the facet prevents or limits forward translation of the upper vertebra.

The role of the zygapophyseal joints in resisting rotational movement had been examined by a number of biomechanical studies (Markolf, 1972; Farfan, 1973; Gunzburg et al, 1991). In general, motion segments which had more medial facet inclinations were found to be stiffer in torsion than those which had more backward inclinations. These studies also demonstrated that removal of the zygapophyseal joints caused a substantial decrease in the torsional stiffness of the specimens.

The above discussion has considered the zygapophyseal joint as motion limiter, but it also has a role in bearing axial (vertical) compressive load. Lorenz et al (1983) attempted to study the load bearing characteristics of the joints in lumbar motion segments. Facet loads were computed from the facet contact pressures and the contact areas which were quantified using pressure sensitive films placed between the articulating surfaces of two facets. However, since the articulating surfaces of the zygapophyseal joint are almost vertical (Panjabi et al, 1993), the facet load measured in this experiment is not representative of the vertical load borne by the joints.

Yang and King (1984) examined the loads on the facets of the motion segments using an indirect technique. Facet load was determined from the difference

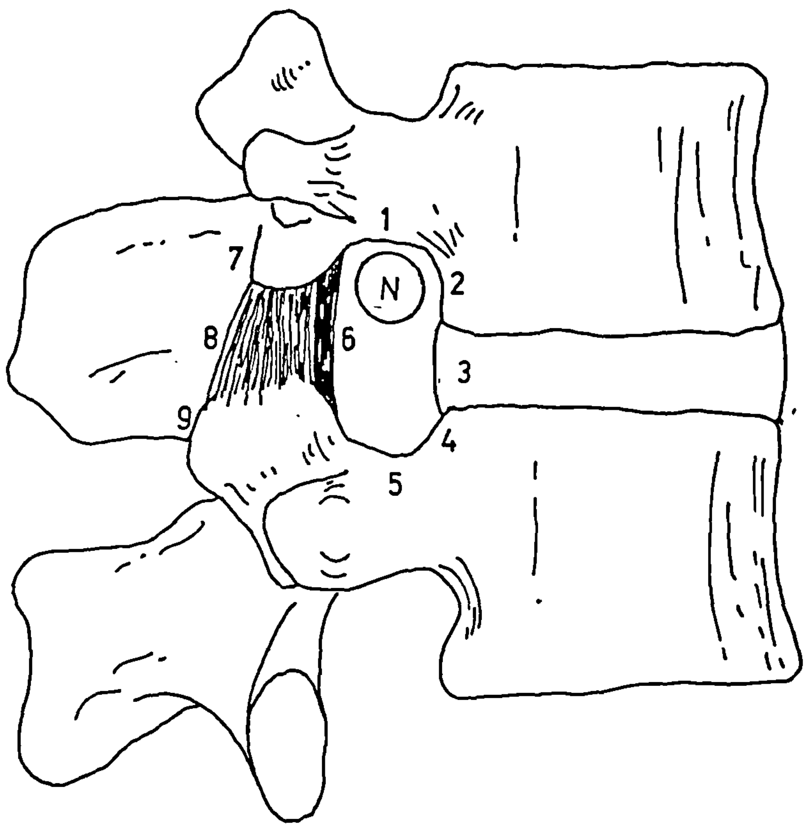


Figure 2.8
Lateral view of the boundaries of an intervertebral foramen and its relation to the radicular complex (After Bogduk and Twomey, 1991).
1- pedicle. 2 - posteroinferior margin of the upper vertebral body. 3 - intervertebral disc. 4 - posterosuperior margin of the lower vertebral body. 5 - pedicle. 6 - ligamentum flavum covering the inferior articular process (7), the zygapophyseal joint (8) and the superior articular facet of the inferior vertebra (9). N - radicular complex leaving the foramen.

between the total compressive load applied and the disc load as measured by a load cell inserted into the vertebra. The results revealed that when maximal load was applied to the centre of the disc, the average facet load was 18% of the total load applied. The average facet load decreased as the compressive load moved anteriorly and increased as it moved posteriorly. In specimens with severe degeneration, facet loads could reach as high as 52% of the total load.

Yang and King (1984) also conducted compressive and tensile tests of isolated zygaophyseal joints which had been dissected from the vertebral bodies. When the joints were loaded to failure in compression, the inferior lumbar facets rotated posteriorly and caused the capsule to rupture without bony fracture. On the other hand, the zygapophyseal joints were much weaker in tension but they also failed due to capsular rupture.

In degenerated motion segments, the zygapophyseal joints may be subjected to excessive compressive or tension during the application of manual therapy techniques and spinal movements. It appears that these may overstretch the richly innervated capsules, eliciting back symptoms.

2.6 THE INTERVERTEBRAL FORAMEN

The structural anatomy and mechanical characteristics of the intervertebral foramen are reviewed in detail in this section, since such informations are not readily available in standard textbooks and the present work would involve investigation of foraminal deformation during traction.

The lumbar intervertebral foramina form a series of cannalized channels through which the lumbar spinal nerves emerge from the vertebral canal. Each foramen is enclosed above and below by the vertebral pedicles with their respective inferior and superior vertebral notches (Anderson, 1985; Bogduk and Twomey, 1991) (figure 2.8). The anterior wall is made up of the intervertebral disc, the adjacent postero-inferior margin of the upper vertebral body and the postero-superior margin of the lower vertebral body. The posterior wall is formed by the ligamentum flavum which covers the anterior aspect of the lamina, the base of the inferior articular process, the zygapophyseal joint and its capsule and the superior articular process of the vertebra below.

Taylor and Twomey (1994) provided a detailed description of the content of the intervertebral foramen and its relationship to the neural tissues. From the lateral recess of the spinal canal (the lateral portion of the canal which is just opposite the pedicle), the nerve roots together with their dural sheath pass obliquely downward and laterally below the pedicle into the upper part of the intervertebral foramen. The lower part of the foramen is occupied by veins, which may be separated from the upper part by a small transforaminal ligament. In the medial portion of the foramen, the ventral root and the dorsal root (with its ganglion) unite to form a mixed spinal nerve and the sleeve of dura becomes continuous with the epineurium of the spinal nerve. This spinal nerve passes out through the upper part of the foramen, behind the lower part of the vertebral body and above the level of the intervertebral disc (figure 2.8). It then immediately divides into ventral and dorsal rami. In addition to the radicular complex described above and two or more quite large veins, each lumbar foramen contains a branch of segmental artery, a small recurrent branch of the spinal nerve called the sinuvertebral nerve, adipose and connective tissues.

The radicular complex may occupy 35% to 50% of the content of the intervertebral foramen (Sunderland, 1974). It occupies the upper half of the foramen above the level of the disc, and is surrounded by rigid bony boundaries. It only comes into contact with the intervertebral disc after it has emerged from the foramen. The hypothesis that disc herniation might compress the neural tissues at the foraminal level and traction therapy could relieve this compression by enlarging the foramen is thus unlikely to be true. Taylor and Twomey (1994) suggested that a herniated disc is more likely to affect the nerve descending to the next intervertebral foramen, in the lateral recess of the spinal canal.

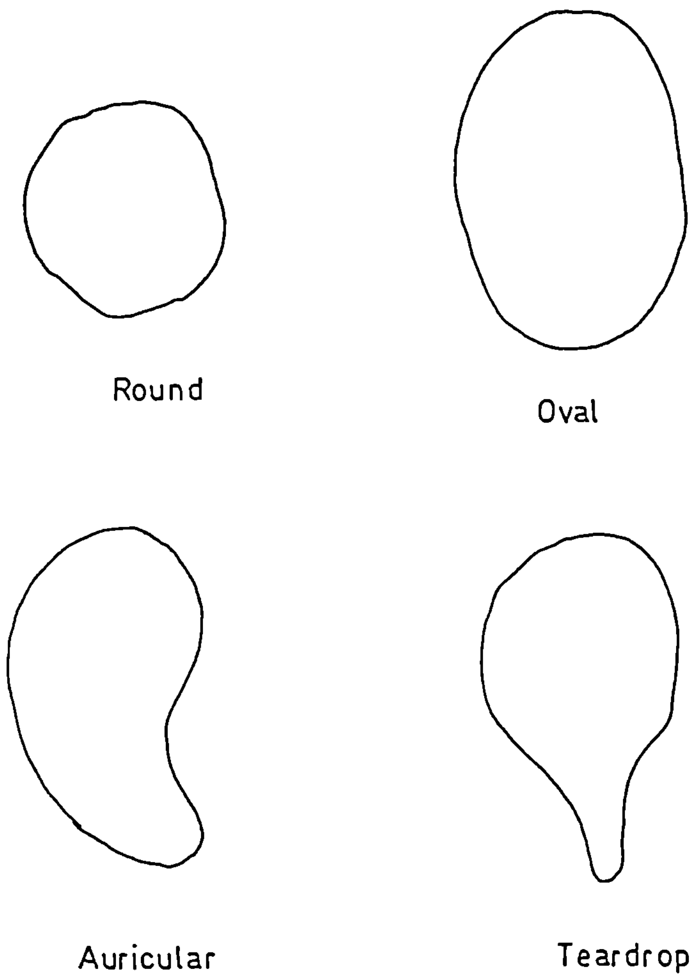


Figure 2.9
The shapes of the intervertebral foramina. (After Stephens et al, 1991)

Although disc herniation may not compress the nerve roots at the foraminal level, a number of other pathological processes will reduce the size of the foramen (by decreasing the foraminal height and/or width) and embarrass the neural tissues (Taylor and Twomey, 1994) described. These include motion segment instability, retrolisthesis of the upper vertebra, disc thinning, vertebral end-plate collapse and osteophytosis of the zygapophyseal joint.

2.6.1 The Shape and Size of the Intervertebral Foramen

Stephens et al (1991) examined the foraminal size and shape of 20 cadaveric lumbosacral spines. All muscles and the contents of the vertebral canal and intervertebral foramina were thoroughly removed. Cotton wool impregnated with silicone rubber was used to take molds of the foramina. After curing, the moulds were sectioned at their waist which represented the narrowest part of the foramina. Foraminal area, height (maximum diameter of the foramen) and width (widest measurement perpendicular to the height) were measured from magnified prints of these cut surfaces.

The shape of the foramen may be described as round or oval, auricular and teardrop-shaped (figure 2.9). It was found that when the associated discs were normal, the majority of the foramina of the upper lumbar segments (L1/2, L2/3 and L3/4) and the L5/S1 segment were oval in outline, but the auricular shape was more common for the L4/5 foramina. When the disc was abnormal, the auricular shape predominated at all segments and the teardrop foramina appeared in greater number. The indentation of the foramina by disc "barrelling" in abnormal segments explained the conversion of the oval foramen into the auricular type. As disc degeneration continued, encroachment into the posterior foraminal boundary by the hypertrophied superior articular facet converted the auricular into a teardrop-shaped foramen.

Stephens et al (1991) reported great variation in foraminal size among different specimens and different levels of the same specimen. The mean foraminal area was

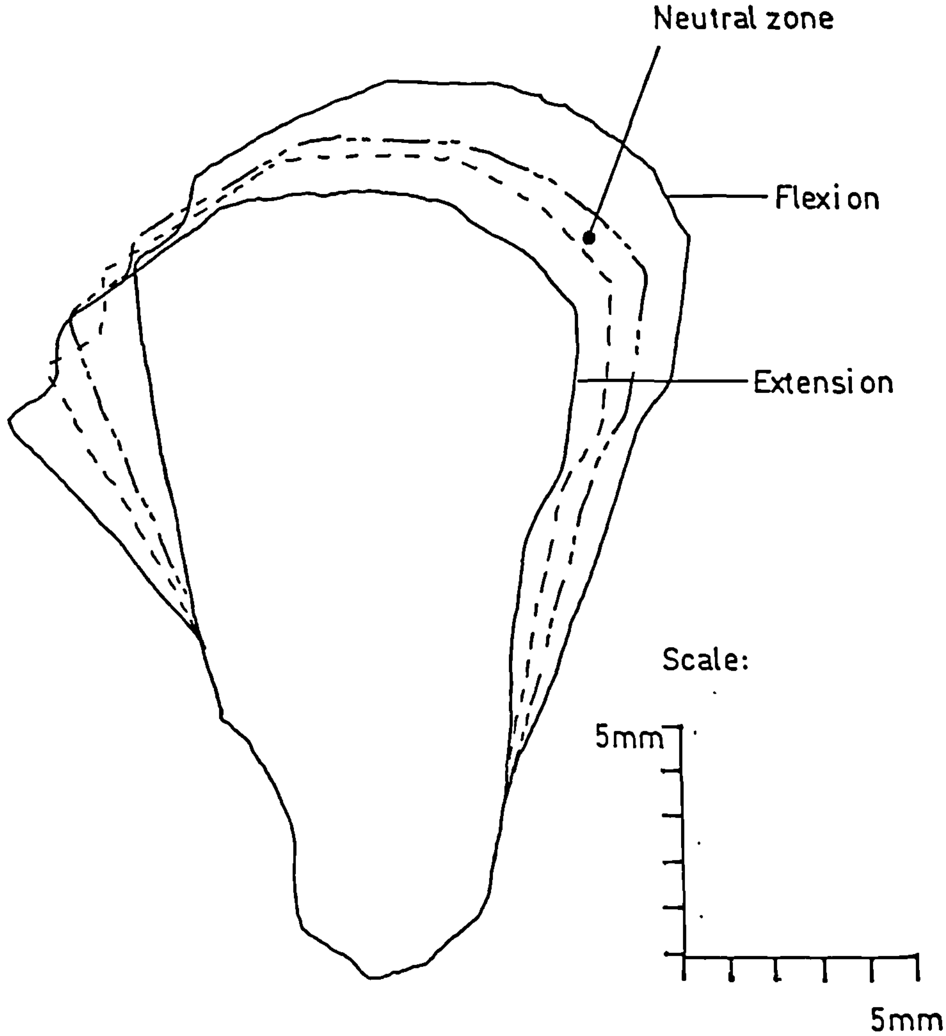


Figure 2.10
Changes in the cross-sectional area of the right intervertebral foramen of a non-degenerated L4/5 motion segments during the application of flexion and extension moment of 7.5Nm. The broken lines define the neutral zone boundaries. (After Panjabi, 1983)

found to be largest at L5/S1, followed by L2/3, L3/4, L4/5 and L1/2. Significant decreases in the foraminal areas were observed when the associated discs were abnormal.

The authors also performed radiographic measurements of the foraminal dimensions and they did not compare closely with the corresponding measurements obtained from the sectioned moulds. This was because the angulation of the foramina materially altered their radiographic size and radiographic measurements did not reflect the minimal foraminal area as measured by the moulds. On the lateral radiograph, the superposition of two foraminal images further added to the difficulty of obtaining accurate radiographic measurements.

2.6.2 Kinematics of Intervertebral Foramen

Panjabi et al (1983) examined the changes in the dimensions of the lumbar intervertebral foramen during physiological movements of the spine. In their experimental study, three dimensional flexibility characteristics of fresh cadaveric motion segments were first determined. These specimens were then fixed and thinly sectioned. The sections were photographed and the projected images of the contour of the foramina digitised. Finally, a mathematical model was used to combine the flexibility and foraminal shape data to compute the changes in the size of the foramina due to the physiological motions.

The results of the experiment revealed that the mean area of the intervertebral foramen increased during flexion and decreased during extension (figure 2.10) and there were significant changes in both the foraminal height and width. Foraminal dimensions did not appear to be affected by lateral bending and axial rotation.

The mean area of the foramen was found to decrease with degeneration. In the degenerated specimens, foraminal dimensions were affected by sagittal rotations as well as axial rotation. The changes in foraminal sizes during movements were much larger than those observed in nondegenerated specimens. Panjabi et al (1983)

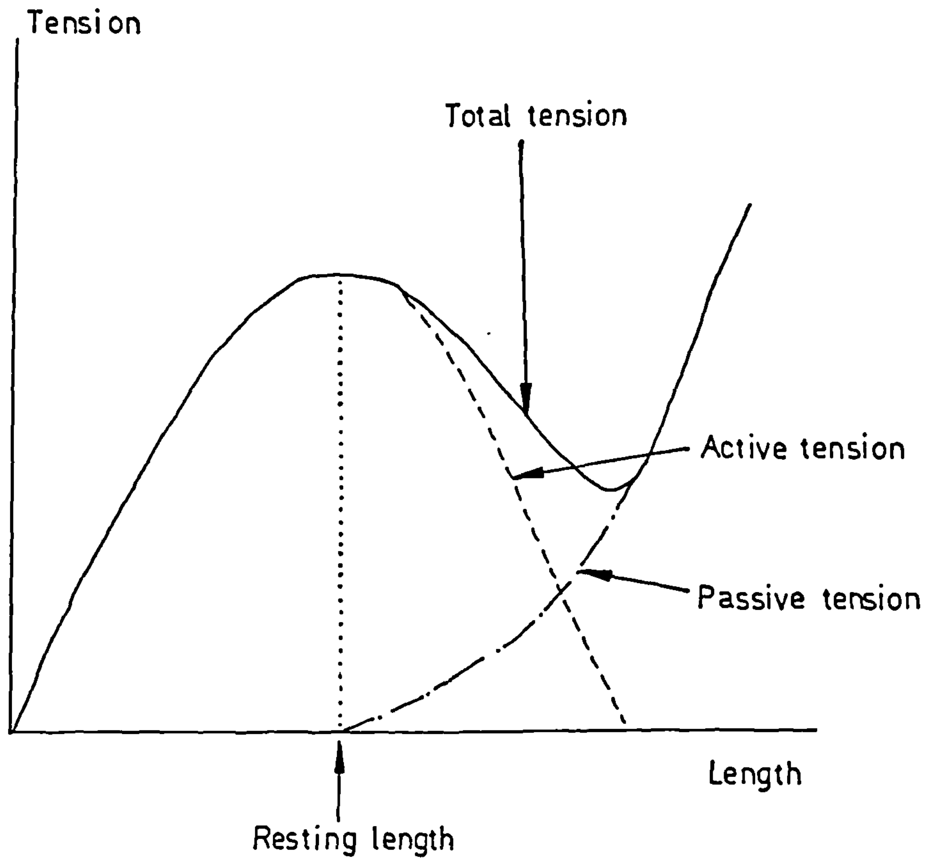


Figure 2.11
A plot of the active, passive and total tension developed in a muscle against its length. (After Winter, 1990)

estimated that for the degenerated segments, the spaces around the nerve root might decrease to only 4.7mm^2 during extension. This will leave very little safety margin and may be enough to cause nerve root compression.

One major limitation of the above experiment is that the mathematical modelling did not take into account the changes in the thickness of the soft tissue boundaries of the foramen during physiological motions, like bulging of the intervertebral disc and the ligamentum flavum. This might affect the validity of the experimental results.

2.7 THE SPINAL MUSCLES

The muscles of the lumbar spine exhibit similar force-length relationship as most other muscles (White and Panjabi, 1990). Such a relationship does not depend only on the muscle fibres, but also the connective tissue network. The total tension in muscle when it is shortened or lengthened is a summation of tension from both the muscular (active tension) and connective tissues (passive tension) (Winter, 1990; Chaffin and Andersson, 1991). Figure 2.11 demonstrates this summation of active and passive tension.

The active tension is maximum at the resting length of the muscle where the number of cross-bridges between the actin and myosin filaments is maximal. With lengthening, the area of overlapping between the two filaments decreases and thus the active tension decreases. Shortening also decreases the active tension because the actin filaments at opposite ends of the sarcomere overlap, interfering with cross-bridges formation. The active tension per unit cross-sectional area that could be produced by the erector spinae muscles at the resting length varies from 35 - 55Ncm^{-2} (McGill and Norman, 1986).

The passive tension curve is nonlinear and resembles the load-deformation curves of any soft connective tissues. As shown in figure 2.11, passive tension plays a role only when the muscle has lengthened beyond its resting length.

During manual therapy, the patient is usually positioned in lying and the musculature relaxed. There is no active tension in the muscles. Traction produces flexion of the spine (Colachis and Strohm, 1969; Twomey, 1985) and therefore lengthens the erector spinae muscles. However, since the amount of flexion produced is small, there will only be little passive tension in the muscles. In the case of posteroanterior mobilisation, there will be no passive tension in the erector spinae muscles as the technique produces extension of the spine (Lee, 1990; Lee and Evans, 1994) and the muscles are shortened. Hence, spinal muscles normally play little role in resisting the movements produced by either traction or mobilisation. This justifies the use of cadaveric materials in the present study where the muscles were removed. However, it should be noted that in patients with back pain, muscle spasm will induce active tension in the muscles and may alter the mechanical characteristics of manual therapy techniques.

■ *Chapter 3*

Movements of the Lumbar Spine

Movements of the Lumbar Spine

Movements of the lumbar spine have been examined in cadaveric spines and living subjects. Cadaveric studies have the disadvantage that they may not accurately reflect the in vivo situation because of post-mortem changes, but they allow precise measurement of movements, the use of conventional material testing machines, and subsequent dissection and histological studies. On the other hand, studies on living subjects will directly reflect the in vivo situation. However, they are limited by the accuracy of the instruments used. Non-invasive surface measurements suffer from errors due to relative movements between the skin and bone and the identification of surface landmarks. Radiographic techniques are commonly used to examine intervertebral movements, but these could also be erroneous due to malalignment of the x-ray beam, poor radiographic film quality and inaccurate identification of radiographic landmarks.

The present work would examine the intervertebral movements produced by mobilisation and traction in both cadaveric and living spines. It was considered that an analysis of sagittal movements alone would be sufficient. This was because both techniques applied loads to the spine in the sagittal plane. As revealed later in this chapter, sagittal loadings did not induce significant movements in the other planes.

The review will concentrate on the mechanics of the sagittal movements of the lumbar spine. Particular attention will be paid to the problems associated with cadaveric testings and radiographic measurements which have been employed in the present work.

3.1 MECHANICAL BEHAVIOUR OF CADAVERIC MOTION SEGMENTS UNDER PHYSIOLOGICAL LOADINGS

3.1.1 The Dependence of Mechanical Properties of Cadaveric Motion Segments on Physiological Conditions

In interpreting the results of cadaveric studies, it is essential to appreciate how mechanical properties may be influenced by changes in physiological conditions after death. The method of specimen storage, the environmental conditions in which the specimens are tested, and post-mortem changes are potential factors that can affect the validity of cadaveric experiments.

Panjabi et al (1985) studied the effect of deep freezing storage on the mechanical properties of the motion segment specimens. Specimens were sealed in double plastic bags and stored at -18°C, and then thawed at the time of testing. It was found that the load-deformation characteristics of the motion segments were not significantly altered, even for very long periods of time of storage (up to 7 months). Earlier authors also demonstrated that this method of specimen storage had no effects on the mechanical properties of the bone, ligamentous tissues and the disc of the motion segments (Sedlin and Hirsch, 1966; Galante, 1967; Hirsch and Galante, 1967; Tkaczuk, 1968).

Galante and colleagues observed that the water content of the intervertebral disc had a significant effect on its mechanical properties (Galante, 1967; Hirsch and Galante, 1967). Immersion of specimens in different solutions (distilled water, 0.9% sodium chloride, plasma and rheomacrodex) all led to swelling of the tissues and significant alteration in these tensile properties. Mechanical testing in a controlled environment of 100% humidity and at a temperature of 25°C was found to minimise the loss of water from the samples. If the specimens were allowed to dry in air, changes in tensile properties were not observed within ten minutes but were significant after one hour.

In view of the above findings, specimens employed for the present study would be stored using the deep freezing method described above. During the mechanical testings, they were covered with moist cotton wool and wrapped in Cling film. This created a high humidity environment and minimised the water loss from the specimens.

Johnstone et al (1992) compared hydration of discs removed at surgery with that of discs taken post-mortem and stored using the deep freezing method. In general, discs taken at surgery were found to have a lower fluid content in the nucleus and a higher fluid content in the outer annulus than discs removed at autopsy. This difference in fluid content was accompanied by a difference in swelling pressure. In discs removed at surgery, the swelling pressure of the nucleus was higher than that of the annulus, whereas in autopsy discs the swelling pressure profile was flat.

The authors explained that after death, fluid exchange via the disc-end-plate margin was blocked and therefore fluid equilibration could only occur within the disc. Fluid would flow from the annulus (high hydration, low swelling and osmotic pressure) to the nucleus (low hydration, high swelling and osmotic pressure) until swelling pressure equilibrated. This redistribution of fluid within the disc explained the results reported.

Post-mortem changes in fluid content and swelling pressure of the intervertebral disc suggest that mechanical properties of the tissues may be altered after death. This was demonstrated by Keller et al (1990) who examined the mechanical properties of spines of anaesthetised pigs before and after they were killed. It was found that after death, there were significant decreases in the stiffness of the spine and the creep rate. The authors also showed the effects of respiration on the mechanical behaviour. Breathing of the animals was controlled by a respirator.

	Posterior element intact	Posterior element destroyed
Mean shear displacement (mm) due to 86N force:		
Anterior shear	0.60	0.94
Posterior shear	0.59	0.84
Right lateral shear	0.67	0.76
Mean shear displacement (mm) due to 145N force:		
Anterior shear	1.21	1.42
Posterior shear	0.85	1.24
Right lateral shear	1.00	1.11

Table 3.1
Mean displacements of motion segments under shear loads in different directions. (Berkson et al, 1979)

Increase in intradiscal pressure was observed with an increase in the breathing volume and a decrease in breathing rate.

Although mechanical properties of spinal tissues were shown not to change with freezing and storage, the studies of Johnstone et al (1992) and Keller et al (1990) revealed that there were differences in the fluid content and mechanical behaviour of the spine between the in vivo and in vitro conditions. These differences existed even when living spines were subjected to a period of bed rest before surgery or to anaesthesia when there was no muscular activity. Results of in vitro studies should be interpreted in the light of these differences.

3.1.2 Compression, Tension and Shear

Previous authors conducted mechanical tests of spinal motion segments in compression, tension and shear (Markolf, 1972; Liu et al, 1975; Lin et al, 1978; Berkson et al, 1979). The motion segments were found to be stiffer in compression than in tension. Markolf (1972) suggested that the higher stiffness in compression was due to the hydrostatic pressure generated within the disc.

The shear stiffness of the motion segments was similar in different directions at shear force of small magnitude (86N) (Berkson et al, 1979) (table 3.1). However, at higher shear force (145N), it was generally found that the segments were more compliant in anterior shear than in posterior with the lateral shear stiffness falling in between (Lin et al, 1978; Berkson et al, 1979). Removal of the posterior elements was found to reduce the shear stiffness in all directions (Liu et al, 1975; Lin et al, 1978; Berkson et al, 1979) (see table 3.1).

3.1.3 Flexion and Extension

A number of studies had examined the mechanical response of the lumbar motion segments under the application of flexion and extension loads (Rolander, 1966; Lin et al 1978; Markolf, 1972; Schultz et al, 1979). They generally

	Posterior element intact	Posterior element destroyed
Mean rotation (degree) due to 4.7Nm moment:		
Flexion	5.13	5.89
Extension	2.12	3.64
Left lateral bending	4.32	4.39
Clockwise torsion	0.69	1.72
Mean rotation (degree) due to 10.6Nm moment:		
Flexion	5.51	5.93
Extension	2.99	*
Left lateral bending	4.90	4.68
Clockwise torsion	1.50	2.28

* not reported by Schultz et al (1979)

Table 3.2
Mean movements of motion segments resulting from the application of bending and torsional moments
(Schultz et al, 1979)

	Percentage of flexion bending moment resisted by the structures	
	at half flexion	at full flexion
Supraspinous/interspinous ligament	8%(±5%)	19%(±7%)
Ligamentum flavum	28%(±10%)	13%(±6%)
Capsular ligament	25%(±8%)	39%(±8%)
Intervertebral disc	38%(±13%)	29%(±12%)

Table 3.3
Mean and standard deviation of the percentage of the applied flexion bending moment resisted by the individual components of the motion segments. (After Adams et al, 1980)

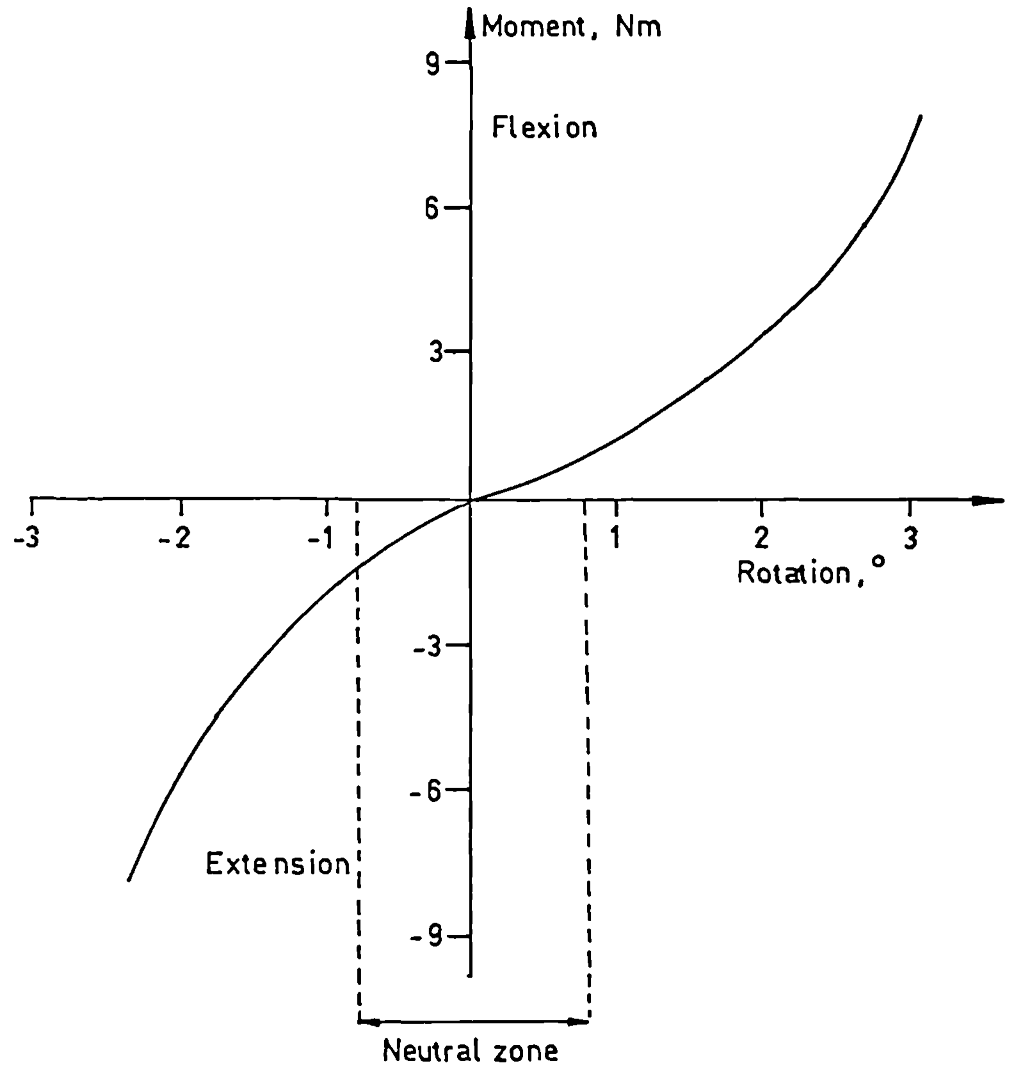


Figure 3.1
Moment-rotation curves for lumbar motion segments under flexion and extension loads. (After Markolf, 1972)

demonstrated that sagittal movements were accompanied by little movements in the other planes. The moment-rotation curves of the motion segments were non-linear and the stiffness increased with an increase in the load applied (figure 3.1). White and Panjabi (1990) called the low stiffness region in the initial phase of the curves the "neutral zone" where the segments were lax.

Previous studies (Rolander, 1966; Lin et al 1978; Markolf, 1972; Schultz et al, 1979) showed that the motion segments were stiffer in extension than in flexion. Excision of the posterior elements was found to decrease sagittal stiffness and the effect was most pronounced in extension. The experimental results of Schultz et al (1979) are presented in table 3.2 for illustration.

3.1.3.1 The role of the anatomical elements in resisting flexion

A number of studies had examined in detail the role of the anatomical elements in resisting lumbar flexion (Adams et al, 1980; Twomey and Taylor, 1983; Goel et al, 1985). The results of these studies are described below. Discrepancies in the results reported by different authors are noted.

Adams et al (1980) simulated physiological flexion by applying a combination of compression, anterior shear and flexion moment. Intact motion segments were loaded to full flexion. The role of the anatomical elements in resisting flexion was determined from the change in bending moment required to achieve the same flexed position after they were dissected sequentially.

The experimental results of Adams et al (1980) showed that at full flexion, the intervertebral disc and the capsules of the zygapophyseal joints provided the most resistance to flexion (table 3.3). In a separate experiment, they observed that there was a marked decrease in the resistance to flexion after cutting the capsular ligaments but very little change in resistance after sawing the zygapophyseal joints. This suggested that the resistance from the zygapophyseal joints was due to tension in the capsular ligament but not bony contact between the articulating facets. Table 3.3 also

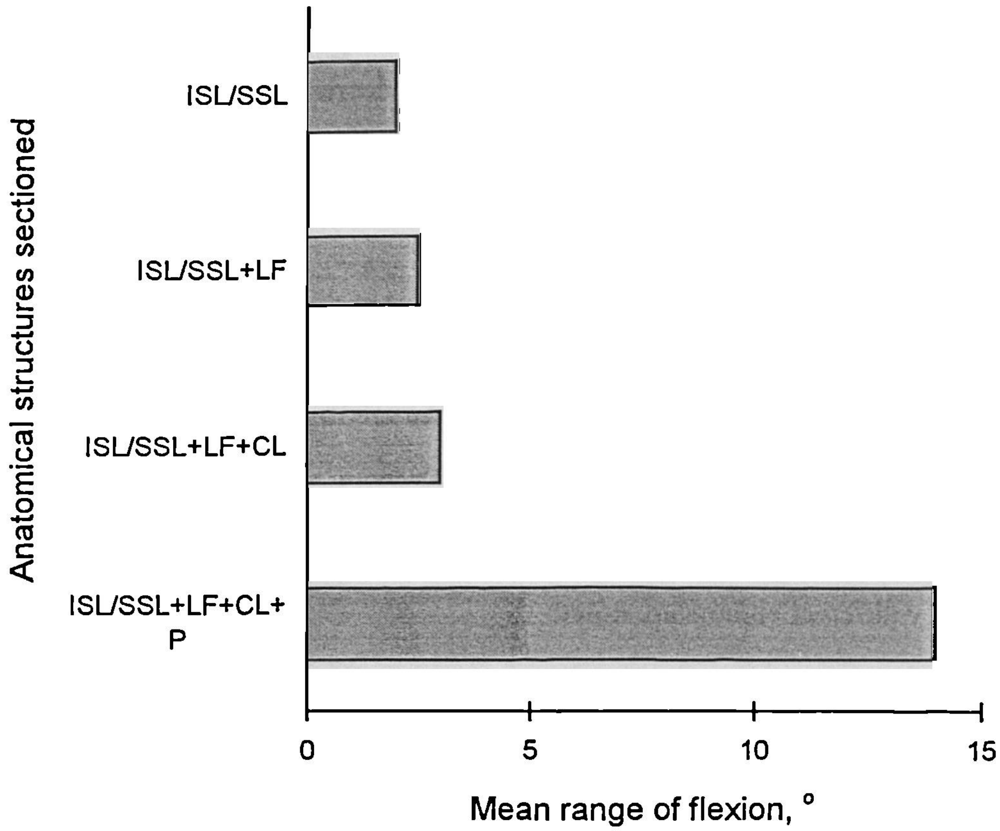


Figure 3.2
Increases in the mean flexion movement of the whole lumbar spine following sectioning of various anatomical structures. (After Twomey and Taylor, 1983)
(ISL/SSL-inter- and supraspinous ligaments, LF-ligamentum flava, CL-capsular ligaments, P-pedicles

shows that the supraspinous and interspinous ligaments played a more significant role but only in the latter half of flexion. This is probably due to the fact that the fibres of these ligaments are lax and buckled in the neutral position.

Twomey and Taylor (1983) examined the contribution of the various structures in whole cadaveric spines. They employed a similar sequential dissection technique to that of Adams et al (1980), and the ranges of motion before and after each dissection were measured. The results (figure 3.2) showed that ligamentous release produced only small increase in motion. The greatest increase in flexion range was recorded after sectioning of the pedicles. The authors thus concluded that zygapophyseal joint apposition had a greater restraining influence on sagittal stiffness than ligamentous factors. This contradicted the finding of Adams et al (1980) who reported that the restraining factor was ligamentous tension.

The differences in the results of the two studies might be explained by the differences in loading conditions. In the study of Twomey and Taylor (1983), a force of 34.3N was applied through the T12 vertebra to generate a flexion bending moment about the lumbar spine. This also produced anterior shear of about the same magnitude at each motion segment and there was no axial compression. The magnitude of flexion moment increased towards the more caudal segments (because of an increase in moment arm). If the distance between the T12/L1 and L5/S1 disc centres were assumed to be 0.18m (Schultz et al, 1973; Chaffin and Andersson, 1991), a maximum flexion moment of 6.17Nm would be produced at the L5/S1 segment. The loads applied in the study of Twomey and Taylor (1983) were relatively small and thus the ligaments would provide very little resistance as they would simply unfold during the subsequent deformation (section 2.4.2). The zygapophyseal joints therefore appeared to have a more significant role than the ligaments. However, in the study of Adams et al (1980), the motion segments were subjected to a mean compressive force of 983N and mean maximum flexion moment of 49.4Nm. These produced much higher deformations than were experienced in the study of Twomey

(a) Ligament force, N	Mean (±1SD)
Supraspinous/interspinous ligament	66.2 (13.6)
Right capsular ligament	14.5 (6.7)
Left capsular ligament	22.8 (10.9)
Right intertransverse ligament	15.0 (8.4)
Left intertransverse ligament	3.4 (1.1)
Right ligamentum flavum	0.0 (0.0)
Left ligamentum flavum	0.0 (0.0)
(b) Disc loads	Mean (±1SD)
Flexion moment, Nm	2.21 (1.15)
Anticlockwise torsion, Nm	0.16 (0.18)
Right lateral bending, Nm	0.29 (0.23)
Anterior shear, N	0.0 (0.0)
Left lateral shear, N	5.89 (4.93)
Axial compression	110.10 (27.10)

Table 3.4
Mean and standard deviation (SD) of the loads induced in the various anatomical elements during the application of 6.9Nm of flexion moment. (After Goel et al, 1985)

and Taylor (1983). Ligamentous slackness was therefore taken up, and the tissues started to offer significant resistance to the movements.

The experimental designs of the above two studies were also different in another important aspect. Adams et al (1980) measured the change in loads required to produce a certain magnitude of motion. Isolated motion segments were used in which structures spanning over more than two vertebrae, such as the supraspinous and longitudinal ligaments, would have lost their continuity and have been only partially functional. Twomey and Taylor (1983), on the other hand, measured the change in motion with a given load and did not destroy the structural integrity of the whole osteoligamentous spine. These differences in experimental designs further explained the discrepancies in the results observed.

The above studies did not allow determination to be made of the simultaneous contribution of the anatomical elements in resisting flexion as the structures were dissected sequentially. Goel et al (1985) employed a semi-experimental approach to solve this problem. The coordinates of the attachments of the ligaments and the disc centre were determined for the initial unloaded position of the specimens. The changes in the ligament coordinates and their lines of actions were then computed based on the movements of the specimens. Using a linear optimisation technique in conjunction with a criterion of minimal anteroposterior shear at the disc, the forces in the ligaments as well as the forces and moments in the disc were computed. This approach made it possible to determine the ligament and disc loads without prior knowledge of the mechanical properties of these components.

The mathematical prediction of Goel et al (1985) is summarised in table 3.4. It was shown that for 6.9Nm of flexion moment, the intervertebral disc provided a resistive moment of 2.2Nm. In other words, the disc had a major role (about 32%) in resisting flexion and this grossly agreed with the finding of Adams et al (1980). In regard to the ligament forces, for 6.9Nm of flexion moment, the supraspinous and interspinous ligament experienced the most force, followed by the capsular ligament and then the intertransverse ligaments (table 3.4). However, Adams et al (1980) showed that capsular ligament had a greater restraining effect than the supraspinous and interspinous ligaments (table 3.3).

This difference in the results was explained by the inaccuracy of the biomechanical model of Goel et al (1985). A closer look at the model revealed that the interspinous ligament fibre direction was mistakenly modelled as posterocaudal, which was at right angle to their true orientation (see section 2.4). This would lead to false estimates of ligament tension. If the interspinous ligament was modelled correctly, the ligament would be predicted to experience a smaller force. This would then be in agreement with the experimental result of Adams et al (1980).

Another limitation of the model of Goel et al (1985) was that the anteroposterior shear was chosen as the objective function so that this force component was always predicted to be zero. This is unrealistic because during forward flexion, there is always an anterior shear force component acting on the disc (Adams et al, 1980; Schultz and Andersson, 1981; Chaffin and Andersson, 1991).

In general, it may be concluded that the disc is the major element that resists lumbar flexion. The relative roles of the zygapophyseal joints and the various ligaments were found to be different in different studies and these have been explained by the differences in experimental methodology.

3.1.3.2 The role of the anatomical elements in resisting extension

The role of the anatomical elements in resisting extension was examined by Adams et al (1988). The technique adopted was similar to that used for examining lumbar flexion as described above (Adams et al, 1980). A combination of extension moment, compression and posterior shear was applied to the motion segments to simulate physiological extension. As in the earlier experiment, the motion segments' resistance to extension was determined initially, and after dissection of each anatomical structure.

	Percentage of extension bending moment resisted by the structures
Spinous processes	46.5% (± 26.2%)
Capsular ligament and ligamentum flavum	5.1% (± 6.7%)
Bony facets	18.6% ± (14.6%)
Intervertebral disc	29.4% ± (14.3%)

Table 3.5
Mean and standard deviation of the percentage of the applied extension bending moment resisted by the individual components of the motion segments. (After Adams et al, 1988)

The study showed that there was large variation in the way the anatomical structures resisted extension. Usually, the spinous processes provided most resistance (table 3.5). This was because the tissue most likely to be affected during extension was the interspinous ligament which would be squeezed between the spinous processes. However, in 6 out of the 44 specimens tested, the bony facets appeared to play a significant role in resisting the extension. These specimens were found to have particularly wide spacing of the spinous processes, and thus most of the loads would be transmitted across the zygapophyseal joints, or from the tips of the inferior articular facets to the adjacent lamina or pedicle. It appears that the morphology of the specimens has a significant influence on the way the structures resist the movement.

3.1.4 Lateral Bending and Axial Rotation

The mechanical response of the lumbar spine in lateral bending and axial rotation is only briefly reviewed as it is not the focus of interest in the present work. More information may be obtained from previous studies (Lovett, 1905; Rolander, 1966; Markolf, 1972; Schultz et al, 1979; Panjabi et al, 1977; Pope et al, 1977; Panjabi et al, 1989; Oxland et al, 1992).

With reference to the experimental results of Schultz et al (1979) which are presented in table 3.2., it was shown that the lumbar motion segments were much stiffer in axial rotation than in lateral bending. The superior articular processes tend to have a medial inclination in the lumbar region. During axial rotation, they impinge on the inferior processes of the adjacent vertebra, thus providing large resistance to the movement (see section 2.5.1).

Excision of the posterior element substantially increased the segmental flexibility in axial rotation (table 3.2). This demonstrated the role of the zygapophyseal joints in resisting the movement. However, posterior release had little effect on lateral bending (table 3.2). This was probably because during lateral bending, there was little impingement of the articular processes (Markolf, 1972).

While flexion and extension are relatively simple movements confined to the sagittal plane, lateral bending and axial rotation are always coupled, that is, lateral bending does not occur without torsion and vice versa. Previous studies had provided a detailed analysis of the motion coupling patterns (Lovett, 1905, Panjabi et al, 1977; Pope et al, 1977; Panjabi et al, 1989; Oxland et al, 1992).

3.1.5 Effects of Compressive Preload on the Mechanical Behaviour of the Motion Segments

In mechanical testings of cadaveric motion segments, it is important to recognise if it is necessary to apply compressive preload. This is because preloading has been shown to have pronounced effects on the mechanical behaviour of the specimens (Panjabi et al, 1977). In order to reproduce the loading conditions during physiological movements in the upright posture, the application of preload is essential so as to simulate the effects of body weight. However, in the cadaveric studies of the present work, no preload application is necessary since mobilisation and traction are generally carried out in lying.

Panjabi et al (1977) showed that compressive preloading increased the flexibility of the specimens in flexion and lateral bending, but the flexibility decreased in the case of axial torsion. No appreciable change in flexibility was observed in extension. The authors did not explain the mechanisms underlying the changes in flexibility with application of preload. It is postulated that preloading reduces the pre-stress of the posterior ligaments which resist flexion and lateral bending and therefore the segments become more flexible in these movements. On the other hand, preloading increases contact forces between the facets, thus increasing the resistance to torsion. Preloading has no effect on extension as it is principally resisted by the spinous process (see section 3.1.3.2).

3.1.6 Influence of Gender, Spinal Level, Degeneration and Ageing on the Mechanical Behaviour of the Motion Segments

In a study of the mechanical behaviour of motion segments, Nachemson and colleagues (1979) demonstrated that female specimens were found to be more flexible than male specimens in response to bending and torsion moment. However, these observed differences were small compared with the large individual variations in mechanical behaviour.

There are no significant differences in the flexibilities of motion segments among different spinal levels (Nachemson, 1979). The L5/S1 segment is an exception. McGlashen et al (1987) showed that this segment was stiffer than the more cranial lumbar segments in flexion, extension and lateral bending. The authors explained that this was due to the extra resistance provided by the iliolumbar ligaments at L5/S1 which did not exist at the more cranial segments. However, in torsion test, the L5/S1 segment was found to be less stiff than the other lumbar segments. The zygapophyseal joint of this segment was aligned further from the sagittal plane than the other segments (see section 2.5). This was expected to result in reduced torsional stiffness.

With disc degeneration, the motion segments tended to be less flexible in flexion, extension and lateral bending but more flexible in torsion. However, such differences in flexibilities were generally small and outweighed by individual differences (Nachemson et al ,1979; Mimura et al, 1994). A more significant change in the mechanical behaviour was found to be the ratio of the neutral zone to the total range of movement (Mimura et al, 1994). With disc degeneration, the ratio increased in all directions of movements, indicating an increase in joint "laxity". In addition, degeneration had significant effects on the viscoelastic behaviour of the motion segments (Kazarian, 1975; Keller et al, 1987). This has been discussed in section 2.3.3.

As pointed out in section 2.3.1, ageing is not inevitably accompanied by disc degeneration. Unlike degeneration, ageing has no pronounced effects on the disc heights and mechanical behaviour of the motion segments (Nachemson et al, 1979; Twomey and Taylor, 1987).

3.2 IN VIVO RADIOGRAPHIC STUDIES OF MOVEMENTS OF THE LUMBAR SPINE

Previous authors had reviewed the various methods which were used to quantify the movements of the lumbar spine in vivo (Pearly, 1986; Helliwell et al, 1992). These included finger-to-floor distance, the skin distraction method (or sometimes called the Schober method), the use of electronic or mechanical devices such as spondylometer, goniometer, inclinometer and flexicurves, various optical and video techniques, and radiographic measurements. More recently, electromagnetic tracking devices had also been used for evaluating spinal movements (Pearcy and Hindle, 1989; Hindle et al, 1990).

Most of the above techniques could measure the total ranges of movements of the lumbar spine only. Radiographic techniques appear to be the most common method for measuring movements of the individual vertebral joints. Other techniques such as the insertion of Steinmann pins into the spinous processes (Gregersen and Lucas, 1967) had also been used for quantifying intervertebral movements. However, this was invasive causing discomfort to the subjects and might not be ethically acceptable.

Radiographic techniques are used in the present work to quantify intervertebral movements. Plain radiograph was considered to be sufficient for assessing sagittal movements which were principally confined to one plane, although accurate measurements of lateral bending and axial rotation would normally require biplanar radiography. A review was undertaken to study the accuracy and reliability of

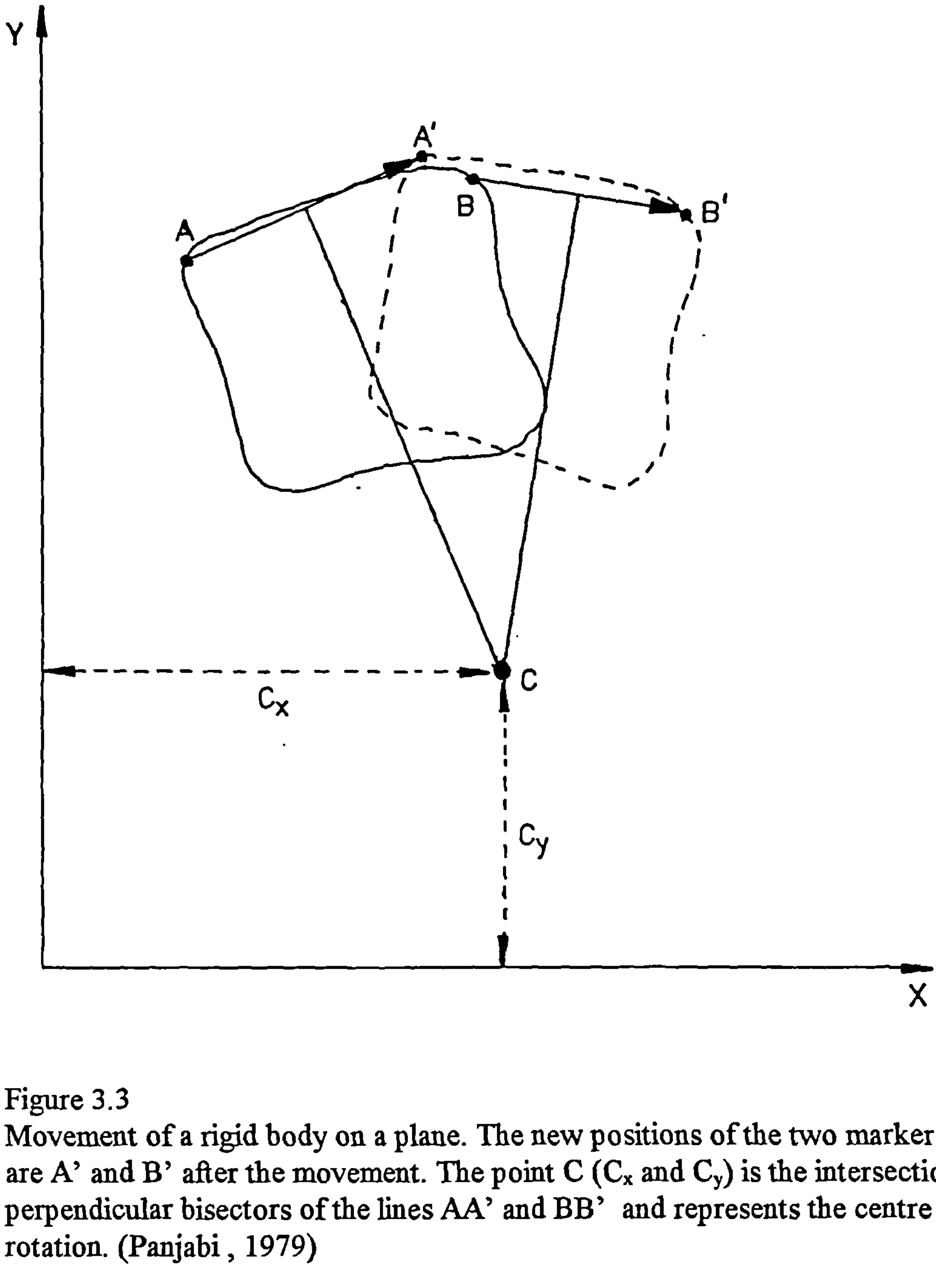


Figure 3.3
Movement of a rigid body on a plane. The new positions of the two markers A and B are A' and B' after the movement. The point C (C_x and C_y) is the intersection of the perpendicular bisectors of the lines AA' and BB' and represents the centre of rotation. (Panjabi , 1979)

radiographic measurements and the findings of previous radiographic studies of movements of the lumbar spine.

3.2.1 Accuracy and Reliability of Radiographic Measurements

3.2.1.1 Identification of anatomical landmarks on radiographs

Determination of vertebral movements from radiographs requires accurate identification of radiographic landmarks. Rab and Chao (1977) determined the coordinates of vertebral landmarks on radiographs and compared them with their true anatomical coordinates. It was shown that the correlation between the two measurements varied with the landmarks selected. The base of the spinous process and the inferior pedicle were least accurate to be located. The endplate centre and the superior pedicle showed close correlation between the radiographic and anatomical measurements, with their absolute positions in space differing by an average of 4mm.

Pearcy and Whittle (1982) showed that the accuracy of identifying bony landmarks could be improved by optimisation technique which adjusted the positions of the landmarks to fulfil the constraint that each vertebra was a rigid body. The authors also demonstrated that the use of metallic markers which were implanted in the vertebra could significantly enhance the accuracy, but this may not be feasible in living spines.

3.2.1.2 Determination of the instantaneous centre of rotation

Instantaneous centre of rotation (C) is often used to document intervertebral joint movement on a plane (Dimnet et al, 1976; Panjabi, 1979; Dimnet, 1980; Panjabi et al, 1982). This generally requires the measurement of coordinates of a minimum of two points in at least two joint positions. Figure 3.3 illustrates that C lies at the intersection of the perpendicular bisectors of the displacement vectors of two markers, A and B, on a rigid body.

As pointed out in section 3.2.1.1, the anatomical landmarks of spinal motion segments might not be readily identified on radiographs. Dimnet (1980) illustrated

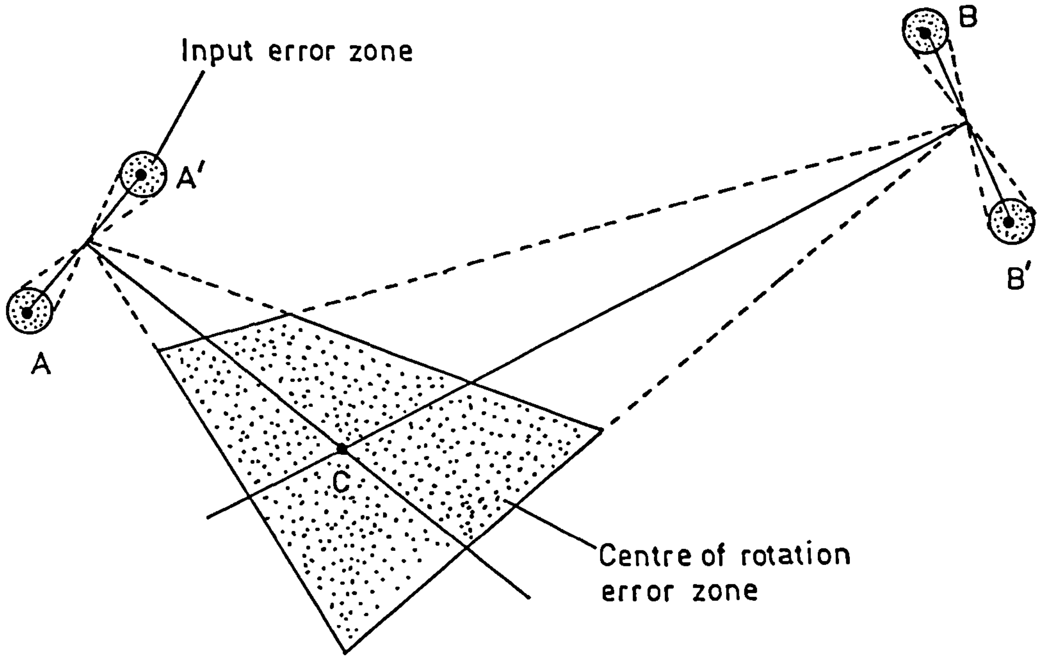


Figure 3.4
Effects of input errors in the coordinates of the landmarks A and B on the errors in the location of centre of rotation C. (After Dimnet, 1980)

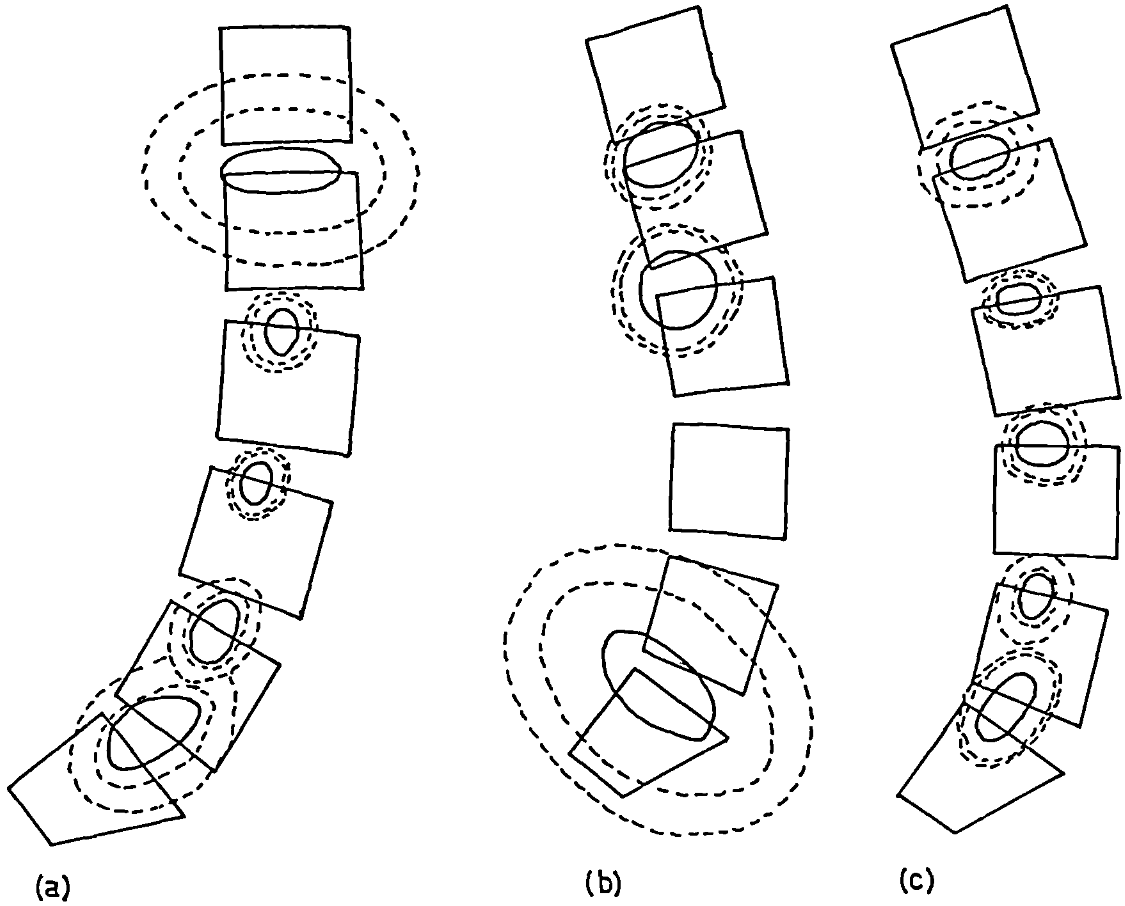


Figure 3.5
The centres of rotation of the movements (a) flexion from upright, (b) extension from upright and (c) flexion from extension. The centres are plotted on a lateral view of the lumbar spine in the upright position. The inner ellipses (with solid lines) depict two standard deviations from the mean centre of rotation of 10 subjects. The intermediate and outer ellipses (with dotted lines) indicate the 95% confidence limits for the within- and between observer errors respectively. (The centres of rotation at L3/4 and L4/5 for extension from upright were subjected to so much error that they were considered to be meaningless, and thus they are not shown in the figure.) (After Pearcy and Bogduk, 1988)

that small errors in measuring the landmark coordinates could lead to relatively large errors in locating the centre of rotation C (figure 3.4).

Previous authors had examined other factors that might affect the accuracy in locating the centre of rotation (Panjabi, 1979; Panjabi et al, 1982). The error was found to be unacceptably large if the magnitude of rotation was less than 5°, and if the markers A and B were located at distances of less than 30mm from the centre of rotation. It was also revealed that the error was minimal when the markers subtended an angle of 90° to each other.

Error in locating the centre of rotation may be reduced when there is a series of radiographs so that a path of C can be determined and then smoothed with a parabolic function (Dimnet et al, 1976). However, such curve fitting procedure will not be possible with only two radiographs when vertebral positions are evaluated at the two extremes of a movement. It was shown that accuracy could also be improved by using more than two markers and making repeated measurements of each marker (Panjabi et al, 1982).

Pearcy and Bogduk (1988) examined the within- and between-observer errors in locating the centre of rotation on lateral radiographs for the flexion and extension movements of the lumbar spine. Figure 3.5 shows their experimental results. There was generally high uncertainty in locating the centres of rotation. Between-observer errors were always greater than within-observer errors. The magnitude of errors varied among different movements. The errors for the movement from full extension to flexion were generally smaller than those for the smaller magnitude movements of flexion or extension alone. The authors believed that only the centres of rotation for the movement from full extension to full flexion could be determined reliably. As in the study of Panjabi (1979), the authors also found that unacceptably large errors occurred for movements which were less than 5°. For the movement extension to flexion, the errors at the L1/2 and L5/S1 segments were found to be greater than the other levels. This was because the radiographs in this study were taken with the beam

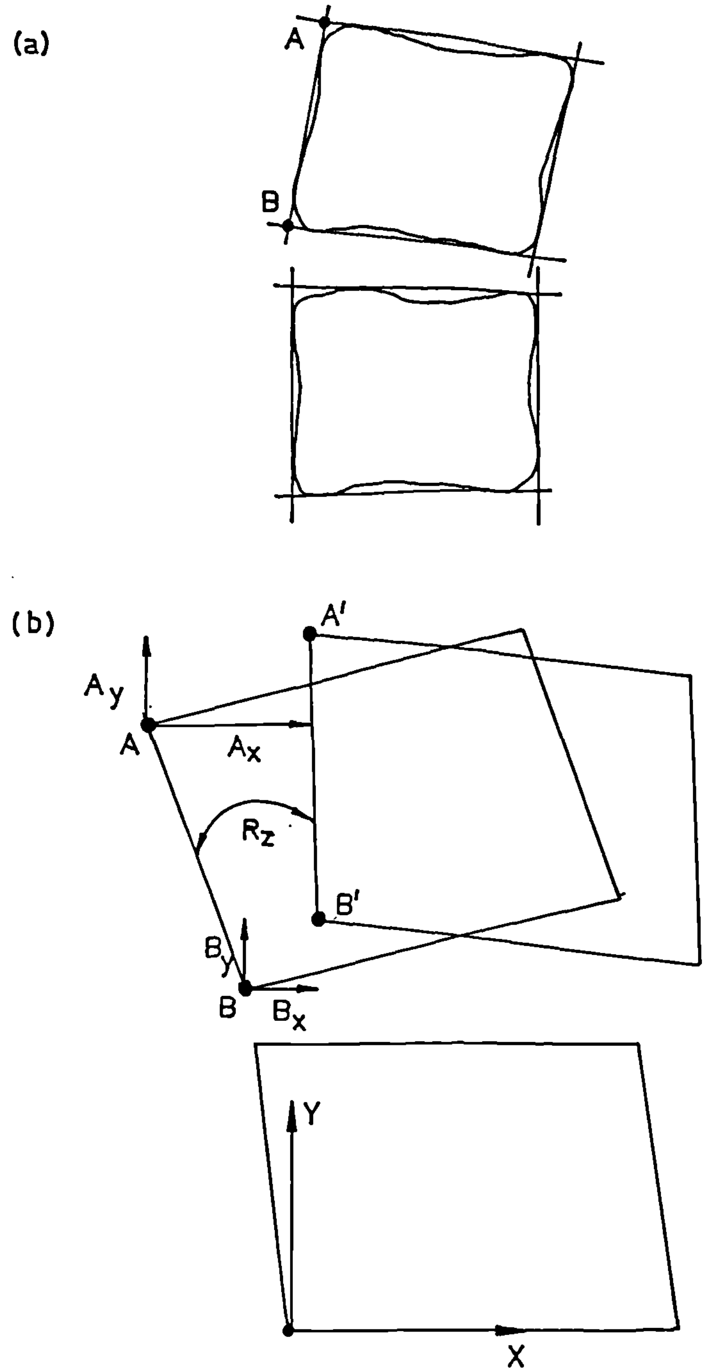


Figure 3.6
Measurement of intervertebral rotation and translations. (a) Each vertebra of the motion segment is enclosed by four lines which are tangential to the vertebral outline. (b) Determination of the vertebral rotation R_z and the translations, A_x, A_y, B_x and B_y at the vertebral corners A and B which have moved to A' and B'. (After Dvorak et al, 1991a)

directed towards the midlumbar region, and consequently the images of the uppermost and the lowest vertebrae were more ambiguous causing larger errors in tracing and superimposing these vertebrae.

In conclusion, the concept of instantaneous centre of rotation proves to be useful only when the magnitude of movement is large. The movements produced by traction and mobilisation would be expected to be small (less than 5°). Hence, the determination of the centre of rotation in this case would be potentially erroneous and might not be meaningful.

3.2.1.3 Measurement of vertebral rotation and translation

As the location of centre of rotation was shown to be prone to large errors, previous authors used the rotatory range of movement and the translation of given points of a vertebra to document intervertebral movement (Dvorak et al, 1991a and 1991b; Panjabi et al, 1992) (figure 3.6).

Any two of the four vertebral body corners can serve as landmarks for computing the intervertebral rotation and translations. However, they are not well defined and might not be identified accurately on radiographs. To overcome this problem, previous authors suggested that four lines, each tangential to one of the faces of the vertebral body image, should be drawn (figure 3.6a). The intersection of these lines could then provide the four points that represented the vertebral body contour (Dvorak et al, 1991a and 1991b; Panjabi et al, 1992). The image of the lower vertebra in the final position was superimposed (either manually or mathematically) onto the corresponding one in the initial position. The displacement of the upper vertebra was documented by computing the intervertebral rotation, R_z, and the translations at the vertebral corners A and B (A_x, A_y, B_x and B_y) (figure 3.6b).

Previous studies had examined the errors in computing the intervertebral rotation and translation (Schaffer et al, 1990; Panjabi et al, 1992). They reported that large errors occurred when digitiser with low resolution and radiographs of poor quality were used, and when there was large out-of-plane movement (>10°). A major

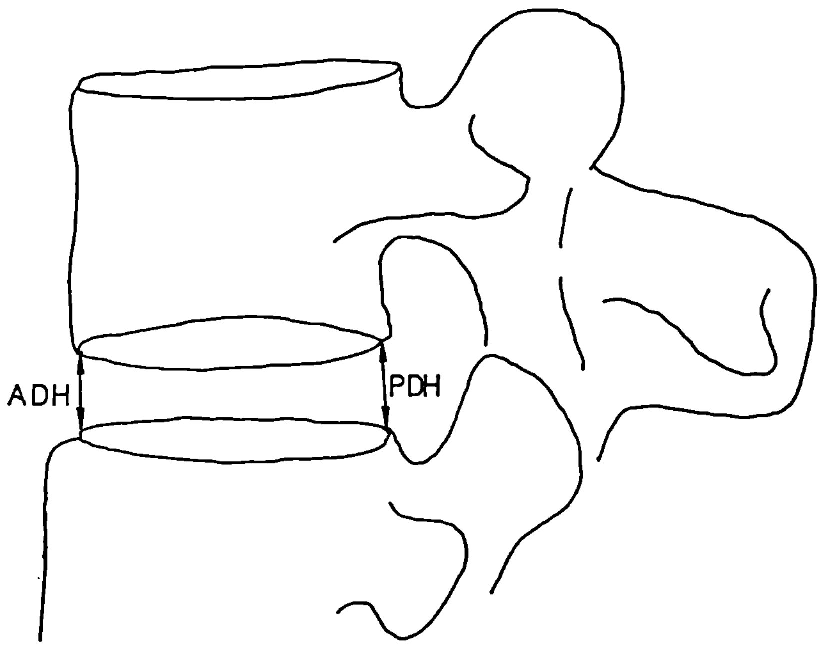


Figure 3.7
Measurement of anterior and posterior disc heights (ADH and PDH). (After Farfan, 1973)

source of error appeared to be the process of marking and superimposition of the radiographic films. In addition, larger errors were associated with the L1/2 and L5/S1 motion segments compared with those in the midlumbar region. As explained earlier, this was due to the fact that the radiographic beam was centred onto the middle of the lumbar spine.

Panjabi et al (1992) documented that the errors involved (defined as one standard deviation of the measurement) in determining intervertebral rotation, translations of the points A and B, and the instantaneous centre of rotation were ±1.25°, ±0.86mm, ±1.06mm and ±4.28mm respectively. This clearly demonstrated that the translations of the points A and B could be determined with much better accuracy than the centre of rotation.

3.2.1.4 Measurement of intervertebral disc height

Intervertebral disc height is frequently evaluated quantitatively from radiographs of the lumbar spine. Pope et al (1977) studied the various methods which were used to evaluate the disc heights. Radiographs were taken of paired vertebrae with plastic intervertebral spaces of known heights, whole cadaveric spines and living subjects. It was shown that the most reliable method of measuring the disc heights was that suggested by Farfan (1973). The method involved identifying the corners of the vertebral bodies and measuring the separation of the anterior and posterior corners of the superior and inferior vertebrae of a motion segment (figure 3.7). However, even with this method, the accuracy was found to be only ±0.5mm. This was not a surprising result because as mentioned earlier, the radiographic images of the vertebral bodies did not have well-defined corners.

Previous studies showed that non-centring of the x-ray beam and out-of-plane motions had significant influence on the accuracy of disc height measurements (Pope et al, 1977; Andersson et al, 1981). The total errors that might result from these sources were considerable, usually about 50% of the measured disc heights. It was

	Mean range of movement									
	Pearcy (1985) (n=11)		Dvorak et al (1991) (n=41)		Putto and Tallroth[Ω] (1990) (n=20)				Hayes et al (1989) (n=59)	
					Method A		Method B			
	F/E[*], °	A/P[#], mm	F/E, °	A/P, mm	F/E, °	A/P, mm	F/E, °	A/P, mm	F/E, °	A/P, mm
Flexion										
L1/2	8	3								
L2/3	10	2								
L3/4	12	2								
L4/5	13	2								
L5/S1	9	0								
Extension										
L1/2	-5	-1								
L2/3	-3	-1								
L3/4	-1	0								
L4/5	-2	-1								
L5/S1	-5	-1								
Total[@]										
L1/2	13	4	11.9	2.6					7	1.9
L2/3	13	3	14.5	3.0	11.5	2.0	11.7	2.4	9	2.4
L3/4	13	2	15.3	3.1	10.6	2.1	12.6	2.8	10	2.5
L4/5	15	3	18.2	2.6	8.6	0.9	12.3	2.8	13	3.0
L5/S1	14	-1	17.0	-0.9	5.9	-0.2	8.9	-0.7	14	1.3

[*] Flexion (positive value) or extension (negative value) movement
[#] Anterior (positive value) or posterior (negative value) translation
[@] The total range of movement from full extension to full flexion
[Ω] Values at L1/2 were not reported by the authors. Method A employed the position of sitting. For method B, sitting with the hips flexed and standing were the positions used for the flexion and extension films respectively.

Table 3.6
Intevertebral movements of the lumbar spine in flexion and extension

concluded that clinical judgement of disc space narrowing should be made with great care.

3.2.1.5 Summary

In summary, radiographic measurements of intervertebral motion are prone to large error if not properly done. There are many factors that may influence accuracy of these measurements. These include the clarity of image, number and positions of chosen landmarks or markers, the process of tracing and superimposition, radiographic quality, obliquity of x-ray beam, digitiser quality, within- and between-observer variance, measurement method and the magnitude of the measured motion. Furthermore, different kinematic parameters can be determined with different precision. Determination of the location of centre of rotation is subject to large error and rather meaningless when the magnitude of the motion is small. The above review had provided important guidelines for the present study in achieving an optimal experimental design.

3.2.2 Flexion and Extension

There were several attempts to quantify the flexion and extension range of movement radiographically (Tanz, 1953; Pennal et al, 1972; Pearcy et al, 1984; Pearcy, 1985; Hayes et al, 1989; Putto and Tallroth, 1990; Dvorak et al, 1991a). Table 3.6 summarises the findings of some of the more recent work. It was shown that during flexion and extension, the only significant coupled motion was anteroposterior translation and the movements were essentially confined to the sagittal plane (Pearcy et al, 1984, Pearcy, 1985). Thus only the primary flexion/extension movement and the coupled translation are shown in table 3.6.

Pearcy (1985) reported separate values for ranges of movements of flexion and extension. Subjects generally flexed more than they extended. As shown in table 3.6, segmental variation in the total range of flexion and extension did not appear to be consistent in different studies. Pearcy (1985) reported that the range was similar for each segmental level (Pearcy, 1985). In some studies, the total ranges were found to be higher at the lower levels (Hayes et al, 1989; Dvorak et al, 1991a), but Putto and Tallroth (1990) reported that L5/S1 showed less mobility compared with the other levels.

In regard to the coupled translation, there was significant amount of anterior displacement of the vertebra during flexion (Pearcy et al, 1984; Pearcy, 1985), but the posterior translation during extension was small in magnitude (Pearcy et al, 1984; Pearcy, 1985). For the movement from full extension to full flexion, the amount of translation at L5/S1 was consistently small. It was frequently reported that there was a posterior displacement of the vertebra at this level (Pearcy, 1985; Putto and Tallroth, 1990; Dvorak et al, 1991a). However, the large error in measuring translation at this level due to quality of the radiographic image (see section 3.2.1.3) cast doubts on the reliability of these measurements.

The differences in the values reported by the different studies (table 3.6) were partly due to the individual variations in spinal mobility. The differences in experimental methodologies also account for the variations observed. For instance, Pearcy (1985) employed biplanar radiographic techniques in the measurements, whereas the other studies used plain radiographs which were more susceptible to errors due to out-of-plane movements. In the studies of Pearcy (1985) and Dvorak et al (1991a), the radiographs were taken with the subjects in standing and the pelvis rigidly fixed. On the other hand, Hayes et al (1989) employed the sitting position with the hips fully flexed for the flexion films and standing without any pelvic fixation for the extension films. Putto and Tallroth (1990) demonstrated that positioning had a significant effect on the intervertebral motion measured (see table 3.6).

In addition, it should be noted that the magnitude of translation was dependent on the coordinate systems used and these were different for the various studies. The translational movement would also be different for different points of the vertebra. The reference landmark for translation was the inferior endplate in the study of Pearcy

	Mean range of lateral bending, ° (positive to the left)					Mean range of axial rotation, ° (positive to the left)	
	Miles and Sullivan (1961) (n=49)		Pearcy (1985) (n=10)		Dvorak et al (1991)* (n=41)	Pearcy (1985) (n=10)	
	Left	Right	Left	Right	Total	Left	Right
L1/2	5.3	-5.3	6	-5	10.4	1	-1
L2/3	5.1	-5.9	6	-5	12.4	1	-1
L3/4	3.9	-4.1	5	-5	12.4	2	-1
L4/5	2.0	-0.8	2	-3	9.5	2	-1
L5/S1	-1.2	0.6	-2	0	5.1	0	-1

* only the total ranges of movements from right to left lateral bending were reported by Dvorak et al (1991)

Table 3.7
Intervertebral movements of the lumbar spine in lateral bending and axial rotation (After Miles and Sullivan, 1961; Pearcy, 1985; Dvorak et al, 1991)

(1985) and the posteroinferior corner of the vertebral body in the other studies. The magnitude of translation in the various studies was thus expected to be different.

In general, gender and race were not found to have significant effect on sagittal mobility (Tanz, 1953; Allbrook, 1957; Hayes et al, 1989). Some authors reported that sagittal intervertebral mobility decreased with advancing age and most of the motion loss occurred between 13 and 35 years of age (Tanz, 1953; Allbrook, 1957). However, Hayes et al (1989) did not demonstrate such a difference.

Clinically, the magnitude of the anteroposterior translation associated with sagittal movements was used as an indicator of vertebral stability (Dupuis et al, 1985; Boden and Wiesel, 1990). However, Schaffer et al (1990) demonstrated that there was a high rate of misdiagnosis of spinal instability when the magnitude of translation was less than 5mm. In addition, the average range of the anteroposterior translation was about 2-3mm in normal spines (table 3.6). Stokes and Frymoyer (1987) showed that a large number of patients with degenerative instability based on clinical symptoms and radiological signs had anteroposterior translation of less than 3mm during flexion. These findings challenged the validity of evaluating the translational motion in the diagnosis of instability.

3.2.3 Lateral Bending and Axial Rotation

The mean ranges of movements in lateral bending and axial rotation as reported by previous radiographic studies are shown in table 3.7 (Miles and Sullivan, 1961; Pearcy, 1985; Dvorak et al, 1991a). Neither in lateral bending nor axial rotation were there significant differences between movements to the right or to the left. The lumbar motion segments generally exhibited very little axial rotation and there were no significant differences between segmental levels. In lateral bending, the lower lumbar segments appeared to be less mobile. Stereoradiographic studies showed that axial rotation and lateral bending were inherently coupled with each other (Pearcy and

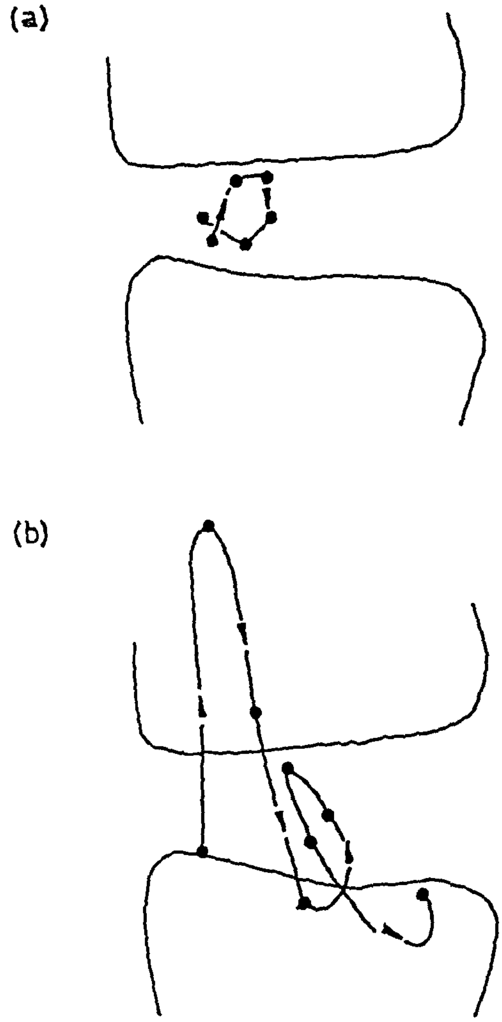


Figure 3.8
The centrode pattern in motion segments with (a) normal disc and (b) degenerated disc. (After Seligman et al, 1984)

	Anterior disc height		Posterior disc height	
	Mean change, mm	Percentage change	Mean change, mm	Percentage change
Flexion:				
L1/2	-2.0	-27	1.5	40
L2/3	-4.0	-35	2.5	60
L3/4	-4.5	-35	2.5	55
L4/5	-4.5	-31	2.5	54
L5/S1	-2.5	-20	1.5	34
Extension:				
L1/2	2.5	30	-1.0	-26
L2/3	1.5	13	-1.0	-20
L3/4	0.5	3	-0.5	-12
L4/5	0.5	6	-0.5	-13
L5/S1	2.0	18	-0.5	-13

Table 3.8
Changes in anterior and posterior disc heights at each spinal segment from the upright to the fully flexed and extended positions. (After Pearcy and Tibrewal, 1984a)

Tibrewal, 1984a; Pearcy, 1985), and this was in agreement with the observations made in cadaveric spines (see section 3.1.4).

3.2.4 Instantaneous Centres of Rotation

Previous authors had determined the locations of the centres of rotation in both cadaveric and living spines (Rolander, 1966; Cossette et al, 1971; Gertzbein et al, 1981 and 1984; Seligman et al, 1984; Ogston et al, 1986; Pearcy and Bogduk, 1988). They generally observed that for flexion, the centre of rotation was normally in the posterior part of the disc and for extension, in the anterior part. In lateral bending, the centre of rotation was found to fall in the region of the disc which was opposite to the side of the movement, whereas in axial rotation, it was normally in the posterior part of the nucleus and tended to move towards the side to which rotation occurred.

Degenerated discs were found to have a very wide spread of the centres of the rotation, that is, an increase in the length of the centrode (the locus of the centres of rotation) (figure 3.8) (Rolander, 1966; Seligman et al, 1984). Ogston et al (1986) demonstrated that the in vivo centrode length was longer than that reported for cadaveric specimens. This was probably due to the effects of body weight and muscle actions which would compress the spines and lengthen the centrode.

3.2.5 Changes in Disc Heights During Lumbar Spinal Movements

Pearcy and Tibrewal (1984a) used the method of Farfan (1973) (see section 3.2.1.4) to measure the anterior and posterior disc heights of normal subjects in the neutral upright position, in full flexion and extension. The results of the study are summarised in table 3.8.

In flexion, there were decreases in anterior disc heights suggesting that the anterior annulus was compressed. The posterior disc heights were found to increase by over 30%. In vitro tensile tests on specimens from annulus fibrosus indicated that they failed between 25% and 30% (Galante, 1967). Thus, if the posterior annulus was

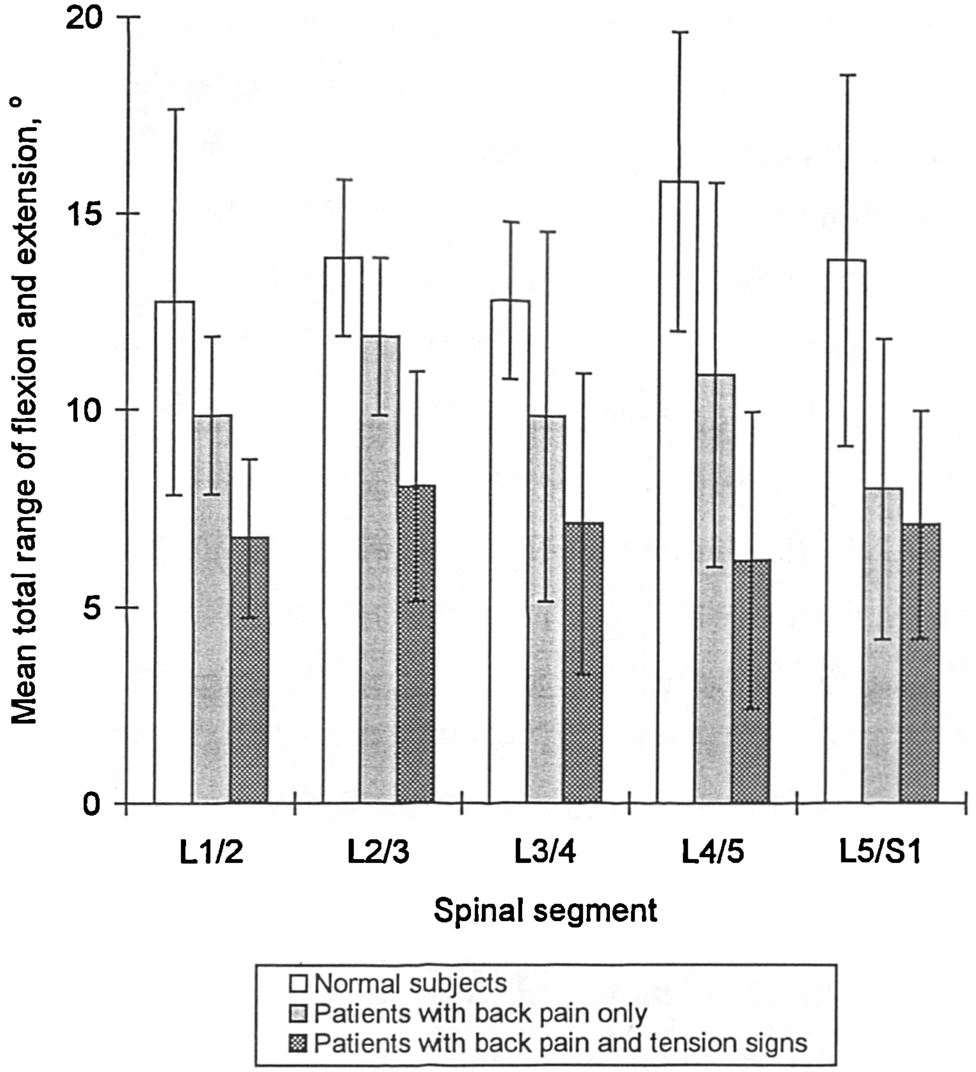


Figure 3.9
The mean total range of flexion and extension at each spinal segment for normals, and patients with back pain only and those with tension signs. (The vertical bars denote ± 1 standard deviation.) (After Pearcy et al, 1985)

stretched by more than 30% during normal flexion, these fibres would be ruptured. This implied that in the upright position, the posterior annulus was in a state of compression.

In extension, the posterior disc heights decreased and thus the posterior annulus was compressed. The anterior disc heights rarely increased by more than 20% (except at L1/2). The anterior annular fibres were thus not stretched to the failure point found in the in vitro tests.

3.2.6 Radiographic Studies of the Effects of Low Back Pain on Lumbar Segmental Mobility

Stokes et al (1980) demonstrated that spinal joints with herniated nucleus pulposus had unequal magnitude of lateral bending to both sides. This degree of asymmetry was significantly higher than that of normal spinal joints. It was believed that the asymmetry was the result of an avoidance of movement to one side which tended to increase pressure on the affected nerve root. Stokes et al (1981) also showed asymmetry of movements of the zygapophyseal joints (defined as the vertical linear motion between each superior facet tip and an adjacent point on the corresponding inferior facet) in patients with disc herniation. Two types of asymmetry were observed. In one type, flexion accompanied lateral bending of the joint to one side. This was thought to prevent narrowing of the intervertebral foramen on that side. In the other type, the painful joint was "splinted" so that no measurable motion occurred in it.

Previous studies had also measured spinal movements in back patients with non-specific diagnosis (Pearcy et al, 1985; Dvorak et al, 1991b). They showed that the intervertebral joints of these patients generally showed hypomobility (figure 3.9). Muscles were probably splinting the joints to reduce or prevent movement. It was also revealed that the range of sagittal movements of patients with neural tension signs was less than that of patients with back pain only (figure 3.9). Interestingly, segmental hypomobility was observed not only at the specific pathological level but throughout the spine (Dvorak et al, 1991b). It appeared that the analysis of segmental motion using flexion/extension radiographs did not aid in diagnosing the level of lesion.

In normal spines, flexion and extension movements were confined to the sagittal plane and only coupled with anteroposterior translation (section 3.2.2). However, sagittal movements were found to be coupled with axial rotation and lateral bending in patients with back pain (Pearcy et al, 1985). It was believed that the occurrence of these coupled movements might be due to unilateral or asymmetrical involvement of the ligaments or muscles when the patient flexed or extended. It was also shown that patients with tension signs had less coupled movements compared to patients with back pain alone (Pearcy et al, 1985). This suggested that the ligaments or muscles were not involved to a great extent.

In conclusion, there were only a few radiographic studies on the effects of back pain on spinal segmental mobility. They generally examined only a few pathologies and movements. Stokes et al (1981) examined only the lateral bending movement whereas Pearcy et al (1985) and Dvorak et al (1991a) studied only flexion and extension. The sample sizes were generally small, ranging from 6 to 23 for each patient group. Further studies on the effects of back pain on spinal segmental mobility are deemed necessary. They are essential to the practice of spinal manual therapy which is often involved in assessing patients' joint mobility and restoring their movements.

■ *Chapter 4*

Previous Studies of Spinal Manual Therapy

Restore vertebrae to normal position
Straighten the spine
Relieve interference with blood flow
Relieve nerve compression
Relieve irritation of sympathetic chain
Mobilise fixed motion segments
Shift a fragment of intervertebral disc
Mobilise posterior joints
Remove interference with cerebrospinal fluid circulation
Stretch contracted muscles, causing relaxation
Correct abdominal somatovisceral reflexes
Remove irritable spinal lesions
Stretching or tearing of adhesions around the nerve root
Reduce distortion of the annulus

Table 4.1
Proposed mechanisms underlying spinal mobilisation. (After Zusman, 1986)

Previous Studies of Spinal Manual Therapy

There have been numerous publications on the hypotheses of mobilisation, traction and other manual techniques, and on the clinical efficacy of these procedures. However, most of the hypotheses lack scientific evidence and much of the published research is flawed by poor experimental design and sub-optimal outcome measures. The present chapter is a critical review of the available literature on spinal manual therapy. The deficiencies in the previous works are highlighted. The review is focused on mobilisation and traction therapy of the lumbar spine which are the subjects of the present investigation.

4.1 SPINAL MOBILISATION

4.1.1 Mechanisms Underlying Spinal Mobilisation

Table 4.1 summarises the various mechanical and neurophysiological hypotheses underlying spinal mobilisation. They are described in detail below.

4.1.1.1 Mechanical mechanisms

Laboratory experiments on rabbit tendons and ligaments showed that immobilisation was deleterious to the healing process (Frank et al, 1984; Woo et al, 1985). Early mobilisation was found to promote cellular activity at the repair site with increased collagen content, improved fibre alignment and decreased scar adherence to surrounding tissues. The strength and stiffness of the early mobilised tissues were also significantly higher than those that were immobilised. Thus it is believed that passive movements such as those produced during mobilisation treatment is essential in stimulating healing and preventing joint contracture (Frank et al, 1984; Woo et al, 1985).

After injury, scarring may occur between the different lamellae of the disc or between moving bundles of the capsules or other soft tissues of the spine (Farfan, 1980; Grover, 1982). Scar tissues are stiffer than normal collagen fibres (Evans, 1973; Farfan, 1980), and thus the joint becomes hypomobile with a shortened toe phase in the load-deformation curve.

Paris (1979) believed that mobilisation movements which are carried out at or to the limit of the joint's available range will produce therapeutic effects. The movements will stretch the tissues by taking them into the area of plastic or permanent deformation of scar tissues (Grover, 1982; Zusman, 1986). The scar tissues may also be sufficiently stretched to its failure limit, but the mobilisation force will not damage the normal collagen fibres as they are stronger than scar tissues (Farfan, 1980). The above effects on the scar tissues will lead to a restoration of joint mobility with an elongation of the toe phase in the load-deformation curve.

The mechanical properties of spinal soft tissues are time-dependent, and mobilisation has been shown to exhibit creep and preconditioning, that is, increase in deformation with sustained or repetitive loadings (Lee, 1990; Lee and Evans, 1991, 1992 and 1994) (see section 4.1.4). It was suggested that these behaviours might explain the improvement in range of movement following mobilisation treatment with the force applied in a sustained or oscillatory manner.

In normal living tissues, the effects of creep and preconditioning were shown to be temporary (Daly, 1966; Kazarian, 1975), that is, over a period of time following load removal, the tissues returned to their original dimensions. Therefore, these mechanisms do not fully explain the clinical improvement in patient's mobility after treatment which is often found to be permanent. Other mechanisms, such as permanent deformation or failure of scar tissues as discussed above, may operate at the same time.

Intense repetitive oscillations of spinal joints, in the face of clinical improvement, have been advocated for patients with documented evidence of disc prolapse and neurological deficit (Corrigan and Maitland, 1983). However, Farfan (1973) showed that rotatory manipulative techniques did not affect the appearance of disc protrusion as seen in a myelogram. He also found that manipulative techniques caused enlargement of the defect of the disc occasionally, and some 30 to 50% of patients experienced relief of symptoms even though the myelographic defect was unaltered in appearance. Corrigan and Maitland (1983) acknowledged that what was actually being achieved was uncertain and that prolapsed disc material could not be returned to its original site.

The "correction" of spinal joint "subluxations" which are said to have compromised spinal neurovascular structures is another mechanical mechanism that has been attributed to spinal mobilisation. However, it has not been shown that spinal joint subluxation is more common in patients with spinal pain compared to pain-free individuals (Zusman, 1986).

4.1.1.2 Neurophysiological mechanisms

Paris (1979) and Wyke (1985) postulated that repetitive oscillatory mobilisation movements would stimulate the mechanoreceptors in the zygapophysial joint capsules causing reflexogenic and pain suppression effects. Wyke (1985) suggested that the reflexogenic effect was one of reciprocally coordinated inhibition of the muscle tone and stimulation of the stretch reflexes in the muscles. This resulted in a reduction in muscle spasm and therefore an improvement in the range of movement and reduction of pain. In addition, the joint capsules are supplied by branches from at least two spinal levels (see section 2.5). Mobilisation of an individual joint therefore not only affects motor activity in the muscle operating over the joint being mobilised but also in more remote muscles.

In a group of back patients and normal subjects, Shambaugh (1987) studied the changes in the electromyographic activities of erector spinae (in the thoracic and lumbar regions) and trapezius muscle groups following chiropractic manoeuvres. The vertebrae T1, T3, T5, L1 and L3 were subjected to manipulative "adjustments". Compared with the control group, there were significant reductions (an average reduction of 25%) in the integrated values of the electromyographic signals of all the muscle groups. There was also a strong correlation between the number of treatments and the reduction in muscle tension. These clinical results provided some experimental support to the hypothesis put forward by Paris (1979) and Wyke (1985).

The gate control theory hypothesised by Melzack and Wall (1965) provided another explanation of the pain suppression effect of spinal mobilisation. The stimulation of large-fibre mechanoreceptors causes a rapid and voluminous transmission of impulses that effectively closes the gate to transmission of small-fibre nociceptive afferents. This most probably happens at the spinal cord level, but is also influenced by higher levels in the central nervous system (Paris, 1979). The closure of the gate blocks the perception of pain, which possibly results in an increase in the range of movement.

There has also been a suggestion that there is a hormonal role in mediating the analgesic response to this kind of therapy. Spinal mobilisation may act as a stressor that activates the hypothalamo-pituitary-adrenal axis to secrete β-endorphins which has a profound analgesic effect. Christian et al (1988) examined this mechanism in back patients and normal subjects as control groups. The plasma levels of β-endorphins, adrenocorticotropic hormone (hormone secreted by the pituitary gland during stress that acts on the adrenal cortex to release its hormones, especially cortisol; commonly abbreviated as ACTH) and cortisol levels were determined before and after spinal manipulative procedures. All patients subjectively reported marked relief of pain after treatment. However, the ACTH and β-endorphins levels showed no change and the cortisol level was found to fall slightly. These findings suggested that

Study	Sample characteristics (sample size)	Treatment group	Control group	Results
Glover et al (1974)	unilateral back and/or leg pain; 2 groups: one with pain less than 7 days and the other more than 7 days; between 16 and 64 years of age (n=87)	rotation technique and detuned short wave diathermy treatment for 4 days	detuned short wave diathermy	Pain relief was greater in the manipulated group 15 minutes after treatment, but at 3 and 7 days and 1 month follow up, both groups showed similar relief of pain.
Doran & Newell (1975)	back/leg pain with painful limitation of movement, aged 20-50 years (n=456)	three groups: (i) mobilisat-ion, (ii) conventional physio-therapy and (iii) corset; treatment given for 4 weeks	analgesics and postural advice	no significant differences between the 4 groups initially, at 3, 6, 12 week and 1 year follow-up
Sims-Williams et al (1978)	non-specific back pain, patients in general practice, aged 20-65 years (n=94)	PA, transverse or rotational mobilisation, manipulation, and traction treatment for 4 weeks	Microwave radiation at the lowest possible setting	Treatment group showed greater improvement in pain, spinal mobility and straight leg raise; but at 12 month follow up, the 2 groups were identical
Sims-Williams et al (1979)	as above except that patients were those referred to rheumatology clinics (n=94)	as above	as above	no differences between the 2 groups immediately after treatment and at 1, 3 and 12 month follow up

Table 4.2
A summary of previous clinical trials of spinal mobilisation.
(to be continued on next page)

Study	Sample characteristics (sample size)	Treatment group	Control group	Results
Coxhead et al (1981)	sciatic pain at least as far as the buttock crease; average age was 42 years and average duration of symptoms 14.3 weeks (n=334)	Factorial designs: 4 treatments (mobilisation, traction, exercise and corset); and thus 16 groups altogether, including the control with no treatment. Patients in all groups received short wave diathermy and back instruction. Treatments were given for 3 weeks.		Each of the treatments, particularly mobilisaton, showed a small degree of benefit over the spontaneously improvement rate. Significant increase in improvement with increase in number of treatment. But there were no beneficial effects of treatment at 4 and 16 months
Zylbergold and Piper (1981)	back patients in out-patient department, 25-65 years of age (n=28)	2 groups: (i) PA or rotational mobilisation and traction (ii) exercise; treatment period was one month	No active treatment, back care instruction only	No differences between the 3 groups in terms of pain, spinal mobility and functional activitiy
Nwuga (1982)	patients with diagnosis of prolapsed intervertebral disc; all females with age between 20-40 (n=51)	rotational mobilisation treatment until pain was no longer present	short wave treatment, exercises, back care instruction	The treatment group showed greater improvement in spinal movements and straight leg raise 4 weeks after treatment, required less treatment time and had less number of patients returning for treatment after 3 months.

Table 4.2
A summary of previous clinical trials of spinal mobilisation.
(to be continued on next page)

Study	Sample characteristics (sample size)	Treatment group	Control group	Results
Farrell and Twomey (1983)	patients with painful movement and straight leg raise; mean age was 42 years; had symptoms of 3 week duration or less (n=48)	PA, transverse or rotational mobilisation treatment for 3 weeks	microwave treatment, exercises, back care instruction	Mobilisation group showed less number of days to become symptom free. There were no difference between the 2 groups in spinal mobility improve-ment at the end of 3 week treatment
Gibson et al (1985)	back pain of more than 2 months but less than 12 months duration; mean age of 36 years (n=109)	2 groups: (i) osteopathic treatment, (ii) short wave diathermy; duration of treatment was 3 weeks	detuned short wave diathermy	Reduction of pain and improvement of spinal movement were similar in the 3 groups.
Hadler et al (1987)	first attack of back pain for less than 1 month, aged 18-40 (n=44)	rotational mobilisation and high velocity thrust techniques	no control group	In the 1st week following treatment, patients treated with high velocity thrusts were found to improve to a greater degree and more rapidly than those treated with mobilisation.
Mathews et al (1988)	2 groups: (i) back pain only, (ii) back pain and limited straight leg raise, 18-60 years of age, pain of less than 3 months (n=434)	rotational or PA mobilisation	infra-red treatment	In all the treated groups there was hastening of pain relief over the control within the first 2 weeks, and this was most maked for patients with limited straight leg raise.

Table 4.2
A summary of previous clinical trials of spinal mobilisation.

spinal manual therapy was not sufficiently stressful to activate the hypothalamo-pituitary-adrenal axis.

Zusman (1986) documented that besides the above-mentioned neurophysiological effects, repetitive oscillatory mobilisation treatments would cause mechanical adaptation (hysteresis) of the joint receptors. The resulting temporary decrease in the peripheral input might be responsible for the improvement in pain and range of movement.

4.1.2 Clinical Trials

The clinical efficacy of spinal mobilisation in the treatment of back pain had been extensively evaluated by previous studies, the results of which are summarised in table 4.2. Some studies reported that there were no differences in the outcome measures between patients treated with mobilisation or manipulation and those untreated or treated with other conservative methods (Doran and Newell, 1975; Sims-Williams et al, 1979; Coxhead et al, 1981; Zylbergold and Piper, 1981; Gibson et al, 1985). Other authors reported positive findings (Glover et al, 1974; Sims-Williams et al, 1978; Nwuga, 1982; Farrell and Twomey, 1983; Hadler, 1987; Mathews et al, 1988). The more long term studies (Glover et al, 1974; Sims-Williams et al, 1978) revealed that spinal mobilisation only led to greater improvement immediately after treatment. Long term follow up (from 3 days to 12 months) did not reveal any persistent benefits over the control group.

Table 4.2 shows that there are large variations among the different studies in the sample characteristics, research methods and outcome measures employed. This explains the different results reported by the studies. An interesting difference in results was noted in the two studies carried out by Sims-Williams et al (1978 and 1979). In the earlier study, short term clinical improvement after mobilisation treatment was observed in back patients in general practice. However, in the later work which was carried out in hospital patients, no definitive advantage was revealed.

The research methods and outcome measures employed in the two studies were identical, and the difference in the clinical findings was due to the difference in sample characteristics. Patients referred to specialist clinics in hospitals would have a longer duration and more severe symptoms that those cared for by the general practitioners, and might be less likely to benefit from mobilisation treatment. The different results in the two studies emphasised the importance of defining the population of patients in conducting any clinical trials.

The discrepancies in clinical observations about the efficacy of mobilisation are complicated by the sub-optimal research designs of the clinical trials reviewed. Although random allocation of patients into the treatment and control groups was found in most studies (Glover et al, 1974; Sims-Williams et al, 1978 and 1979; Zylbergold and Piper, 1981; Farrell and Twomey, 1983; Gibson et al, 1985; Matthews et al, 1988), the samples were often not homogeneous. Glover et al (1974) attempted to solve the problem by dividing the sample into subgroups according to the duration of symptoms and the number of attacks. Some studies employed statistical methods to demonstrate that the treatment and control groups were similar (Doran and Newell, 1975; Farrell and Twomey, 1983; Zylbergold and Piper, 1981; Nwuga, 1982).

All the trials reviewed related only to short courses of treatment. The duration of treatment was often restricted to a few days and generally not more than 4 weeks. None of the studies were double blind trials. This was difficult because patients were aware of whether they had manual therapy or not, and Doran and Newell (1975) reported that the assessing physician inadvertently discovered the treatment in about 10% of the cases. In addition, a proper control was essential in clinical trials because of the natural resolution of back pain and the placebo effect associated with the treatment (Difabio, 1986; Koes et al, 1991). In the study of Hadler et al (1987), no control group was employed. Although control groups were employed in many other studies, they were generally improper (Doran and Newell, 1975; Nwuga, 1982; Farrell and Twomey, 1983, Mathews et al, 1988). The so called placebo treatment consisted of analgesics or thermal therapy which might also produce therapeutic effects.

There was also generally a lack of standardisation of the treatment methods. Patients were often treated by different combinations of mobilisation techniques. The choice of techniques was at the discretion of the therapists and would vary from one patient to another. Manual techniques were also sometimes combined with traction and thermal therapy. These made the determination of the clinical efficacy of manual techniques difficult. The efficacy and indications of a particular technique such as posteroanterior mobilisation was not examined.

Literature producing negative results had been rebutted by proponents of mobilisation and manipulation who criticised that the therapist providing the treatment did not have adequate skills (Ottenbacher and Difabio, 1985). None of the studies reviewed in the present survey reported the qualification and experience of the manual therapist providing the treatment.

Another deficiency of previous clinical trials was that there was a lack of objective outcome measures. Most of the works were based on subjective pain relief reported by the patients either orally or in questionnaires (Glover et al, 1974; Doran and Newell, 1975; Coxhead et al, 1981; Hadler et al, 1987; Mathews et al, 1988). A few studies had employed simple objective criteria such as improvement in spinal mobility and straight leg raise (Sims-Williams 1978 and 1979; Zylbergold and Piper, 1981; Nwuga, 1982; Farrell and Twomey, 1983; Gibson et al, 1985). Functional outcome measures were reported only in the study of Zylbergold and Piper, 1981).

None of the clinical trials reviewed proves to be wholly satisfactory. In a review conducted by Koes et al (1991), previous clinical trials were scored according to the principles of intervention research which included sampling method, treatment intervention, outcome measures and data analysis. None of the trials was found to score more than 60 points (maximum score 100), suggesting that they were generally of poor quality. The authors identified similar methodological problems as discussed above.

Ottenbacher and Difabio (1985) also performed a "quantitative" review of nine clinical trials of manual treatments. They only included trials with control comparisons and appropriate outcome measures and those which reported statistical results in sufficient details. Statistical indices were calculated to assess the efficacy of mobilisation and manipulation. The results indicated that the beneficial effects of manual treatment were observed only when it was provided in conjunction with other forms of treatment and when the treatment effects were measured immediately following therapy.

Despite the deficiencies identified in previous clinical trials, it may be generally concluded that in the short term, spinal mobilisation appears to hasten the rate of improvement in pain, spinal mobility and functional impairment. There is, however, still much controversy over the long term effect of the treatment, the efficacy of a particular mobilisation technique, the indication and duration of treatment.

4.1.3 Reliability of Clinical Manual Examination

Manual examination of segmental compliance of the spine is often performed by therapists using mobilisation techniques such as posteroanterior pressure. The objective is to identify the painfully stiff motion segment by perceiving subjectively the intervertebral movement produced by the applied mobilisation force (Maitland, 1986).

The reliability of manual evaluation of intervertebral motion was examined by many previous authors (Gonnella et al 1982; Jull and Bullock, 1987; Phillips and Twomey, 1993; Maher and Adams, 1994). In the study of Gonnella et al (1982), segmental mobilities of 5 subjects were assessed manually by 5 therapists, and graded according to an ordinal rating scale of 13 grades. The authors performed only descriptive reliability analysis. They found that the intertherapist reliability was reasonably good, but consistency among therapists was not demonstrated. They also reported that the L5/S1 had the greatest variability, probably because of the difficulty in palpating the S1 spinous process.

Jull and Bullock (1987) further evaluated the reliability of manual examination with a larger sample size (n=20) and statistical treatment of data. The posteroanterior mobility of the lumbar spine was examined manually on 2 different days for all physiological movements and posteroanterior mobility. Motion was rated on a five point rating scale. Suprisingly good intra- (r=0.98) and inter-examiner (r=0.94) reliability was found. They obtained a much better result than Gonnella et al (1982). This was probably because they had used a much simpler grading system. This might indicate a drop in reliability as the task complexity increased.

More recently, Phillips and Twomey (1993) showed that manual examination was unreliable when no verbal communication was allowed between the patient and the therapist. In their study, the segmental level which the therapists felt was responsible for the patients' pain was compared with the symptomatic level determined by spinal block procedure. It was found that there was 100% agreement between palpation findings and spinal block procedures when patients were allowed to report the pain response during palpation. However, the agreement fell to 60% when no verbal pain reponse was allowed.

Maher and Adams (1994) also demonstrated that the pain response could be more reliably assessed than segmental mobility. Manual examination was carried out by 6 manual therapists with a minimum of 5 years of experience. The segmental stiffness and pain response of back patients were assessed by rating scales. The reliability of stiffness palpation was low, with intraclass correlation coefficient ranging from 0.03 to 0.37 and agreement scores ranging from 21% to 29%. The intraclass correlation coefficient values of pain judgement ranged from 0.67 to 0.72, with agreement scores ranging from 31% to 43%.

It may be concluded from the above studies that manual examination of intervertebral mobility is unreliable, particularly when a complicated rating scale is employed. The level of lesion cannot be accurately identified by palpation of mobility and the patient's pain response appears to be a useful criterion in clinical diagnosis or decision making.

"Movement diagrams" are often drawn by therapists to depict joint behaviour during clinical manual examination (Maitland, 1986). This involves the palpation of the mobilisation force applied and the intervertebral motion so produced. This is a much more complicated task than assessment of segmental mobility using a rating scale. The intervertebral motion produced by mobilisation is probably small, in terms of a few mm or degrees, and it is highly improbable that the therapist can perceive the motion (Jull, 1987). Similarly, the quantity of tissue resistance and the mobilisation load applied at various stages in the range are also difficult to feel. The reliability of constructing the movement diagram is thus doubtful.

Trott et al (1989) attempted to examine the reliability of constructing the movement diagram by assessing the ability of therapists to detect changes in force and displacement. Therapists were asked to apply vertical force to a "palpation simulator" which started to resist the movement with a linear stiffness at a certain point (R_1). They then attempted to detect a change of stiffness of the simulator at another predetermined point (R_{1a}). It was shown that the error of palpation, that is, the distance between the perceived and true positions of R_{1a}, ranged from 0.39mm to 1.47mm. This error size was considered to be large compared with the magnitude of movements produced by mobilisation. The accuracy of locating R_{1a} was found to decrease when the distance between R_1 and R_{1a} was small (less than 1 mm) and when the change of stiffness was small (the ratio of stiffness in the second region to that in the initial region was less than 2.0).

It should be noted that the experimental condition of Trott et al (1989) was artificial. The palpation simulator had only two linear stiffness regions and there was a sudden change of stiffness from one linear region to another. However, the load-deformation characteristic of soft tissues is known to be non-linear and stiffness

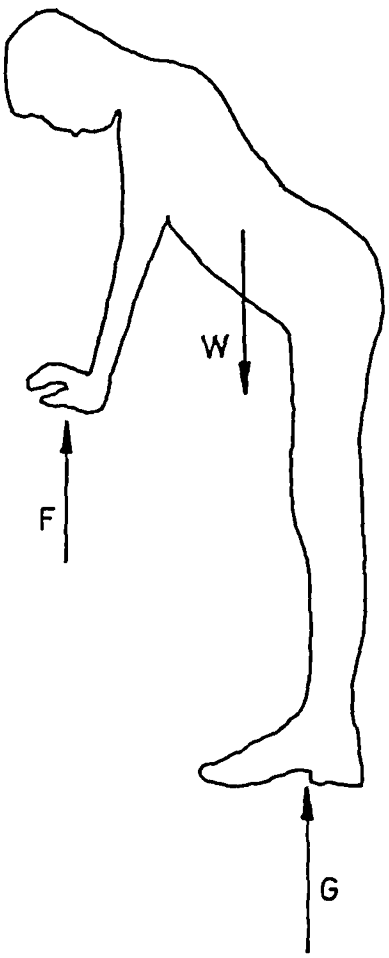


Figure 4.1
Vertical forces acting on the therapist while performing posteroanterior mobilisation (inertial force not shown). (After Matyas and Bach, 1985)
(F-mobilisation force, W-weight of therapist, G-ground reaction force)

changes gradually and continuously until it reaches the linear phase (see section 2.4.1). Detection of stiffness changes in living subjects would therefore be even more difficult and unreliable.

In view of the poor reliability and uncertainty of manual examination, instrumentation has to be developed to objectively quantify the mobilisation force applied and the movement so produced. This allows accurate assessment of the load-deformation characteristics or the so called "movement diagram". There were several attempts to measure these variables experimentally and they are described in the following section.

4.1.4 Previous Biomechanical Studies

Figar and Krausova (1975) used a pressure transducer to measure the posteroanterior mobilisation force delivered to cervical motion segments, and it was probable that such a method could also be used for the lumbar spine. Matyas and Bach (1985) suggested another simple procedure to measure the posteroanterior force applied. The therapist performs mobilisation of the lumbar spine of a patient while standing on forceplate. Figure 4.1 is a free body diagram showing the vertical forces acting on the therapist in such situation. If acceleration of the centre of gravity of the therapist is ignored, the mobilisation force F is given by

$$F = W - G$$

where W is the weight of the therapist and G the ground reaction force measured by the forceplate.

Watson et al (1989) attempted to measure the spinal motion produced by posteroanterior mobilisation using a video motion analysis system. The therapist's thumb was fitted with a reflective marker, the movement of which was detected by the system. The movement of the thumb was used to estimate the movement of the spine. The authors assessed only one cervical and one lumbar level on one normal subject.

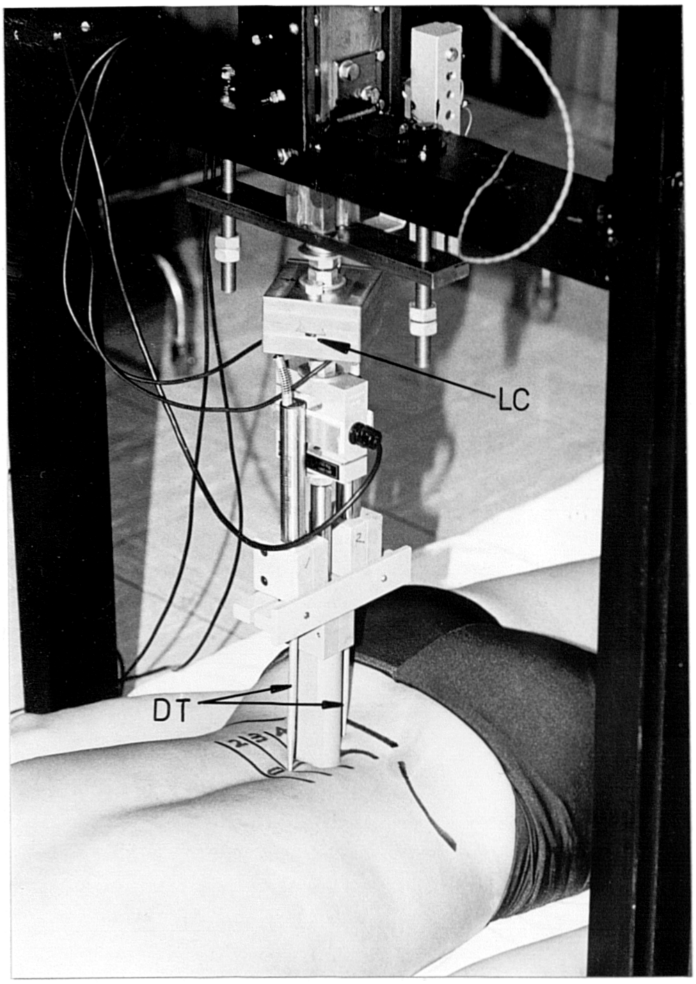


Figure 4.2
Instrument developed for measuring the load-deformation characteristics of posteroanterior mobilisation. (After Lee, 1990)
(LC=load cell; DT=displacement transducer)

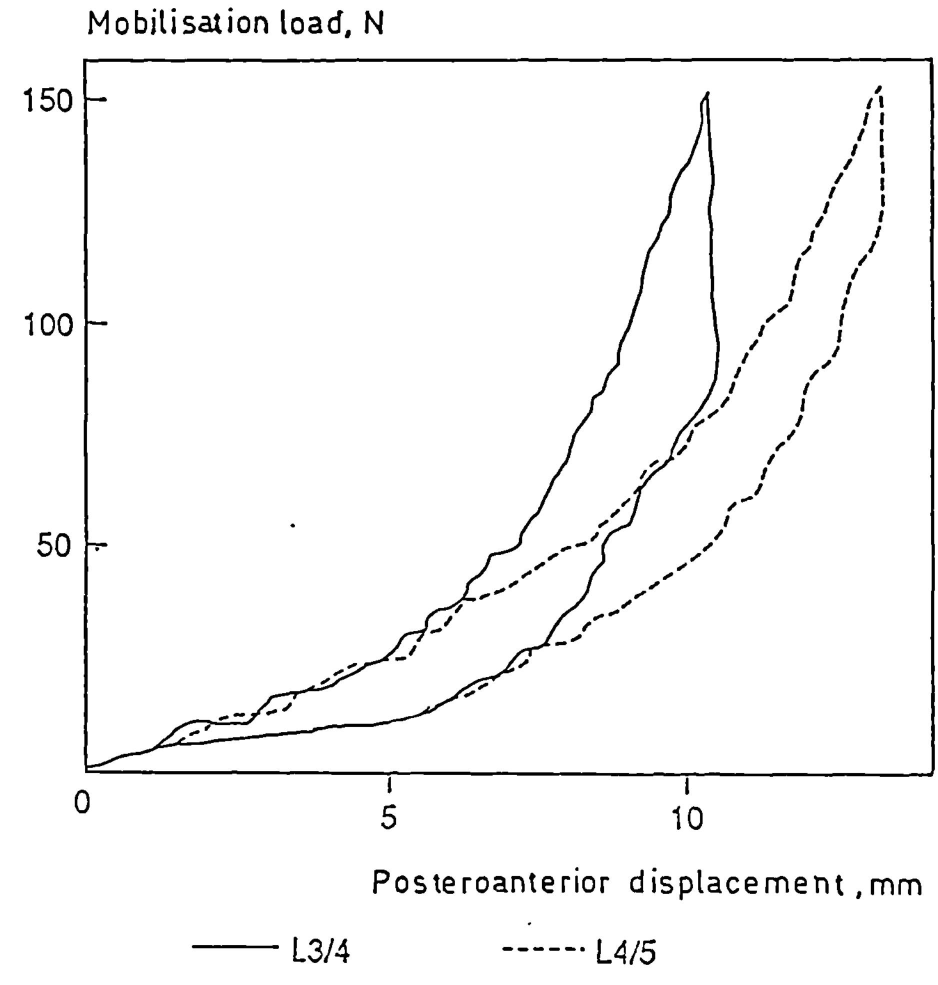


Figure 4.3
Load-displacement curves for posteroanterior mobilisation of L4 obtained from a 23 year old normal man. (After Lee, 1990; Lee and Evans, 1991 and 1992)

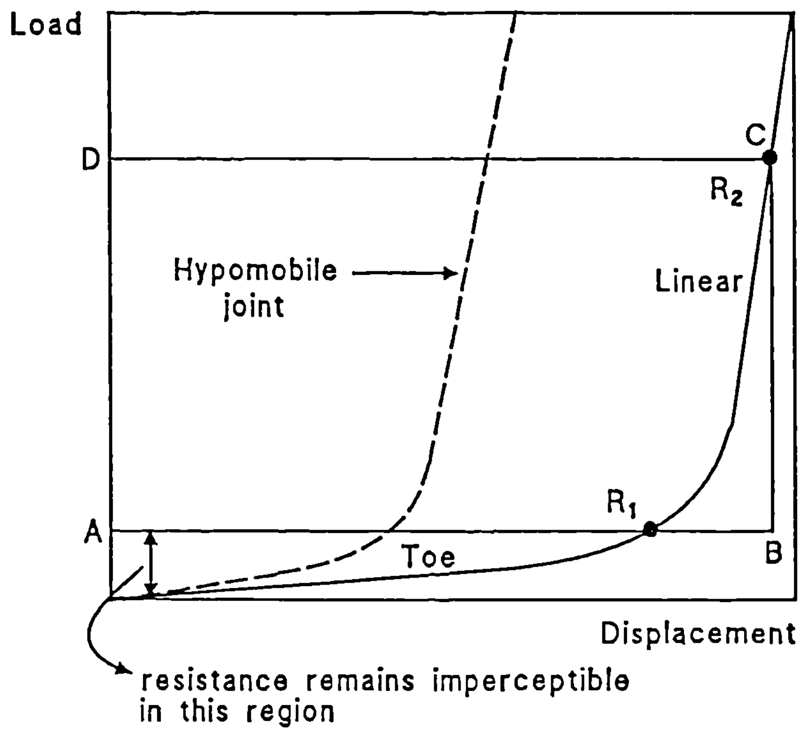


Figure 4.4
Relationship of movement diagram (ABCD) to the load-displacement curve. (After Lee and Evans, 1994)
(R_1 - the point where resistance is first felt; R_2 - the maximum resistance that the therapist is prepared to "push through")

The study was thus very preliminary, and the reliability of the measurement method uncertain.

The above studies measured either the mobilisation load applied or the displacement produced. However, both variables have to be measured simultaneously if the movement diagram is to be constructed. A number of researchers had developed instrumentation to quantify the load-deformation characteristic of posteroanterior mobilisation (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and Evans, 1991 and 1992). The instrument generally consisted of an applicator which was centred over the spinous process of the vertebra to be mobilised and fitted with a load cell to measure the posteroanterior force applied (figure 4.2). Displacement transducers were employed to record the posteroanterior displacements produced, that is, the vertical displacements of the skin overlying the spinous processes of the two adjacent vertebrae relative to that of the mobilised vertebra. Lee (1990) reported that such system provided highly repeatable data (r=0.98 in a test-retest study) and for mobilisation of L4, the maximum error in recording the posteroanterior displacements at L3/4 and L4/5 were ±0.7mm and ±0.8mm respectively.

It was found that the load-deformation characteristic of PA mobilisation was always non-linear (Lee, 1990; Lee and Evans, 1991 and 1992). Figure 4.3 shows a typical load-displacement curve obtained from a normal subject during a posteroanterior loading of L4. There are two curves in the figure, one representing the segment above (L3/4) and the other the segment below (L4/5). The loading and unloading curves for the two segments are different exhibiting hysteresis. In addition, the displacement at L4/5 is larger than that at L3/4 for a give mobilisation force.

Lee and Evans (1994) documented that movement diagrams are related to the load-deformation curves. It can be seen that the resistance curve of the movement diagram (ABCD) is essentially a part of the load-deformation curve (figure 4.4). Stiffness in the toe region is minimal and may remain imperceptible to the therapist. R_1, where resistance is first felt, is probably the transition point between the toe and

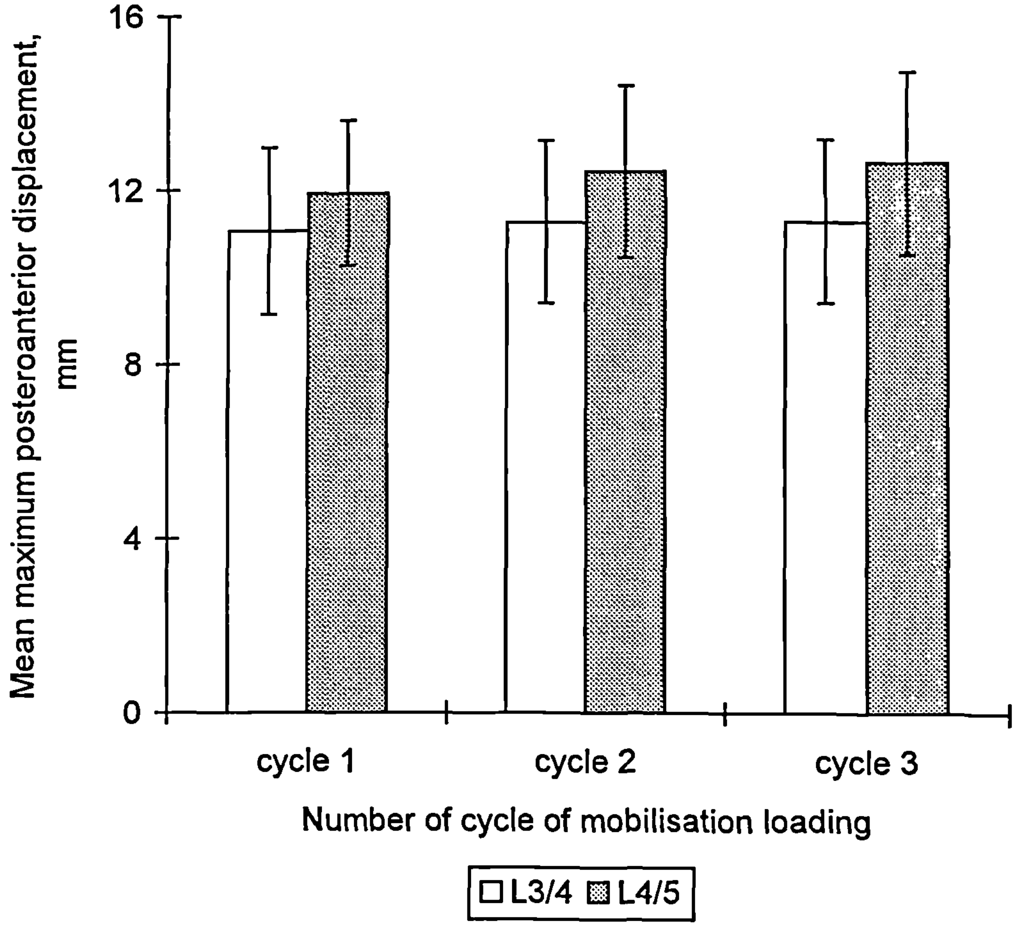


Figure 4.5
Mean maximum posteroanterior displacements at L3/4 and L4/5 during cyclic mobilisation loading of L4. (n=28) (After Lee, 1990; Lee and Evans, 1992) (The vertical bars denote ± 1 standard deviation.)

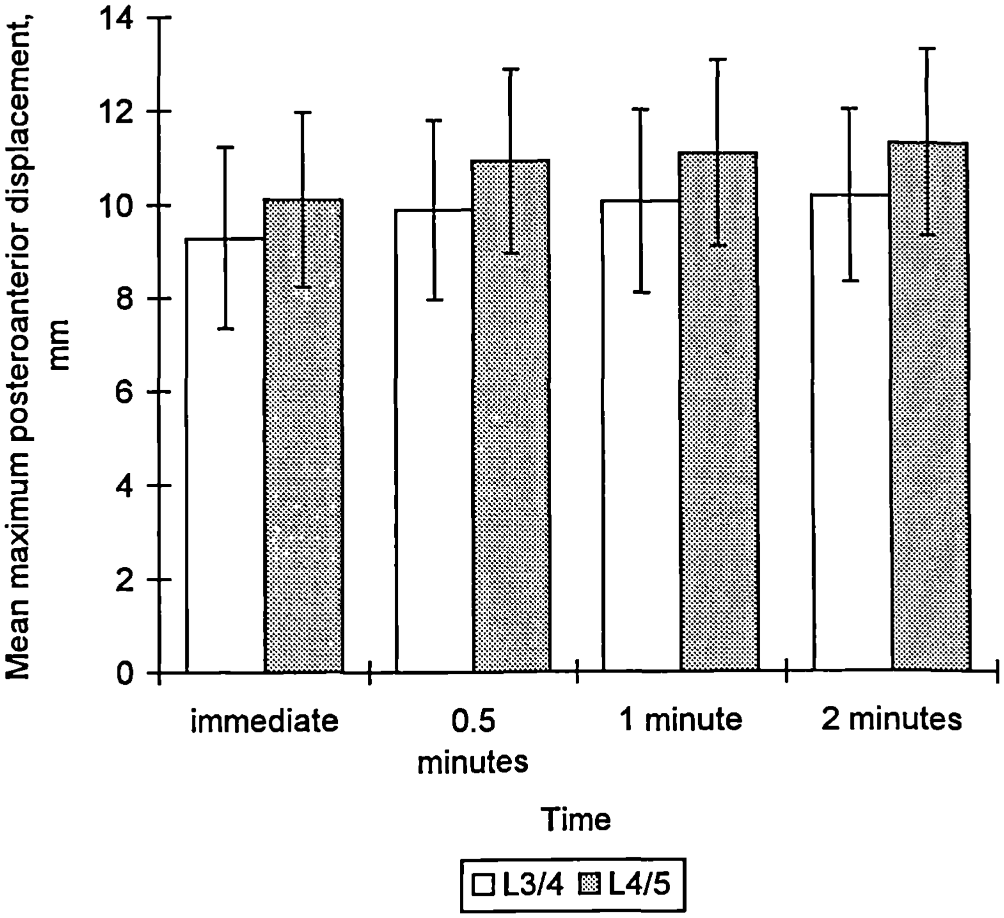


Figure 4.6
Effect of sustained mobilisation loading of L4 on the mean posteroanterior displacements at L3/4 and L4/5. (n=28) (After Lee, 1990; Lee and Evans, 1992) (The vertical bars denote ± 1 standard deviation.)

linear regions, where the abrupt change of resistance can be readily perceived by the therapist. R_2 is the maximum resistance that the therapist is prepared to "push through" and is well below the failure point on the load-deformation curve. Figure 4.4 also shows the resistance curve for a hypomobile joint. The toe phase is shortened with the point R_1 shifted to the left. In other words, strong resistance is felt by the therapist in the early part of the range.

It was shown that the load-displacement characteristics of posteroanterior mobilisation could be altered by repetitive and sustained loadings (Lee, 1990; Lee and Evans, 1991 and 1992). When the L4 vertebra was subjected to cyclic application of mobilisation force, significant increases in posteroanterior displacements at the L3/4 and L4/5 segments were observed with each loading cycle (figure 4.5). Such preconditioning effect decreased with each loading cycle. Posteroanterior mobilisation was also shown to exhibit creep during sustained loading of L4 vertebra (figure 4.6). The creep rate fell with time, and in fact, most of the observable creep effect occurred in the first half minutes. These findings were consistent with those of previous mechanical studies which showed that the mechanical properties of spinal soft tissues were time-dependent (Kazarian, 1975; Keller et al, 1987; Twomey and Taylor, 1982).

The preconditioning and creep effects were believed to be one of the mechanical hypotheses underlying spinal mobilisation (see section 4.1.1.1). It might account for the improvement in posteroanterior mobility and in the ranges of active physiological movements which were often observed immediately after mobilisation treatment (Lee, 1990; Lee and Evans, 1991 and 1992; Lee, 1994).

It should be noted that the additional effects of preconditioning and creep decrease with each loading, or with time, and finally reach a steady state. Thus the rate of improvement with subsequent mobilisation would be expected to slow down. When improvement had effectively halted, the therapist should consider increasing the mobilisation force applied, so that further preconditioning or creep was allowed to occur.

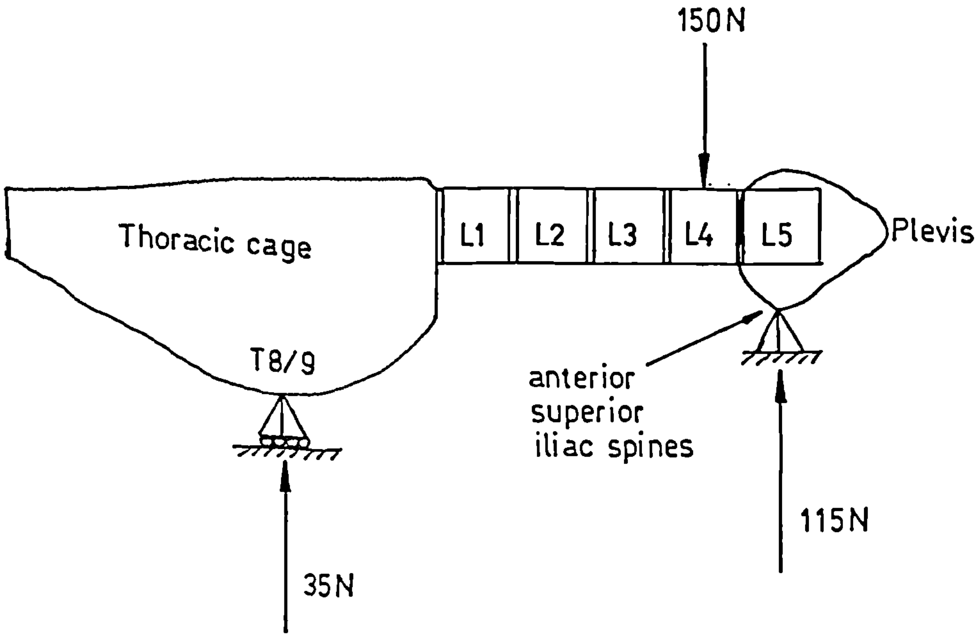


Figure 4.7
Posteroanterior mobilisation as three point bending of the lumbar spine. (After Lee, 1990)

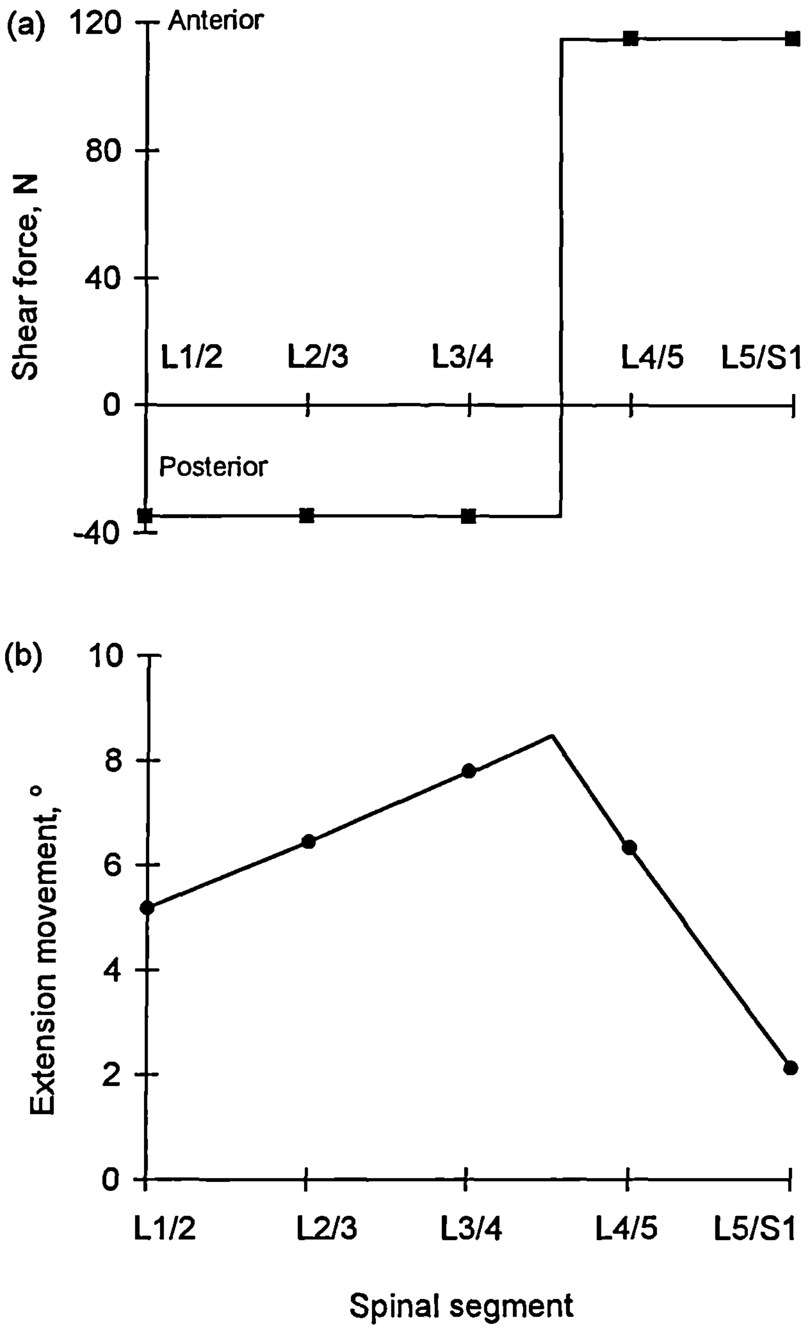


Figure 4.8
Shear force and extension moment produced at the lumbar motion segments when L4 vertebra is mobilised with 150N posteroanterior mobilisation force. (After Lee, 1990).

Motion segment	Vertebra being mobilised	Anterior shear produced at the inferior segment, N	Extension moment produced at the inferior segment, Nm
L1/2	L1	62.3	10.4
L2/3	L2	79.6	10.3
L3/4	L3	97.4	8.9
L4/5	L4	115.2	6.3
Mean		**88.7**	**9.0**

Table 4.3
The magnitude of anterior shear and extension moment produced at the motion segments below the mobilised vertebra.

Posteroanterior mobilisation might be approximated as three point bending of the lumbar spine which was considered to be a beam supported over the anterior superior iliac spines and the anterior surface of the thoracic cage (Lee, 1990; Lee and Evans, 1991 and 1994) (figure 4.7). The model assumed that there were no significant compressive forces along the spine and that the deformation of the spine was small. Based on the model, the loads produced at the motion segments were predicted for a static posteroanterior force of 150N applied to L4 (figure 4.8).

The three point bending model predicted that the motion segments above the level of mobilisation (L4) were subjected to posterior shear forces and segments below L4 to anterior shear forces. The shear forces above and below the mobilised vertebra differed not only in direction, but also in magnitude. The magnitude of the force was larger in the segments below L4. This prediction agreed with the experimental finding that posteroanterior displacement at the segment below was larger than that above (figure 4.3). It was also interesting to note that the preconditioning and creep effects were more more pronounced at the segment below the mobilised vertebra (figures 4.5 and 4.6). The above findings indicate that mobilisation produces more mechanical and perhaps therapeutic effects at the inferior segment.

In addition, it was shown that posteroanterior mobilisation produced extension bending moment at all motion segments and this was greatest at the point of force application. This implied an increase in lordosis of the lumbar spine during mobilisation.

Posteroanterior mobilisation is generally believed to produce anterior "gliding" of one vertebra upon another at the segment below the mobilised spinal level where most of the mechanical effects are produced (Mennell, 1960; Maitland, 1986; Grieve, 1988). The three point bending model does not provide support to the belief. The mobilised vertebra is subjected not only to anterior shear but also extension moment. Table 4.3 shows the magnitude of these loads for mobilisation of different spinal levels as predicted by the model. The mean anterior shear and extension moment was calculated to be 90N and 9Nm respectively. An understanding of how the anatomical structures resist these loads is of clinical importance. It will shed light on how posteroanterior mobility may be altered when a structure is injured and when it heals, providing guidelines on the indications and contraindications of the technique. However, such information is not available in the literature.

Since posteroanterior mobilisation produced extension moment at the motion segments, it was suggested that it had a potential effect on the extension movement of the lumbar spine (Lee, 1990). McCollam and Benson (1993) used the double inclinometer method to measure the sagittal mobility of normal subjects before and after nine minutes of mobilisation. They demonstrated that mobilisation increased the lumbar extension mobility significantly and had no effect on the range of movement in flexion.

Lee et al (1993) examined the effect of muscular contraction on the response of the lumbar spine to posteroanterior mobilisation. It was shown that voluntary contraction of the erector spinae muscles increased the posteroanterior stiffness of the lumbar spine. These muscles produced extension of the spine and thus preloaded the spine before the mobilisation force was applied. This shifted the initial position into the later part of the load-deformation curve. The spine would thus behave in a stiffer manner during subsequent posteroanterior force application. It should be pointed out that the study of Lee et al (1993) might not truly reflect the situation of muscle spasm in which case the muscular contraction was small and sustained over a period of time.

4.1.5 Summary

Clinical manual examination of posteroanterior mobility was shown to be unreliable. Instrumentation was thus developed to objectively quantify the load-deformation characteristics of mobilisation or the "movement diagram". Previous studies had provided data which supported the mechanical mechanisms underlying the technique, such as preconditioning and creep.

The literature review showed that there was a lack of understanding of the anatomical basis of posteroanterior mobilisation. It was uncertain how the various structures resisted the mobilisation loads as predicted by the three point bending model. Such information was clinically important as it would indicate the effects of injuries on posteroanterior stiffness. One of the objectives of the present work was to fill this knowledge gap.

The instruments developed by the previous authors provided a useful clinical means of indicating posteroanterior mobility or stiffness (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and Evans, 1991 and 1992). However, they did not actually measure the true intervertebral movements produced. They were simply surface measurements recording the displacement of the skin over the spinous processes. Such measurements would be subjected to large errors due to skin deformation and rotation of the vertebra. The present study will address this limitation of the previous studies. The intervertebral movements will be fully quantified in terms of segmental rotation and translations. The data obtained will also help validate the three point bending model described above.

4.2 SPINAL TRACTION

4.2.1 Mechanisms Underlying Spinal Traction

As pointed out in chapter 1, traction should be regarded as a form of mobilisation applied along the longitudinal axis of the spine (Maitland, 1986). It should therefore produce similar mechanical and neurophysiological effects as mobilisation, including stretching of scar tissues, preconditioning, creep, stimulation of mechanoreceptors and the closing of the pain gate. However, there was no

Development of a negative pressure in the disc that "sucks" back any protrusion
Production of posterior longitudinal ligament tension to reduce disc herniation
Flattening of the lumbar lordosis
Increase in intervertebral disc heights
Enlargement of the intervertebral foramen
Relieving nerve root impringement
Separation of the zygapophyseal joints
Release of entrapped synovial membrane
Stimulation of mechanoreceptors in the discs, ligaments and zygapophyseal joints
Stretching of spinal muscles and ligaments
Relaxation of muscle spasm
Mobilisation of hypomobile joints as in spinal mobilisation

Table 4.4
Hypothesized mechanisms of spinal traction therapy. (Wyke, 1976; Cyriax, 1978; Saunders, 1979; Cailliet, 1981; Hinterbuchner, 1985; Grieve, 1988; White and Panjabi, 1990)

substantial scientific evidence to support the notion that these mechanisms also operated during traction.

Table 4.4 summarises the specific effects that might be produced by lumbar traction (Wyke, 1976; Cyriax, 1978; Saunders, 1979; Cailliet, 1981; Hinterbuchner, 1985; Grieve, 1988; White and Panjabi, 1990). A number of previous studies had attempted to examine some of these effects.

4.2.1.1 Reduction of a disc prolapse

Many authors recommended the use of traction for the treatment of lumbar disc prolapse (Masturzo, 1955; Yates, 1972; Cyriax, 1978; Saunders, 1979). Cyriax (1978) believed that a negative pressure developed in the disc during traction which "sucked" back a protrusion, but this hypothesis was shown to be unfounded by the study of intradiscal pressure by Andersson et al (1983) (see section 2.3.4).

Previous studies examined the effects of traction on patients whose epidurographs showed undulations of the dural sacs indicating disc protrusions (Mathews, 1968; Gupta and Ramarao, 1978). They generally reported disappearance or reduction in size of the radiographic defects during and immediately after traction, although the radiographic defects were sometimes found to persist.

The results of the epidurographic studies are difficult to interpret. Firstly, these studies did not employ any control groups and thus the changes could have happened without the application of traction and simply with, for example, bed rest. The changes in radiographic defects during traction were unreliable because the absorption and flow of the contrast medium could cause an apparent change in the size of the radiographic defects. Furthermore, no measures were taken to ensure that the radiographic magnification of successive measurements was the same, and thus comparison of the sizes of the defects in different films was potentially erroneous. Finally, these studies did not provide direct evidence on migration of disc materials but were based on changes in epidurographic defects which might not be produced necessarily by disc protrusion.

More recently, Onel et al (1989) further examined the effects of traction on disc protrusion. They employed computerised tomographic (CT) technique to directly examine the movement of disc materials. CT scans were taken before and after traction therapy. It was demonstrated that the prolapsed nuclear materials had retracted in 78% of patients with central protrusion, 67% of those with posterolateral protrusions, and 57% in cases with lateral protrusions. In two cases, however, there were an increase in the amount of prolapsed nuclear materials extruding into the spinal canal. The authors concluded that the effects of traction were variable depending on the direction of protrusion.

Like the earlier epidurographic studies, the study of Onel et al (1989) also had inherent problems in the experimental methodology. Firstly, no control subjects were employed. The use of CT scans to evaluate the changes in the size of protrusion might be unreliable. It was demonstrated that traction produced an elongation of the spine and a flattening of the lordosis (Colachis and Strohm, 1969; Twomey, 1985). It would therefore be difficult to ascertain whether the CT scans before and after traction were taken from the same section and at the same angles.

In summary, the results of previous studies had not provided convincing evidence to support the mechanical hypothesis that traction could reduce a disc prolapse. The notion that it would develop a negative pressure and suck back a protrusion was rebutted by the work of Andersson et al (1983).

4.2.1.2 Increase in disc height and reduction in lordosis

Traction was hypothesised to produce separation of the vertebrae, changes in intervertebral disc heights and a flattening of the lordosis. Colachis and Strohm (1969) carried out a radiographic study on ten normal subjects to examine these effects. Traction force of 445N was delivered in the Fowler position with an angle of pull of 18° with the horizontal. The subjects were given 15 minutes of intermittent traction (10 seconds of "hold" and 5 seconds of "rest") and then 5 minutes of continuous traction after a period of rest. The changes in anterior and posterior disc heights were

Level		Mean changes in disc heights, mm			
		Fowler position@	Inter-mittent traction, 445N#	Continuous traction, 445N#	10 minutes after traction#
L1/2	A*	0.40	0.60	0.85	0.40
	P*	0.35	0.40	0.55	0.30
L2/3	A	0.30	-0.75	-0.80	-0.35
	P	0.80	1.50	1.25	0.50
L3/4	A	-1.60	-0.65	-0.55	-0.05
	P	1.35	1.40	1.30	0.40
L4/5	A	-2.90	-0.45	-0.90	0.25
	P	2.70	1.55	1.85	0.6
L5/S1	A	-2.75	0.15	-0.40	-0.05
	P	0.8	0.10	0.30	-0.05
Total$^{\Psi}$	A	-6.55	-1.10	-1.80	0.20
	P	6.00	4.95	5.25	1.75

* A denotes anterior disc height and P posterior disc height, positive values indicate increases and negative values decreases

@ Changes in disc heights compared with those values in supine lying

\# Additional changes in disc heights, i.e. changes relative to the values obtained in the Fowler position

$^{\Psi}$ Total is the summation of the changes of the five lumbar motion segments

Table 4.5
Mean changes in anterior and posterior disc heights with the adoption of the Fowler position and the traction applications. (After Colachis and Strohm, 1969)

determined after the adoption of the Fowler's position and the different traction applications.

It was shown that the adoption of the Fowler position generally caused reduction in anterior disc heights and increases in posterior disc heights at the lower lumbar segments, but the changes at L1/2 and L2/3 were small (table 4.5). During both intermittent and continuous traction, further reduction in anterior heights and increases in posterior heights were generally observed. These changes were also more pronounced at the lower segments. The observed changes in disc heights suggested that traction produced flexion of the spine and a flattening of the lordosis. In addition, the authors showed that the disc heights almost returned to their initial values 10 minutes after the traction applications (table 4.5), indicating that the mechanical changes produced by traction was temporary.

Table 4.5 shows that the additional changes in disc heights observed after the application of traction were smaller than the initial changes after the adoption of the Fowler position. It appears that hip positions have significant effects on the disc heights produced by traction. Reilly et al (1979) reported that traction produced greater changes in disc height as the angle of hip flexion increased. The changes were largest when the hips were flexed to 90° as in the Fowler's position. Hence, it may be concluded that the adoption of the Fowler's position is clinically important if the objective is to increase vertebral separation.

The results of Colachis and Strohm (1969) should be interpreted with great caution. As discussed in section 3.2.1.4, with the use of an accurate digitiser, the error involved in the measurement of disc height might be as much as 50% of the measured value. Colachis and Strohm (1969) employed engineering callipers to measure the disc heights on radiographs. This instrument was much less precise than a digitiser, and their measurements were thus subjected to large errors.

Twomey (1985) examined the deformation of the lumbar spine produced by traction in cadaveric lumbosacral spines. A traction force of 90N was applied to the specimens via a metal rod through the L1 intervertebral foramina. He showed that there was a mean elongation of 7.5mm for the whole spines immediately after the application of traction. A further mean increase of 1.5mm was observed after creep loading for 30 minutes. The study thus showed that traction exhibited similar time-dependent characteristics as spinal mobilisation. The mean residual deformation 30 minutes after continuous traction was found to be only 0.5mm. This further supported the fact that the mechanical effects of traction were temporary.

Twomey (1985) also demonstrated the effects of ageing and disc degeneration on the deformation produced by traction. Older spines showed a more rapid rate of creep and more residual deformation. This might suggest that traction had a longer lasting mechanical effect in elderly patients. It was also shown that the amount of elongation (11-12mm) was greater in spines with healthy discs (grades of degeneration 0 and 1), and substantially less (3-5mm) in spines with degenerated discs (grades 2 and 3).

The study of Twomey (1985) had several limitations. The cadaveric study did not accurately reproduce the in vivo loading conditions. The loads imposed by hip and knee flexion or the adoption of the Fowler position were not taken into account. The traction force was applied axially through the intervertebral foramina. In living patients, traction was applied through a harness fastened to the pelvis and along the skin surface of the back. The traction force should therefore be applied posterior to the discs rather than axially as in Twomey's study. In addition, the magnitude of traction force used (90N) was small compared to those used in the clinical setting (about one third of body weight; Grieve, 1988).

4.2.1.3 Relaxation of muscle spasm

It was hypothesised that traction produced relaxation of muscle spasm in patients with low back pain (Saunders, 1979; Cailliet, 1981; Hinterbuchner, 1985). Previous studies (Hood et al, 1981; Weatherell, 1987; Letchuman and Deusinger, 1993) demonstrated that electromyographic activities of the erector spinae muscles

Study	Sample characteristics (sample size)	Treatment group	Control group	Results
Hood and Chrisman (1968)	back patients with diagnoses of ruptured disc; aged 22-63 (n=40)	intermittent traction on a split table in the Fowler position , traction force was between 295-318N	No control group	About 50% of patients showed great improvement or disappearence of symptoms immediately after treatment and at 1-2 years follow up
Weber (1973)	radiating pain with neurological signs and a radiculogram showing indentation of the dural sac or occluded root pocket; aged 30-60 (n=86)	Intermittent traction on a split table in the Fowler position for 20 minutes, traction force was one third of body weight; duration of treatment was 5-7 days	Sham traction with a force of 70N only	No significant differences between the 2 groups in terms of improvement in pain, spinal mobility, straight leg raise, neurological signs.
Mathews and Hickling (1975)	sciatica of at least 3 weeks duration, aged 20-60 (n=27)	static traction on a plain couch with a force of 363-610N for 30 minutes; duration of treatment was 3 weeks	Sham traction with 91N (just enough to overcome friction between couch and patient)	Greater improvement in pain and straight leg raise in the traction group immediately after the course of treatment
Weber et al (1984)	radiating pain with neurological signs and a radiculogram showing indentation of the dural sac or	4 groups: (i) intermittent traction by motorised machine, (ii) traction in a metal frame and force is	Sham traction and isometric exercises	After the course of treatment (ranging from 7 days to 2 weeks), all the treated groups showed similar improvement in pain, spinal

Table 4.6
A summary of previous clinical trials of spinal traction.
(to be continued on next page)

Study	Sample characteristics (sample size)	Treatment group	Control group	Results
	occluded root pocket; aged 30-60 (n=215)	applied by the patient's arms, (iii) same as (ii) but on a multiplane table, (iv) traction applied manually by therapists		mobility, straight leg raise and neurological signs when compared to the control group. Manual traction appeared to provide immediate temporary relief of symptoms which were not obtained with exercises in the control group.
Pal et al (1986)	patients admitted to hospital with back pain and sciatica, mean age was 38 years old (n=39)	Hospital traction with a force of 55-82N applied continuously in supine lying on a tilted bed	Sham traction with a force of only 14-18N	There were no significant differences between the 2 groups in terms of pain, analgesic consumption and straight leg raise after 1, 2 and 3 weeks of treatment. The length of stay in hospital was also the same in the two groups.
Mathews et al (1988)	back pain with limitation of spinal movements and straight leg raise, duration of symptoms less than 3 months, 18-60 years of age (n=233)	intermittent traction on a split table in the Fowler position for 30 minutes; traction force was usually about 450N	Infra-red treatment with the same of freqency of attendance of the treated group	A significantly higher proportions of treated patients improved in the first two weeks of treatment. Outcome evaluation was based on subjective pain relief.

Table 4.6
A summary of previous clinical trials of spinal traction.

actually increased during the first few minutes (2-8 minutes) of traction application. The muscular activities gradually returned to its resting level thereafter. Significant decreases in muscular activity were never observed. Thus the rationale for prescribing traction to relax muscle spasm appears to be unfounded. The increase in muscular activity in the early stage of traction was probably due to a reflex muscle response to stretching. After the initial increase, the muscles probably fatigued and allowed vertebral separation.

4.2.1.4 Other mechanisms

Many other mechanisms have been proposed through which spinal traction was believed to offer therapeutic benefits. These include the widening of the intervertebral foramina, relief of nerve root compression, release of entrapped synovial membrane, and stimulation of the mechanoreceptors. They have not been investigated to the same extent as those described above. Further research in these areas is thus strongly indicated.

4.2.2 Clinical Trials

There were a number of studies which had examined the clinical efficacy of traction in the treatment of low back pain (Hood and Chrisman, 1968; Weber, 1973; Mathews and Hickling, 1975; Weber et al, 1984; Pal et al, 1986; Mathews et al, 1988), and the details of these studies are summarised in table 4.6. The table shows that as in clinical trials of other therapeutic procedures like spinal mobilisation, there are large variations among the different studies in the sample characteristics, research methods and outcome measures employed. These explain the differences in clinical results observed.

Most of the studies reported that there were no significant differences in improvement in signs and symptoms between patients treated with traction and those untreated. Only the works of Mathews and Hickling (1975) and Mathews et al (1988) showed that traction therapy was beneficial. In addition, none of the studies reviewed had provided data on the long term clinical effects of traction. Most of them examined the therapeutic benefits up to 3 weeks only.

Clinical trials of traction therapy suffered similar methodological flaws as those of spinal mobilisation. Random allocation of patients into the treatment and groups was found in only two studies (Mathews and Hickling, 1975; Pal et al, 1986). With the exception of the work of Mathews et al (1988), the sample size was often small and most studies thus lacked statistical inference power. Weber et al (1984) appeared to have used a large sample size but there were several treatment groups and only 21-37 patients in each group. A close examination of the characteristics of the samples also revealed that they were often not homogeneous. No considerations were made in regard to the age and gender of patients, duration of symptoms and the number of attacks.

The earlier study of Hood and Chrisman (1968) did not employ any control group. In some studies (Weber et al, 1984; Mathews et al, 1988), the use of control was improper. The control patients were given either therapeutic exercises or infra-red therapy which might have therapeutic effects on back pain. In addition, only two studies had adopted a double blind approach (Mathews and Hickling, 1975; Pal et al, 1986).

Another deficiency of previous clinical trials was that the outcome measures employed were rather limited. Most of the works were based on subjective pain relief reported by the patients and the improvement in spinal mobility, straight leg raise and neurological signs. None of the works measured the improvement in functional ability and pyschosocial measures to fully characterise the therapeutic benefits of traction.

It is also shown in table 4.6 that different methods of traction were employed in previous studies. Pal et al (1986) examined the effects of hospital traction which were continuously applied for several days with a small magnitude of force (55-82N). Most other studies examined the effects of either sustained or intermittent traction for a duration of 15-20 minutes with a force ranging from 300-600N. The force might be applied manually, in metal frames or by motorised machines. There were no consensus on the choice or indications of the various forms of traction therapy.

Van der Heijden et al (1995) evaluated the quality of previous works on traction according to a scoring system. The evaluation criteria included study sample characteristics, treatment interventions and outcome measures. They reported that only 12.5% of the studies reviewed scored more than 50 points (maximum score=100 points), indicating that as discussed above, most of the works were of poor quality. The authors concluded that there were no clear indications that traction was an effective therapy.

4.2.3 Previous Biomechanical Studies

A number of biomechanical studies examined the hypotheses underlying spinal traction therapy (Mathews, 1968; Colachis and Strohm, 1969; Gupta and Ramarao, 1978; Reilly et al, 1979; Hood et al, 1981; Andersson et al, 1983; Twomey, 1985; Weatherell, 1987; Onel et al, 1989; Letchuman and Deusinger, 1993). They have been described in detail in the earlier section 4.2.1.

Judovich and Nobel (1957) analysed the friction between the body and the couch when traction was applied. The friction between disarticulated lower body segments of cadavers (below L3/4) and the traction couch was found to be 27% of the total body weight. Thus, it was concluded that a traction force of less than this value would not be enough to overcome friction and be ineffective to produce elongation of the spine. Judovich and Nobel (1957) recommended the use of the split table to eliminate the friction force and thus considerably reduce the traction force required. This has now become a standard practice in clinical traction therapy.

4.2.4 Summary

The therapeutic mechanisms underlying traction have not been supported by data from well-conducted studies. The studies of Colachis and Strohm (1969) and

Twomey (1985) had provided basic information on the spinal deformations produced by traction, but they had several limitations. Colachis and Strohm (1969) used callipers to measure disc heights and their results were thus subjected to large errors. The cadaveric work of Twomey (1985) measured only the total elongation of the spine and failed to reproduce the in-vivo loading conditions. The intervertebral movements had not been fully quantified in terms of the rotation and translation parameters. Furthermore, although traction was believed to enlarge the intervertebral foramina, this had also not been studied experimentally. In view of these, it is felt that there is a need to re-examine the deformations produced by traction, taking into account the limitations of previous studies. The data obtained would shed light on the mechanical hypotheses underlying spinal traction. The hypotheses of reduction of disc prolapse and enlargement of intervertebral foramina would be evaluated in particular.

In clinical practice, traction is often applied in an intermittent manner with periods of "hold" and "rest". Twomey (1985) did not evaluate this form of traction, but examined the creep characteristics of sustained traction. Colachis and Strohm (1969) studied the residual deformation after the spines were subjected to both intermittent and continuous traction. The residual deformation after a single application of intermittent traction and the changes in deformation with each loading cycle have not been studied. It appears that there is insufficient information on the time-dependent characteristics of intermittent traction, and the present work would attempt to address this issue.

4.3 CONCLUSION

It is concluded that the clinical efficacy of mobilisation and traction has not been supported by previous studies. There are major deficiencies in previous biomechanical studies which examined the movements produced by these techniques and the underlying mechanical hypotheses. Areas that will require further investigations are pointed out in the above review. These include

1. the role of the anatomical structures in resisting posteroanterior mobilisation loads,
2. the true intervertebral movements produced by mobilisation,
3. the intervertebral movements and the deformation of the intervertebral foramina produced by traction, and
4. the time-dependent characteristics of traction.

The present study would attempt to fill the gaps in knowledge and examine the areas identified above. The similarities and differences of the mechanical effects of mobilisation and traction would be compared.

■ *PART II*

THE BIOMECHANICAL BASIS OF SPINAL MOBILISATION

■ *Chapter 5*

Mechanical Behaviour of the Lumbar Motion Segments Under Posteroanterior Mobilisation Loads

---

Mechanical Behaviour of the Lumbar Motion Segments Under Posteroanterior Mobilisation Loads

5.1 INTRODUCTION

Posteroanterior mobilisation is generally believed to produce anterior "gliding" of the mobilised vertebra upon its neighbour (Mennell, 1960; Maitland, 1986; Grieve, 1988). The three point bending model clearly shows that the belief is not entirely true (see section 4.1.4). The mobilised vertebra is subjected not only to anterior shear but also extension moment. An understanding of how the anatomical structures resist these loads is of clinical importance. It will allow the clinicians to predict the change in posteroanterior mobility when a structure is injured and when it heals. However, such information is not available in the literature.

A cadaveric motion segment study was thus carried out to determine the intervertebral movements produced by mobilisation loads and the effects of dissection of the anatomical structures on these movements. As explained in chapter 2, an analysis of the sagittal movements alone would be sufficient as only anterior shear and extension moment were applied to the motion segments. Although movements in the coronal and transverse planes were not studied, an attempt was made to determine the errors due to any out-of-plane movements.

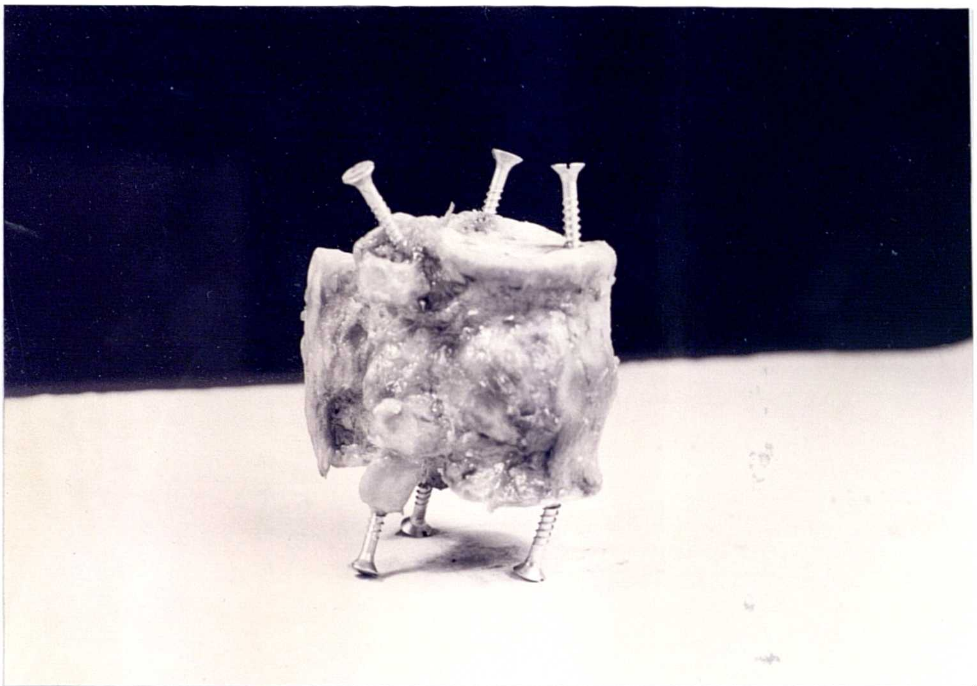


Figure 5.1
A lumbar motion segment specimen (with wood screws inserted into the bones).

Spine No.	Sex	Age (years)	Body height (m)	Motion segments tested and their grade of disc degeneration (specimen number)
120	F	63	1.63	L1/2 grade 2 (specimen 01) L3/4 grade 2 (specimen 02)
119	F	78	1.55	L2/3 grade 2 (specimen 03) L4/5 grade 2 (specimen 04)
121	M	49	1.70	L1/2 grade 1 (specimen 05) L3/4 grade 2 (specimen 06)
123	F	55	1.50	L2/3 grade 3 (specimen 07) L4/5 grade 3 (specimen 08)
124	M	88	1.75	L2/3 grade 2 (specimen 09) L4/5 grade 3 (specimen 10)

Table 5.1
Personal particulars of the cadavers from which the specimens were taken and the degree of degeneration of the motion segments tested

5.2 MATERIALS AND METHODS

5.2.1 Materials

Five complete lumbar spines (L1-L5 inclusive, 2 males and 3 females, aged between 49 and 88) were obtained at routine necropsies within 24 hours of death, and double-wrapped in plastic bags at -20^{o}C until required for testing. This method of specimen storage has been shown not to significantly affect the mechanical properties of bone and soft tissues (Sedlin and Hirsch, 1966; Galante, 1967; Hirsch and Galante, 1967; Tkaczuk, 1968; Panjabi et al, 1985) (see section 3.1.1).

In preparation for testing, each spine was thawed overnight at 3^{o}C. It was then left to equilibrate with room temperature for a few hours before it was taken out of the plastic bags for dissection. Psoas major and the spinal muscle groups were removed from each specimen. Care was taken not to damage the intervertebral disc or the ligaments. The spine was then dissected into two motion segments, each consisting of two vertebrae and the intervening disc and ligaments (figure 5.1). Altogether 10 motion segments were prepared for testing.

The specimens were examined carefully by visual inspection to determine if there were any signs of bone pathology and anatomical anomalies. After testing, the intervertebral discs of all motion segments were sectioned through their mid-transverse planes. The degree of degeneration of the discs was evaluated on an integer scale of 0 to 3 according to the criteria proposed by earlier authors (Nachemson, 1960; Rolander, 1966; Galante, 1967) as described in section 2.3.1. None of the discs tested was found to be grade 0, one was grade 1, six were grade 2 and three were grade 3.

The sex, age, whole body height and grade of disc degeneration of the specimens are presented in table 5.1. The body weight of the cadavers was not available from the mortuary records.

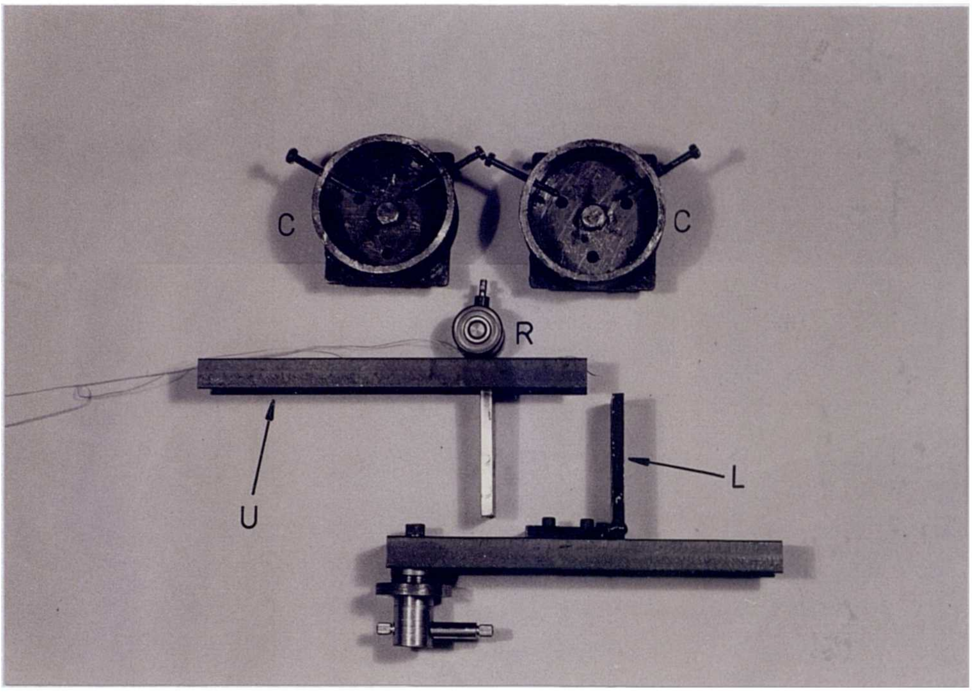


Figure 5.2
The moulding cups and the loading jigs.
(C=moulding cups, R=roller to be attached to the load cell, U=upper loading jig, L=lower loading jig to be atttached to be crosshead of the Instron machine)

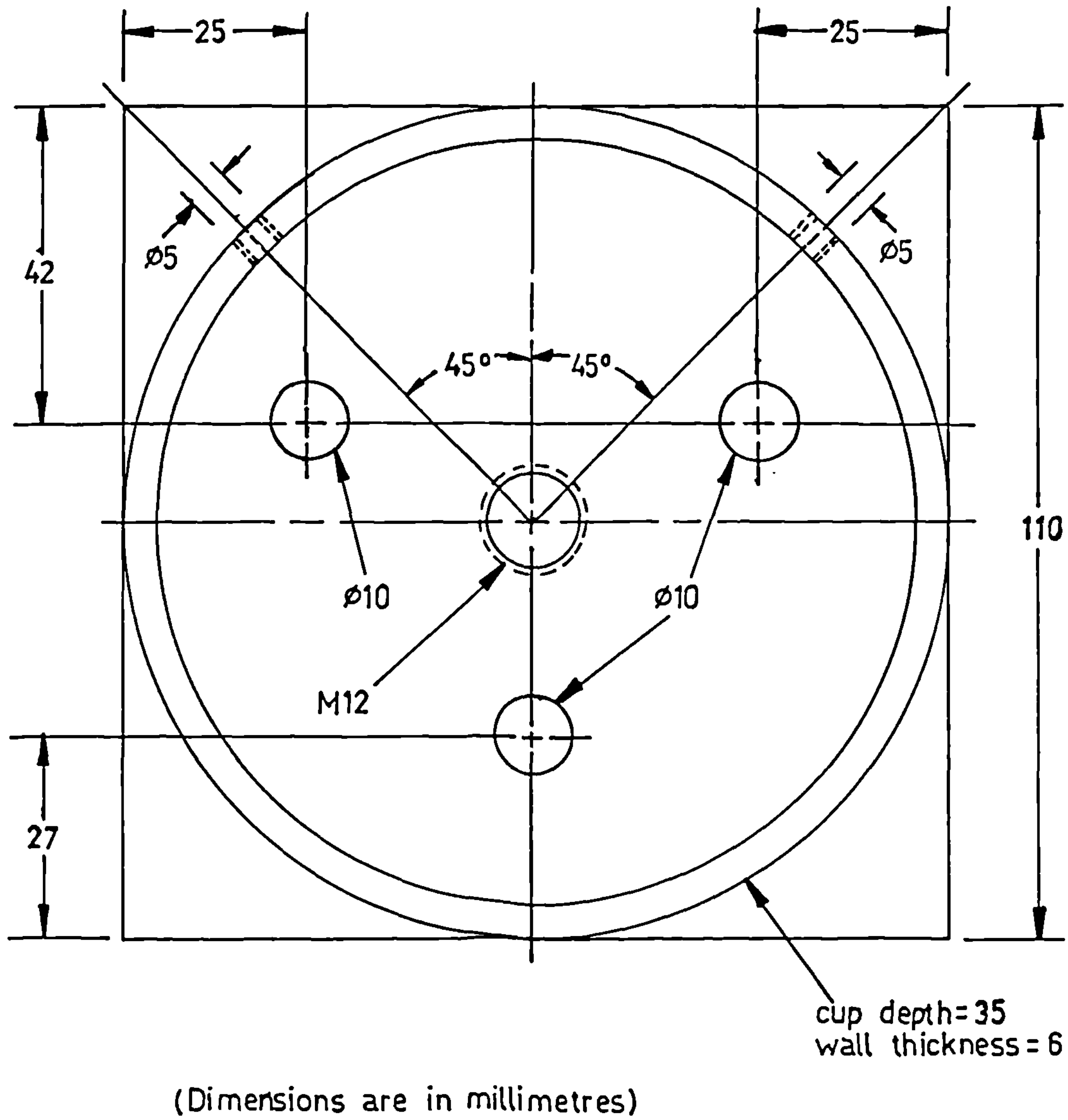


Figure 5.3
A schematic diagram of the moulding cup (top view) showing its dimensions.

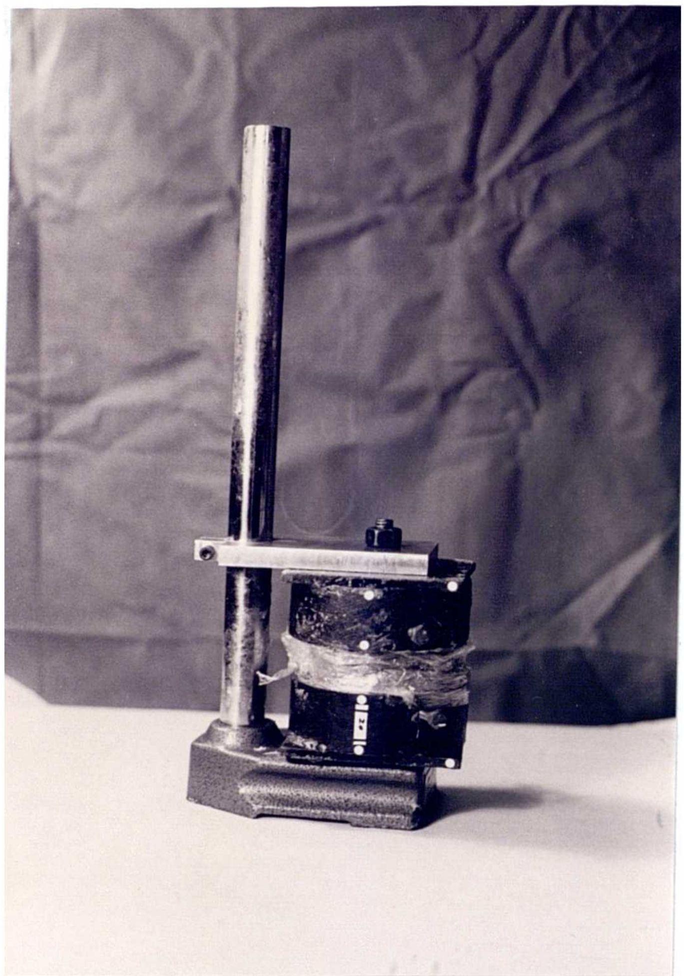


Figure 5.4
The alignment device used to hold the moulding cups during setting of the Isopon. It ensured that the bases of the moulding cups were horizontal.

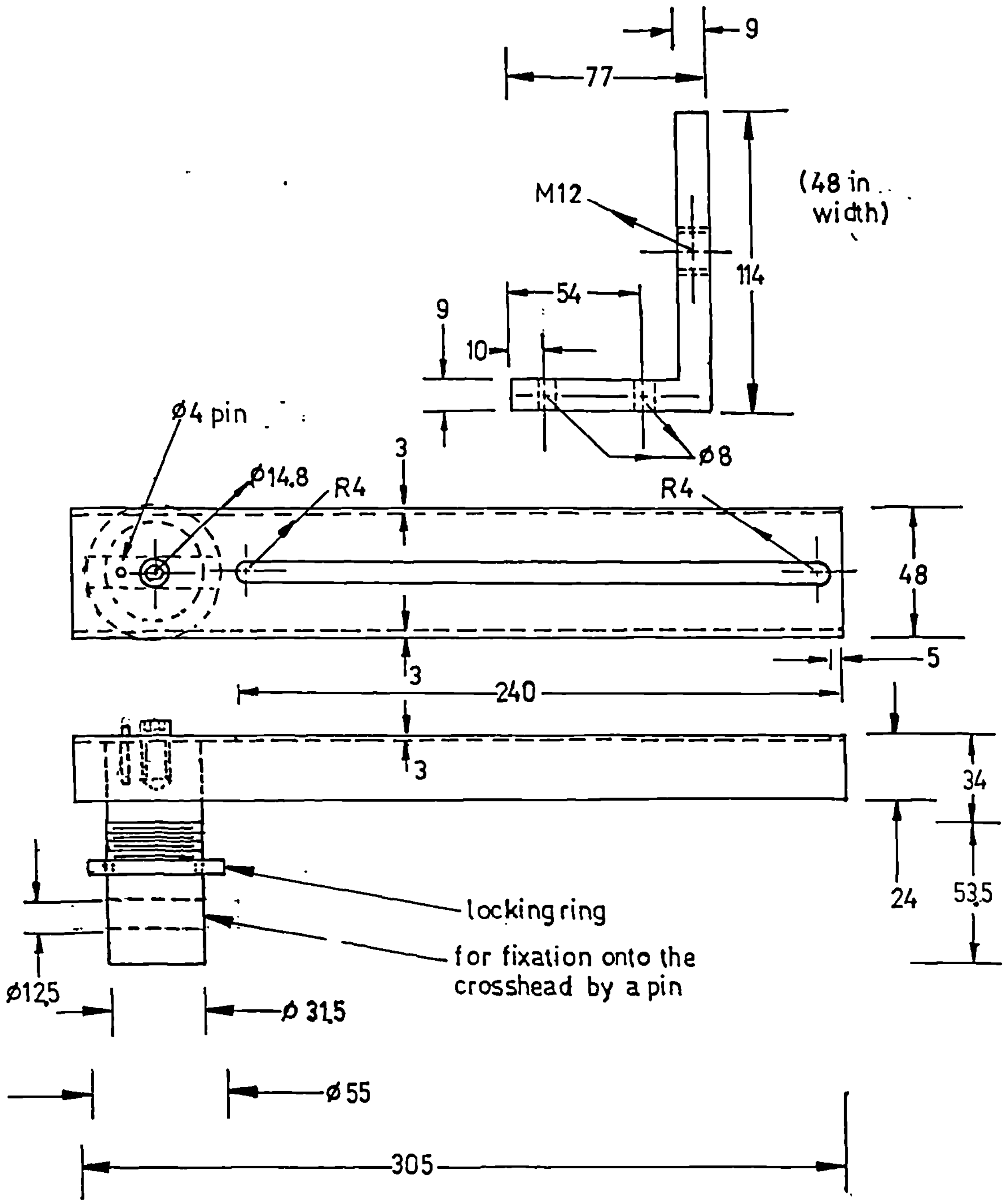
5.2.2 Preparation of the Motion Segments

Two stainless steel cups (figures 5.2 and 5.3) were made to hold the motion segment securely for attachment to the loading system. They had an outer diameter of 0.100m, a wall thickness of 0.006m and a height of 0.035m. The inferior vertebra of each motion segment was set in Isopon (i.e. epoxy adhesive with filler incorporated; Plastic Padding Limited, High Wycombe, Bucks HP10 OPE, UK) in one of the cups. A relatively fluid mix was used as it had a good flow characteristic which ensured good contact with the bones. It also had a conveniently short curing time. The preparation required about 20 minutes to set, after which the motion segment was turned upside down and the superior vertebra mounted in the other holding cup in an identical fashion.

To enable secure gripping of the segments, wood screws were inserted into the vertebral bodies and the posterior elements to create additional contact surfaces (figure 5.1). Through the two threaded holes in the wall of each holding cup (figures 5.2 and 5.3) were passed tapered locking screws. These were tightened so as to penetrate the Isopon and the specimen and thus rigidly hold the ensemble together. In addition, there was a 0.012m diameter threaded hole in the centre of the base plate (figures 5.2 and 5.3), through which the attachment bolt was inserted. The bolt head was also cast into the mould to provide further fixation of the ensemble.

In order that the moulding cups could be reused, their inner surfaces were covered by a thin coat of lubricating grease to prevent permanent adhesion between the hardened Isopon and the cup. Three 0.01m diameter holes were bored through the base plate of the cup (figures 5.2 and 5.3) so that the ensemble could be pushed out easily when the locking screws had been removed. All the holes were covered by cellophane tape during setting of the Isopon.

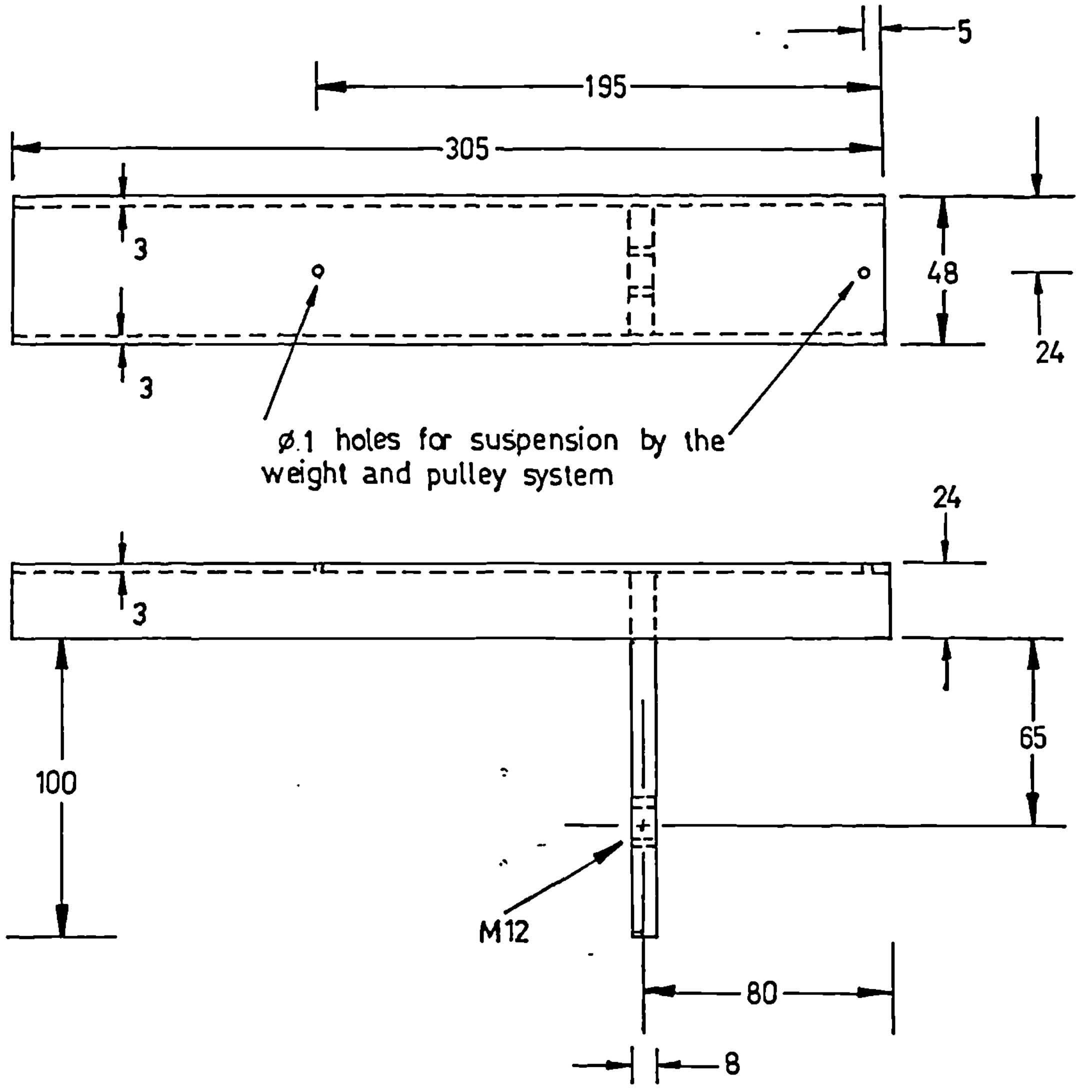
An alignment device (figure 5.4) was used to hold the specimen and the cups during the setting process so that the mid-plane of the disc was parallel to the base



The L bracket was fixed onto the horizontal beam by two bolts and nuts.

(Dimensions are in millimetres)

Figure 5.5a
A schematic diagram of the lower loading jig showing its dimensions.



(Dimensions are in millimetres)

Figure 5.5b
A schematic diagram of the upper loading jig showing its dimensions.

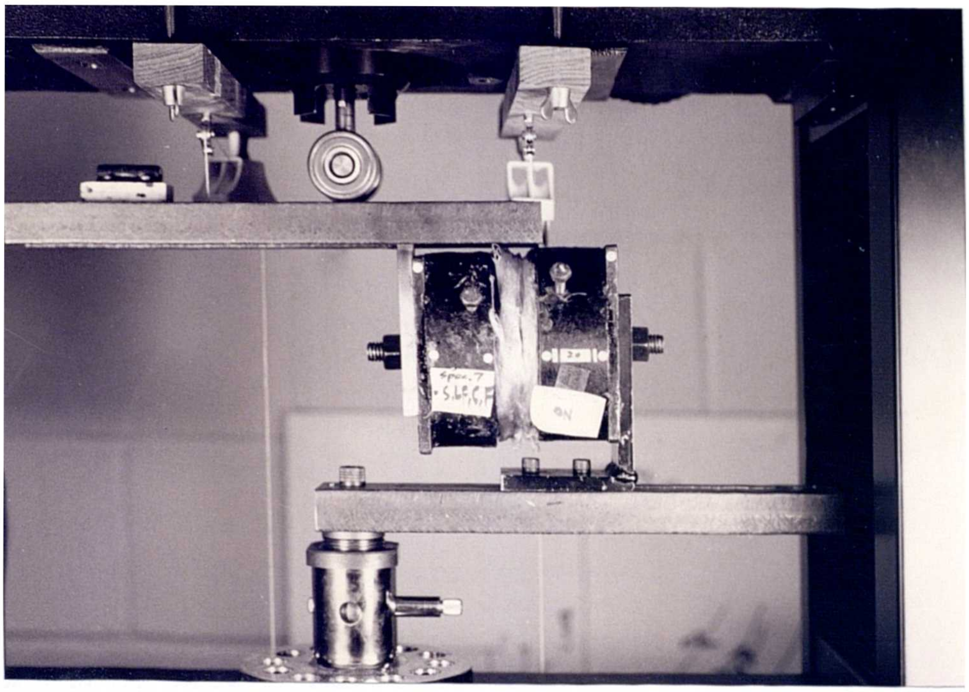


Figure 5.6a
The experimental arrangement of the moulding cups (painted black), loading jigs, roller and markers (white).

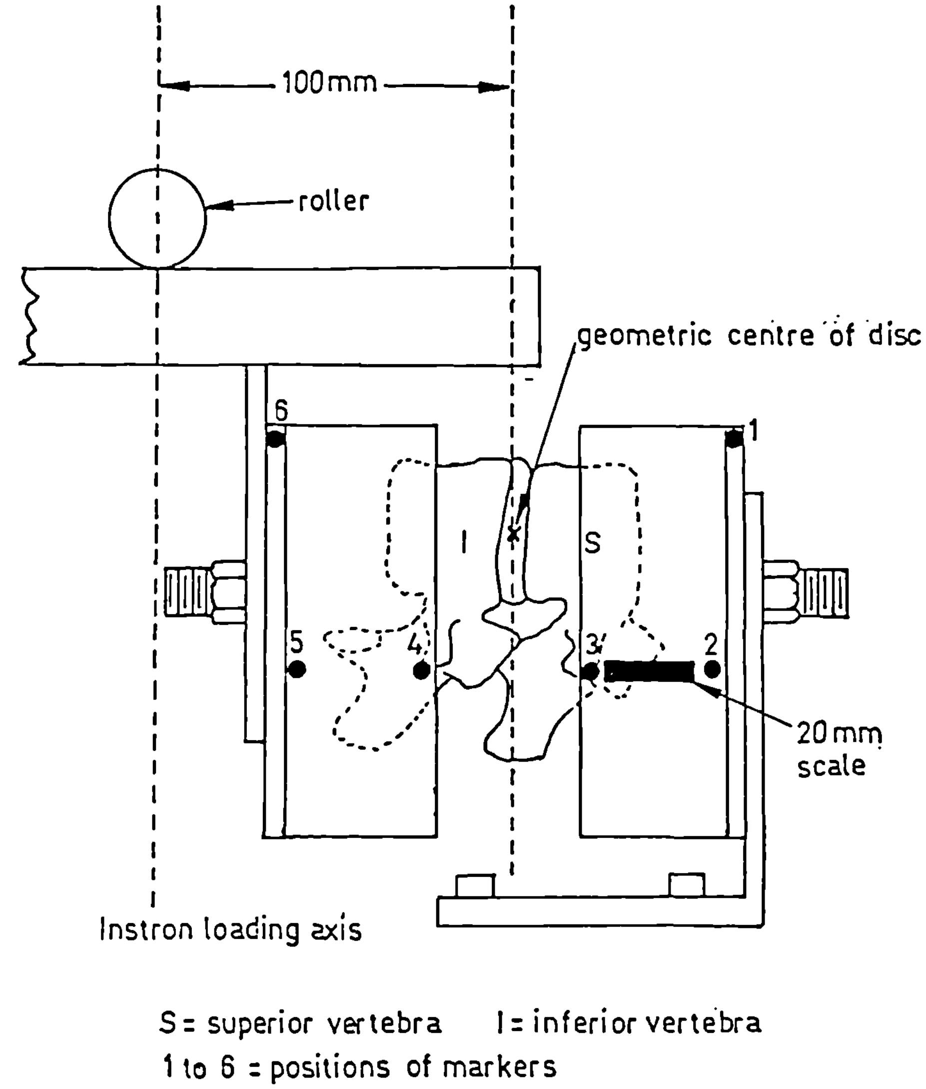


Figure 5.6b
The experimental arrangement showing the position of the motion segment inside the moulding cups and the location of the disc centre in relation to the loading axis of the Instron machine. The markers and the 20mm scale are also shown.

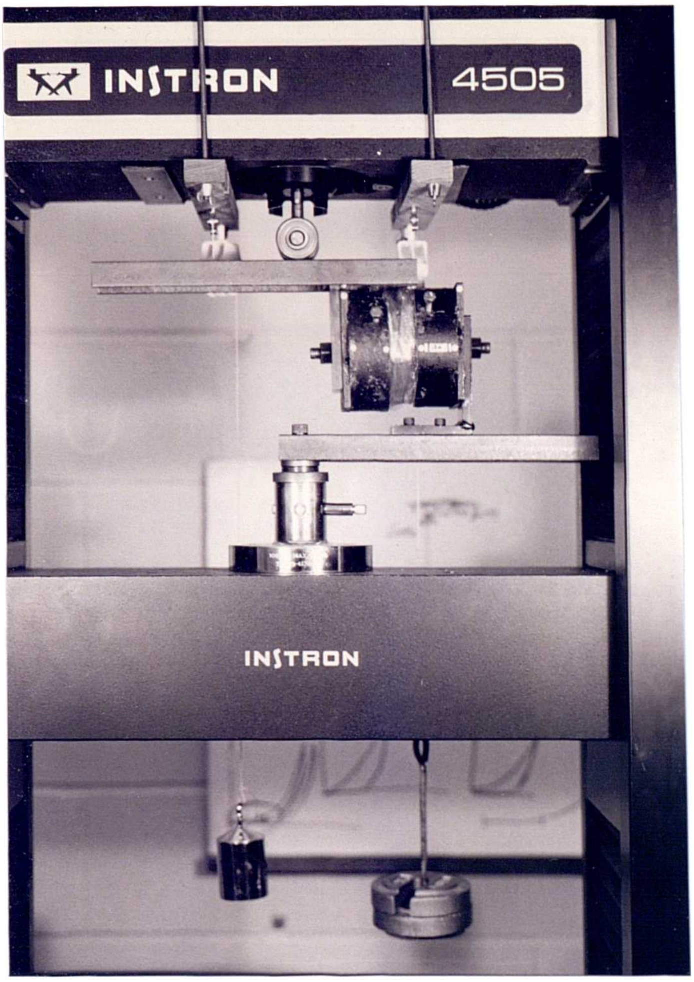


Figure 5.7
The weight of the upper loading jig was counterbalanced by weight and pulley systems which were attached to the frame of the Instron machine.

plates of the cups. During setting, the specimen was covered with moistened cotton wool and Cling Film to reduce moisture loss.

5.2.3 Experimental Apparatus

All experiments were carried out on an Instron testing machine (Model 4505, Instron Limited, High Wycombe, Bucks HP12 3SY, UK). In order to simulate the loading conditions of PA mobilisation, loading jigs (figures 5.2 and 5.5) were specially designed to convert the uniaxial force produced by the Instron machine into the required combination of anterior shear and extension moment. A schematic diagram of the loading jigs is shown in figure 5.5.

One of the loading jigs comprised an inverted U beam and an L-plate (figures 5.2 and 5.5a). It was fitted onto the crosshead of the Instron machine and locked into place by a shear pin and a locking ring (figures 5.6). The moulding cup containing the superior vertebra of the motion segment was attached to the L-plate. The specimen was arranged so that the spinous processes were facing downwards.

The cup containing the inferior vertebra was attached to the other loading jig which consisted of an inverted U beam and a vertical plate (figures 5.2, 5.5b and 5.6). Should the weight of this jig have been applied as a preload to the specimen, this would have caused a significant error since the simulated mobilisation loads to be applied were relatively small in magnitude. The jig weight was therefore balanced by two weights using pulley systems as shown in figure 5.7. The positions of these weights were adjusted so that the beam of the jig was horizontal. This was checked by spirit level.

A roller with a diameter of 0.041m was fitted onto the load cell of the Instron machine by a shear pin (figure 5.6). It applied a posteriorly directed force to the inferior vertebra through the upper beam when the crosshead was driven vertically upwards. The superior vertebra of the motion segment was thus subjected to anterior shear. The position of the L-plate of the lower jig was adjusted so that the geometric

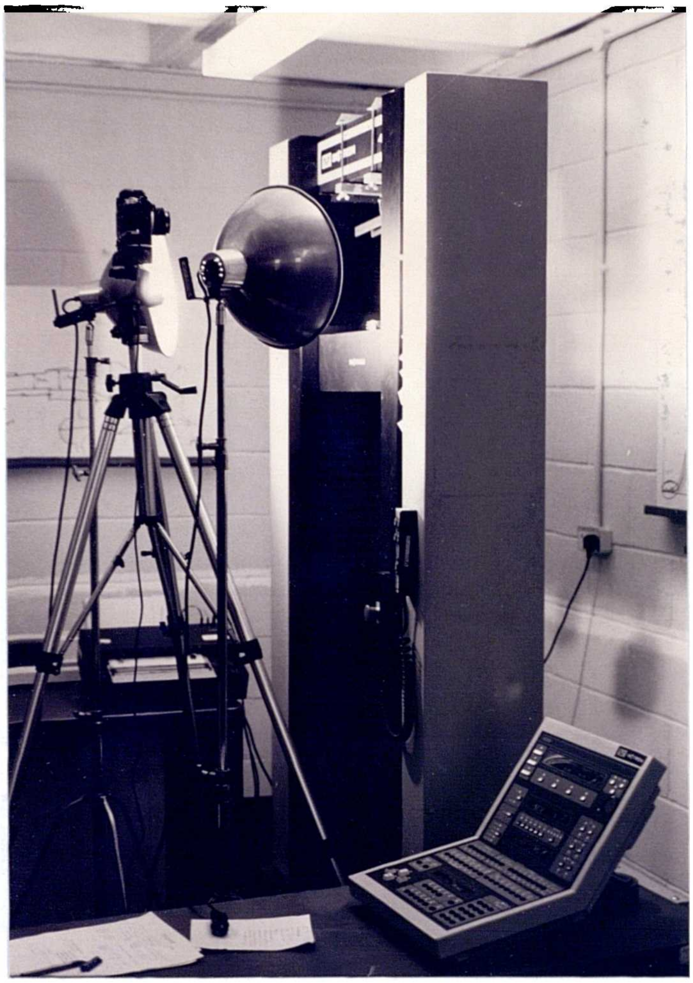


Figure 5.8
The experimental set-up showing the positioning of the camera and the photographic flood lamps in relation to the Instron machine.

centre of the disc was offset from the loading axis of the Instron machine by 0.1m (figure 5.6). The maximum shear force applied in this experiment was 90N. Offsetting the disc centre would therefore produce maximum extension moment of 9Nm. The magnitude of these loads would be the same as the average mobilisation loads predicted by the three point bending model as shown in table 4.3 (section 4.1.4).

A Nikon "F" series 35mm single lens reflex camera fitted with a 55mm MicroNikon lens was used to study the movements of the vertebrae. After a trial, it was found that the best results were achieved with a speed of 1/60s and an aperture of f8, using an Ilford FP4 Plus ASA125 black and white film. The camera was fitted with a motor drive for remote, single frame advancement. It was placed at a distance of 0.5m from the specimen and mounted onto a tripod. Spirit levels were employed to ensure that the camera was vertical and that the optical axis was perpendicular to the loading plane of the specimen. Two 500W photographic flood lamps were used to provide lighting and placed in optimal positions where shadowing could be reduced to a minimum.

Three 0.004m diameter circular markers (white in colour) were placed on the outer surface of each holding cup. The markers were labelled 1 to 3 for the superior vertebra and 4 to 6 for the inferior vertebra (figures 5.6). The outer surface of the cup was painted dark to provide good contrast. A 0.02m scale was also fixed to the cup containing the superior vertebra to allow for correction of photographic magnification (figure 5.6).

The entire experimental set-up is shown in figure 5.8.

5.2.4 Experimental Procedure

The specimen was covered by moist cotton wool and Cling Film during the mechanical test to reduce moisture loss. The crosshead was first driven upwards so that the roller was just in contact with the upper beam of the jig. The Instron machine was set in the "load control" mode. The anterior shear and extension moment loads

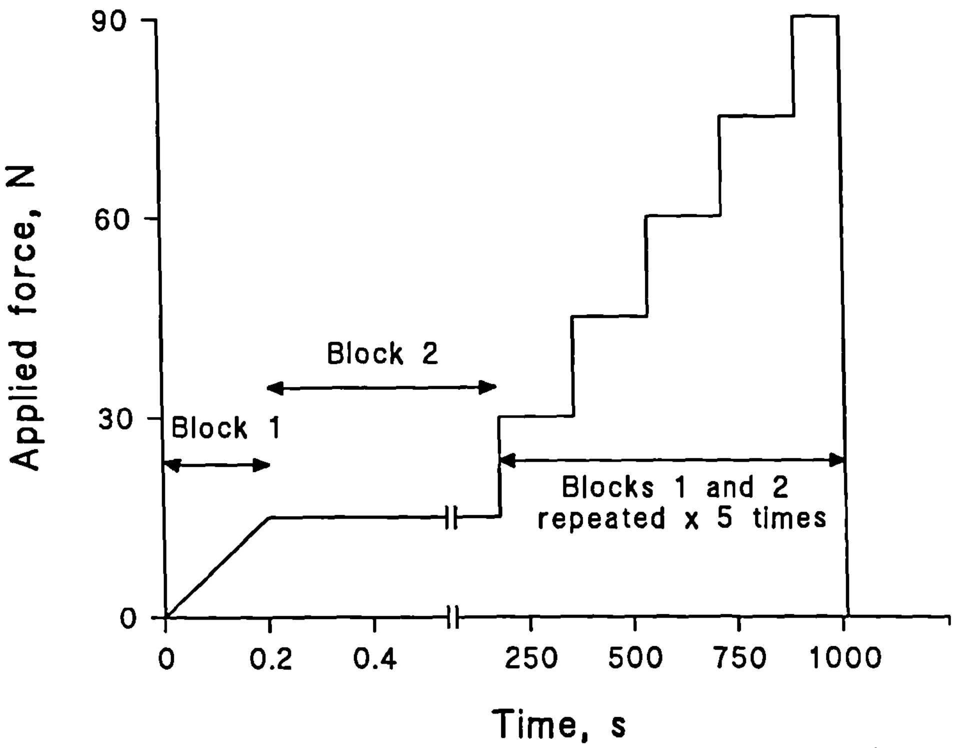


Figure 5.9
The loading cycle of the experiment.

were applied in 6 equal increments (i.e. increments of 15N and 1.5Nm). For each load increment, a ramp waveform was first applied at a rate of $75Ns^{-1}$ up to an amplitude of 15N (block 1). This loading rate was similar to the mobilisation rate observed in a normal clinical session (Lee, 1990). This was then followed by a ramp input with zero amplitude (i.e. fixed force) and a dwell time of 3 minutes to allow the specimen to creep (block 2). The machine was programmed to repeat blocks 1 and 2 five more times (figure 5.9). When the loading sequence was completed, the specimen was immediately unloaded.

Before the test proper, a loop shaping procedure was carried out to minimise any loop overshooting and to optimise the waveform response. This was conducted by running a pre-test and observing the load-time characteristics on a plotter. The appropriate proportional, integral and derivative gain settings were found which varied with different specimens according to their flexibilities.

The positions of the markers were recorded by the Nikon camera before any load was applied and after each load increment. A photograph was taken after each load increment at the end of the 3 minute creep period (i.e. at the end of the block 2 of each load increment) to ensure repeatable measurements. Initial testing had shown that there was no further significant creep 3 minutes after loading.

After the initial test was completed, the various posterior elements were dissected sequentially and tests repeated at each stage. The order of dissection was:

1. the supraspinous and interspinous ligaments
2. the ligamentum flava
3. the capsules of the left and right zygapophyseal joints
4. the left and right zygapophyseal joints

The supraspinous and interspinous ligaments were divided between each spinous process, down to the level of the root of each process. The part of the ligamentum flavum medial to the zygapophyseal joints was sectioned by passing the scalpel horizontally between the laminae at each level. The joint capsules were cut along the lateral joint margins, and a scalpel blade was passed between the joint surfaces to section the medial aspect of the capsule, which also cut the lateral parts of the ligamentum flava that blend with the capsule. The zagapophyseal joints were cut close to the junctions of the articular facets and the laminae using a necropsy saw.

The mechanical testing procedure was repeated after dissection of each of the above anatomical structures. The experimental sequence was usually completed within 3 hours.

Upon completion of the tests, the disc was cut in half. After the grade of degeneration was determined, the disc materials of the two vertebrae were removed so as to expose the endplates. The most anterior and posterior parts of the two exposed endplates were marked. Their positions relative to the markers were measured by a digital caliper with the assistance of an L-shaped bracket. The measurements were repeated three times to minimise error.

After the experiment, the photographic negatives were projected (using a Kodak Carousel S-AV2010 Projector) onto a digitising tablet (Digi-pad Type 5A, Gtco Corporation, Columbia, Maryland, USA) where the image was magnified by about 2.7 times. Since the image was magnified, the error involved in the measurements was reduced by the magnification factor. For each negative, the positions of the six markers were digitised in the same order. The two ends of the scaling bar were also digitised to correct for magnification.

5.2.6 Treatment of Data

5.2.6.1 Error analysis

An effort was made to quantify the errors in the present experiment. Sources of errors included digitisation, measurement of the coordinates of anatomical landmarks, malalignment and lens distortion of the camera and slide projector, and out-of-plane movements. Error was defined as the standard deviation of a repeated measure.

In order to determine the error due to digitisation, the photographic image of a marker was digitised ten times. The standard deviations of the horizontal and vertical coordinates were computed and then divided by the magnification factor of the image.

A projected image of concentric circles was utilised to help align the projector so that the lens and the digitising pad were in the same plane. This was checked by symmetry of the circles. The radius of the outermost circle was determined using ten different points of the circumference. Error due to malalignment and lens distortion of the slide projector was defined as the standard deviation of these ten different measurements of the radius.

The positions of the most anterior and posterior parts of the two endplates with respect to the markers were measured by calliper ten times in one of the specimens. The standard deviations of these measurements were computed.

The distances between various markers should remain constant in all photographic slides if there were no out-of-plane movements and optical errors. They were calculated by

$$D_{i(i+1)} = \sqrt{(M_{ix} - M_{(i+1)x})^2 + (M_{iy} - M_{(i+1)y})^2}$$

where

$D_{i(i+1)}$ is the distance between the ith and (i+1)th markers,

M_{ix} and $M_{(i-1)x}$ are respectively the x coordinates of the ith and (i-1)th markers, and

M_{iy} and $M_{(i-1)y}$ are respectively the y coordinates of the ith and (i-1)th markers.

Four intermarker distances (D_{12}, D_{23}, D_{45} and D_{56}) were measured. Since there were five sets of data (one for the intact specimens and one after each dissection) and seven films for each set (one for zero load and one for each load increment), the measurement of each intermarker distance could be repeated 35 times. The means and standard deviations of the intermarker distances were then computed.

The superior vertebra was rigidly fixed to the crosshead of the Instron machine. Markers 1 to 3 therefore always moved in the same plane. Any differences in the distances D_{12} and D_{23} were unlikely due to out of plane motions but represented error due to malalignment and lens distortion of the camera.

The inferior vertebra was free to move in all planes and markers 4 to 5 were thus subject to out of plane motions. Concomitant coronal motions of the specimen would affect distance D_{45} while concomitant axial rotations distance D_{56}. The standard deviations of these distances represented both optical error and error due to out of plane motions.

Since mean distances D_{12}, D_{23}, D_{45} and D_{56} were different in magnitude, their standard deviations cannot be directly compared to show the relative amount of errors due to concomitant coronal and axial rotations and optical distortion. The coefficients of variations of these distances which expressed the standard deviations as percentages of the means were thus obtained.

5.2.6.2 Computation of the movements of the motion segments

The various experiments of this thesis employed the same coordinate system and similar computational procedures in determining the movements of the motion segments. These are fully described in appendix I (figure A1). The inferior vertebra is considered to be fixed while the superior one moves relative to it. Sagittal movements of the superior vertebra, after each load increment, was described in terms of rotation about the mediolateral axis, and translations at the anteroinferior and posteroinferior corners (i.e. the most anterior and posterior parts of the inferior endplate) along the anteroposterior (x) and superoinferior (y) axes (figure A1).

The first step in the data analysis was to correct the marker coordinates for magnification. The spatial relationships between the anatomical landmarks (the most anterior and posterior parts of the two exposed endplates) and the markers determined by the calliper measurements as described above. These would be unchanged in different photographic negatives if the vertebrae were assumed to be rigid bodies and there were no relative movements between the moulding cups and the vertebrae. The coordinates of the landmarks in the photographic negatives could then be computed from the positions of the markers. The computational steps are explained in section A1.2.

The coordinates of the anteroinferior and posteroinferior corners of the superior vertebra were determined with respect to the reference coordinate system (see section A1.1). The most anterior and posterior parts of the superior endplate of the inferior vertebra served as the reference points for the coordinate transformations. The x- and y-translations at the two corners of the superior vertebra were then determined (see section A1.4).

Sagittal rotation of the motion segment could be calculated from the change in inclination of a line joining any two of the three markers attached to the superior vertebra (see section A1.3). Three independent computations of the movement were performed - from markers 1 and 2, 2 and 3 and 1 and 3. The three results were then averaged to provide the best evaluation of the sagittal rotation movement.

The rotation and translations of the motion segments were computed at different load levels, and in the intact and dissected specimens. The computation was performed by the computer program "PAMOB.PAS" (Appendix II) which was especially written for this study.

5.2.6.3 Statistical analysis

One way analyses of variance (ANOVA) were carried out to examine the effect of dissection on the mean maximum movements produced by the mobilised loads . The maximum rotation and translational movements represented the dependent variables, and the specimen conditions (intact or dissected) the independent variable. The data were statistically analysed by the "Statistical Package for the Social Sciences for the Personal Computer" (SPSS/PC+).

Distance between markers	Size of error, mm (1 standard deviation)	Mean, mm	Coefficient of variation, %*
D_{12} (reflects optical error)	0.53mm	52.87mm	1.00
D_{23} (reflects optical error)	0.32mm	31.02mm	1.03
D_{45} (reflects optical error and error due to concomitant lateral bending)	0.35mm	29.80mm	1.17
D_{56} (reflects optical error and error due to concomitant axial rotation)	0.53mm	50.77mm	1.04

* (Coefficients of variation = standard deviations expressed as percentages of means.)

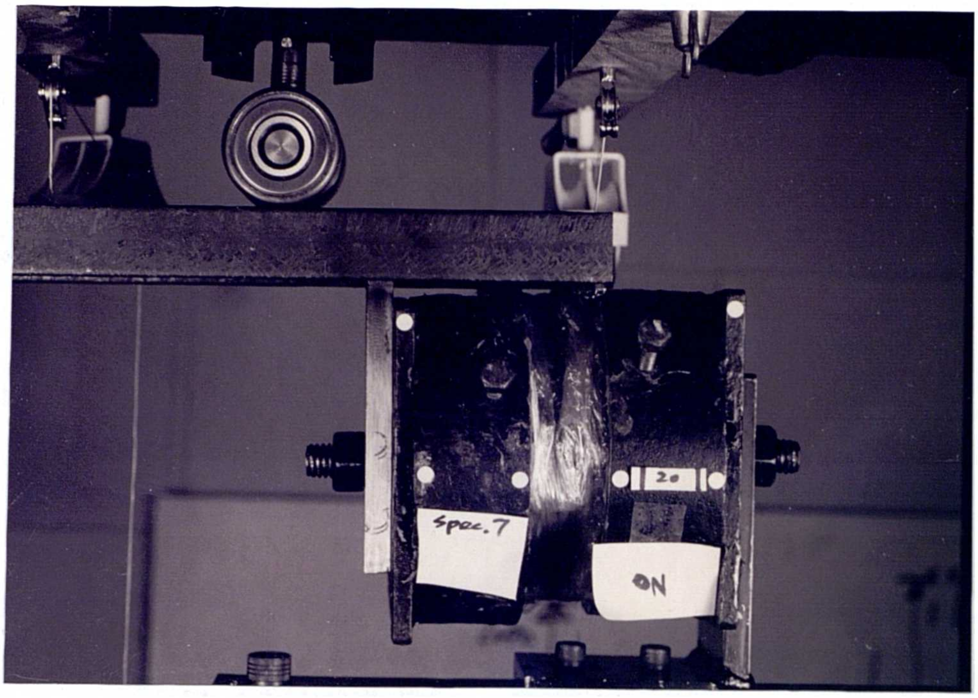
Table 5.2
Errors in the distances between markers.

	Mean (±S.D.) maximum movements of the motion segments (n=10)				
Applied loads	Extension of the motion segment, °	Translation of the anteroinferior corner of the superior vertebral body, mm		Translation of the posteroinferior corner of the superior vertebral body, mm	
		x-translation*	y-translation#	x-translation*	y-translation#
0N + 0Nm	0	0	0	0	0
15N + 1.5Nm	0.38 (±0.35)	0.12 (±0.38)	0.15 (±0.36)	0.03 (±0.42)	-0.19 (±0.22)
30N + 3.0Nm	1.06 (±0.67)	-0.02 (±0.53)	0.45 (±0.63)	-0.22 (±0.59)	-0.30 (±0.28)
45N + 4.5Nm	2.04 (±1.10)	0.14 (±0.61)	1.08 (±0.73)	-0.28 (±0.74)	-0.43 (±0.23)
60N + 6.0Nm	3.05 (±1.38)	0.29 (±0.79)	1.78 (±0.90)	-0.36 (±0.88)	-0.52 (±0.35)
75N + 7.5Nm	4.12 (±1.62)	0.37 (±0.87)	2.43 (±1.58)	-0.53 (±1.06)	-0.69 (±0.49)
90N + 9.0Nm	4.99 (±2.03)	0.29 (±0.98)	3.00 (±1.32)	-0.86 (±1.18)	-0.81 (±0.59)

* positive values denote posterior translations and negative anterior translations
# positive values denote superior translations and negative inferior translations

Table 5.3
Mean movements of the motion segments produced by simulated posteroanterior mobilisation loads

(a)



(b)

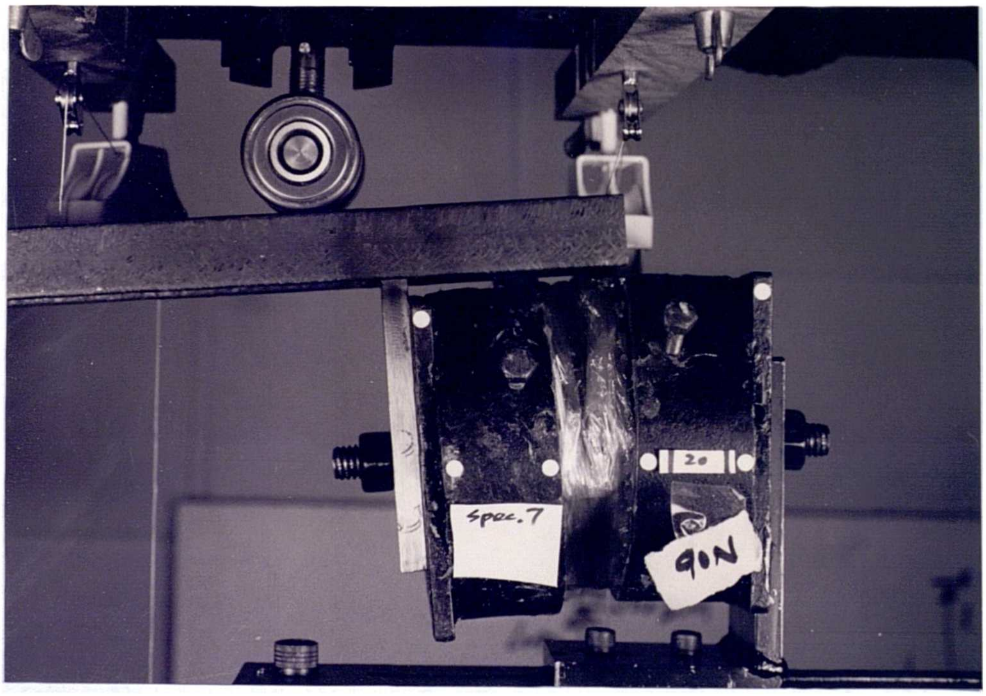


Figure 5.10
Photographs of an intact specimen (specimen 07, female, aged 55, grade of degeneration=3) showing the movements produced by simulated mobilisation loads. (a) when no load was applied (b) when anterior shear of 90N and extension moment of 9Nm was applied.

5.3 RESULTS

5.3.1 Error Analysis

In relocating and repositioning the digitiser cursor over the marker image, the average error due to digitisation was found to be ±0.038mm. Error due to malpositioning of the projector, as revealed by the standard deviation of the radius measurements, was ±0.37mm. The calliper was found to have a mean error of ± 0.60mm in determining the positions of the most anterior and posterior parts of the two endplates with respect to the markers. This appeared to be the major source of error in this experiment.

Errors in the distance between the markers are shown in table 5.2. The coefficients of variations of D_{45} and D_{56} (1.17% and 1.04% respectively) were only slightly larger than those of D_{12} and D_{23} (1.00% and 1.03% respectively). The additional errors in D_{45} and D_{56} which were due to out-of-plane movements were thus small. Most of the errors in the intermarker distances were due to optical problems and these errors were negligible (about 1%).

In summary, the measurement method was found to be accurate, and there was very little out-of-plane movements.

5.3.2 Movements of the Intact Specimens Produced by Simulated Posteroanterior Mobilisation Loads

Figure 5.10 shows the movements of an intact specimen (specimen 07, L2/3, female, aged 55, grade of degeneration=3) produced by the simulated posteroanterior mobilisation loads as appeared in the photographs taken. Table 5.3 summarises the experimental results of the ten specimens tested. It was shown that the most significant movements observed were extension of the specimens and the y-translation of the anteroinferior corner of the vertebral body which was in the superior direction. For an applied force of 90N, i.e. a combined loading of anterior shear of 90N and an

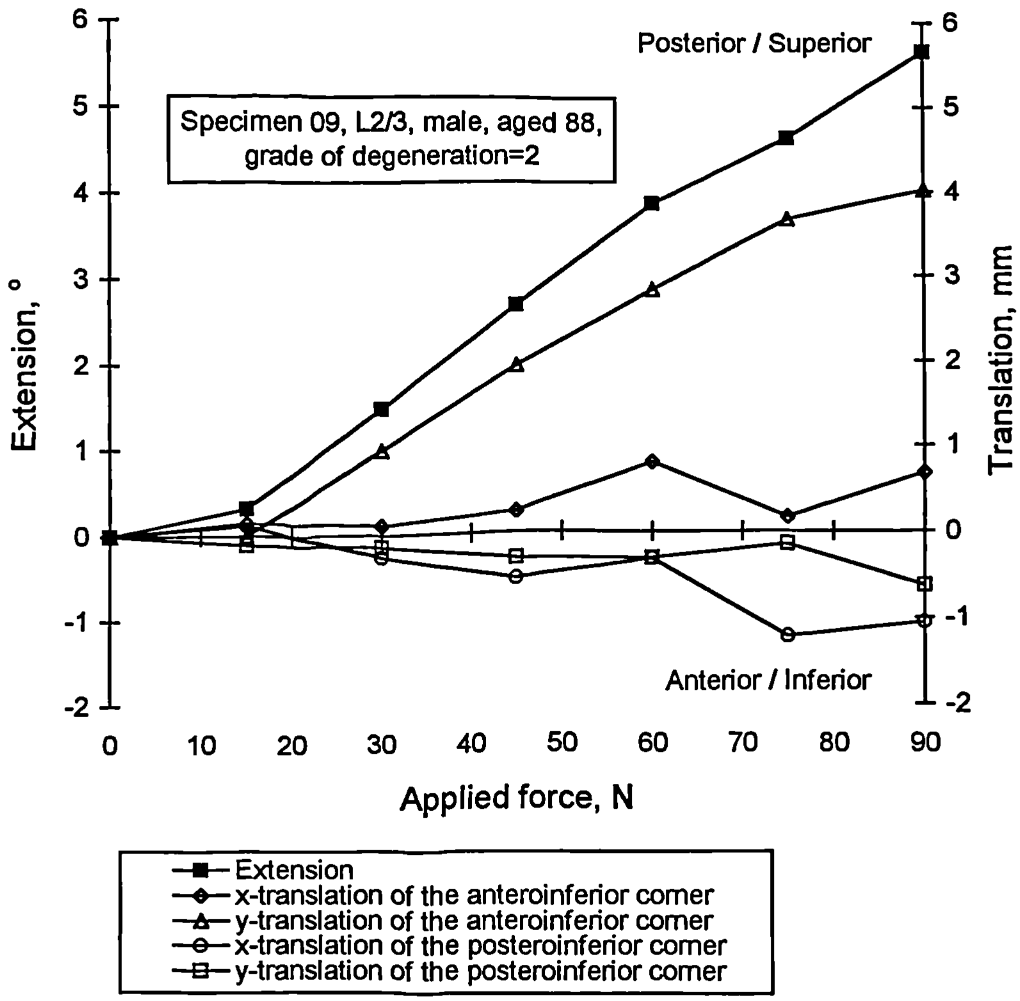


Figure 5.11
The typical movement pattern produced by mobilisation loads in a L2/3 motion segment (specimen 09, male, aged 88, grade of degeneration=2).

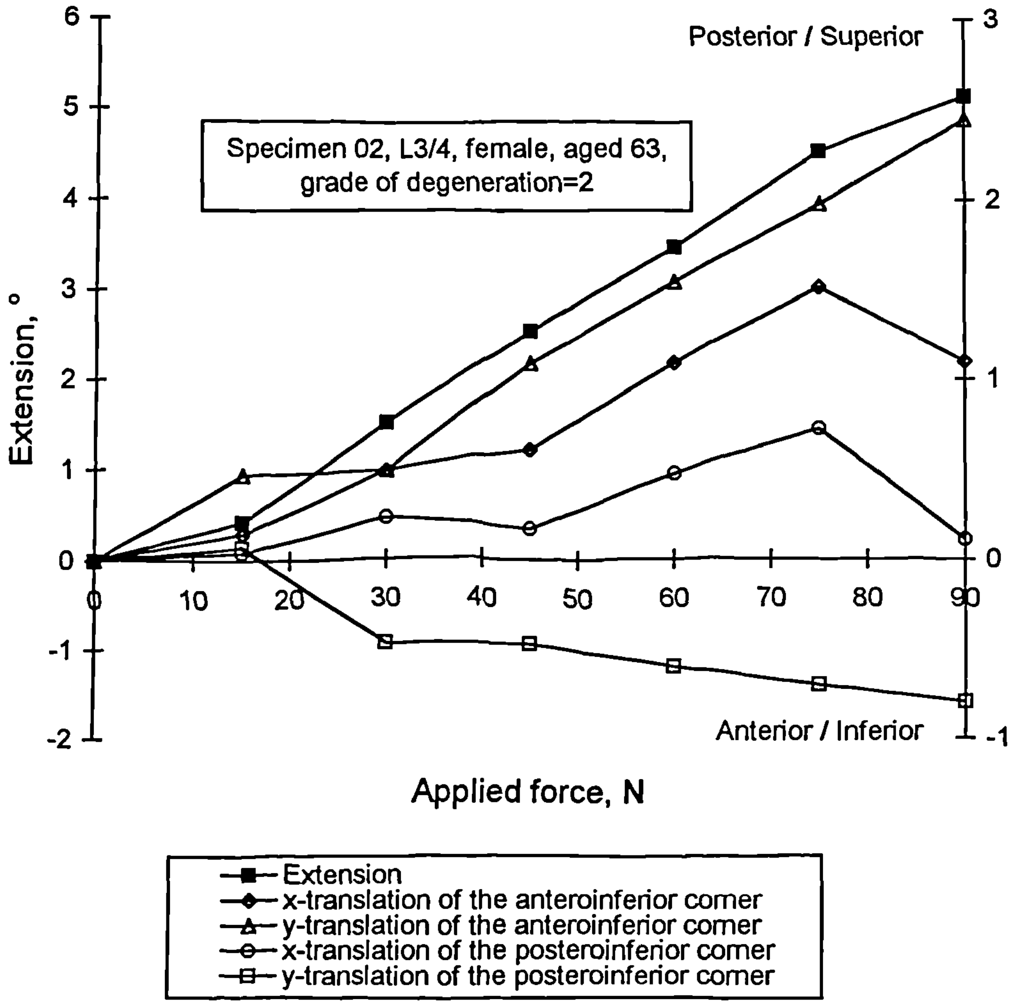


Figure 5.12
The movement pattern of specimen 02 (L3/4, female, aged 63, grade of degeneration=2).

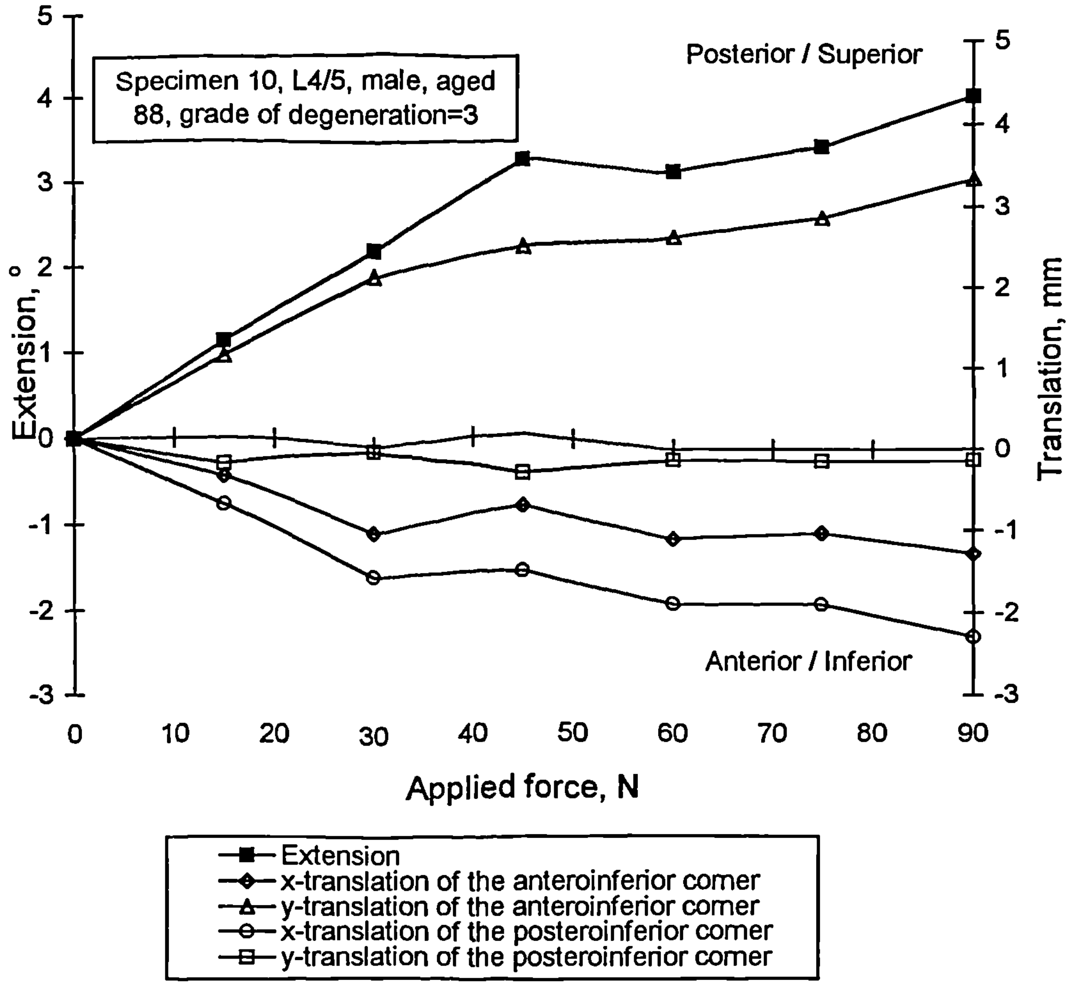


Figure 5.13
The movement pattern of specimen 10 (L4/5, male, aged 68, grade of degeneration=3).

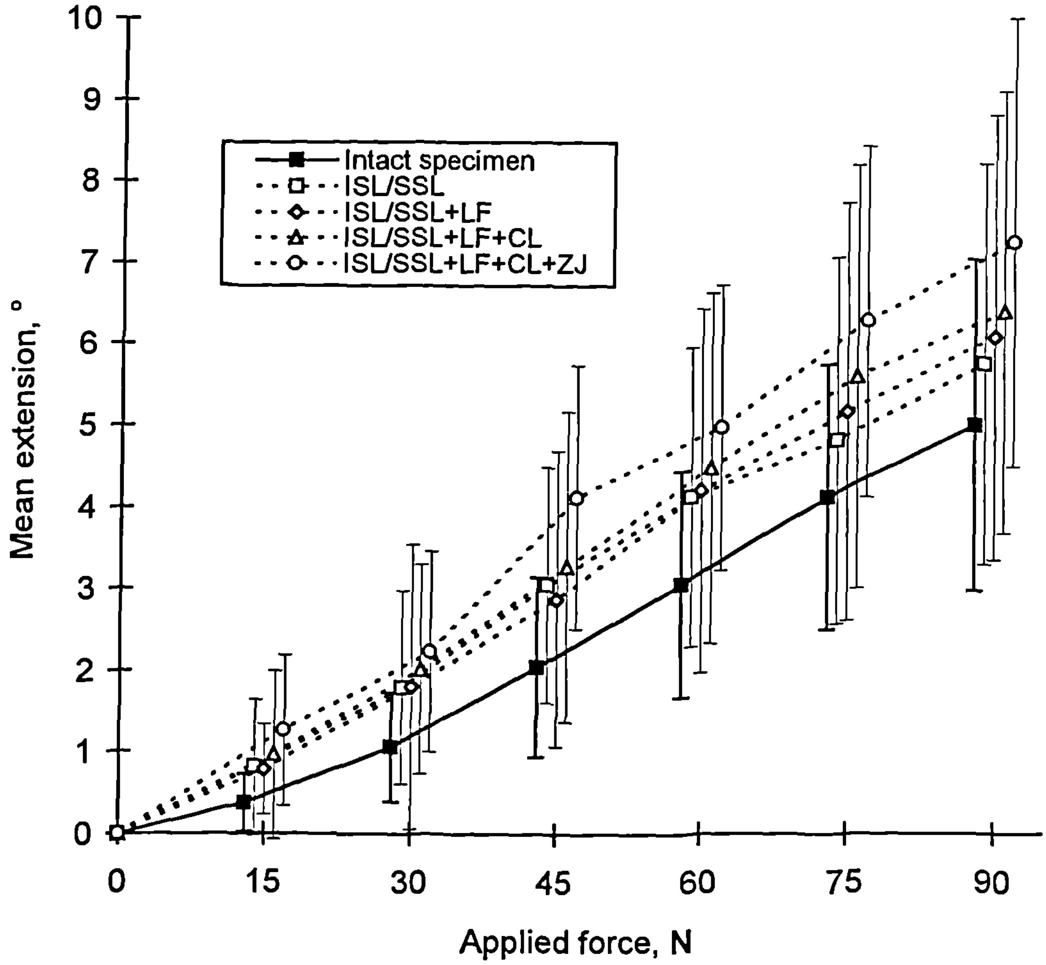


Figure 5.14
Effect of sequential dissections on the mean extension movement. (n=10)
(ISL/SSL - inter- and supraspinous ligaments cut; LF - ligamentum flava cut; CL - joint capsules cut; ZJ - zygapophyseal joints removed)
(The vertical bars denote ± 1 standard deviation for the 10 specimens.)

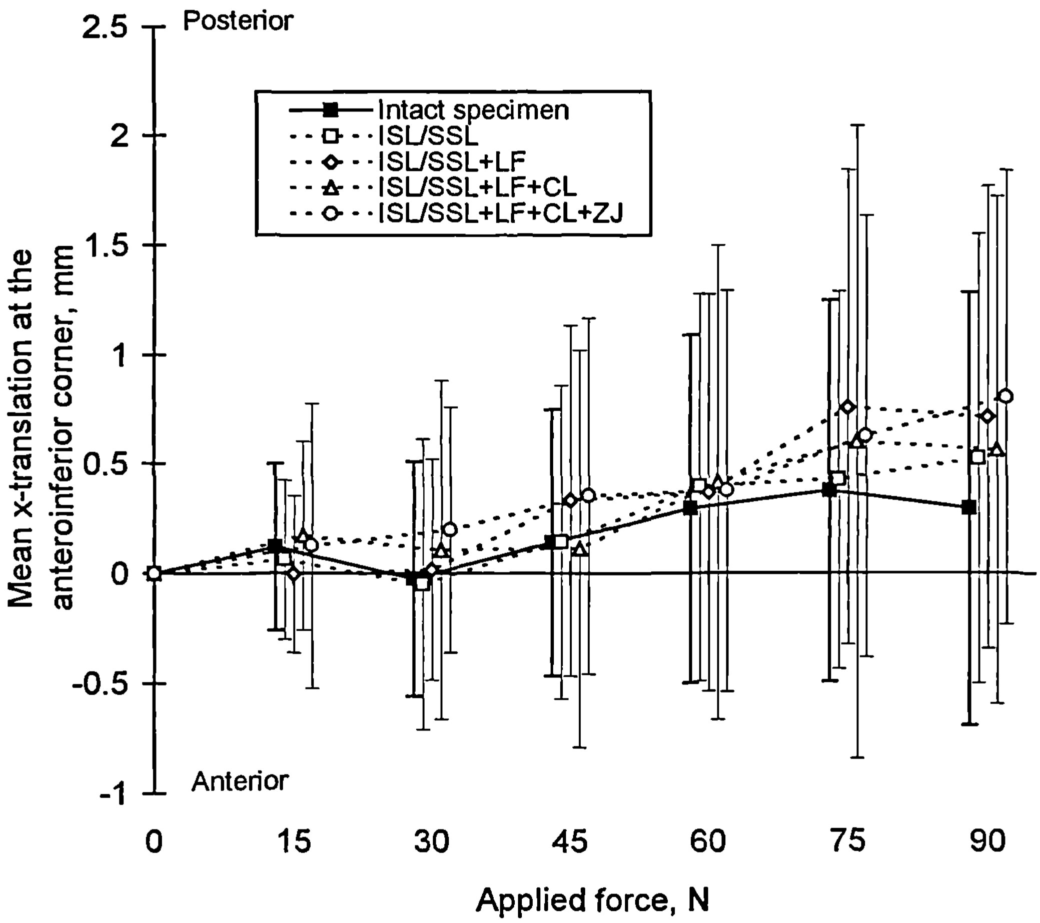


Figure 5.15
Effect of sequential dissections on the mean x-translation of the anteroinferior corner of the superior vertebral body. (n=10)
(ISL/SSL - inter- and supraspinous ligaments cut; LF - ligamentum flava cut; CL - joint capsules cut; ZJ - zygapophyseal joints removed)
(The vertical bars denote ± 1 standard deviation for the 10 specimens.)

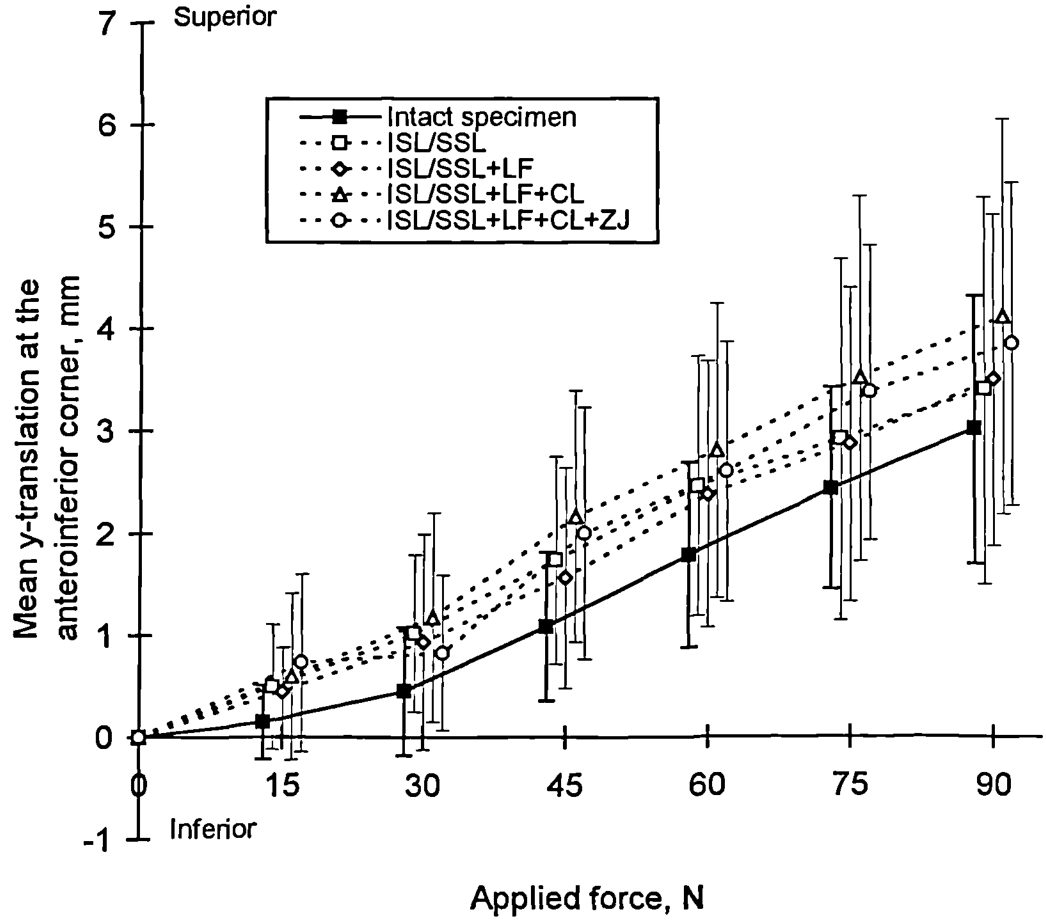


Figure 5.16
Effect of sequential dissections on the mean y-translation of the anteroinferior corner of the superior vertebral body. (n=10)
(ISL/SSL - inter- and supraspinous ligaments cut; LF - ligamentum flava cut; CL - joint capsules cut; ZJ - zygapophyseal joints removed)
(The vertical bars denote ± 1 standard deviation for the 10 specimens.)

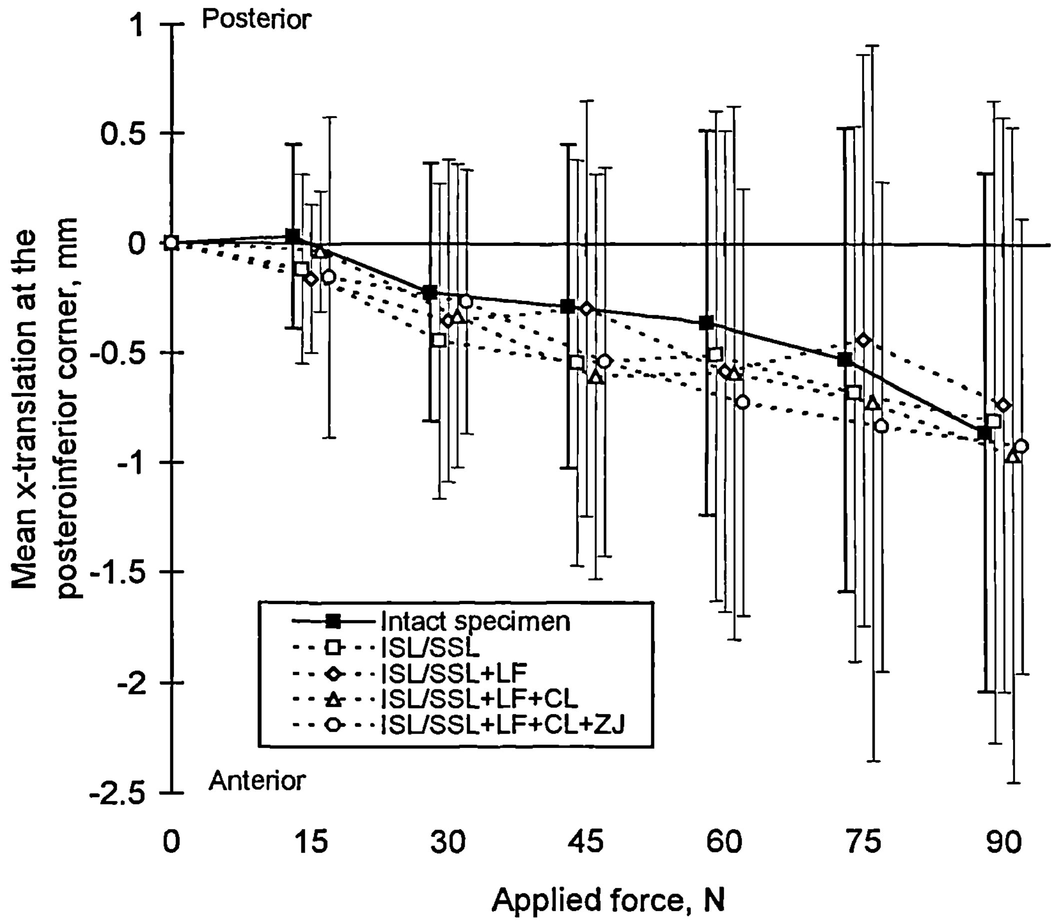


Figure 5.17
Effect of sequential dissections on the mean x-translation of the posteroinferior corner of the superior vertebral body. (n=10)
(ISL/SSL - inter- and supraspinous ligaments cut; LF - ligamentum flava cut; CL - joint capsules cut; ZJ - zygapophyseal joints removed)
(The vertical bars denote ± 1 standard deviation for the 10 specimens.)

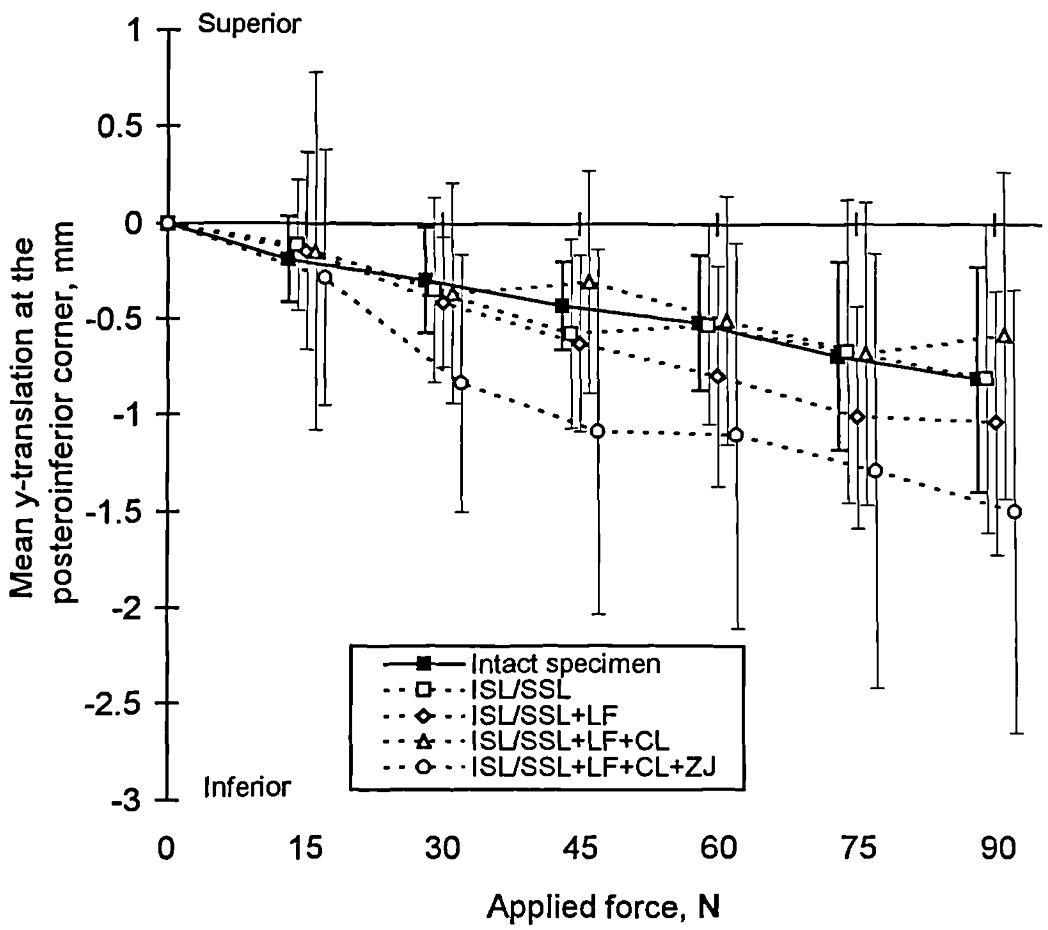


Figure 5.18
Effect of sequential dissections on the mean y-translation of the posteroinferior corner of the superior vertebral body. (n=10)
(ISL/SSL - inter- and supraspinous ligaments cut; LF - ligamentum flava cut; CL - joint capsules cut; ZJ - zygapophyseal joints removed)
(The vertical bars denote ± 1 standard deviation for the 10 specimens.)

extension moment of 9Nm, the mean extension was 4.99° ± 2.03° and the mean superior translation of the anterior corner was 3.00 ± 1.23mm. However, the y-translation of the posteroinferior corner and the x-translations were generally small in magnitude (less than 1mm).

Table 5.3 also shows that there were large variations in the magnitude of movements observed among the different specimens. The variations were most pronounced in the translational motions, and in the case of x-translations, the standard deviations well exceeded the mean values.

The typical movement pattern of an intact L2/3 motion segment (specimen 09, male, aged 88, grade of degeneration=2) is shown in figure 5.11. It can be seen that the specimen extended and this was accomplished by the superior translation of the anteroinferior corner of the superior vertebral body and the inferior translation of the posteroinferior corner. The anterior corner was also found to translate posteriorly and the posterior corner anteriorly.

The directions of sagittal rotation and y-translation movements were consistent in all specimens. However, the directions of x-translations of a few motion segments were different from the pattern of specimen 09 described above. Three specimens (specimen 02, 07 and 08) displayed posterior translation at both the anterior and posterior corners, indicating that the motion segment, as a whole, translated posteriorly. Figure 5.12 shows the movement pattern of specimen 02 for illustration. Two motion segments (specimen 03 and 10) were found to translate anteriorly at both corners showing that the whole specimen translated anteriorly and this is illustrated by specimen 10 in figure 5.13.

5.3.3 Effects of Dissection on the Movements Produced by Simulated Mobilisation Loads

Figures 5.14-5.18 show the effect of dissection of various anatomical structures on the mean sagittal rotation and translation movements produced at

Mean (±S.D.) maximum movements of the motion segments (n=10)

	Extension of the motion segment, °	Translation of the anteroinferior corner of the superior vertebral body, mm		Translation of the posteroinferior corner of the superior vertebral body, mm	
		x-translation*	y-translation#	x-translation*	y-translation#
Intact	4.99 (±2.03)	0.29 (±0.98)	3.00 (±1.31)	-0.86 (±1.18)	-0.81 (±0.59)
ISL/SSL	5.75 (±2.45)	0.52 (±1.02)	3.39 (±1.89)	-0.81 (±1.46)	-0.81 (±0.80)
ISL/SSL+LF	6.07 (±2.72)	0.71 (±1.05)	3.49 (±1.62)	-0.73 (±1.31)	-1.04 (±0.69)
ISL/SSL+LF +CL	6.38 (±2.71)	0.56 (±1.15)	4.11 (±1.93)	-0.95 (±1.49)	-0.58 (±0.85)
ISL/SSL+LF +CL+ZJ	7.24 (±2.75)	0.80 (±1.03)	3.84 (±1.58)	-0.92 (±1.04)	-1.50 (±1.16)

* positive values denote posterior translations and negative anterior translations
# positive values denote superior translations and negative inferior translations

ISL/ SL - inter- and supraspinous ligaments cut, LF - ligamentum flava cut, CL - joint capsules cut, ZJ - zygapophyseal joints removed

Table 5.4
Effect of dissection on the mean maximum movements produced by simulated posteroanterior mobilisation loads.

Results of Analyses of Variance

Movement	Degree of freedom	Between groups mean squares	Within groups mean squares	F ratio	F Prob-ability
Mean maximum extension	4,45	6.80	6.49	1.05	0.39#
Anteroinferior vertebral body corner					
Mean maximum x-translation	4,45	0.38	1.10	0.34	0.85#
Mean maximum y-translation	4,45	1.83	2.83	0.65	0.63#
Posteroinferior vertebral body corner					
Mean maximum x-translation	4,45	0.08	1.71	0.05	0.99#
Mean maximum y-translation	4,45	1.21	0.70	1.72	0.16#

No significant difference

Table 5.5
Results of the analyses of variances of the mean maximum extension and the mean maximum translations at the two vertebral body corners by groups (5 groups: intact, supra- and interspinous ligament cut, ligamentum flavum cut, joint capsule cut and removal of the zygapophyseal joints).

different magnitudes of mobilisation loads. Table 5.4 provides the summary statistics of the maximum movements produced after dissection of each structure.

The mean extension of the motion segments increased after sequential dissection of each anatomical structure (figure 5.14). This trend was consistently observed in all specimens. At the maximum loads applied, removal of the various anatomical structures resulted in a total increase of 2.25° in the mean maximum extension (table 5.4). Dissection of the inter- and supra-spinous ligaments and the zygapophyseal joints produced the largest changes in the movement among the various dissection steps. However, analysis of variance showed that the increases in the mean maximum extension were found to be statistically insignificant ($p=0.39$) (table 5.5).

The effects of sequential dissections on the mean x- and y-translations at the anteroinferior and posteroinferior vertebral corners were small and rather variable (figures 5.15-5.18, table 5.4). No consistent trends were demonstrated among the specimens. Furthermore, as shown in table 5.5, none of the changes in the mean maximum translations were found to be statistically significant ($p>0.01$ in all cases).

5.4 DISCUSSION

5.4.1 Movements Produced by Simulated Posteroanterior Mobilisation Loads

No previous in-vitro study has been made of the intervertebral movements produced by mobilisation loads nor of the effects of dissection on these movements. The present study demonstrated that simulated mobilisation produced extension of the motion segments in all the ten specimens examined. This was principally achieved by the superior translation of the anteroinferior corner of the vertebral body, that is, an increase in anterior disc height. There was also a small magnitude inferior translation at the posteroinferior corner (mean value 3.5 times less compared to that at the anterior corner).

At a load magnitude of 9Nm, the mean extension of the motion segment was found to be 4.99 ± 2.03° in this experiment. This value is comparable to those reported by Schultz et al (1979) and Adams et al (1988). Schultz et al (1979) found that an extension moment of 10.6Nm produced a rotation of 3.00° whereas Adams et al (1988) found that that under a bending moment of 11Nm to 50Nm, 3-7° of extension was observed.

The relatively small difference in the values observed among these studies is to be expected. Previous authors had shown that the differences in behaviour of cadaveric motion segments were large (Markolf, 1972; Schultz et al, 1979). The differences could also be due to the different loading conditions. The present study attempted to simulate posteroanterior mobilisation and thus a combined loading of extension moment and anterior shear was applied. No compressive preload was used as mobilisation is normally carried out in prone lying. The works of Schultz et al (1979) and Adams et al (1988) attempted to simulate physiological extension in the upright posture, and an axial compressive force (400-800N) was applied to the specimens during the tests to account for the effects of body weight. Schultz et al did not apply any shear force whereas Adams et al combined his loads with an anterior shear of 0-150N.

Under the simulated mobilisation loads, the specimens were found to translate superiorly at the anteroinferior corner of the superior vertebra. There was also generally a relatively small amount of inferior translation at the posteroinferior corner. It appears that the extension movement of the segments was accomplished by these movements.

The present study showed that the x- or posteroanterior translations of the specimens were rather inconsistent. Some specimens moved anteriorly (2 specimens) or posteriorly (3 specimens) as a whole, while most (5 specimens) moved posteriorly at the anterior corner of the vertebral body and anteriorly at the posterior corner.

The direction of x-translation at a given point of the vertebra is dependent not only on the anterior shear force applied but also on the accompanying extension movement. The shear force will tend to produce anterior translation, whereas the extension movement will bring about posterior translation. In most specimens, the anterior part of the segment was found to have large deformation due to the extension movement, and would thus be made to move posteriorly. The effect of shear was relatively small at this point. At the posterior part of the segment, there was only very little deformation due to extension. The posterior vertebral corner would be made to move anteriorly by the shear force applied. In specimens where the shear stiffness was low, the translation that might be brought about by the applied anterior shear force would become significant at both corners, and thus the whole vertebra will move anteriorly. On the other hand, in specimens where the shear stiffness was very high, there would be negligible translation due to shear, and the vertebra would move posteriorly as a whole due to the extension movement. Hence, the apparently inconsistent observation in x-translation may be explained by differences in stiffness of the segments in the anterior shear direction.

Large scatter in the magnitude of the movements of the motion segments was observed in this experiment. Previous authors (Schultz et al, 1979) attempted to normalise the segmental motion data on the basis of the morphology of the specimens, such as the disc cross-sectional areas, heights and widths. However, as this did not appear to reduce the scatter of data, such an approach was not adopted in the present study.

5.4.2 Effects of Dissection on the Movements Produced by the Simulated Mobilisation Loads

Dissection of the anatomical structures produced inconsistent changes in the translational movements produced by the mobilisation loads. These changes were small and comparable to the error size reported in section 5.3.1. The structures dissected seemed to play only a small role in resisting the translational movement components under the loading conditions imposed. However, some increase in extension was consistently observed when the various tissues were removed sequentially. The large variations in the data overwhelmed the small changes in extension, and thus statistical tests did not demonstrate these changes to be significant.

Adams et al (1988) demonstrated that dissection of the inter- and supraspinous ligaments reduced the resistance to the approximation of the spinous processes during extension (see section 3.1.3.2). This effect is most marked if the spinous processes are very close to each other as in the extreme of extension. In the present study, the ligaments were also found to provide some resistance to extension. This resistance was not great as the specimens were not loaded to full extension.

The zygapophyseal joints appeared to play a role in resisting the extension movement produced by mobilisation. During extension, the inferior facets moved downwards and force was transmitted from the facet tips to the adjacent lamina or pedicle. Thus, removing the joints would reduce the resistance to extension. However, strong resistance would be produced only when bony contacts occurred during extreme hyperextension. Hence, in the case of simulated mobilisation, the resistance provided by the joints would not be of great magnitude.

The ligamentum flavum and the capsular ligament provided slight resistance to the extension moment produced, although their role appeared to be less significant compared to the inter- and supra-spinous ligaments and the zygapophyseal joints. As discussed in section 2.4.1, ligamentum flavum possessed a small amount of resting tension (Nachemson and Evans, 1968). Removal of this ligament should lead to a small amount of flexion and/or increase in posterior disc height. There would thus be a slight apparent increase in the extension movement. In addition, extension of the motion segment would cause downward sliding of the inferior facets stretching the capsular ligaments (Yang and King, 1984). Thus, removing this ligament would slightly reduce the resistance to extension.

As shown in this experiment, none of the tissues dissected resisted the mobilisation loads to any great extent. After dissection of these tissues, only the intervertebral disc and the anterior longitudinal ligament were left to resist extension. The other soft tissues of the motion segments, that is, the posterior longitudinal and intertransverse ligaments, are unlikely to resist the movement. This is because they are posterior to the centre of rotation which is slightly anterior to the geometric centre of the disc for the extension movement (Rolander, 1966; Gertzbein et al, 1981 and 1984; Seligman et al, 1984; Ogston et al, 1986; Pearcy and Bogduk, 1988).

The role of the anterior longitudinal ligament in resisting the movement was not examined by dissection technique in this study as the ligament blends with anterior annular fibres and it is not technically possible to cut this ligament without damaging the disc. However, a simple calculation of the strain and force in this ligament will reveal that it is too weak to resist the extension movement. The mean resting length of the ligament between the edges of the vertebral bodies was reported to be 12.3mm in the lumbar region (Chazal et al, 1985). The present experiment showed that at maximal loading, the mean x- and y-translations at the anteroinferior corner of the superior body were 0.29mm and 3.00mm respectively. Therefore, the new length of the ligament after loading would be $\sqrt{(12.30+3.00)^2+(0.29)^2}\,\mathrm{mm}$ = 15.30mm and the strain in the ligament (15.30-12.30)/12.30 = 0.24.

The force produced in the anterior longitudinal ligament, F_{lgt}, was then determined using the equation provide by McGill (1988).

$$F_{lgt} = aC\,e^{b\varepsilon} + P$$

where a ($\mathrm{Nmm^{-2}}$) and b are constants, C the cross-sectional area of the ligament ($\mathrm{mm^2}$), ε the strain and P the ligament pretension (N). The values of a and b had not been determined experimentally for the anterior longitudinal ligament. However, previous authors (Shirazi-Adl et al, 1986; McGill, 1988) had assumed the values for the ligament to be the same as those for the interspinous ligament ($a=0.0139Nmm^{-2}$; $b=0.272$) since their elastin-collagen content and stress-strain characteristics were similar after normalisation by their cross-sectional area. The cross-sectional area of the ligament ($66mm^2$) was obtained from the work of Chazal et al (1985) and the pretension value (1.8N) from that of Tkaczuk (1968).

Hence, the force developed in the ligament was $0.0139*66*e^{(0.272*0.24)} + 1.8 =$ 2.8N. If the sagittal diameter of the disc was assumed to be 0.0342m (Shirazi-Adl et al, 1986), the restorative flexion moment generated by the ligament about the centre of the disc would be 2.8*0.0171=0.048Nm. It was thus shown that the anterior ligament had negligible effect in balancing the 9.0Nm moment applied. It should be noted that the above calculation did not take into account the change in disc bulging during extension. The ligament was simply modelled as a line joining the edges of the vertebral bodies. The purpose was not to accurately predict the ligamentous load but to illustrate the fact that the ligament was likely to be incapable of offering significant resistance to the movements induced.

From the above discussion, it may be concluded that none of the structures examined appears to provide resistance to any of the motion components to a great extent. The intervertebral disc is the principal structure resisting the mobilisation loads. The result implies that degeneration and injuries of the disc will lead to an increase in the movement produced by mobilisation, and scarring of the annular fibres which occurs during healing will increase the posteroanterior stiffness. The clinical implication of this study will be fully addressed in chapter 9.

5.4.3 Errors Involved in the Measurements of Movement

The error analysis reported in section 5.3.1 showed that the experimental technique was sufficiently reliable for the present analysis. Errors due to digitisation and optical errors of the camera and the slide projector (due to malalignment and lens distortion) were shown to be minimal.

Determination of the positions of the most anterior and posterior parts of the two endplates with respect to the markers appeared to be the major source of error in this experiment. However, such error was consistent because the same relationship between the markers and the anatomical landmarks measurement was used to compute the locations of the landmarks in the various photographic negatives. The error would not therefore alter the movement patterns observed in the present experiment.

Analysis of the error in the distances between markers reveals that there were very little out-of-plane movements in this experiment. This supports the assumption that posteroanterior mobilisation produced movements which are basically confined to the sagittal plane and thus a three-dimensional analysis to compute movements in the other planes is considered unnecessary.

Only two markers are required to calculate the extension rotation of the specimen. Since three markers had been attached to the superior vertebra of the specimen, three independent calculations of the extension movement are possible. These calculations should produce the same answers if the spatial relationship between the markers on the sagittal plane is the same and there are no out-of-plane movements. In the present study, the three calculations were averaged to reduce the error involved in computing the extension movement.

In theory, the moment applied to the specimens would be smaller than expected as the beam of the upper loading jig rotated under loading. However, this error would be minimal as the magnitude of the extension movement was small. At maximal loading which produced an extension of 4.99°, the actual moment applied would be cosine(4.99°) of the nominal load, that is, 8.97Nm. This constitutes an error of less than 0.4%.

Another source of error would be relative movement between the specimen and the moulding cup. Ideally, the markers should be rigidly attached to the specimens, but this was not possible in the present experimental set up due to the inaccessibility of the mounting cups. Since vertebral movements were computed by measuring the movements of the cups, it was important to obtain rigid fixation of the specimens and this was achieved by the use of locking screws and Isopon producing a strong mould which did not yield upon loading. The specimens were also carefully inspected after application of each load increment for signs of slippage which might be accompanied by an audible sound and a large increase in displacement.

5.4.4 Experimental Validity

In the present experiment, simulated mobilisation loads were applied to the motion segments. This involved the application of 90N anterior shear at a rate of $75Ns^{-1}$ and a corresponding extension moment of 9Nm. These parameters were chosen to be similar to those used in clinical practice. Since the magnitude of loads involved was small, the weight of the loading jigs alone would have applied relatively large loads to the specimens. The test apparatus employed weight and pulley systems to negate such loading effects.

The various anatomical structures were dissected in a predetermined sequence in this experiment. The resistance provided by a given structure may change when another tissue has been dissected. Thus, the order of dissection should ideally be randomised in different specimens to eliminate such problem. However, this is not possible in the present experiment. The scalpel cannot reach the ligamentum flavum until the inter- and supraspinous ligaments are dissected. Sawing the zygapophyseal joints will remove their capsules at the same time. Therefore, the dissection has to be carried out in a particular order. In the present experiment, the change in movement after each dissection step was small, and thus a change in the dissection order would be unlikely to alter the observed pattern of change in movement.

The experimental results should be interpreted in the light of post-mortem changes in the mechanical properties of biological tissues (Keller et al, 1990; Johnstone et al, 1992) (see section 3.1.1). Great care was taken to minimise such changes. The specimen storage method had been shown not to influence the properties significantly (Sedlin and Hirsch, 1966; Hirsch and Galante, 1967; Tkaczuk, 1968; Panjabi et al, 1985) and Cling film was used to maintain the water content of the tissues which had significant effects on the mechanical properties (Galante, 1967; Hirsch and Galante, 1967).

One limitation of the present study was that it was performed on isolated motion segments and the mechanical behaviour of the whole lumbar spine in vivo would likely be different. In the living spines, the structural integrity was not destroyed and the motion segments move within the constraints imposed by their neighbours. Chapter 6 will describe a separate experiment which examined the in-vivo behaviour. Nevertheless, testing of cadaveric motion segments has enabled dissection techniques to be employed to help determine the role of the anatomical structures in resisting the mobilisation load.

5.5 CONCLUSION

The motion segment study had provided important data on the vertebral movements produced by posteroanterior mobilisation. The experimental results refute the general belief that posteroanterior mobilisation is a simple "gliding" of one vertebra upon another, and the movement pattern produced is rather complex. The principal movement was found to be extension. This was accompanied by an increase in anterior disc height and a relatively smaller decrease in posterior disc height. There was only very small amount of translation along the posteroanterior direction and the trend was rather inconsistent. Most specimens showed posterior translation at the anteroinferior vertebral body corner and anterior translation at the posteroinferior corner, but in some cases, the vertebra translated anteriorly or posteriorly at both corners.

The study had also produced data which should provide a better understanding of the anatomical basis of posteroanterior mobilisation. It was shown that the intervertebral disc is the structure that provides the major resistance to the movements produced by mobilisation. The other soft tissues were found to resist the movements to a small extent.

The photographic technique employed in this study had been shown to be accurate and repeatable for examining small magnitudes of movements. A similar method was thus used in a later study for measuring the intervertebral movements produced by lumbar traction (see chapter 7). One limitation of the present study is that it may not truly reflect the in-vivo situation when the whole lumbar spine is subjected to mobilisation. This was examined by a separate experimental study which will be discussed in chapter 6.

■ *Chapter 6*

An In Vivo Study of the Intervertebral Movements Produced by Posteroanterior Mobilisation

---

An In Vivo Study of the Intervertebral Movements Produced by Posteroanterior Mobilisation

6.1 INTRODUCTION

Earlier studies (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and Evans, 1991 and 1992) employed displacement transducers to record the vertical displacements of skin overlying spinous processes as measurements of posteroanterior stiffness (see section 4.1.4). Such a surface technique was subjected to errors due to skin deformation and rotation of the vertebra, and did not provide data on the true intervertebral movements produced. The study in chapter 5 had studied these movements in isolated motion segments, but the movements produced in vivo are likely to be different. The cadaveric study also did not reveal the movements of the adjacent motion segments which were not directly mobilised.

The present work would therefore further examine the mechanical effects of posteroanterior mobilisation in living subjects. The purpose of the investigation was to measure

1. the sagittal rotation and translations of the lumbar and lumbosacral motion segments, and
2. the displacements of the lumbar spinous processes

when the L4 vertebra was subjected to posteroanterior mobilisation.

Radiographic measurements were employed to measure the above movements at an intervertebral level, and special precautions were taken to enhance the accuracy. The cadaveric study described in chapter 5 confirmed the assumption that movements

Subject's initial	Age, years	Body height, m	Body weight, N
FTC	22	1.65	608.2
HYS	42	1.60	696.5
PCY	20	1.68	637.7
LCK	21	1.83	735.8
MKM	36	1.62	657.3
LKS	21	1.72	539.6
WMC	29	1.68	588.6
LKC	20	1.65	500.3
TCK	25	1.62	569.0
YIK	32	1.75	784.8
YAU	33	1.64	735.8
CHF	22	1.62	686.7
Mean	26.9	1.67	645.0
Standard Deviation	7.4	0.07	86.9

Table 6.1
The age, body weight and height of the subjects.

produced by posteroanterior mobilisation were principally confined to the sagittal plane. It was thus considered sufficient to take lateral radiographs only and measure the sagittal movements from the data derived.

6.2 SUBJECTS

Twelve male normal volunteers under the age of 45 were recruited for the study. Subject age, height and weight are shown in table 6.1. Inclusion in the study required a normal physical examination of the lumbar spine. Subjects were excluded if they had

- any history of back pain, leg pain, anaestheisa or paraesthesia which had necessitated their seeking advice (medical, physiotherapeutic, osteopathic, chiropractic, naturopathic, from Chinese bonesetter or acupuncturist, or from any practitioners) or time off from work within the past 12 months,
- past history of fracture, dislocation and surgery of the lumbar spine,
- any medical conditions which may affect the musculoskeletal system,
- any radiographic abnormalities subsequently found in the study, including spondylolisthesis, narrowed disc spaces, osteophytes, transitional lumbosacral vertebrae,
- any history of significant radiation exposure within the last 12 months, and
- any disorders that may contraindicate spinal mobilisation

The experimental procedure was explained to all the subjects accepted and their written consent obtained before the commencement of the radiographic investigation.

Approval was obtained from the Ethics Committee of the Hong Kong Hospital Authority to perform the work which involved the exposure of normal individuals to x-rays. The study was carried out at the Radiography Department of the Duchess of Kent Orthopaedic Hospital, Sandy Bay, Hong Kong. The potential risks associated with radiation were fully explained to the subjects and every effort was made to

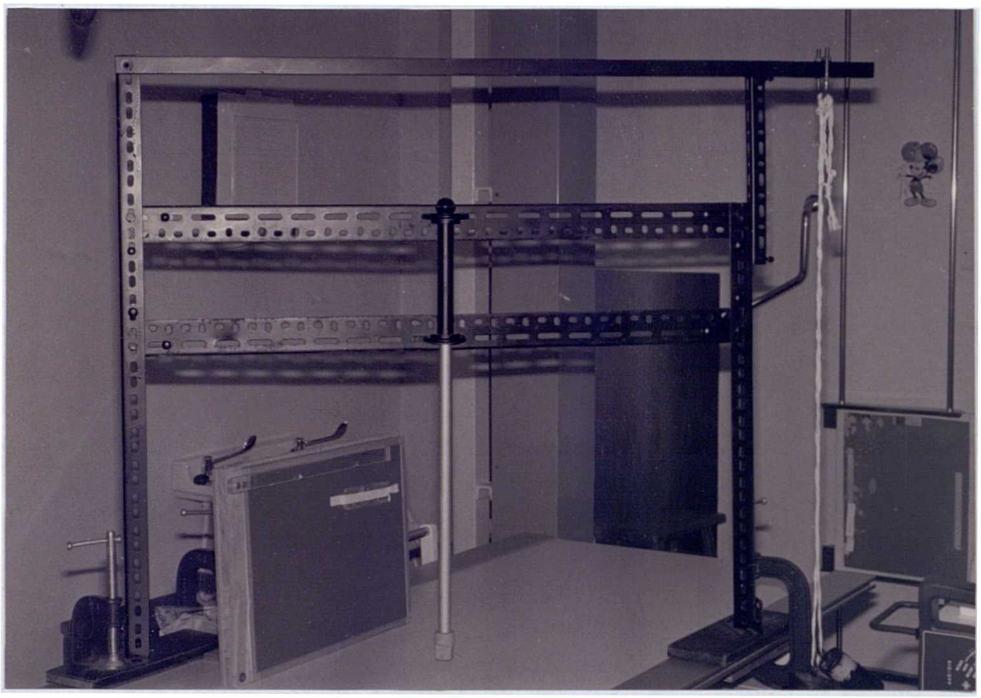


Figure 6.1.
The loading mechanism which comprised of a loading frame and an applicator.

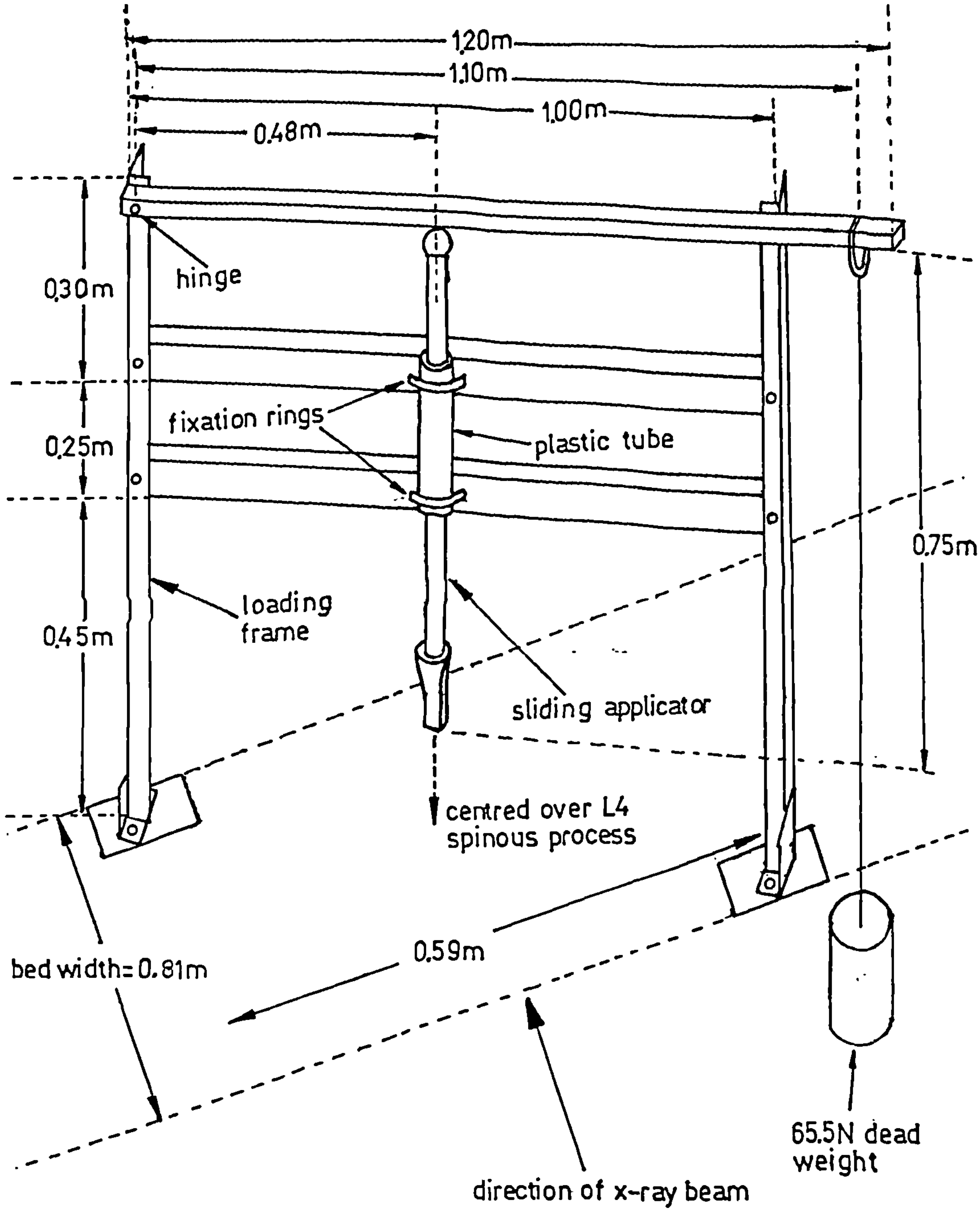


Figure 6.2
A schematic diagram showing the dimensions of the loading frame and the applicator.

minimise the radiation used. The x-ray beam was tightly coned to the lumbosacral spine and gonadal shields were applied to all the subjects. The study was limited to a small sample of male subjects only as the reproductive organs could not be effectively protected in females. Each subject experienced only two standard lateral exposures (Each exposure was taken at about 90kV and 40mAs at a focus-film distance of 1.00m, equivalent to a skin entry dose of approximately 1×10^{-3}Gy each).

6.3 METHODS

6.3.1 Apparatus

Posteroanterior mobilisation force was applied to the L4 spinous process by a specially designed loading mechanism which consisted of an applicator supported by a loading frame (figures 6.1 and 6.2). The applicator was allowed to slide freely and vertically inside a plastic tube which was held in place by two metallic rings. The lower end of the applicator was fitted with a tapered plastic adaptor which was rectangular in cross-section at its end. The shorter longitudinal length of the cross-section (1.5cm) enabled the load to be localised to one specific spinal level and the longer transverse length (2.5cm) provided a large surface area. The use of a soft pad helped minimise the discomfort caused to the subject. A metal pin was inserted into the inferior end of the adaptor. This would appear on the radiograph so that the investigator could check if the applicator was centred over the desired spinous process.

The loading frame was made from Dexion (L-shaped metallic beam with a cross section of 0.058mx0.030m). It had three horizontal beams. The lower two were 1.00m long and rigidly supported by two vertical columns of 1.00m in height. The uppermost was 1.20m long and could be moved about a hinge at one of the columns. The vertical columns had wide wooden bases so that they could be rigidly clamped onto the sides of the bed on which the subject received the mobilisation load.

The lower two horizontal beams allowed fixation of a plastic tube through which the loading applicator could freely slide. The plastic tube was fixed onto the beams by two metallic rings to ensure that the applicator moved vertically. The lowest beam was 0.45m from the bed surface so that there was enough room for the subject to get underneath. The beams were made longer than the width of the bed (0.81m) so that the frame could be arranged diagonally across the bed. This produced a distance of 0.41m between the vertical poles on the lateral view and ensured that the poles would not be blocking the lateral x-ray beam.

The uppermost moving horizontal beam applied the posteroanterior force to the spine via the applicator. The top end of the applicator was fitted with a hemi-spherical adaptor so as to minimise friction between the beam and the applicator, and to ensure that the force was applied along the centre of the applicator at all times. One end of the beam was loaded with a dead weight of 65.5N. The weight was not applied directly onto the applicator but at the free end of the horizontal beam. This ensured subject safety in case the weight were to drop.

The applicator axis and the point of weight suspension were at horizontal distances of 0.48m and 1.10m from the hinge. By taking moments about the hinge, it can be shown that a force of 65.5x1.1/0.48=150N was applied through the applicator. This was about the same as the maximum force that would be used clinically, as estimated by Thompson (1983) and Lee (1990).

The x-ray tube used was General Electric Portable Radiographic Machine AMX-4 (General Electric Co., Wisconsin, USA). The tube permitted adjustment in all directions and could be locked in the desired position. It could also project a light beam with a cross wire image which helped alignment. Fuji RX medical x-ray film (Fuji Photo Film Co., Tokyo, Japan) was used and enclosed in an Okamoto 30cmx40cm cassette (Okamoto Manufacturing Co., Tokyo, Japan) which was equipped with a Buckey grid to reduce scattering and an intensifying screen to reduce the necessary exposure to radiation. The film-screen speed was 210ASA.

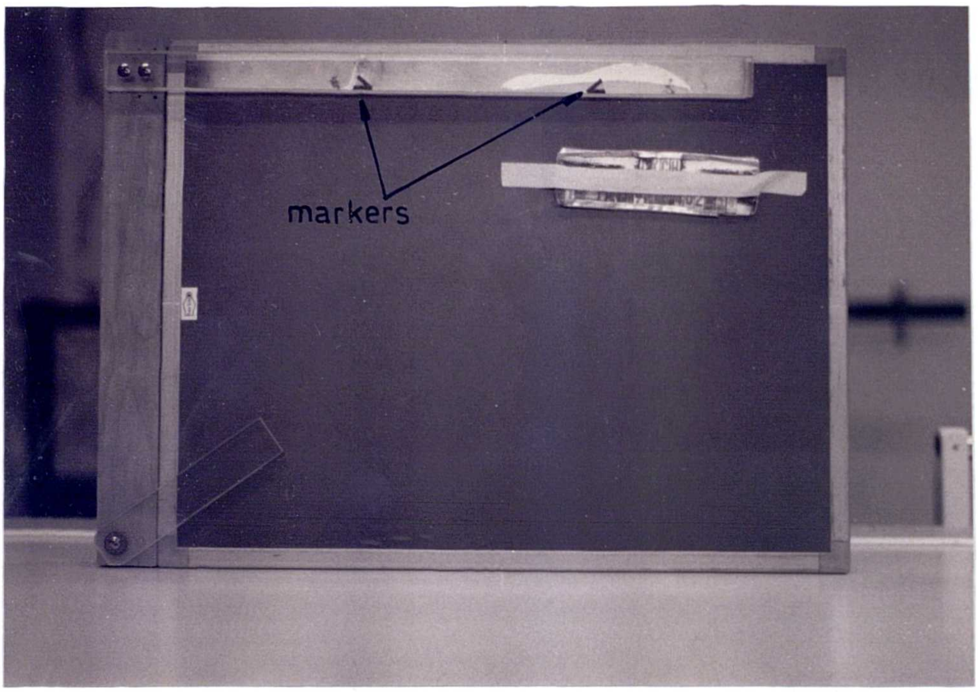


Figure 6.3.
The x-ray cassette holder.
It had two arms which were fixed in position. Reference markers were attached to the upper horizontal arm.

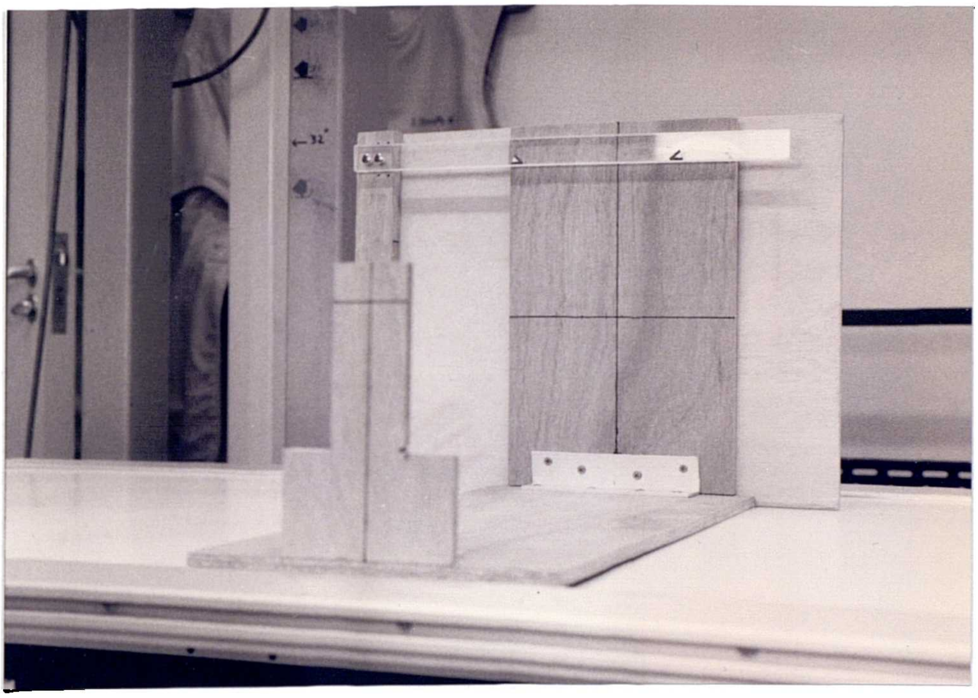


Figure 6.4.
The device used for alignment of the x-ray tube.

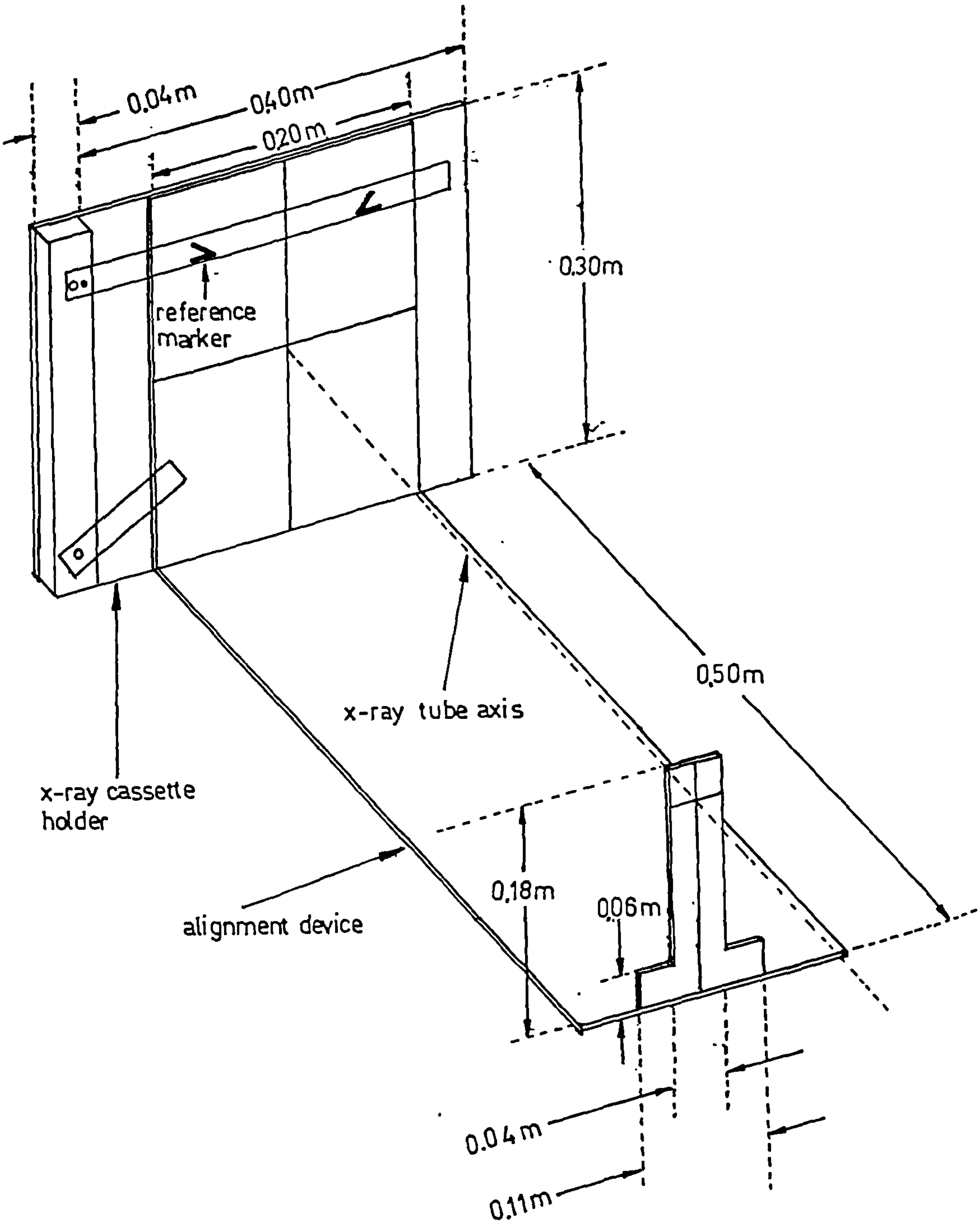


Figure 6.5
A schematic diagram showing the dimensions of the x-ray cassette holder and the alignment device. The dotted line denotes the axis of the x-ray tube which is in line with the centres of the two crosses on the vertical plates of the alignment device.

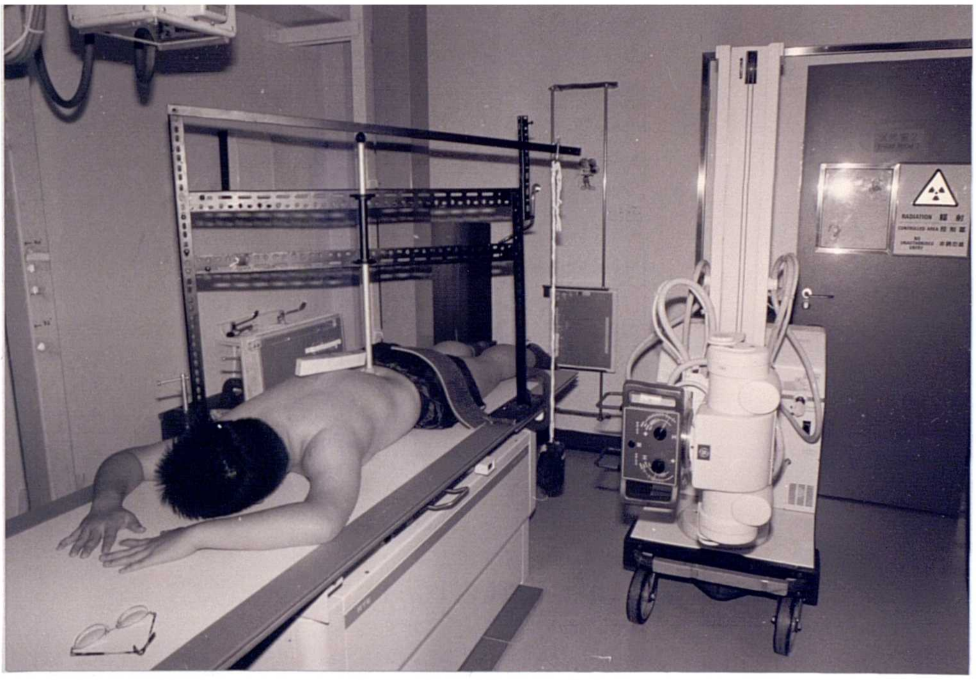


Figure 6.6.
The experimental set up.

A wooden x-ray cassette holder (figure 6.3) was specially made for this study. It was rigidly clamped onto the side of the bed and had two fixed plexiglass arms that locked the cassette securely in place. Two radioopaque markers were attached to one of the holding arms in front of the x-ray cassette to define the horizontal axis and served as reference points. All radiographic measurements were made in relation to them. Any errors due to different positioning or tilting of the cassette among different exposures were thus avoided.

An alignment device (figures 6.4 and 6.5) was made to ensure that the x-ray tube was perpendicular to the cassette. It was important to align the tube properly in order to avoid producing a distorted image of the spine. The device consisted of two vertical wooden plates which were arranged parallel to each other and had crosses drawn on them. A line joining the centres of the two crosses as shown in figures 6.4 and 6.5 was perpendicular to the plates. The device was placed directly in front of the x-ray cassette holder so that the crosses of the plates were over the centre of the cassette. Therefore, when the cross-wire image of the x-ray tube overlapped with the two crosses, the tube would be perpendicular to the cassette and the x-ray beam directed at the centre of the cassette.

Figure 6.6 shows the experimental set up when mobilisation load was applied to one of the subjects.

6.3.2 Procedure

The x-ray cassette holder was fixed to one side of the bed by clamps and the x-ray tube placed on the other side. The focus-film distance was 1.00m. As described in section 6.3, the light beam from the x-ray tube and the alignment device were used to position the tube so that it was perpendicular to the cassette. The alignment device was then removed and the x-ray cassette inserted into the holder.

The loading frame was placed over the bed. Its position was adjusted until the applicator was at the level of the middle of the x-ray cassette. The investigator checked to ensure that the vertical columns of the frame would not be blocking the x-ray beam. When the position of the frame was found to be satisfactory, it was locked in place by two clamps.

After the experimental procedure was explained to the subjects, they were requested to lie face down on the bed under the loading frame as shown in figure 6.5. The arms of the subjects were fully flexed since they would block the x-ray if they were by the sides of the trunk. The L4 spinous process was carefully identified by palpation and its location marked on the skin. The position of the subject was adjusted so that the applicator was directly over the marked spinous process. The L4 spinal level was chosen for this study as it is a site commonly implicated in low back pain (Grabias and Mankin, 1980).

A scaling bar with radioopaque markers attached was also placed over the spinous processes of the subject just above the level of the applicator. This allowed the magnification of the radiographic image to be determined. The magnification factor was found to be about 1.2.

With the help of the light beam from the x-ray tube, the x-ray beam was coned down to include only the five lumbar vertebrae and the upper half of the sacrum. The area of exposure was minimised so as to reduce the amount of radiation used and also the scattering of the x-ray beam which could degrade the radiographic image. The subject was instructed to take a deep breathe, exhale completely and then hold the breathe while the lateral radiograph was taken. It was found that with this set up, a dose of 90kV and 40mAs would generally produce a good image. The dose was adjusted for subjects of extreme body builds.

The radiograph was developed immediately. Any radiographic abnormalities were then screened. One of the subjects recruited was found to have an old untreated fracture of the L4 vertebral body and subsequently excluded from the study. The investigator also checked if the radiographic image of the metal pin inserted into the inferior end of the applicator was accurately located over the L4 spinous process. In

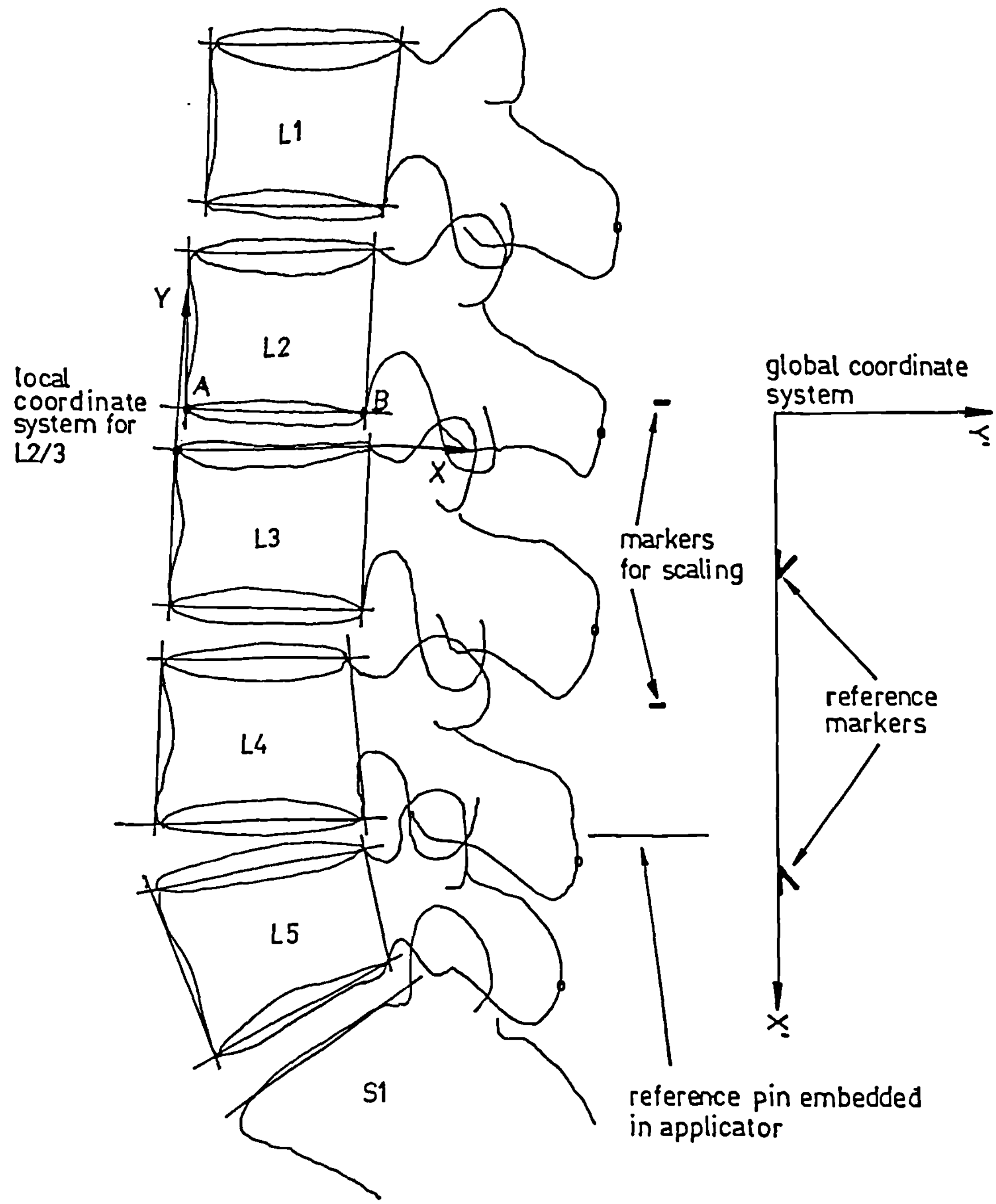


Figure 6.7
Tracing of a typical radiograph.
The four corners of the vertebral body and the most posterior parts of the spinous processes are marked. The markers for scaling and the reference markers of the cassette holder, and the local and global coordinate systems are also shown in the figure.

most cases, location by palpation was found to be accurate, and in cases where the spinous process was not properly identified and the applicator was malaligned, the position of the subject was adjusted and the above procedure repeated.

The dead weight was then attached so as to apply the 150N posteroanterior force to the L4 spinous process. The spine was allowed to creep for 3 minutes. The earlier work of Lee (1990) showed that no significant deformation occurred after this period of creep. A second lateral radiograph was taken in full exhalation at the end of the 3 minute period and the load was removed immediately afterwards. Between the two radiographic exposures, the positions of the x-ray tube, the cassette holder and the subject were not disturbed. All the x-ray films in this study were taken by qualified radiographers.

6.3.3 Radiographic Measurement Method

Intervertebral movements were computed by comparing the positions of anatomical landmarks on the two radiographic images of the vertebra before and 3 minutes after the mobilisation load was applied. The technique of tracing and superimposition of the radiographic images was adapted from the method employed by earlier authors which had been shown to be reasonably accurate (Dvorak et al, 1991a and 1991b; Panjabi et al, 1992) (see section 3.1.2.3). Figure 6.7 shows the tracing of a typical radiograph.

The four corners of the vertebral body were chosen to be the reference landmarks. As these corners appeared to be rounded, the following technique was employed in order to locate them precisely. Four lines were drawn around the vertebral body image with a fine (0.1mm) transparency marker (figure 6.7). Each line was drawn as a tangent to the two most prominent points on one of the vertebral body margins. The four lines formed a quadrangle which represented the vertebral body and the intersections of these lines were taken to be the corners of the body.

The process of fitting a quadrangle was not possible in the case of the S1 vertebra as it did not have a quadrangular shape. The anterosuperior and posterosuperior corners of this vertebra, which served as the reference points, were identified as the first point of inflection of either the anterior or posterior margin of the vertebra where it turned towards the superior margin (figure 6.7). A line was then drawn joining these two corners. Thus, only two reference points were employed for the S1 vertebra.

Marking of the vertebral images as described above was first performed on the radiograph taken before the application of the posteroanterior load. The second radiograph taken after load application was then placed on top of the marked radiograph over an x-ray film view box. The two images of each vertebra were carefully superimposed and the four intersection points of the image of the first radiograph were traced onto the second one. Since the images were sometimes not absolutely identical, a decision had to be made about what constituted the best superimposition of the two images and the following criteria were adopted. The superior and inferior margins of the vertebral body were superimposed as closely as possible while the anterior margins were fully superimposed. The posterior margins were less clearly evident and therefore not used as a criterion.

The vertical displacements of the five lumbar spinous processes were examined in this study. These would show the change in back surface configuration and that of L4 would directly reflect the movement of a therapist's hand were it to apply the posteroanterior force. The most posterior part of the spinous process was rounded, ill-defined and difficult to be marked consistently in different films. This had been shown by Rab and Chao (1977) who reported that identification of the spinous process on the radiograph was subjected to unacceptably large error (see section 3.2.1.1). Therefore, the most posterior point of a spinous process was visually identified and marked only on the first radiograph (figure 6.7). Its identical position in the second radiograph was then predicted mathematically based on the change in the positions of the vertebral body corners (see section 6.4.2). This avoided the difficulty in identifying the same point on the second radiograph.

In fact, the images of the vertebral body and the spinous processes were not marked directly on the radiographs, but actually on a transparent film placed on top of it. The film was removed after the marking was completed. The process of marking and superimposition was repeated two more times with other transparent films. This procedure helped reduce the errors in measurement by providing three independent data sets.

Twenty seven anatomical landmarks (4 for each lumbar vertebral body, 2 for the sacrum and 5 for the lumbar spinous processes) were marked on the transparent film for the first radiograph and twenty two landmarks for the second radiograph (excluding the spinous processes). The following points were also marked on each film (figure 6.7):

- the positions of the two markers of the scaling bar
- the positions of the two reference markers attached to one of the arms of the x-ray cassette holder (i.e. horizontal reference)

There were thus altogether 31 points marked for the first film and 26 for the second film.

The transparent film was placed on top of a digitising tablet (Digi-pad Type 5A, Gtco Corporation, Columbia, Maryland, USA) and the various points were digitised in a specified order.

The digitisation process was repeated three times. Hence, together with three times of marking and tracing, nine sets of data were obtained for each subject. Theoretically, if there were no errors, the intervertebral movements calculated from different sets of data would be identical. Multiple sets of data would thus permit an estimation of the errors involved and averaging of data would reduce the errors.

6.4 TREATMENT OF DATA

Data analysis was performed by the computer program "PAXRAY.PAS" (Appendix II). The first step in the analysis was to transform the coordinates of all the digitised points from the digitiser coordinate system to a global coordinate system (the X'Y' system) established by the two reference markers of the cassette holder (see figure 6.7). This was performed by using the coordinate transformation equation (A1.2) described in appendix I. The X'Y' system was fixed in space with the x'-axis formed by the horizontal line joining the two reference points and the y'-axis vertical.

The coordinates of the digitised points were scaled according to the magnification factor computed from the coordinates of the two markers of the scaling bar.

6.4.1 Computation of Intervertebral Movements

The local coordinate system of the motion segment which was used for computing the intervertebral movements is described in appendix I (see figure A1) and also shown in figure 6.7.

6.4.1.1 Sagittal rotation

The mathematical procedures used in determining the intervertebral sagittal rotation are explained in section A1.3 of Appendix I. Two intersection points of the quadrangle which represented the superior vertebra were required for the *computation. The movement was* determined from the change in the inclination of the line joining these two points.

Since four reference points (the four intersection points of the quadrangle) were available, six different computations of the rotational movement could be generated. As described earlier, nine sets of coordinate data are produced for each subject as a result of repeated marking and digitisation. Hence, in total, 54 independent computations of sagittal rotation were performed and these results were averaged to provide the best evaluation of the movement.

6.4.1.2 Sagittal translations

The coordinates of the anteroinferior and posteroinferior vertebral body corners with respect to the local coordinate system of the motion segment were determined using the coordinate transformation equation A1.2 (Appendix I). The posteroanterior (x) and superoinferior (y) translations at the two corners were then determined from the differences in their x- and y-coordinates between the unloaded and loaded conditions. The mathematical procedures are described in section A1.4 of Appendix I.

Since there were nine data sets as a result of repeated marking and digitisation, nine independent computations of the x- and y-translations at the anteroinferior and posteroinferior vertebral corners were performed. These computational results were averaged to reduce the error size.

6.4.2 Displacements of the Spinous Processes

The horizontal and vertical displacements of the spinous processes were determined from the differences in the horizontal (x') and vertical (y') coordinates of most posterior part of the spinous processes between the first and second radiographs.

For the first radiograph, the locations of the spinous processes were marked, and their coordinates with respect to the global X'Y' system were measured by digitisation (see section 6.3.3). Their coordinates in the second radiograph were obtained by mathematical prediction. Assuming that the vertebra is a rigid body, there is a fixed spatial relationship between the positions of the spinous process and the corresponding vertebral body. The relationship was determined mathematically from the first radiograph where both the spinous process and the vertebral body were marked. With a knowledge of this relationship, the new position of the spinous process in the second radiograph was determined based on the new position of the

Sagittal rotation, °

Spinal segment	Mean	SD	Min	Max
L1/2	-1.68	1.12	-4.15	-0.06
L2/3	-2.39	1.10	-5.26	-1.25
L3/4	-1.18	1.06	-3.05	0.18
L4/5	-1.23	1.23	-2.81	0.86
L5/S1	0.71	1.65	-1.79	4.17

Translations at the anteroinferior corner, mm

	posteroanterior				superoinferior			
Spinal segment	Mean	SD	Min	Max	Mean	SD	Min	Max
L1/2	0.75	0.42	0.20	1.55	0.40	0.63	-0.63	1.35
L2/3	0.43	0.55	-0.25	1.31	1.00	0.59	-0.12	1.90
L3/4	0.29	0.49	-0.43	1.22	0.50	0.60	-0.54	1.81
L4/5	0.17	0.39	-0.58	1.08	0.56	0.51	0.07	1.46
L5/S1	-0.75	0.86	-2.26	0.66	-0.02	0.66	-0.83	1.47

Translations at the posteroinferior corner, mm

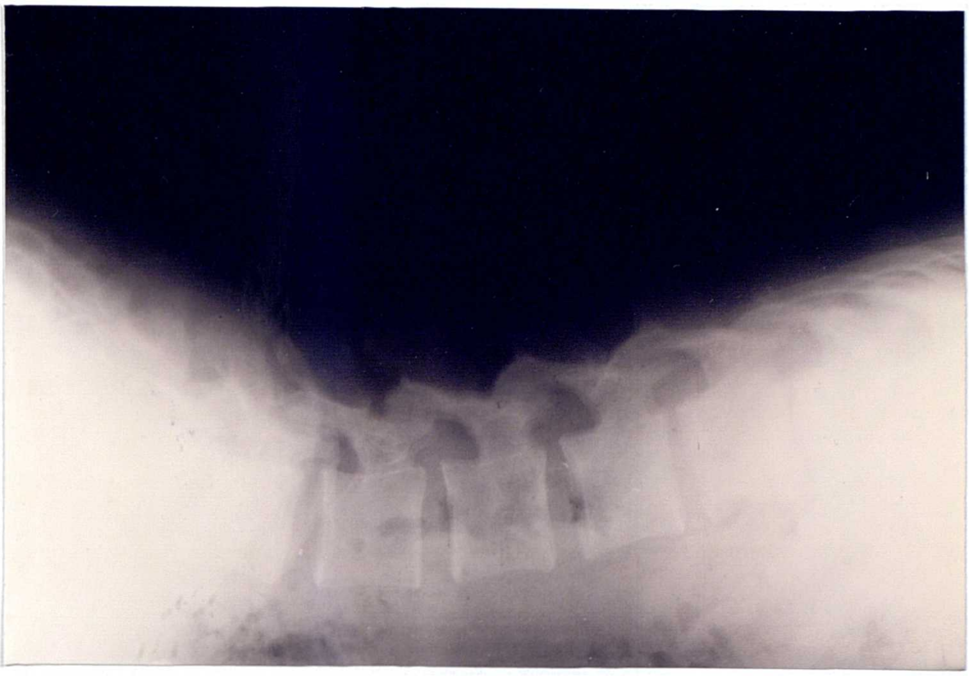
	posteroanterior				superoinferior			
Spinal segment	Mean	SD	Min	Max	Mean	SD	Min	Max
L1/2	0.58	0.50	0.01	1.64	-0.42	0.60	-1.02	1.26
L2/3	0.15	0.44	-0.57	0.81	-0.44	0.39	-1.13	0.24
L3/4	0.08	0.49	-0.69	1.02	-0.21	0.37	-0.73	0.60
L4/5	-0.01	0.42	-0.83	0.81	-0.16	0.52	-1.06	0.92
L5/S1	-0.67	0.87	-2.33	0.50	0.37	0.82	-0.98	1.74

Displacements of the spinous processes, mm

	vertical (upward/doward)				horizontal (cranial/caudal)			
Spinal level	Mean	SD	Min	Max	Mean	SD	Min	Max
L1	-10.22	2.89	-14.70	-5.65	-2.91	3.46	-9.02	2.85
L2	-11.47	2.79	-16.57	-6.49	-1.47	3.13	-5.32	4.36
L3	-11.01	2.85	-15.39	-5.93	0.60	2.69	-2.73	6.32
L4	-10.12	2.65	-14.44	-5.01	1.56	2.70	-2.98	6.59
L5	-8.80	2.55	-12.38	-3.59	2.64	3.25	-3.36	8.38

Table 6.2
Summary statistics of the intervertebral movements and the displacements of the spinous processes produced by the application of posteroanterior mobilisation load. (n=12) (SD=standard deviation, Min=minimum, Max=maximum)
(Sagittal rotation is positive for flexion and negative for extension. Translation is positive in the directions of posterior and superior, and negative in the directions of anterior and inferior. Upward and cranial displacements of spinous processes are positive, and downward and caudal negative)

(a)



(b)

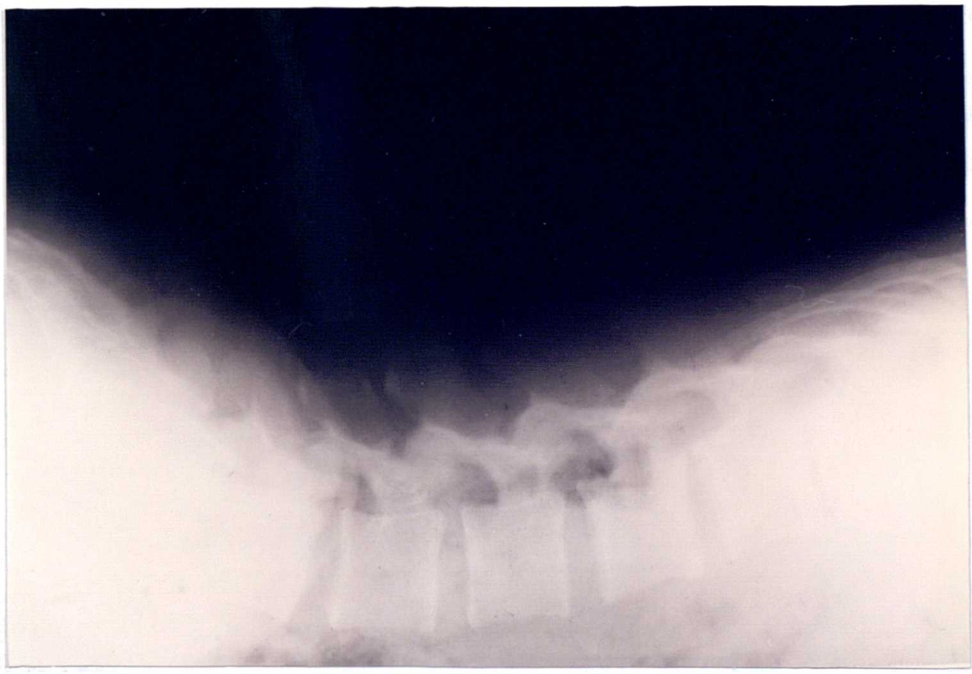


Figure 6.8
Radiographs of a subject (subject LKS, male, aged 21) showing the intervertebral movements produced by posteroanterior mobilisation. (a) before mobilisation (b) after mobilisation.

vertebral body. The mathematical procedures involved in the above computations are explained in section A1.2 of Appendix I.

The prediction of the coordinates of the spinous process in the second radiograph required only two reference points, that is, any two intersection points of the quadrangle representing the vertebral body. With the four corners of the vertebral body (and therefore 6 choices of intersection points) and the nine sets of data, a total of 54 independent calculations was carried out and the results averaged.

6.4.3 Analysis of Errors in Radiographic Measurements

The present study involved repeated marking and digitisation of radiographic images. This allowed an assessment of the errors involved in radiographic measurements. The standard deviations of the kinematic parameters (sagittal rotation and translations, and displacements of the spinous processes) obtained from the nine sets of data were determined. These represented errors due to marking, superimposition and digitisation, and were separately calculated for the different motion segments so as to assess the effect of spinal level. Errors in measuring sagittal rotation and spinous process displacements with different choices of vertebral corners as reference points were also evaluated.

6.5 RESULTS

Figure 6.8 shows the intervertebral movements produced by posteroanterior mobilisation as appeared in a pair of radiographs. The summary statistics of the experimental results are provided in table 6.2.

6.5.1 Sagittal Rotation

It was shown that the application of the posteroanterior load generally produced extension at the upper four lumbar motion segments and flexion at the

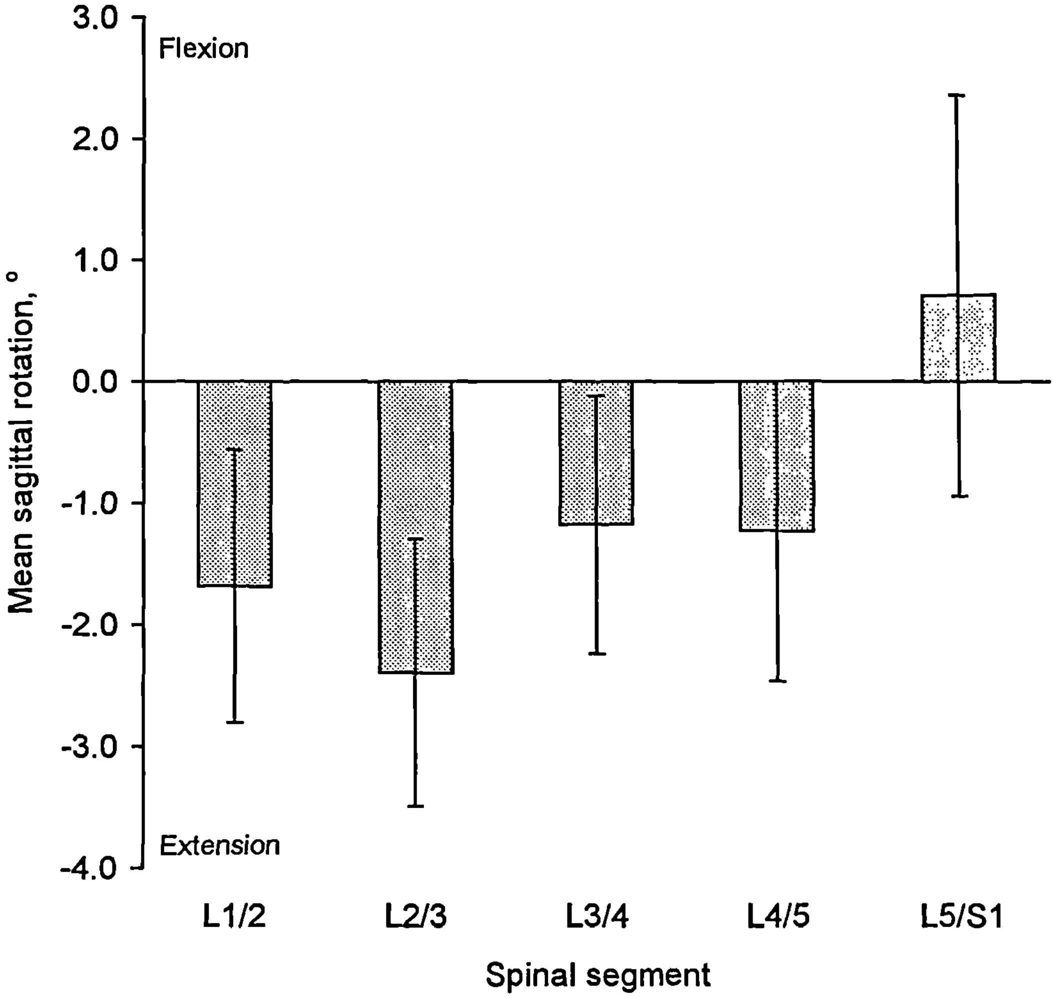


Figure 6.9
Mean sagittal rotation of the motion segments produced by posteroanterior mobilisation. (n=12) (The vertical bars denote ± 1 standard deviation for the 12 subjects.)

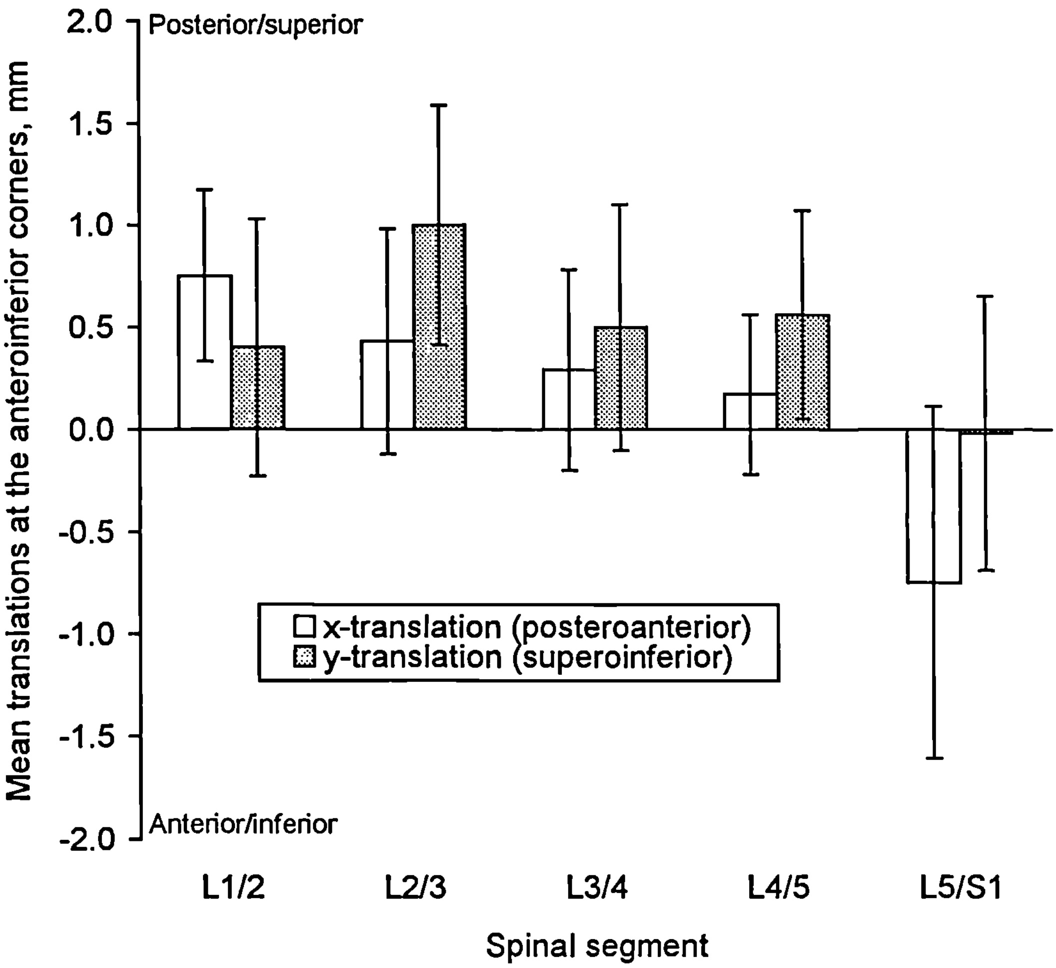


Figure 6.10
Mean x- and y-translations at the anteroinferior corners of the superior vertebral bodies of the motion segments. (n=12) (The vertical bars denote ± 1 standard deviation for the 12 subjects.)

L5/S1 segment (figure 6.9 and table 6.2). The mean extension movements ranged from 1.2° to 2.4°, maximally at L2/3, and the mean flexion at L5/S1 was 0.7°.

Extension was consistently observed in all subjects at L1/2 and L2/3. While most subjects exhibited extension at L3/4 and L4/5, flexion was found to occur in one subject at L3/4 and in two subjects at L4/5. However, the magnitude of flexion, if observed, was never large (less than 0.2° for L3/4 and less than 0.9° for L4/5, table 6.2). The tendency to produce flexion was higher at L4/5.

The observation at L5/S1 was less consistent, although it generally tended to flex. Flexion was observed in six of the subjects with a maximum value of 4.2°. The segments of two of the subjects showed no measurable rotation, and those of the remaining four subjects extended.

6.5.2 Sagittal Translations

The translational movements observed were generally small in magnitude. The magnitude of the movement was never more than 2.0mm in the various subjects. There were large individual variations in the magnitude of translations observed, as shown by the large standard deviations and ranges in table 6.2.

6.5.2.1 Translations at the anteroinferior vertebral body corners

In general, the anteroinferior corners of the superior vertebral bodies of the upper four motion segments were found to translate posteriorly and superiorly. The mean posterior translation ranged from 0.2-0.8mm and the mean superior translation from 0.4-1.0mm (figure 6.10 and table 6.2). The movement patterns at the upper four segments, in particular L1/2 and L2/3, were fairly consistent among the subjects. At each motion segment, there were less than three subjects who exhibited translations in directions which were opposite to those of the general trend.

The anteroinferior corner of the L5/S1 segment exhibited a mean anterior translation of 0.8mm (figure 6.10 and table 6.2). There was no obvious trend for translation along the y-axis. There were equal numbers of subjects showing translation

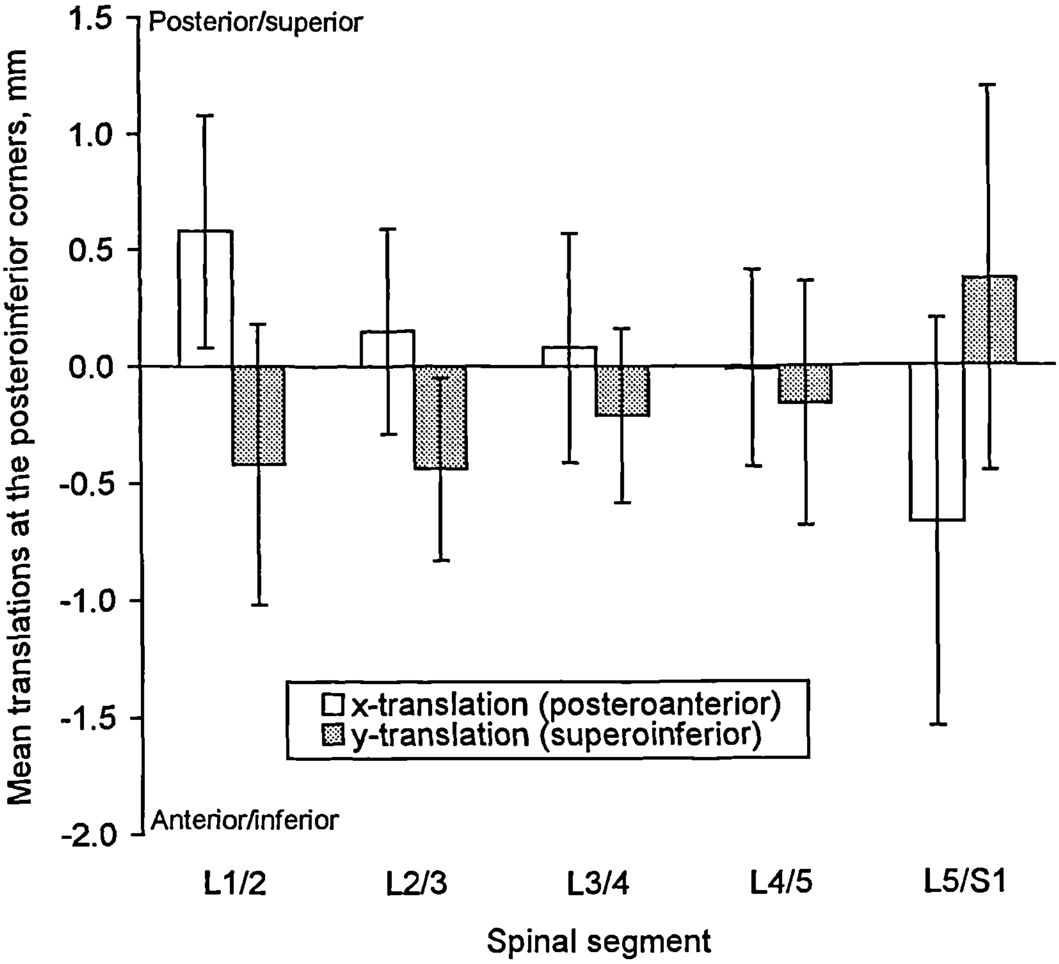


Figure 6.11
Mean x- and y-translations at the posteroinferior corners of the superior vertebral bodies of the motion segments. (n=12) (The vertical bars denote ± 1 standard deviation for the 12 subjects.)

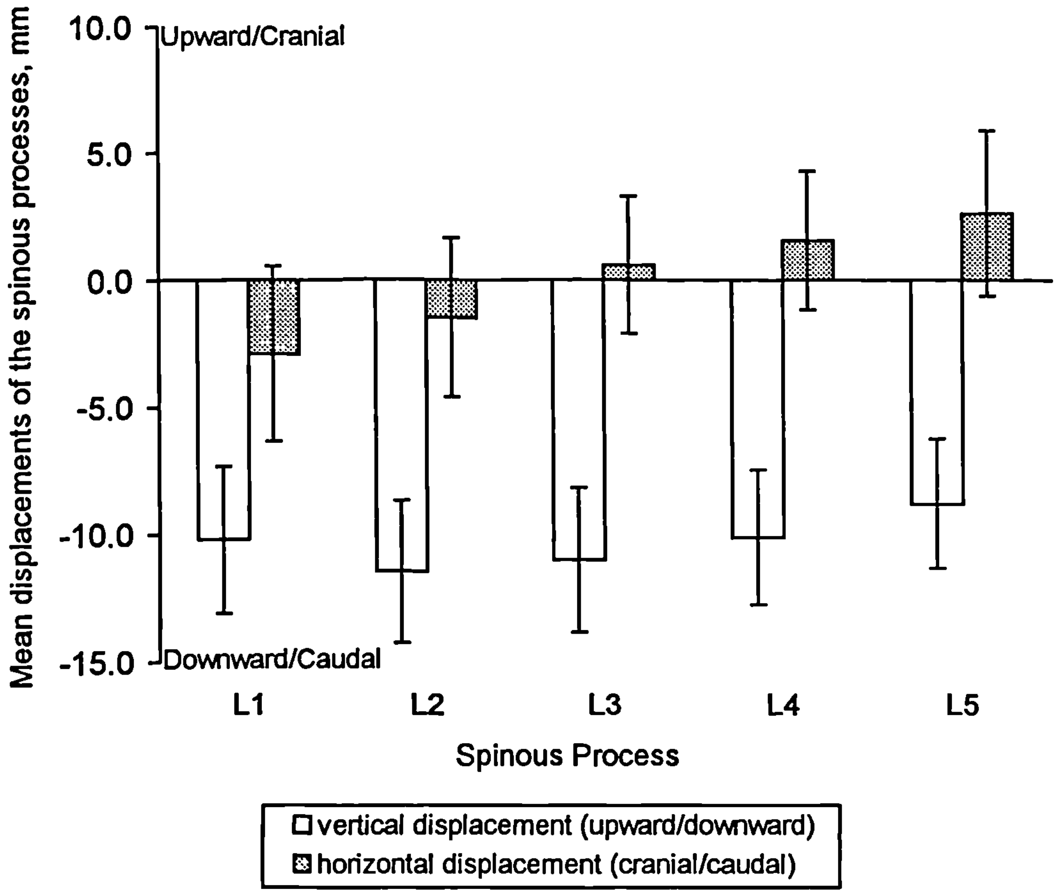


Figure 6.12
Mean vertical and horizontal translations of the spinous processes. (n=12) (The vertical bars denote ± 1 standard deviation for the 12 subjects.)

in both the superior and inferior directions, resulting in a very small mean value (0.02mm).

6.5.2.2 Translations at the posteroinferior vertebral body corner

The posteroinferior vertebral corners were generally found to translate posteriorly (mean=0.6mm at L1/2, 0.2mm at L2/3) and inferiorly (mean=0.4mm at L1/2 and L2/3) at the upper two motion segments. At L5/S1, the general trend was opposite with anterior (mean=0.7mm) and superior (mean=0.4mm) translations (figure 6.11 and table 6.2). The above patterns were fairly consistent with less than three subjects showing deviations at each segment.

The L3/4 and L4/5 appeared to be the transitional segments. There were no obvious trends in the directions of x- and y-translations at the posteroinferior corners (figure 6.11 and table 6.2). There were about equal numbers of subjects in translations of opposite directions and thus the mean translations at these levels were small (not more than 0.2mm).

6.5.3 Displacements of the Spinous Processes

All the five lumbar spinous processes moved in the downward direction under the application of the posteroanterior load (figure 6.12, table 6.2). This was consistently observed in all subjects. The mean downward displacements at the various levels ranged from 8.8mm to 11.5mm. The displacement was greatest at L2/3 *and least at L5/S1. The mobilised* L4 spinous process did not appear to move to any greater extent than the other spinous processes.

The upper two lumbar spinous processes generally moved caudally (a mean of 1.5mm for L1 and 2.9mm for L2) and the lower two cranially (a mean of 1.6mm for L4 and 2.6mm for L5) (figure 6.12, table 6.2). Most subjects showed this general pattern and movements in the opposite directions were observed in less than three subjects for each level. Finally, the L3 spinous process generally showed very little

	L1/2	L2/3	Spinal Segment L3/4	L4/5	L5/S1	
Errors in sagittal rotation, °						
Reference points						*Row mean*
1 and 2	1.03	0.80	0.87	0.94	1.41	*1.01*
3 and 4	0.98	0.82	0.79	1.02	1.34	*0.99*
1 and 4	1.06	0.81	0.80	1.00	1.34	*1.00*
2 and 3	0.92	0.79	0.87	1.01	1.45	*1.00*
1 and 3	0.92	0.71	0.79	0.91	1.29	*0.92*
2 and 4	1.00	0.78	0.76	0.95	1.36	*0.97*
Column mean	*0.98*	*0.78*	*0.81*	*0.97*	*1.37*	*Overall mean =0.98*
Errors in sagittal translations, mm						
						Mean
T_{ax}#	0.41	0.52	0.50	0.52	1.33	*0.66*
T_{ay}#	0.34	0.40	0.42	0.46	0.76	*0.48*
T_{bx}#	0.38	0.46	0.47	0.48	1.14	*0.59*
T_{by}#	0.48	0.42	0.43	0.50	0.92	*0.55*
Errors in vertical (upward/doward) displacement of spinous process, mm						
Reference points						*Row mean*
1 and 2	0.59	0.50	0.53	0.52	0.63	*0.55*
3 and 4	0.55	0.46	0.53	0.49	0.65	*0.54*
1 and 4	0.56	0.49	0.53	0.48	0.63	*0.54*
2 and 3	0.53	0.41	0.46	0.48	0.62	*0.50*
1 and 3	0.53	0.45	0.53	0.48	0.61	*0.52*
2 and 4	0.56	0.43	0.47	0.49	0.63	*0.52*
Column mean	*0.55*	*0.46*	*0.51*	*0.49*	*0.63*	*Overall mean=0.53*
Errors in horizontal (cranial/caudal) displacement of spinous process, mm						
Reference points						*Row mean*
1 and 2	1.10	1.09	1.04	0.97	1.03	*1.05*
3 and 4	1.06	0.95	0.94	0.88	1.05	*0.97*
1 and 4	1.25	1.11	1.11	1.10	1.10	*1.13*
2 and 3	1.12	1.06	1.08	0.97	0.93	*1.03*
1 and 3	1.11	1.07	1.04	0.96	0.95	*1.03*
2 and 4	1.02	0.88	0.90	0.87	0.96	*0.93*
Column mean	*1.11*	*1.03*	*1.02*	*0.96*	*1.00*	*Overall mean=1.02*

Table 6.3
Errors in measuring the various kinematic parameters.
(# T_{ax}, T_{ay}, T_{bx} and T_{by} denote respectively the x- and y- translations at the anteroinferior and posterior corners of the superior vertebral body.)
(The reference points 1, 2, 3 and 4 from sagittal rotation and spinous process displacements were determined were respectively the anteroinferior, posteroinferior, posterosuperior and anterosuperior corners of the superior body.)

movement (a mean of 0.6mm) in the craniocaudal direction, and there were about equal numbers of subjects showing movements in the cranial and caudal directions.

6.5.4 Errors in Determining the Kinematic Parameters

The errors associated with marking, superimposition and digitisation in determining the various kinematic parameters are illustrated in table 6.3. The variations with different segmental levels and choices of reference points are also shown.

It was found that with the present method of marking and superimposition, the mean error involved in determining sagittal rotation and translation, and displacement of the spinous processes was 1.0°, 0.6mm and 0.8mm respectively. The error size was shown to be dependent on the segmental level examined. Table 6.3 clearly illustrates that the determination of sagittal rotation and translation was subjected to the largest error at the L5/S1 segment. However, the table shows that there were no major differences in the choice of reference points in determining sagittal rotation and displacement of the spinous process. Thus, any two corners of the vertebral body would produce equally accurate results.

It should be noted the standard deviation represents the error that might occur randomly by chance with one single measurement. The accuracy of the data reported in this study would be expected to be better than that observed in the error analysis. This was because the result of each subject was the average of repeated calculations (54 times in the case of sagittal rotation and spinous process displacement, and 9 times in the case of sagittal translation) instead of a single measurement. Hence, it was considered that the experimental technique was sufficiently accurate for the purpose of this study.

6.6 DISCUSSION

6.6.1 Intervertebral Movements Produced by Mobilisation

The present investigation examined the intervertebral movements produced by posteroanterior mobilisation in a group of living subjects. Such data have never been reported by any previous studies (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and Evans, 1991 and 1992). The work was performed in a small group of male volunteers, but the observations were sufficiently consistent to enable conclusions to be drawn on the mechanical effects of mobilisation and to establish a baseline for the comparison of movements in patients with low back pain.

It was demonstrated that posteroanterior mobilisation produced extension of the lumbar motion segments with the exception of L5/S1 which generally tended to flex. This finding was in agreement with the prediction of the three-point bending model proposed by Lee (1990) (see section 4.1.4). The model predicted that extension moment was produced at the lumbar motion segments. In addition, the predicted extension moment ranged from 2.1Nm to 7.8Nm. Given that the lumbar disc has a mean rotational stiffness of 3.6Nm deg^{-1} (Markolf, 1972), a simple calculation reveals that the magnitude of extension produced will range from 0.6° to 2.2°. These values are remarkably close to the mean extensions observed in the present investigation (ranging from 1.2° to 2.4°).

The three-point bending model also predicted that the extension moment was maximal at the point of posteroanterior application. This moment decreased caudally and cranially and became zero above the two supporting points of the spine which were at the thoracic cage and the anterior superior iliac spines (ASIS) of the pelvis. The present study did not show maximal extension movement at L3/4 where maximal extension moment was predicted. However, it should be noted that the segmental rotation depends on the stiffness of the motion segments as well as the moment imposed. The stiffness of the various segments will be different and thus maximal movement might not necessarily occur where maximal moment was experienced.

The direction of sagittal rotation at L5/S1 was found to be less consistent, although the segment generally tended to flex. It is believed that this will depend on the relative locations of the ASIS and the L5/S1 disc centre. The highest point of the iliac crest of the pelvis is generally at the same level as the L4/5 disc centre (Romanes, 1981; Williams et al, 1989) and thus the ASIS and the L5/S1 disc centre will be very close to each other.

If the L5/S1 disc centre is over the supporting points ASIS, no moment will exist at this segment and thus there will be no local movement. If L5/S1 is superior to the ASIS, a small amount of extension will be produced. On the other hand, if L5/S1 is more inferior, the lumbar spine will tend to be tilted upwards at the lower end. If the pelvis and thus the sacrum are relatively fixed in position or restrained from rotation, the L5/S1 will be forced into flexion. Hence, the less consistent observation at L5/S1 is a reflection of the individual variations in the relative locations of L5/S1 and the ASIS and of restraints on pelvic rotation.

In the present study, in order to reduce the amount of scattering and exposure, the x-ray beam was tightly coned to the lumbar spine and the posterior half of the sacrum. As the ASIS was not revealed in the films, it was not possible to confirm the above explanation.

It appeared that the extension movements at the upper lumbar motion segments were achieved by superior translation at the anteroinferior corner of the vertebral body and inferior translation at the posteroinferior corner. This observation was consistent with that found in the cadaveric study reported in chapter 5. The pattern was somewhat reversed at L5/S1 where flexion of the segment was accomplished by superior translation at the posteroinferior corner. The anteroinferior corner of the L5/S1 segment had no obvious trend for translation along the superoinferior axis.

Another interesting finding was that there were generally posterior translations at both the anteroinferior and posteroinferior corners of the superior bodies of upper lumbar motion segments (L1/2 and L2/3). This indicated that the segments translated posteriorly as a whole. For L5/S1, the reverse was true where there were anterior translations at both corners of the vertebra. The middle motion segments (L3/4 and L4/5) appeared to be the transitional segments, where the above trends were less obvious.

These findings are also in remarkable agreement with the prediction of the three-point bending model that the segments above the point of posteroanterior force application were subjected to posterior shear and those below to anterior shear (see section 4.1.4). The L3/4 and L4/5 segments are just adjacent to the point of force application where the change in the direction of shear occurs. This explains the transitional nature of these segments. The change in the direction of translation may occur at L3/4 or L4/5 and thus movements in both directions may be found at these segments, producing less consistent results.

The three-point bending model predicted that the posterior shear produced at the upper segments was 34.8N and the anterior shear at the lower segments 115.2N. It was reported that the mean stiffness was 214Nmm^{-1} for posterior shear loading of the lumbar segments (Panjabi et al, 1984) and 81Nmm^{-1} for anterior shear loading of the L5/S1 segment (McGlashen et al, 1987). Based on these data, it can be shown that the posterior translation at the upper segments will be 0.2mm and the anterior translation at L5/S1 1.4mm. These values are close to those observed in the present investigation. At L1/2 and L2/3, the mean posterior translation observed was 0.4-0.8mm at the anteroinferior corner and 0.3-0.6 at the posteroinferior corner, whereas at L5/S1, the mean anterior translation was 0.8mm and 0.7mm at the anteroinferior and posteroinferior corners respectively.

A comparison of the results of the present study with those of the earlier work on cadaveric motion segments (chapter 5) was made. The cadaveric study simulated the application of mobilisation loads to the segments inferior to the mobilised vertebrae. The segments showed a mean extension of 5° which was accomplished by mean superior translation of 3.0mm at the anteroinferior corner and mean inferior translation of 0.8mm at the posteroinferior corner (table 5.4). The same pattern was observed in this study although the magnitude of movement was smaller. In the present case of L4 mobilisation, the L4/5 segment had mean extension of only 1.2° with translations of less than 0.2mm (see table 6.2). The differences in magnitude are not surprising. This is because isolated cadaveric motion segments are not restrained from movements by adjacent segments and the passive resistance of the musculature and therefore, will be expected to exhibit more movement. Furthermore, in regard to translations along the posteroanterior axis, both studies showed that the direction of translation varied among different specimens and individuals.

6.6.2 Displacements of the Spinous Processes

It was shown that all the five lumbar spinous processes moved in the downward directions under the application of posteroanterior loads. This supports the belief that the technique produces a sagging of the whole lumbar spine. It appears that the sagging is achieved by extension of the lumbar motion segments, that is, an increase in lordosis. There are no major differences in the downward displacements among the different spinous processes.

Another finding was that the L1 and L2 spinous processes moved caudally and those of L4 and L5 cranially. The L3 process had very little craniocaudal displacement. This indicates a decrease in the longitudinal length of the posterior lumbar spine. The extension of the motion segments and the increase in lordosis have brought the two ends of the lumbar spine closer together.

Previous biomechanical studies had attempted to examine the downward displacements of the spinous processes by displacement transducers resting on the skin overlying them (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and

Evans, 1992) (see section 4.1.4). Lee (1990) reported that during mobilisation of L4, the mean downward displacement of the L3 spinous process relative to the L4 process was 11.1mm and that of L5 relative to L4 12.3mm. The L5 process generally showed more relative displacement than L3.

The present radiographic study revealed that the relative displacement of L3 and L5 (with respect to L4) was 0.9mm and 1.3mm respectively. As reported in the previous study, the L5 process also showed greater relative displacement. However, the previous results obtained from surface measurements were about ten times larger in magnitude than those in the present work. This indicated that surface measurements were subjected to large errors due to skin deformation. In addition, in the previous studies, the posteroanterior force applicator was fitted with a soft pad to reduce the discomfort of the patient. The deformation of the pad would also contribute to the error in surface measurements.

6.6.3 Repeatability of Radiographic Measurements

Previous studies generally employed functional radiographs taken in routine examination. There were no vigorous checks on the alignment of the x-ray beam, the positions of the patients (which were normally in standing and could be different in different films), the identification of landmarks and the superimposition of films (Miles and Sullivan, 1961; Tanz, 1963; Pennal et al, 1972; Hayes et al, 1989; Putto and Tallroth, 1990).

Radiographic measurements were generally found to be inaccurate. Tanz (1963) documented that the error associated with sagittal rotation measurement was as high as 2°. Schaffer et al (1990) also found that the measurement of vertebral translation was unreliable when the movements were less than 5mm and the accuracy could be decreased by poor film quality and out-of-plane movements.

Previous authors found that the accuracy of radiographic measurements could be improved by using tangent lines and their intersections to represent the vertebral body corners and computing the movements mathematically (Dvorak et al, 1991a and 1991b; Panjabi et al, 1992) (see section 3.1.2.3). The errors in determining sagittal rotation and translation with such a technique were 1.3° and 0.9mm respectively and therefore more acceptable.

The radiographic technique employed in the present work incorporated the suggestions of the earlier authors described above. Vigorous attempts were made to further improve the accuracy. These included rigid fixation of the x-ray cassette, the use of reference markers (to eliminate error due to different positioning of the cassette in different exposures), accurate and proper alignment of the x-ray beam using specially made device, repeated digitisation, marking and superimposition of radiographs. The errors in determining the rotational and translational movements were further reduced to 1.0° and 0.6mm. It should be emphasised that the error size of the data reported in this study would be expected to be even smaller than these because the result was the average of repeated computation instead of a single measurement.

The error analysis demonstrated that the L5/S1 segment was shown to be particularly susceptible to error due to marking and superimposition of images. This was because the x-ray beam was directed towards the midlumbar region and the image of the sacrum was somewhat ambiguous because of the obliquity of the incident x-rays. The pelvic shadow and the absorption and scattering of x-ray caused by the pelvis also made the sacral image ambiguous. Large errors would therefore occur with measurements of the L5/S1 movements.

However, the choice of reference points had no effect on the accuracy of determining sagittal rotation and the spinous process displacement. Therefore, in clinical practice, any two corners of the vertebral body could be chosen for measuring these movements. To minimise the errors involved, as in the present study, the kinematic parameters could be computed six times and the average taken using the different combination of reference points.

The kinematic parameters chosen in the present study are sagittal rotation and the translations at the anteroinferior and posteroinferior corners of the vertebral body. The instantaneous centre of rotation was not examined since it had been shown that radiographic measurement of this parameter in living subject without the use of implanted metallic markers was subjected to very large errors (Pearcy and Bogduk, 1988; Panjabi et al, 1992) (see section 3.2.1.2). Panjabi et al (1992) showed that the error size of the centre of rotation could be as large as 4.3mm for measurements based on functional radiographs.

6.7 CONCLUSION

The radiographic study demonstrated that posteroanterior mobilisation produced extension of the upper four lumbar motion segments and generally flexion at the lumbosacral joint. There were also posterior translations of the vertebrae above that mobilised and anterior translations below. These movements were generally small in magnitude. The movement patterns observed were in good agreement with the predictions of the three-point bending model proposed by Lee (1990).

It was also shown that all the lumbar spinous processes moved in the downward direction, indicating that there was a sagging of the spine during mobilisation. The superior displacements of the spinous processes of the upper lumbar vertebrae and the inferior displacements of the lower ones indicated that there was a decrease in the longitudinal length of the spine. The sagging and shortening of the spine was produced by the increase in lordosis as a result of the extension of the motion segments.

The present study has established normative data with which future studies on back patients can be compared. It remains to be seen whether clinical pathology or pain has any effects on the segmental movements produced by posteroanterior mobilisation.

■ *PART III*

THE BIOMECHANICAL BASIS OF SPINAL TRACTION

■ *Chapter 7*

Deformations of the Spine Produced by Lumbar Traction

Deformations of the Spine Produced by Lumbar Traction

7.1 INTRODUCTION

The literature review showed that there was insufficient information on the deformations of the spine produced by lumbar traction (see section 4.2.4). A critical reappraisal of the mechanical effects of traction was considered necessary. Although the results of previous radiographic studies (Colachis and Strohm, 1969; Reilly et al, 1979) could be erroneous (see section 4.2.1.2), they did provide fundamental information on the in-vivo deformation of the spine. Further radiographic examination was initially considered to be unwarranted due to the risks associated with radiation. In order to examine the nature and magnitude of segmental movements, cadaveric spines were subjected to simulated traction loads.

In cadaveric study, rigid attachment of markers is made possible to enhance the accuracy of measurement. In vitro study also had the advantage that it would allow the examination of the therapeutic hypotheses of traction which would not be possible in living subjects. It could examine more directly the restoration of the anatomy of osteoligamentous spine, such as that of the intervertebral foramina.

The line of action of traction force is generally considered to be in the sagittal plane. As in the case of posteroanterior mobilisation, such loading should not produce significant out-of-plane movements. An analysis of the sagittal movements was considered sufficient. Although the study did not quantify the out-of-plane

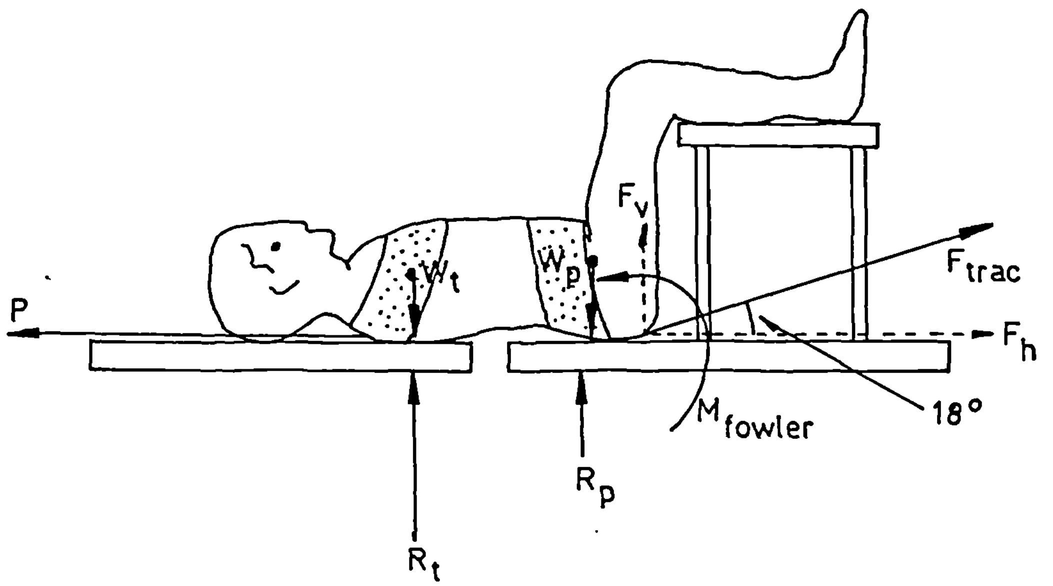


Figure 7.1.
Loads acting on the trunk and pelvis during traction therapy on a split table with the subject in the Fowler position (i.e. the hips and knees flexed and the legs supported on a stool).
(F_{trac} = traction force, F_h = horizontal component of the traction force, F_v = vertical component of the traction force, R_p = reaction force at the pelvis, R_t = reaction force at the upper trunk, W_p = weight of the pelvis and the abdominal contents, W_t = weight of the trunk above the pelvis, P = force provided by the thoracic harness)

movements, an attempt was made to identify the extent of these movements and to quantify any errors introduced through these movements.

More specifically, the purposes of this cadaveric study were to determine

- the sagittal rotation and translations of each individual motion segment under the application of traction loads, and
- the associated changes in the sizes of the intervertebral foramina.

7.2 LOADS ACTING ON THE LUMBAR SPINE DURING TRACTION

Figure 7.1 shows the clinical application of traction with the patient in the Fowler's position (i.e. with the hips and knees flexed, and the legs supported on a stool) and the loads acting on the trunk.

Clinically, a traction force of up to 350N (about one half of body weight) may be applied and the angle of pull is generally about 18° to the horizontal (Colachis and Strohm, 1969; Saunders, 1975; Hinterbuchner, 1985). As shown in figure 7.1, the traction force (F_{trac}) can be resolved into horizontal (F_h) and vertical (F_v) components, which will be 350xcos18°=333N and 350xsin18°=108N respectively.

The horizontal component of the traction force provides the effective mechanical pull. This will be counteracted by an equal but opposite force (P) provided by the thoracic harness which fixes the upper body in position. Since traction is normally applied through a pelvic harness along the skin surface of the back, F_h will be acting posterior to the centres of rotation of the motion segments which are approximately at the geometric centres of the discs (Rolander, 1966; Gertzbein et al, 1981 and 1984). Previous authors measured the distances between the disc centres and the overlying skin in the mid-sagittal plane on magnetic resonance imaging scans (Tracy et al, 1989). The mean distances for the L2/3 and L5/S1 segments were found to be 0.082m and 0.088m respectively. Therefore, the 330N horizontal force will also produce significant flexion moment on the lumbar spine. This will be about 27Nm and 29Nm at L2/3 and L5/S1 respectively.

The vertical component of the traction force (108N) will partially counterbalance the weight of the pelvis and the abdomen which will be about 196N (28% of the body weight of a 70kg person; Winter, 1990). This will reduce the normal reaction force at pelvis (R_p) and thereby the frictional resistance by about 55%. However, a split table is generally employed in modern physiotherapy departments (Colachis and Strohm, 1969; Saunders, 1975; Hinterbuchner, 1985) and the friction to traction is effectively eliminated. In this case, the vertical component will be practically redundant. It is therefore unclear why previous authors have recommended an angle of pull of 18° even with the use of a split table. It appears that the traction force should be applied horizontally so as to minimise the total force required.

The weights of the shank and thigh would be supported primarily by the stool (figure 7.1). They should not impose a significant bending moment on the spine. However, Yoon and Mansour (1982) showed that the hip joint was not entirely free to rotate. A resistive moment at the hip was produced when the joint was passively flexed with the knees bent. This moment will be transmitted to the lumbar spine, tending to flatten its lordosis. Therefore, the adoption of the Fowler's position during traction therapy will impose a flexion moment (M_{fowler}) on the spine, but no information is available in the literature regarding the magnitude of the moment.

Consequently, a pilot study was carried out to measure the moment produced by passive hip and knee flexion. Seventeen normal subjects (9 males and 8 females, aged between 20 and 24, mean height=1.63m, mean weight=541N) were studied. Each subject was requested to lie supine with the trunk supported on a wooden board which was placed on top of a force plate (Advanced Mechanical Technology Inc., Massachusetts, USA). The pelvis was placed over the edge of the board. The straightened legs were supported on another independent surface which was at the same level as the board. The hips and knees were then passively flexed to 90° and

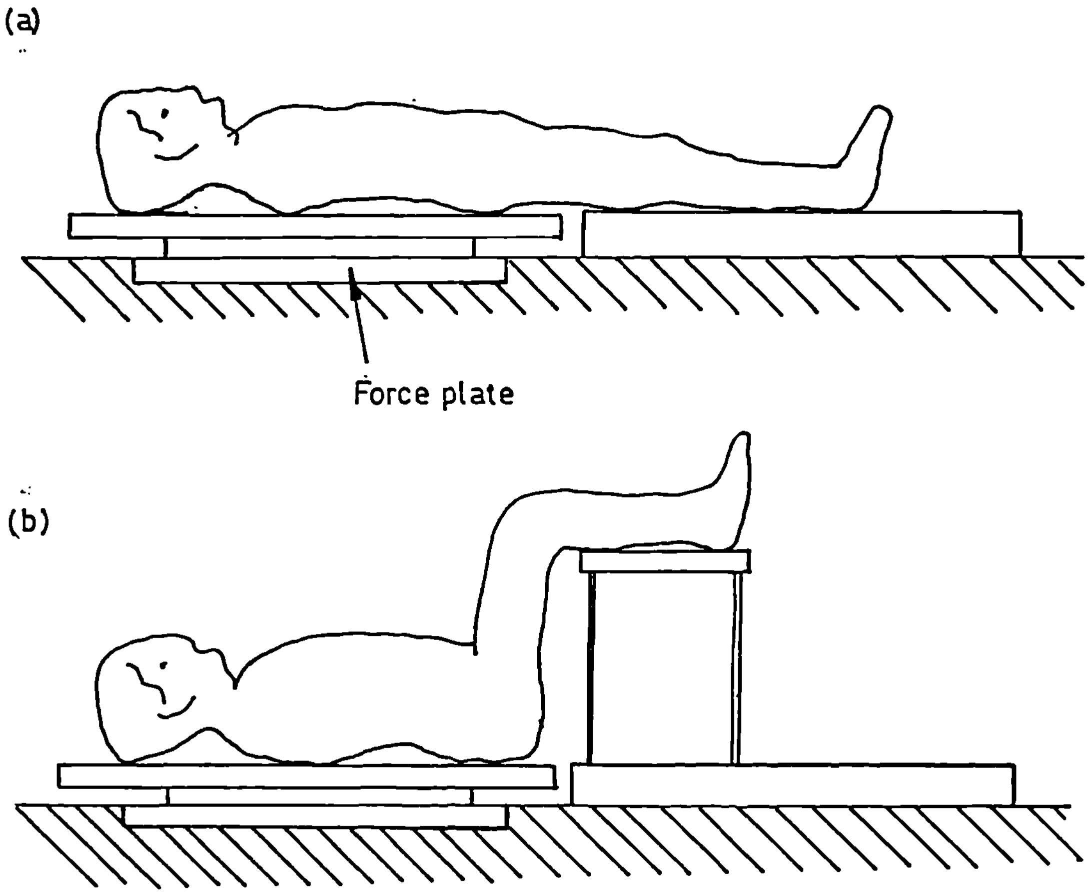


Figure 7.2
Experimental measurement of the flexion moment imposed on the spine as a result of the adoption of the Fowler's position. This is given by the difference in the sagittal moment recorded by the force plate between positions (a) and (b).

supported on an adjustable height stool. Figure 7.2 shows the experimental arrangement.

A mean flexion moment of 24±6Nm (range=15-41Nm) was recorded by the force plate as a result of the adoption of the Fowler's position. This represented the moment imposed on the lumbar spine due to tissue resistance produced by passive hip and knee flexion. The force plate also recorded a mean initial vertical force of 421±41N which was slightly larger than the weight of the head, arms and trunk of the subjects (mean 383±40N, estimated from body weight data using segmental weight properties provided in Winter, 1990). The vertical force decreased to 401±38N after the adoption of the Fowler's position. The data suggested that the wooden board initially carried a small percentage of the weight of the legs which was then largely supported by the stool in the Fowler's position. It was noted that the hip joints were 0.19±0.02m from the force plate centre. Assuming that the weight of the legs acted through the hip joints, the decrease in the vertical force (20±6N) would produce an increase of 4±1Nm in the flexion moment and thus the moment produced by the Fowler's position would have been overestimated by 16±4%. The error is considered to be acceptable in view of the non-linear response of the spine in bending and the degree of bending induced. The force plate result is also reasonably consistent with the finding of Yoon and Mansour (1982) who reported that 60° of hip flexion and 51° of knee flexion would generate a moment of 15Nm at the hip. The observed value was higher in the present study as the hips and knees were flexed to 90°.

It should be noted that as shown in figure 7.1, during traction, there will be an increase in the reaction force (R_t) at the upper trunk to counteract the flexion moment produced by the traction force and the adoption of the Fowler's position.

A theoretical analysis is presented in this section to examine the loads acting on the lumbar spine during traction. In the present in vitro study, an attempt was made to reproduce these loads to simulate the in-vivo conditions.

Specimen number	Sex	Age	Body height, m	Body build	Cause of death
1	Male	56	1.70	average	Lung carcinoma
2	Male	62	1.96	average	Chronic obstructive airway disease
3	Female	60	1.57	average	Aspiration pneumonia
4	Male	74	1.88	heavy	Bronchopneumonia
5	Female	49	1.93	heavy	Hepato-renal failure
6	Male	66	1.65	average	Myocardial infarction
7	Male	68	1.98	heavy	Bronchopneumonia

Table 7.1.
Personal particulars and the cause of the death of the cadavers from which the specimens were taken.

Specimen number	1	2	3	4	5	6	7
Morphological abnormalities of the disc as revealed by discograms							
L1/2	N*	N	N	d$^{\phi}$	N	d	N
L2/3	N	N	N	d	a	c$^{\Omega}$	N
L3/4	a#	N	N	d	a	N	c
L4/5	a	b@	N	d	a	b	b
L5/S1	a	b	N	d	a	b	b
Macroscopic examination of the grade of degeneration							
L1/2	1	1	1	2	2	3	1
L2/3	1	1	1	3	2	1	1
L3/4	1	1	1	2	2	1	2
L4/5	3	1	1	2	2	2	2
L5/S1	3	3	1	3	2	2	3

Table 7.2.
Morphological abnormalities and the degree of degeneration of the intervertebral discs of the specimens as revealed by discograms and macroscopic examinations.
(* N = normal discogram; #a = disc space narrowing; @b=posterior tears; $^{\Omega}$c=anterior tears; $^{\phi}$d=generalised diffuse tears)

7.3 MATERIALS AND METHODS

7.3.1 Materials

Seven complete lumbosacral spines (T12-sacrum inclusive, 5 males and 2 females, aged between 49 and 74) were tested in this experiment. They were obtained from routine necropsies within 24 hours of death and double wrapped in plastic bags at -20°C until required for testing. This method of storage had been shown not to significantly affect the mechanical properties of the specimens (Sedlin and Hirsch, 1966; Galante, 1967; Hirsch and Galante, 1967; Tkaczuk, 1968; Panjabi et al, 1985) (see section 3.1.1). Prior to testing, each spine was thawed overnight at 3°C. It was then left to equilibrate with room temperature for a few hours before it was taken out of the plastic bags for dissection.

Table 7.1 shows the personal particulars and the cause of the death of the cadavers from which the specimens were taken. The body weights of the cadavers were not available from the mortuary records but the build of the subjects was indicated. All specimens were examined carefully by visual inspection to determine if they were any signs of bone pathology and anatomical anomalies.

After testing, contrast medium (about 2ml of Niopam 300) was injected into the nucleus pulposus of all the intervertebral discs of the specimens. Discograms were then taken to assess the morphology of the disc. The radiographic assessment was performed by a qualified radiologist. In addition, the discs were dissected at the mid-transverse plane and their degree of degeneration evaluated according to the criteria proposed by earlier authors (Nachemson, 1960; Rolander, 1966; Galante, 1967). The grading of disc degeneration has been described in section 5.2.1. The findings of the radiographic and macroscopic examinations are presented in table 7.2.

The discograms showed that among the 35 discs examined, 14 of them were normal, 7 showed disc space narrowing, 6 had posterior tears, 2 had anterior tears and 6 demonstrated generalised diffuse tears indicating gross disc degeneration. In

Figure 7.3
A dissected specimen of the osteoligamentous lumbosacral spine.

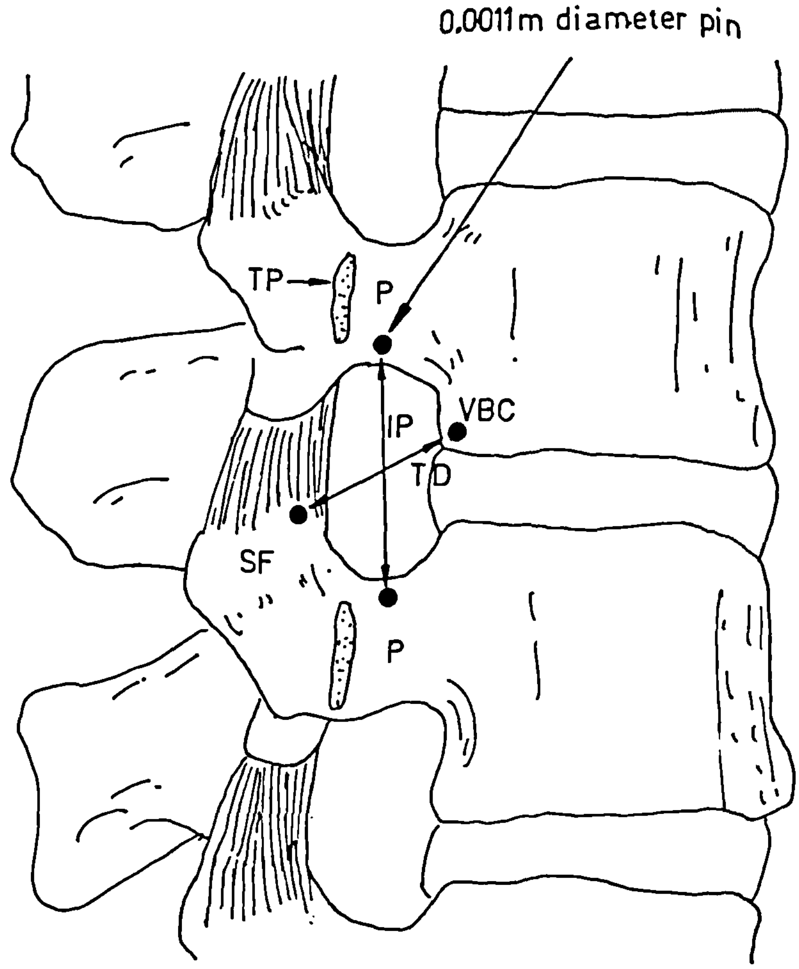
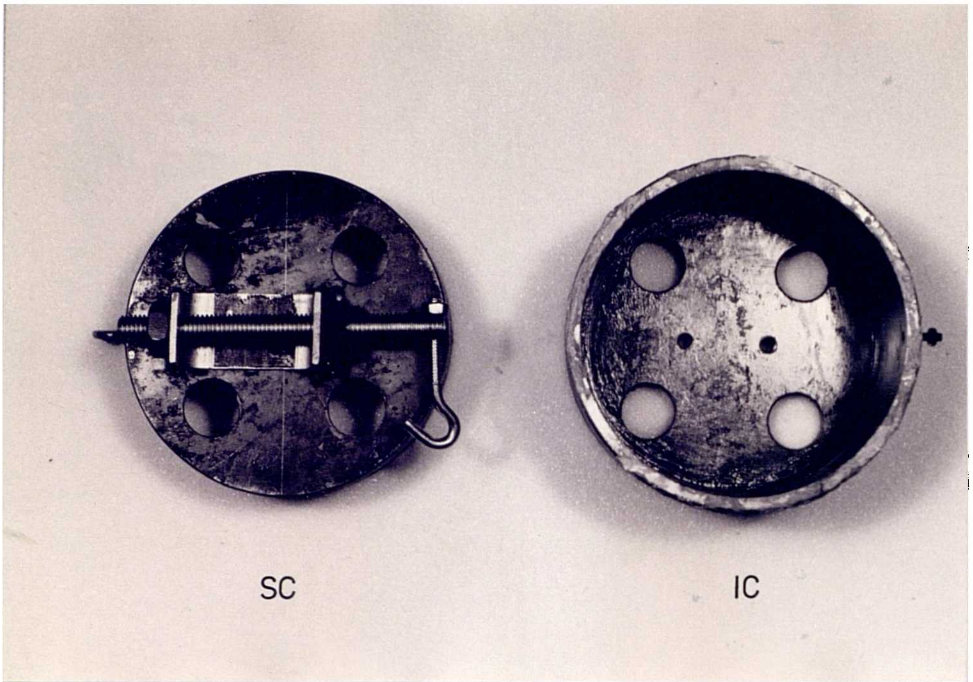


Figure 7.4
Measurement of the interpedicular (IP) and transverse foraminal (TD) distances using pins as reference points.
(TP = cut end of transverse process, P = pedicle, SF = superior articular facet of the inferior vertebra of the motion segment, VBC = posteroinferior corner of the body of the superior vertebra.)

Top view



Side view

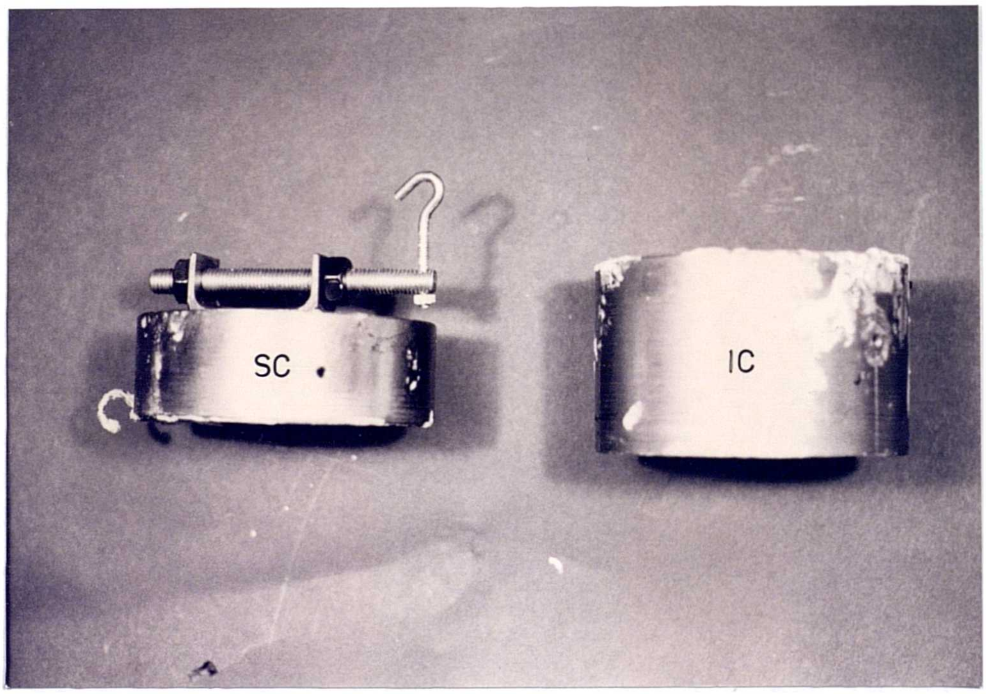


Figure 7.5a
The superior (SC) and inferior (IC) moulding cups (top and side views).

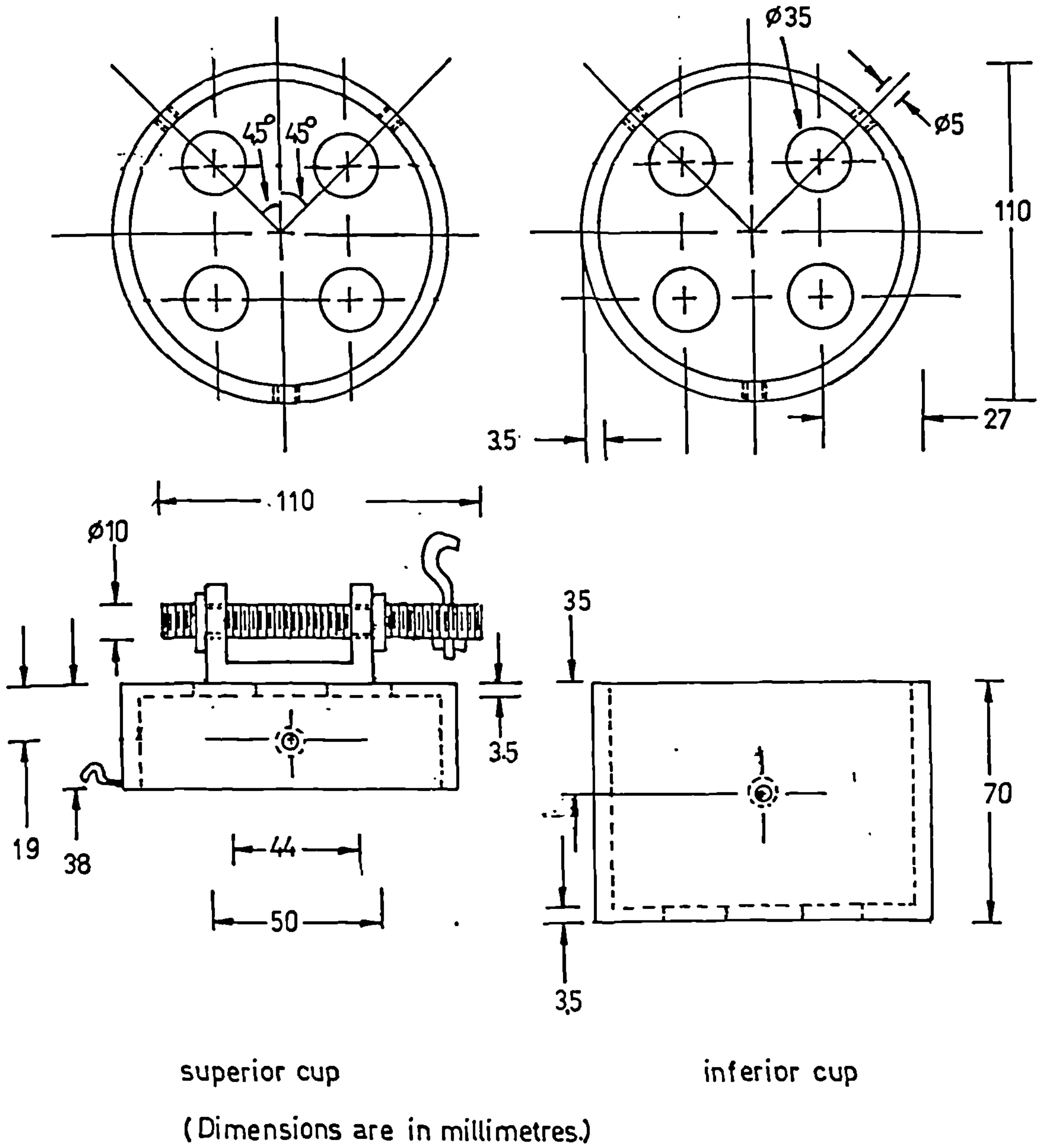


Figure 7.5b
The dimensions of the superior and inferior moulding cups.

addition, the macroscopic examination revealed that none of the discs was grade 0, 16 were grade 1, 12 were grade 2 and 7 were grade 3. Discs which were found to be more degenerated in the macroscopic examination had higher incidence of annular tears in their discograms (see table 7.2). In general, the specimens examined in the present work were found to be moderately degenerated.

7.3.2 Preparation of the Specimens

The specimen was dissected as shown in figure 7.3 in preparation for testing. Each spine was stripped of psoas major and the surrounding spinal muscles. Great care was taken not to damage the intervertebral disc or the ligaments. The contents of all the intervertebral foramina were cleared, and this involved removing the nerve tissues, the spinal vessels and the associated adipose and connective tissues.

The osteoligamentous boundaries of the foramina were clearly exposed. In order to examine the foraminal sizes, four 0.0011m diameter metal pins were inserted into the bony parts of the boundaries of each foramen (figure 7.4). The interpedicular distance of the foramen was defined by the two pins inserted into the lower edge of superior pedicle and the upper edge of the inferior pedicle. The two pins inserted into the posteroinferior corner of the vertebral body and the anterior edge of the superior articular facet defined the transverse distance. The transverse processes of vertebrae were removed so as to facilitate the measurement of the distances between the pins by a digital calliper.

Two moulding cups (figure 7.5) were made to hold the specimen securely for attachment to the loading system. The sacrum was set in Isopon (Plastic Padding Limited, High Wycombe, Bucks HP10 0PE, UK) in one of the cups (the inferior cup) which had an inner diameter of 0.105m, a wall thickness of 0.0035m and a height of 0.070m. The two lateral sides and the lower one-fourth of the sacrum were cut by a saw so that the bone could be fitted into the moulding cup. The other moulding cup

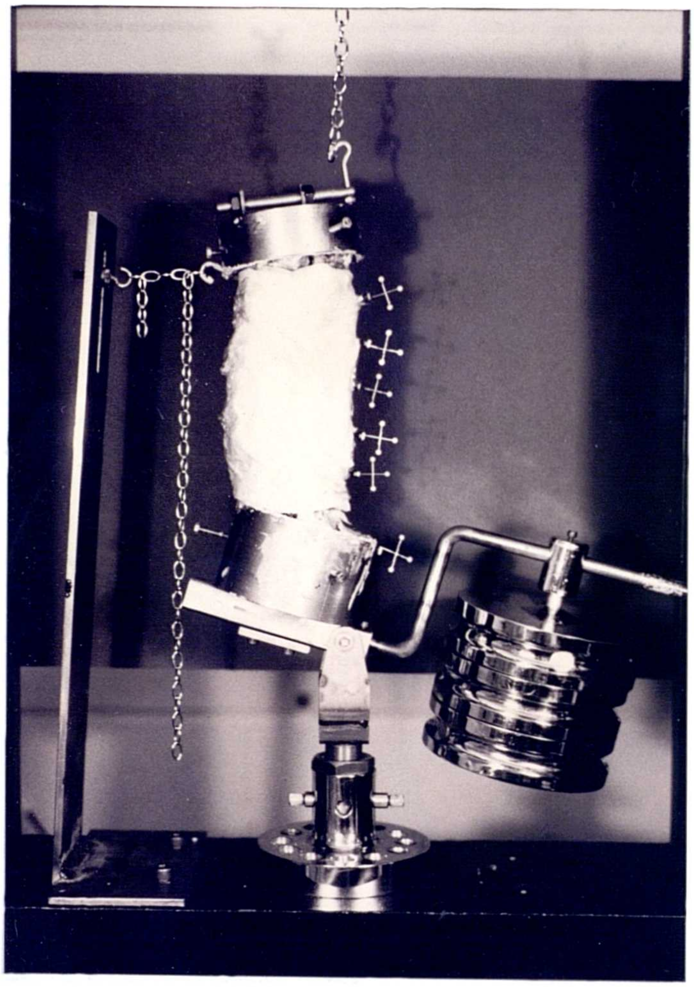
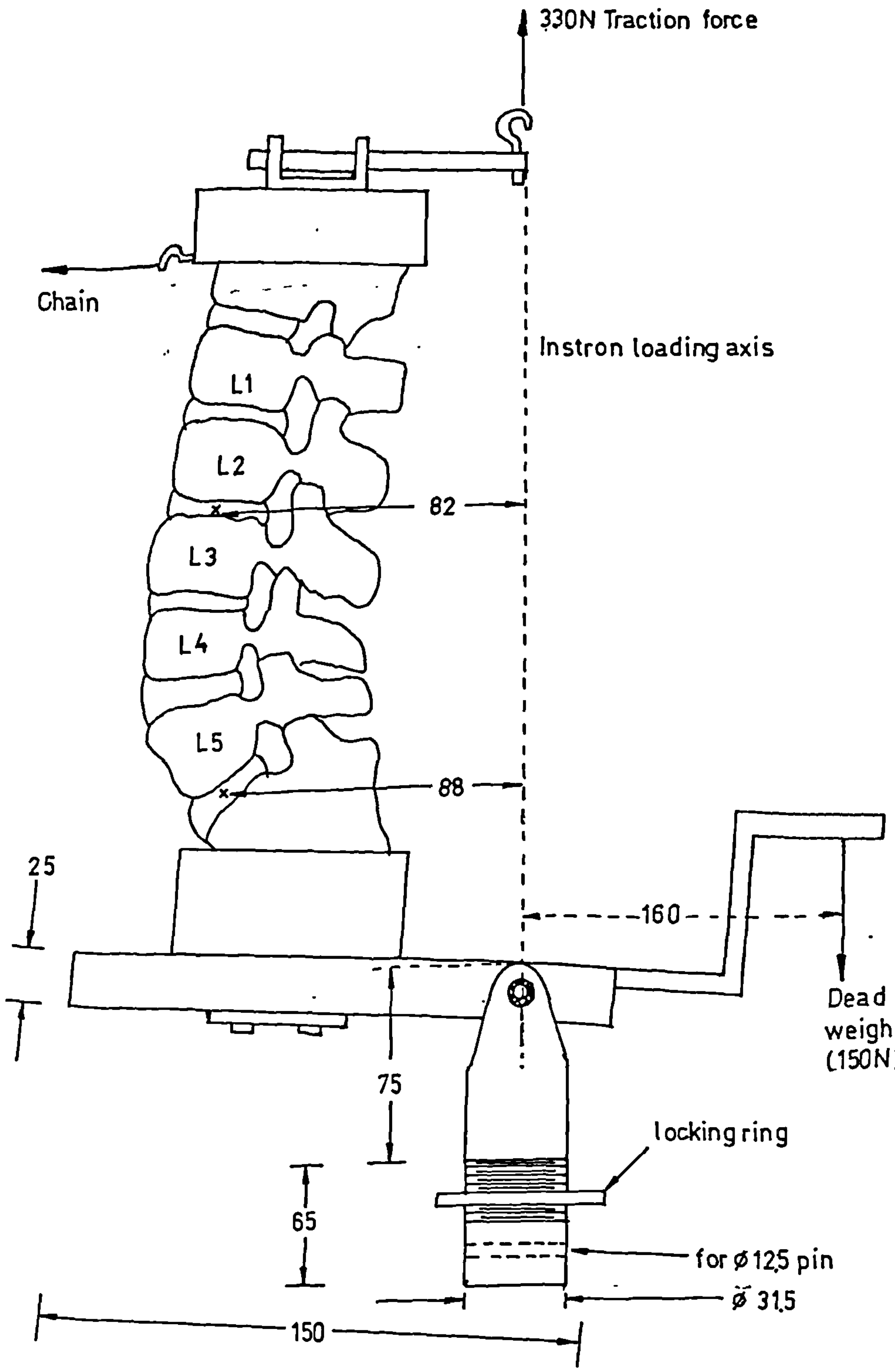


Figure 7.6
A specimen attached to the Instron machine with its moulding cups showing the arrangement of the dead weight, the chains and the markers.



(Dimensions are in millimetres)

Figure 7.7
A schematic diagram showing the position of the specimen in relation to the loading axis of the Instron machine and the dimensions of the beam on which the inferior moulding cup rested.

which contained the T12 vertebra (the superior cup) had the same dimensions but a shorter height of 0.038m because of the smaller vertebral size.

The setting technique was the same as that employed for the motion segment study described in chapter 5 (see section 5.2.2). This involved the insertion of wood screws into the bones to create additional contact surfaces. There were three threaded holes in the wall of each cup through which locking screws were inserted into the Isopon and the bone to rigidly hold the mould in the cup. In order to reduce the amount of Isopon used, the mould was impregnated with chips of mahogany wood. In addition, the bases of both moulding cups had four 0.035m diameter holes. These allowed the mould material to be pushed out of the cup after testing.

The alignment device (figure 5.4) used in the earlier motion segment study was used to ensure that the bases of the moulding cups were parallel to each other and that the specimen was in an anatomical position with the mid-plane of the L3 disc horizontal.

7.3.3 Experimental Apparatus

The experiments were carried out on an Instron testing machine (Model 4505, Instron Limited, High Wycombe, Bucks HP12 3SY, UK). Figure 7.6 show a specimen with its moulding cups attached to the machine.

The superior moulding cup was connected to the load cell of the testing machine by a metal chain which provided the required traction force. The base of the was fitted with a hook (figures 7.5 and 7.6) for attachment of this chain. Anteriorly, it had another hook which was connected to a fixed L-shaped plate by a second chain (figures 7.5 and 7.6). This reproduced the reaction force acting on the body which counteracted the flexion moment imposed on the spine, thus keeping the loading system in equilibrium (see section 7.2).

The inferior moulding cup rested on a beam of 0.15m in length (figures 7.6 and 7.7). The beam was free to rotate about a hinge at one end. The hinge was formed by

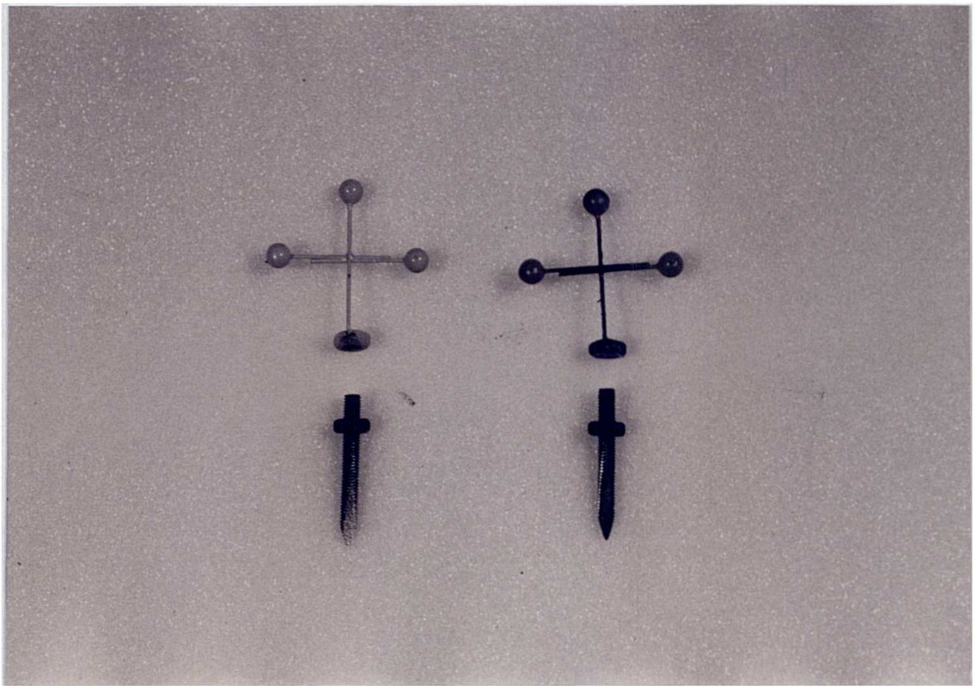


Figure 7.8
The markers which were attached to the spinous processes by tapered screws.

a pin which passed through the beam and two vertical plates, and in line with the loading axis of the testing machine and. The vertical plates were attached to an adaptor which was fitted onto the crosshead of the machine and locked into place by a shear pin and a locking ring.

The position of the hook at the base of the superior cup could be adjusted. The inferior cup could slide along slots drilled in the beam and be fixed in the desired position by screws. These adjustments allow positioning of the specimen so that the L2/3 and L5/S1 disc centres were 0.082m and 0.088m from the loading axis respectively (figure 7.7). The spine was offset from the loading axis because the line of action of the traction force was posterior to the spine (see section 7.2).

A flexion moment of 24Nm about the hinge of the beam was produced by using a dead weight of 150N which was placed at a distance of 0.16m from the hinge (figures 7.6 and 7.7). This was to simulate the moment imposed on the spine due to passive hip and knee flexion during the adoption of the Fowler's position.

Marker systems (figures 7.6 and 7.8) were attached to the spinous processes of all the five vertebrae and the inferior moulding cup which contained the sacrum. Three 0.0034m diameter spherical markers were used for each vertebra. They were connected to each other by pins and then attached to a tapered screw which was inserted into the spinous process. The markers of different vertebrae were painted with different colours to help identification. In order to provide a good contrast, fluorescent paint was used and a black cardboard placed behind the specimen.

The positions of the markers were recorded by a Nikon "F" series 35mm single lens reflex camera with a 28mm lens. Kodak photographic colour slides (film speed ASA 200) were used. After a trial, it was found that the best results were achieved with a shutter speed of 1/125s and an aperture of f4. The camera was placed at a distance of 0.65m from the specimen and mounted onto a tripod. Spirit levels were employed to ensure that the camera was vertical and that the optical axis was

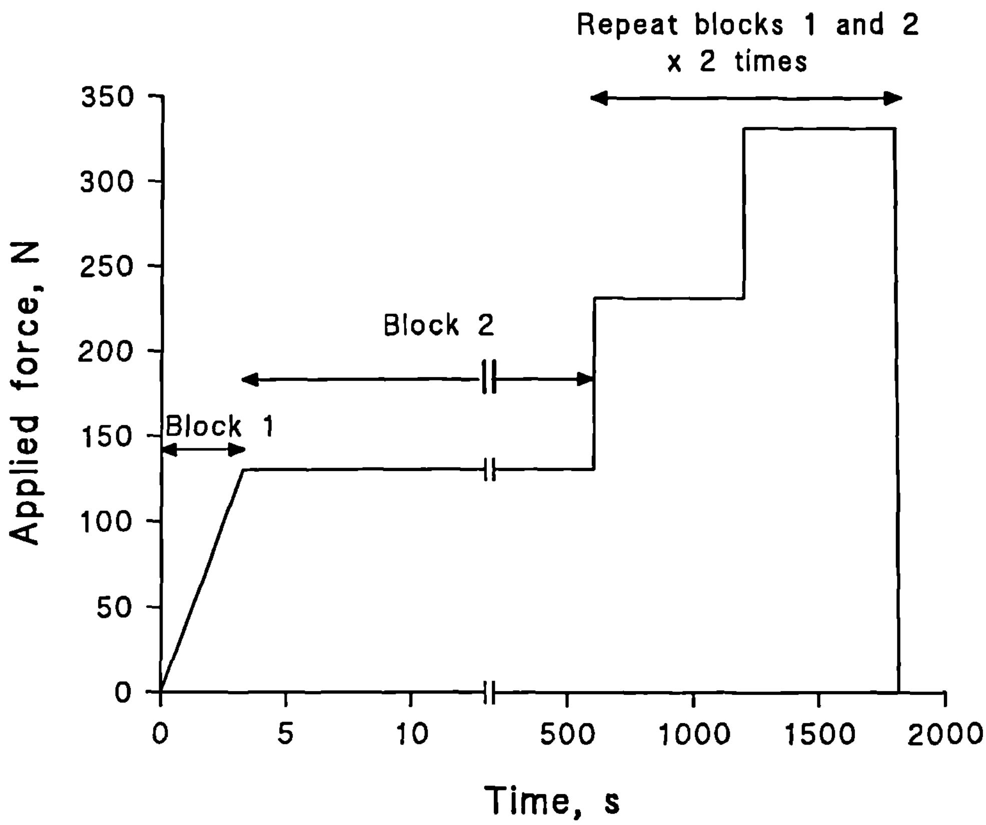


Figure 7.9
The loading cycle of the experiment.

perpendicular to the loading plane of the specimen. Two 500W photographic flood lamps were used to provide lighting and positioned so as to minimise shadowing.

7.3.4 Experimental Procedure

The experimental technique employed in the earlier motion segment study (chapter 5) involved the use of still photography and digitisation of markers' positions to determine the intervertebral motions. It had been shown to provide repeatable data, and thus a similar technique was used in this experiment.

Prior to the mechanical testing, the positions of the markers were recorded by the Nikon camera so as to determine the positions of the motion segments in the unloaded position. The interpedicular and transverse distances of the left and right intervertebral foramina of all motion segments were also measured by a digital calliper. These measurements were taken at the bases of the pins which were inserted into the rigid elements of the foraminal boundaries.

The experimental apparatus was then set up as shown in figure 7.6. Another photographic slide and foraminal size measurements were taken of the specimen 10 minutes after the application of the dead weight (which produced the required 24Nm flexion moment). Initial testing had shown that no further significant change in position occurred after this period of creep.

The Instron machine was set in the "load control" mode. Traction force was applied in three increments up to 130N, 230N and 330N by driving the crosshead downwards (figure 7.9). For each increment, a ramp waveform was first applied at a rate of $40Ns^{-1}$ up to the required force (block 1). Pilot testing showed that commercial traction machines used in hospital applied the force at a similar rate. This was then followed by a ramp input with zero amplitude (effectively "load hold") and a dwell time of 10 minutes to allow the specimen to creep (block 2). At the end of the 10 minute period, the positions of the markers were photographed, and the interpendicular and transverse distances of the intervertebral foramina measured. In order to minimise any loop overshooting and optimise the waveform response, the machine was set with the appropriate proportional, integral and derivative gain values.

The testing machine was programmed to repeat the above two blocks until the maximum force 330N was reached. When the loading sequence was completed, the specimen was unloaded at the same rate as in loading ($40Ns^{-1}$).

A 0.25m ruler was placed alongside the specimen at all times. This allowed the magnification factor to be determined from the photographic slides. In addition, in order to reduce moisture loss of the specimens, they were always covered by moist cotton wool and Cling film. They were only exposed briefly for foraminal size assessment and when the photographic slides were taken.

After the experiment, the photographic negatives were projected (using a Kodak Carousel S-AV2010 Projector) onto a digitising tablet (Digid-pad Type 5A, Gtco Corporation, Columbia, Maryland, USA) where the images were magnified by about 1.9 times. Since the image was magnified, the error involved in measurements was reduced. There were five slides for each specimen - in the unloaded condition, after the application of 24Nm flexion moment, and when 130N, 230N and 300N traction force were applied. For each slide, the positions of the 18 markers and the two ends of the ruler were digitised.

The four corners of the vertebral body images on the first photographic slide (the one when the specimen was unloaded) were identified and marked. In the case of the sacrum, only the anterosuperior and posterosuperior corners were located. These anatomical landmarks were digitised, and thus the spatial relationships between the markers and the landmarks could be determined.

7.3.5 Treatment of Data

The interpedicular and transverse foraminal distances were directly measured by callipers with pins as reference points, and thus no further data reduction was necessary. Determination of the intervertebral movements are described below.

7.3.5.1 Computation of sagittal rotation and translations

The coordinate system employed and the kinematic parameters derived were the same as those used in the earlier chapters and are described in appendix I. The mathematical procedures and equations are also given in the appendix. The computer program used for data reduction was TRACTION.PAS (appendix II).

The coordinates of the digitised points were scaled according to the magnification factor obtained. The spatial relationships between the markers and the four corners of the vertebral bodies were determined from the first photographic slide. If the vertebrae were assumed to be rigid body, this relationship would be fixed in all slides. Based on the relationship, the coordinates of these anatomical landmarks in the other slides were then computed mathematically from the movements of the markers (section A1.2). The anatomical landmarks were not digitised in every slide as this would lead to accumulation of errors involved in identification of landmarks.

The next step in the analysis was to determine the coordinates of the anteroinferior and posteroinferior vertebral body corners with respect to the local coordinate systems. The posteroanterior and superoinferior translations at these corners were calculated (see section A1.4). Sagittal rotation was obtained from the change in the inclination of the line joining the corners (see section A1.3).

It should be noted that only two markers were required for predicting the new positions of the anatomical landmarks after loadings. Since three markers were available for each motion segment, three independent predictions were possible. This allowed three different computations of the various kinematic parameters to be made. These results were averaged to provide the best evaluation of the movements.

7.3.5.2 Error analysis

The previous motion segment study had analysed the error involved in the digitisation of a marker's position, and the optical error due to malalignment and lens distortion of the projector and the camera (see sections 5.2.6.1 and 5.3.1). The error sizes due to these sources were expected to be similar in this study.

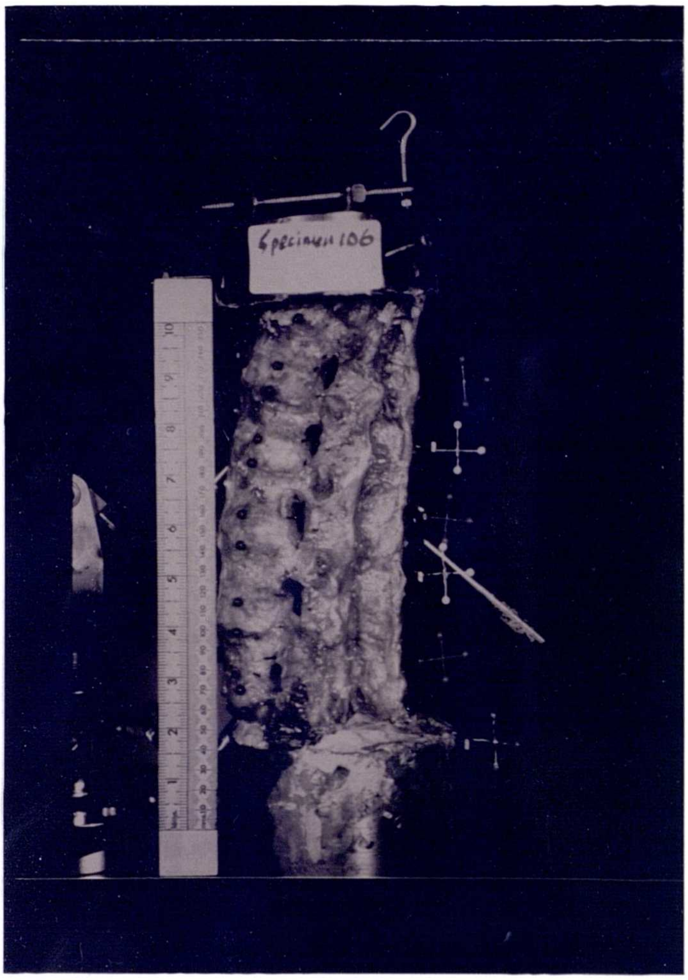


Figure 7.10
Photographs of a specimen (specimen no. 6, male, aged 66) showing the movements produced by the simulated traction loads.
(a) when no load was applied

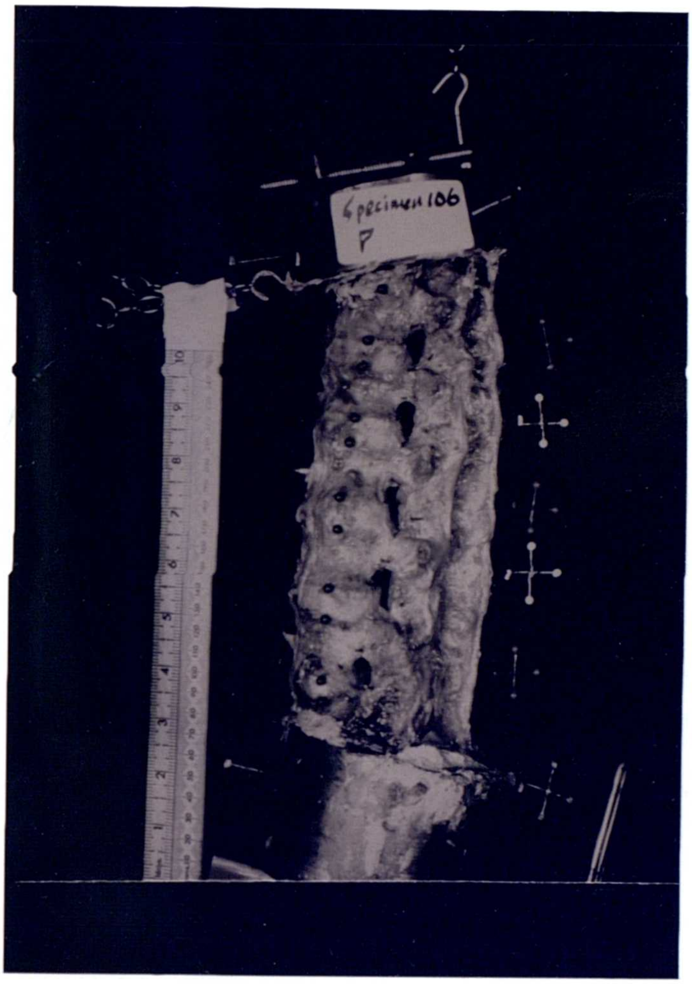


Figure 7.10
Photographs of a specimen (specimen no. 6, male, aged 66) showing the movements produced by the simulated traction loads.
(b) when a flexion moment of 24Nm was applied

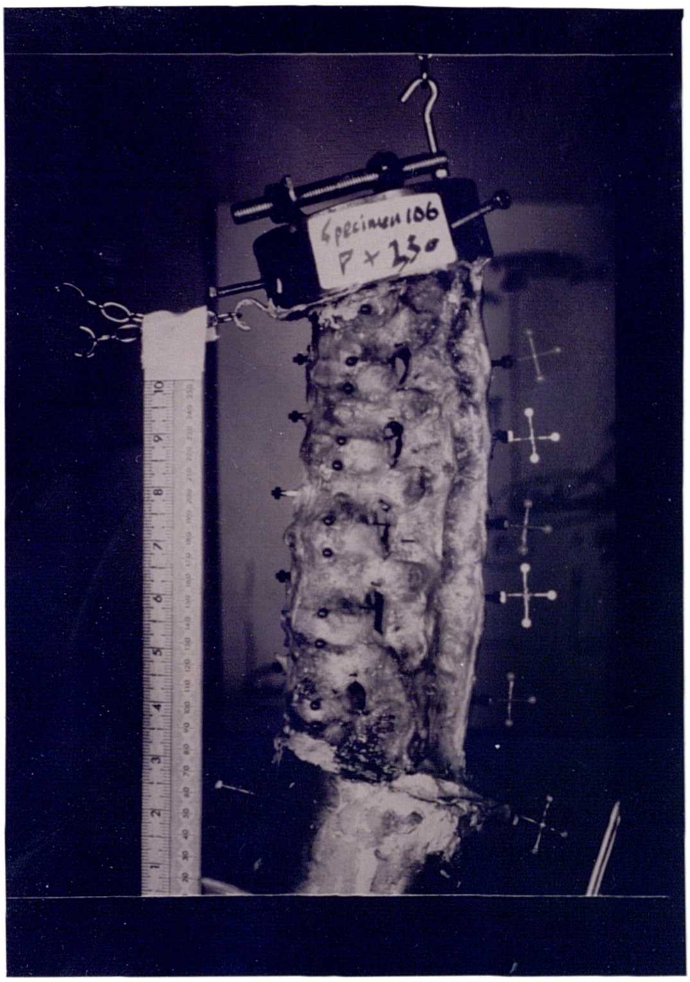


Figure 7.10
Photographs of a specimen (specimen no. 6, male, aged 66) showing the movements produced by the simulated traction loads.
(c) when a flexion moment of 24Nm and a traction force of 330N was applied

	Sagittal flexion, °									
	L1/2		L2/3		L3/4		L4/5		L5/S1	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	2.67	1.54	4.47	1.76	5.03	3.16	8.09	3.60	7.90	6.25
24Nm +130N	3.51	1.62	4.97	1.79	5.86	3.72	8.37	3.60	8.32	6.42
24Nm +230N	3.90	1.74	5.46	1.91	6.12	3.66	8.56	3.62	8.67	6.58
24Nm +330N	4.12	1.86	5.39	1.73	6.35	3.39	8.85	3.57	9.05	6.87

Table 7.3
Mean and standard deviation (SD) of sagittal flexion produced by the 24Nm flexion moment and the application of 130N, 230N and 330N of traction force. (n=7)

	L1/2 Mean	L1/2 SD	L2/3 Mean	L2/3 SD	L3/4 Mean	L3/4 SD	L4/5 Mean	L4/5 SD	L5/S1 Mean	L5/S1 SD
Posteroanterior translations at the anteroinferior vertebral body corners, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	-0.11	1.82	-1.29	1.02	-1.87	1.68	-1.86	2.23	-1.44	3.08
24Nm +130N	-0.50	2.02	-1.28	1.17	-2.15	1.73	-1.85	2.28	-1.46	3.16
24Nm +230N	-0.62	1.99	-1.51	1.30	-2.37	1.56	-2.00	2.47	-1.68	3.36
24Nm +330N	-0.76	2.13	-1.52	1.29	-2.37	1.45	-2.03	2.21	-1.99	3.57
Superoinferior translations at the anteroinferior vertebral body corners, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	-1.66	1.93	-2.65	1.97	-2.21	3.69	-3.29	2.37	-4.30	4.37
24Nm +130N	-1.98	1.81	-2.76	2.21	-3.06	4.62	-3.10	3.22	-4.29	4.50
24Nm +230N	-2.20	1.89	-3.27	2.15	-2.95	4.32	-3.10	3.14	-4.32	4.53
24Nm +330N	-2.37	2.20	-2.82	2.36	-3.06	4.11	-3.34	3.10	-4.64	5.10
Posteroanterior translations at the posteroinferior vertebral body corners, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	-0.05	1.81	-1.04	0.95	-1.62	1.68	-0.86	1.53	-0.97	2.83
24Nm +130N	-0.43	2.01	-1.02	1.09	-1.90	1.74	-0.87	1.68	-0.98	2.94
24Nm +230N	-0.55	1.98	-1.23	1.21	-2.10	1.59	-0.98	1.84	-1.18	3.06
24Nm +330N	-0.70	2.11	-1.24	1.23	-2.08	1.48	-1.02	1.69	-1.50	3.29
Superoinferior translations at the posteroinferior vertebral body corners, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	0.71	1.32	1.14	0.80	2.31	1.61	1.76	1.40	1.59	2.12
24Nm +130N	0.93	1.24	1.14	0.71	2.47	1.77	2.17	2.06	2.10	1.77
24Nm +230N	1.10	1.03	1.35	0.75	2.63	1.52	2.30	1.97	2.18	1.38
24Nm +330N	1.21	1.27	1.63	0.89	2.64	1.62	2.25	1.94	2.29	2.01

Table 7.4.
Mean and standard deviation (SD) of sagittal translations produced by the 24Nm flexion moment and the application of 130N, 230N and 330N of traction force. (n=7) (Translations are denoted as positive in the superior and posterior directions, and negative in the inferior and anterior directions.)

	L1/2 Mean	L1/2 SD	L2/3 Mean	L2/3 SD	L3/4 Mean	L3/4 SD	L4/5 Mean	L4/5 SD	L5/S1 Mean	L5/S1 SD
Increases in the left interpedicular distances, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	0.74	0.48	1.56	0.36	1.68	1.32	1.97	0.86	1.98	1.59
24Nm +130N	1.15	0.39	1.75	0.37	1.80	1.34	2.52	1.37	2.05	2.00
24Nm +230N	1.31	0.35	2.00	0.32	1.86	1.38	2.63	1.44	2.24	2.07
24Nm +330N	1.36	0.36	2.18	0.35	2.15	1.30	2.82	1.68	2.28	2.08
Increases in the right interpedicular distances, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	0.84	0.39	1.45	0.34	1.44	1.16	1.94	1.29	1.77	1.49
24Nm +130N	1.24	0.50	1.68	0.25	1.72	0.90	2.50	1.66	2.04	1.79
24Nm +230N	1.30	0.55	1.67	0.44	1.85	1.06	2.66	1.81	2.19	1.81
24Nm +330N	1.50	0.32	2.15	0.55	2.00	1.01	3.06	1.94	2.32	1.91
Increases in the left transverse foraminal distances, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	0.53	0.45	1.17	0.56	1.82	1.34	2.04	1.48	1.33	1.29
24Nm +130N	0.63	0.64	1.25	0.57	1.92	1.46	2.27	1.48	1.45	1.41
24Nm +230N	0.64	0.87	1.43	0.59	2.04	1.68	2.44	1.61	1.52	1.42
24Nm +330N	0.93	0.69	1.59	0.60	2.33	1.56	2.59	1.53	1.61	1.64
Increases in the right transverse foraminal distances, mm										
Unloaded	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
24Nm	0.54	0.40	1.14	0.80	1.47	1.19	1.68	1.09	1.31	1.12
24Nm +130N	0.68	0.42	1.22	0.74	1.78	1.08	1.98	1.27	1.48	1.42
24Nm +230N	0.93	0.44	1.37	0.81	1.91	0.98	2.26	1.30	1.77	1.92
24Nm +330N	1.02	0.49	1.48	0.72	2.11	1.11	2.45	1.43	1.79	1.98

Table 7.5.
Mean and standard deviation (SD) of the increases in the left and right interpedicular and transverse distances of the intervertebral foramina after the application of the 24Nm flexion moment, and 130N, 230N and 330N of traction force. (n=7)

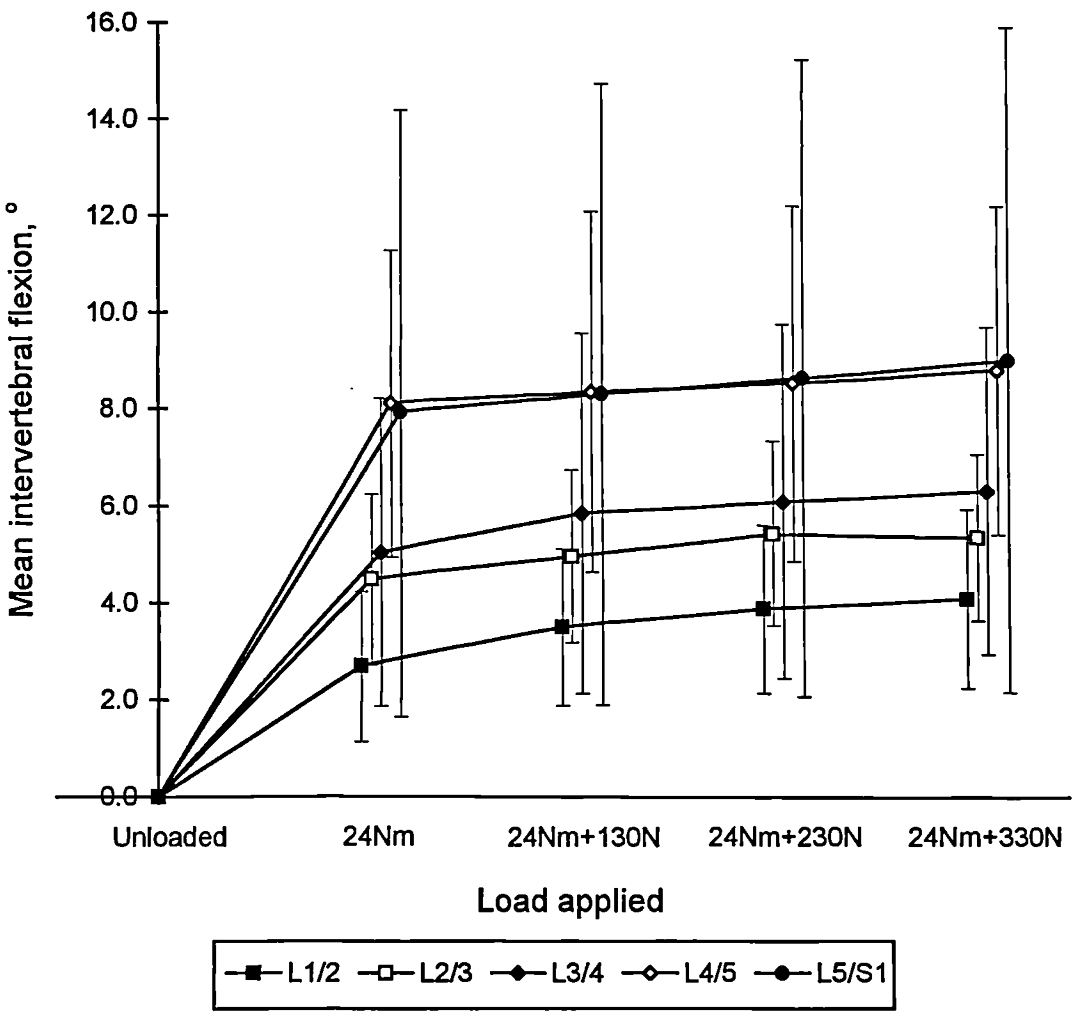


Figure 7.11
Mean intervertebral flexion produced by the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

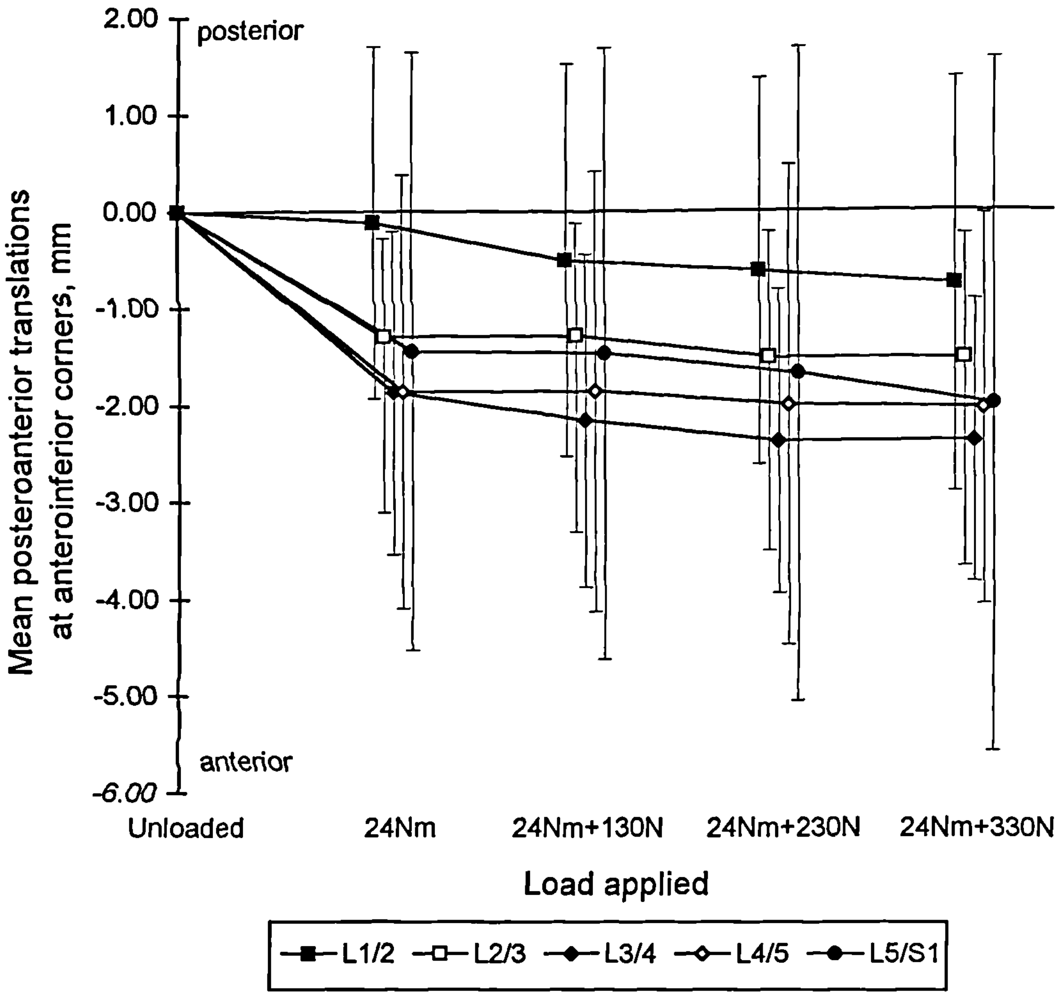


Figure 7.12
Mean posteroanterior translations at the anteroinferior vertebral body corners under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

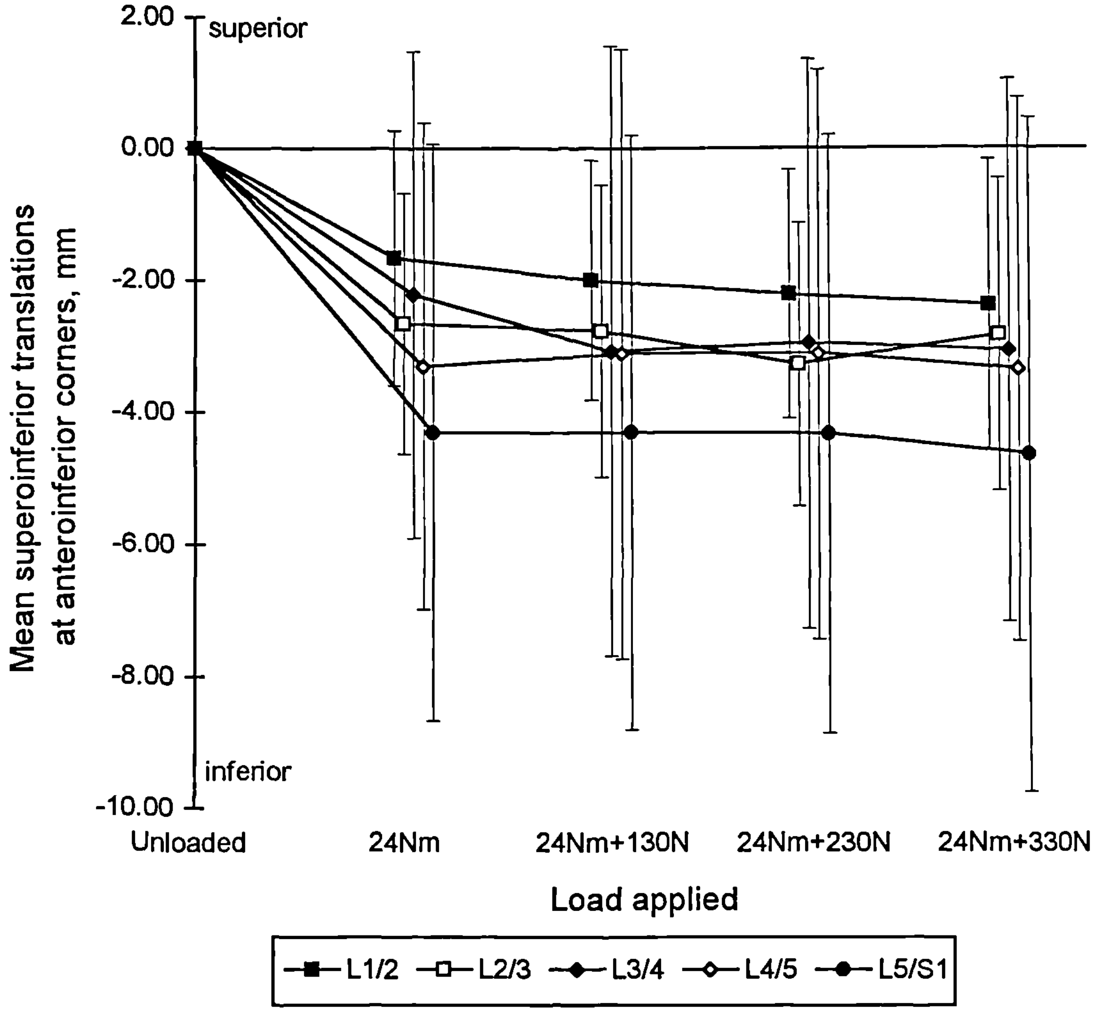


Figure 7.13
Mean superoinferior translations at the anteroinferior vertebral body corners under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

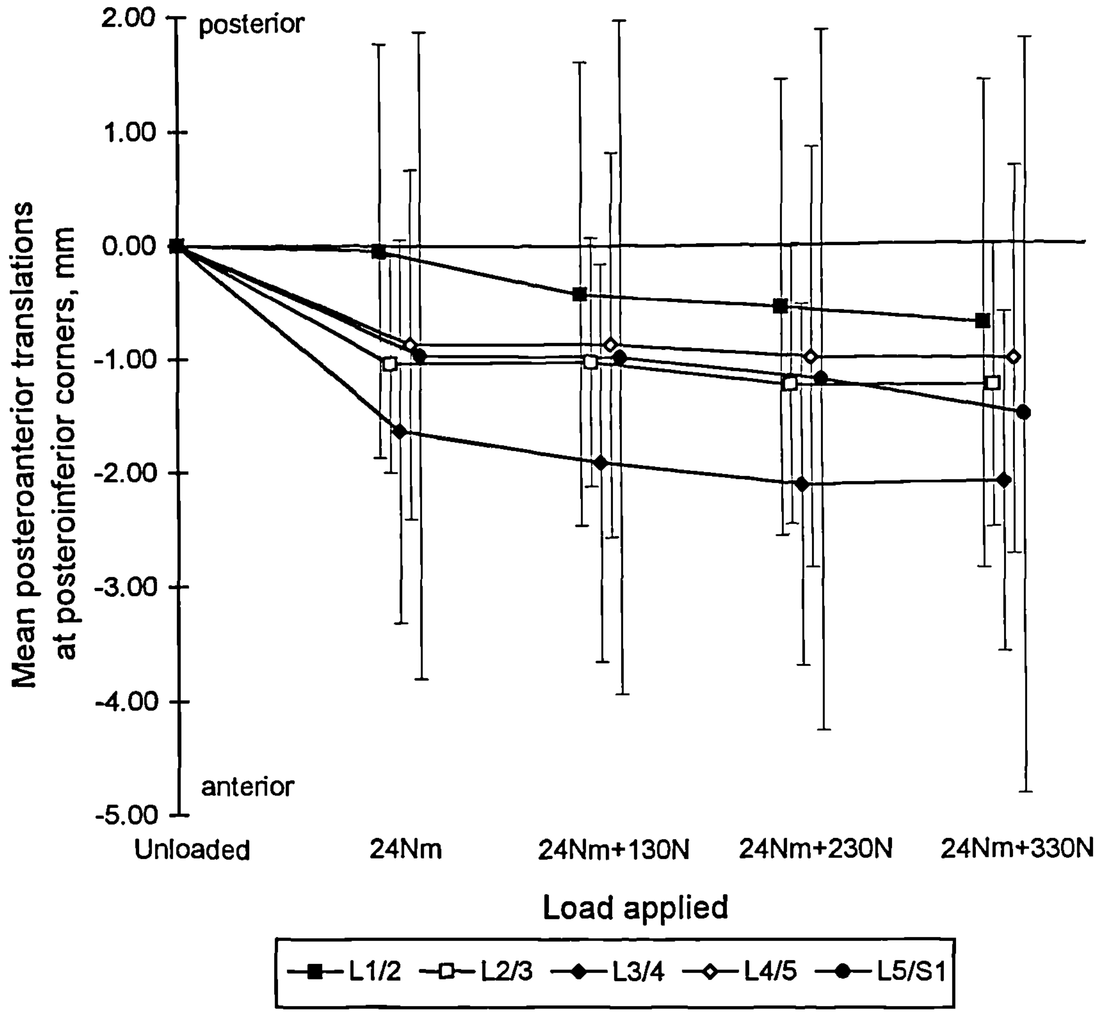


Figure 7.14
Mean posteroanterior translations at the posteroinferior vertebral body corners under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

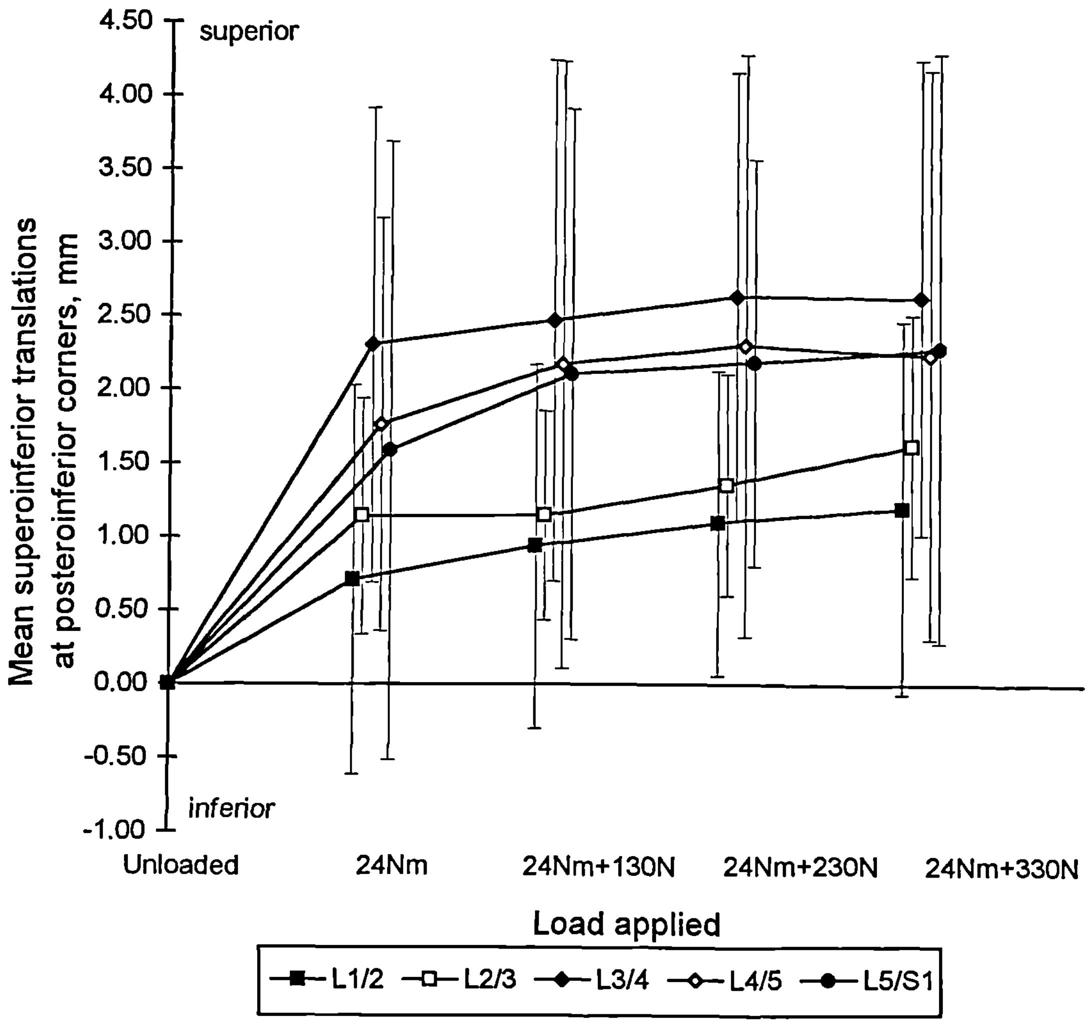


Figure 7.15
Mean superoinferior translations at the posteroinferior corners under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7) (The vertical bars denote ± 1 standard deviation for the 7 specimens.)

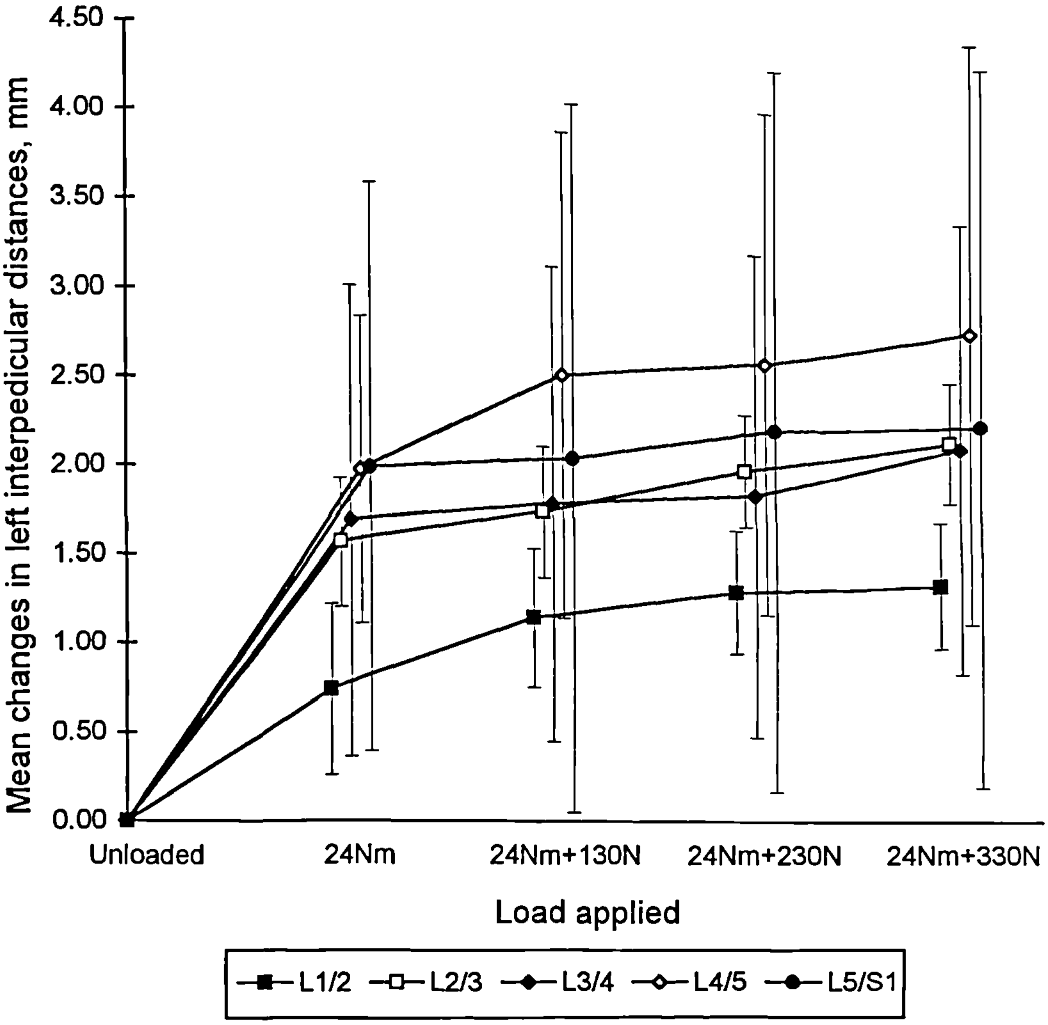


Figure 7.16
Mean changes in the left interpedicular distances under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

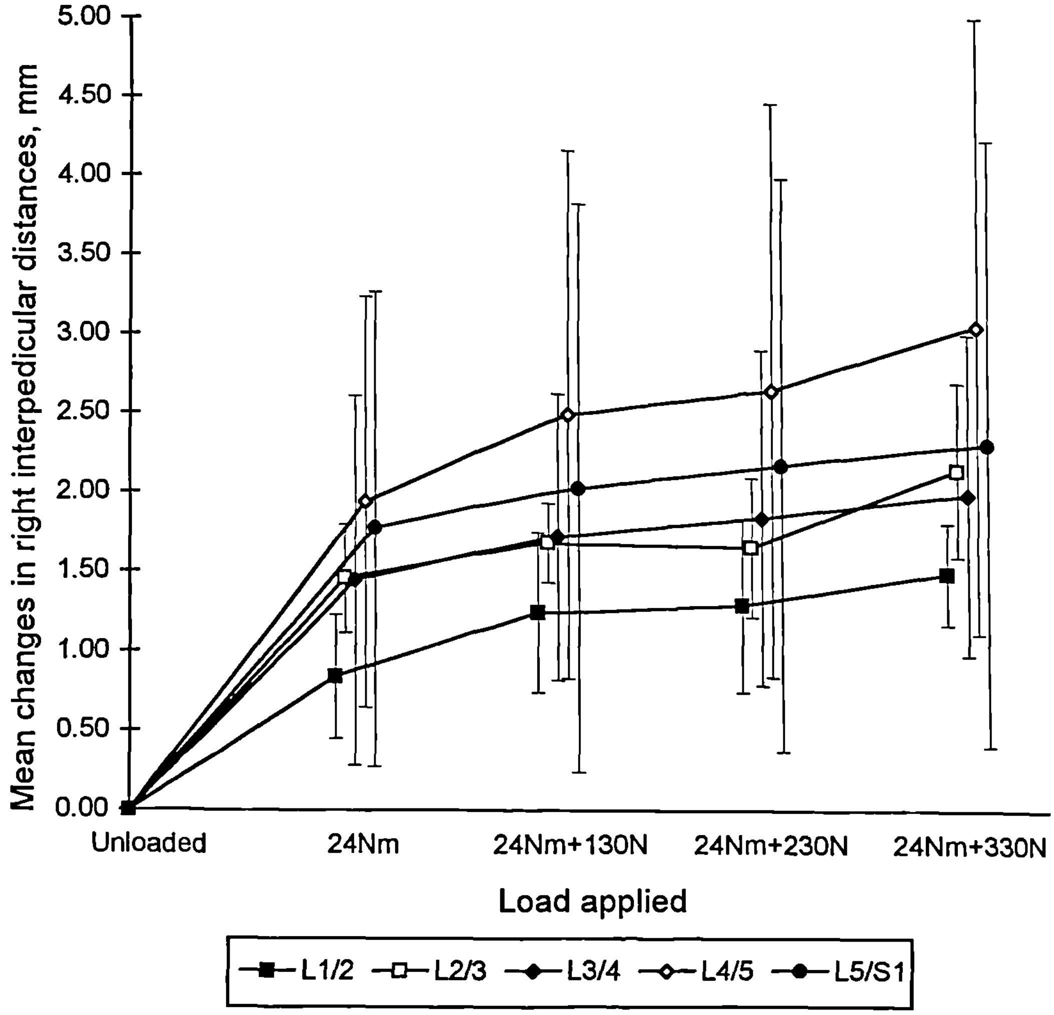


Figure 7.17
Mean changes in the right interpedicular distances under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

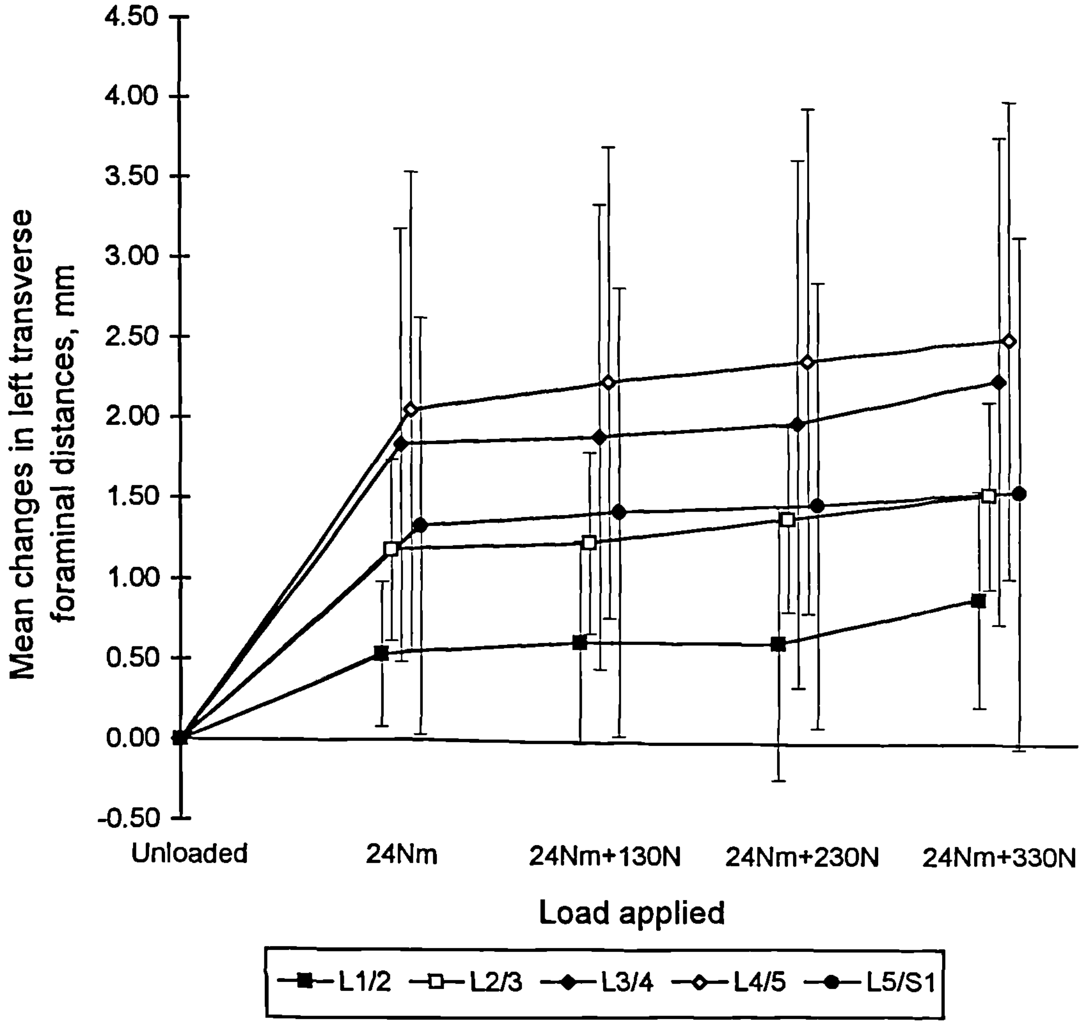


Figure 7.18
Mean changes in the left transverse foraminal distances under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

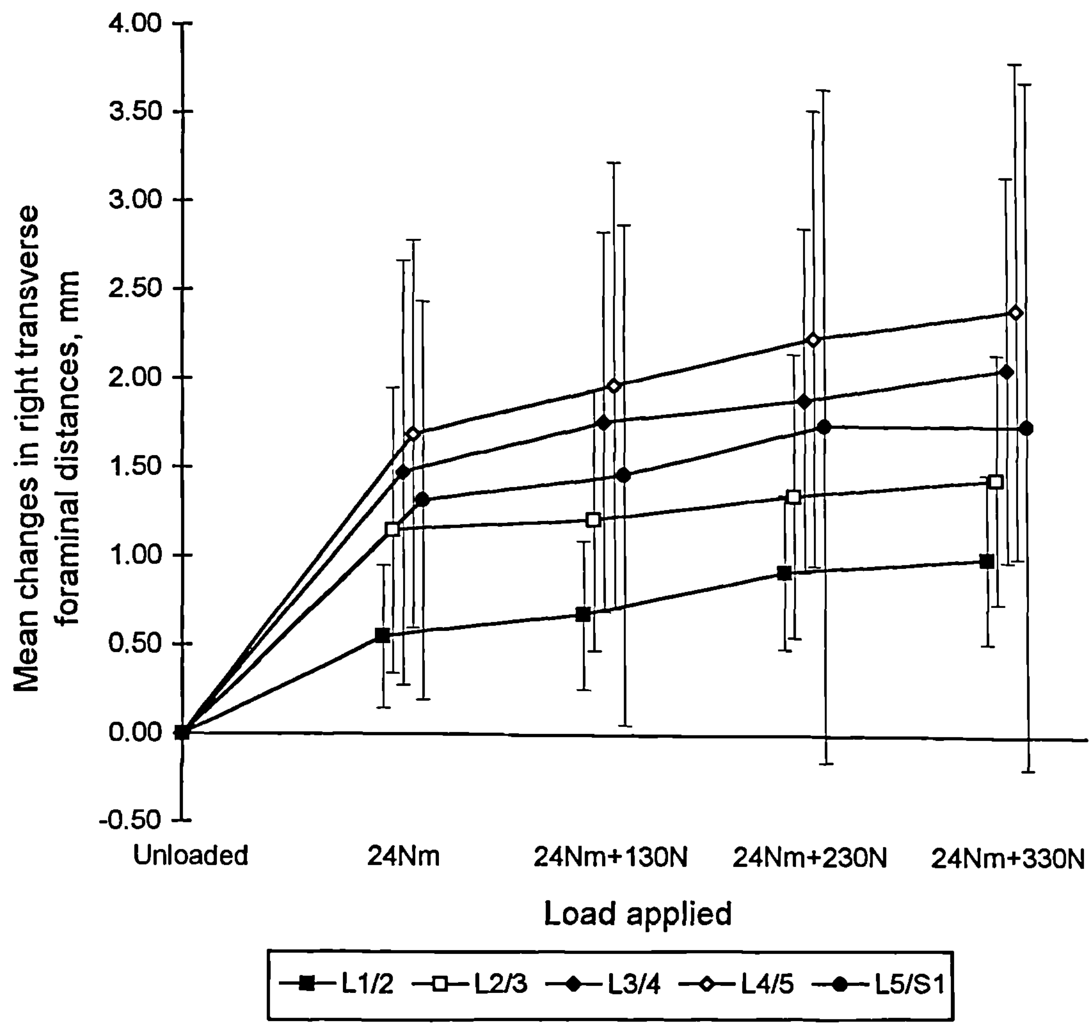


Figure 7.19
Mean changes in the right transverse foraminal distances under the application of the 24Nm flexion moment and the 130N, 230N and 330N traction force. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

The error associated with the marking of the vertebral corners on photographic slides had not been subjected to analysis previously. The marking process was thus repeated five times in one of the specimens. The standard deviations of these repeated measurements, which represented the error in identifying the landmarks, were computed. The standard deviations of the kinematic parameters determined from these five sets of data were also computed. This allowed assessment of the effects of the error on the kinematic parameters.

The repeatability of measurements of interpedicular and transverse foraminal distances was also evaluated. The measurement of the foraminal distances in the unloaded state was repeated two times in one of the specimens. The root mean square error of the measurement was determined and the correlation of the two sets of data assessed by determining the intraclass reliability coefficient (R).

The extent of out of plane movements was checked by computing the distances among the markers of the vertebrae which should remain constant in all photographic slides (see section 5.2.6.1). Three intermarker distances were computed for each vertebra of the specimen as there were three markers. The errors in the intermarker distances (i.e. standard deviations) and the coefficients of variation (i.e. standard deviations expressed as percentages of the mean intermarker distances) were determined.

7.4 RESULTS

Figure 7.10 shows the spinal movements of a specimen (specimen no. 6, male, aged 66) produced by the traction loads as appeared in the photographs. The descriptive statistics are shown in tables 7.3-7.5 and figures 7.11-7.19.

7.4.1 Sagittal Rotation

Traction loads were found to produce flexion of the motion segments at all levels. Mean flexion of 2.7°-8.1° was observed at the segments after the application of the 24Nm moment which was employed to simulate the effects of passive hip and knee flexion (table 7.3 and figure 7.11). This accounted for about 80% of the total movements observed. The addition of the 330N traction force further increased the movements by 0.7°-1.4° only. The above findings were consistently observed in all specimens.

In general, the L1/2 segment showed much less flexion movement (mean maximum flexion=4.1°) than the other segments. The L4/5 and L5/S1 segments generally exhibited the greatest amount of flexion. The mean maximum flexion at these segments was 8.8° and 9.0° respectively.

7.4.2 Sagittal Translations at the Anteroinferior Vertebral Body Corners

Table 7.4 and figures 7.12-13 show the sagittal translations produced at the anteroinferior corners of the vertebral bodies. Generally speaking, the corners were found to translate anteriorly and inferiorly. The mean maximum anterior translations of the various segments ranged from 0.8mm to 2.4mm, and the mean maximum inferior translation from 2.4mm to 4.6mm. There were no major differences in the magnitude of the mean anterior and inferior translations among the different spinal segments.

Mean anterior translation of 0.1-1.9mm and inferior translation of 1.7-4.3mm occurred after the application of the 24Nm flexion moment. As in the case of the flexion movement, these were about 80% of the total movements produced. Hence, the 330N traction force produced little further increases in the movements.

There were some exceptions to the general trends described above. Seven out of the 35 motion segments examined exhibited posterior translations (L1/2, L2/3 and L4/5 of specimen no.1, L1/2 and L5/S1 of specimen no.4, and L1/2 and L4/5 of specimen no.5). The translations at the L1/2 segments in the posteroanterior direction were least consistent (4 specimens showed anterior translation and 3 posterior) and thus the mean magnitude at this segment level tended to be small compared with the other levels. In addition, there were three segments which showed superior translations rather than inferior translations (L2/3 of specimen no.1, L3/4 of specimen no.2 and L4/5 of specimen no.6).

7.4.3 Sagittal Translations at the Posteroinferior Vertebral Body Corners

The sagittal translations at the posteroinferior corners of the vertebral bodies are shown in table 7.4 and figures 7.14-7.15. The general trend was that the specimens showed anterior and superior translations at these corners. The mean maximum anterior translation ranged from 0.7mm to 2.1mm and the mean maximum superior translation from 1.2mm to 2.6mm.

Like the other kinematic parameters, most of the anterior and superior translations at the posteroinferior corners occurred after the application of the 24Nm moment. The magnitude of the mean anterior translation was generally small. There were no major segmental variations in the magnitude. In regard to the mean superior translation, the L1/2 and L2/3 segments tended to exhibit less movement and there were no major differences in the magnitude among the other segmental levels.

The observation of superior translation at the posteroinferior corners was consistent. None of the specimens showed significant inferior translation (more than 1mm). However, some specimens departed from the general trends in regard to translation in the posteroanterior direction. Seven of the motion segments examined showed posterior translation (L1/2, L2/3 and L4/5 of specimen no.1, L1/2 and L5/S1 of specimen no.4, L1/2 and L4/5 of specimen no.5). Interestingly, these segments were the same as those which exhibited posterior translations at the anteroinferior corners.

7.4.4 Changes in the Foraminal Sizes

Increases in the interpedicular and transverse distances of the intervertebral foramina were observed after the application of the traction loads (table 7.5; figures

	Error in measurements (= 1 standard deviation), mm			
Vertebra	Anteroinferior corner	Posteroinferior corner	Anterosuperior corner	Posterosuperior corner
x-coordinate				
L1	0.17	0.04	0.23	0.09
L2	0.40	0.47	0.11	0.10
L3	0.61	1.08	0.54	0.83
L4	0.59	0.30	0.71	0.24
L5	0.15	0.73	0.46	0.51
S1			0.19	0.42
y-coordinate				
L1	0.28	0.30	0.48	0.33
L2	0.19	0.23	0.22	0.35
L3	0.37	0.36	0.45	0.28
L4	0.25	0.09	0.21	0.16
L5	0.76	0.30	0.10	0.42
S1			0.33	0.62
Mean	0.38	0.39	0.34	0.36
	Overall mean error = 0.37mm			

Table 7.6.
Error in identifying the vertebral body corners on photographic slides (= 1 x standard deviation of five repeated measurements).
(The x- and y-coordinates refer to those of the coordinate system of the digitiser. The values given have been corrected for magnification.)

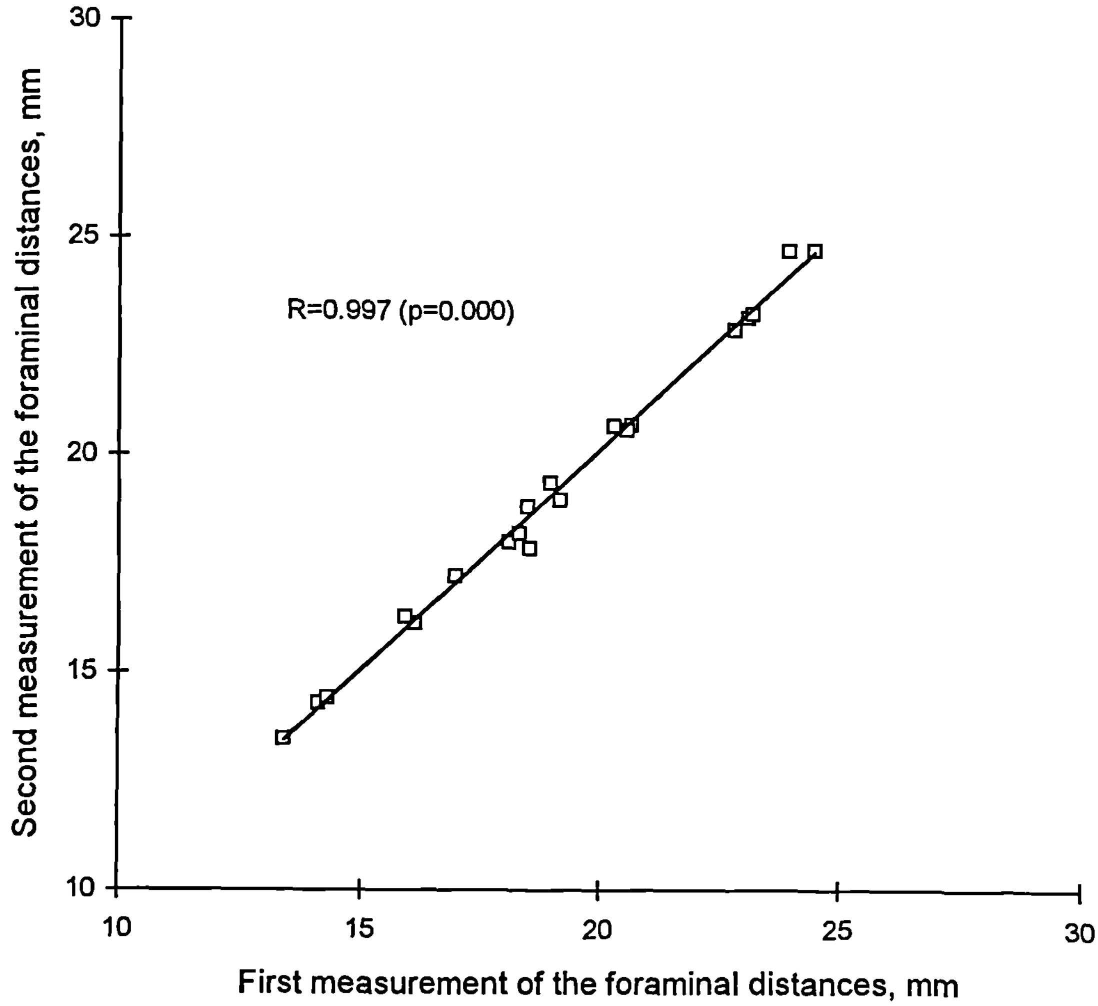


Figure 7.20
Correlation between two independent measurements of the foraminal distances of a specimen.

	Mean error in distances between markers, mm (mean coefficient of variation, %)						
Distance between markers	L1	L2	L3	L4	L5	S1	*Row mean*
1 and 2	0.16 (1.0%)	0.21 (1.3%)	0.24 (1.5%)	0.24 (1.5%)	0.21 (1.3%)	0.25 (1.6%)	*0.22*
2 and 3	0.26 (1.6%)	0.30 (1.8%)	0.20 (1.2%)	0.27 (1.7%)	0.23 (1.4%)	0.25 (1.5%)	*0.25*
1 and 3	0.26 (1.1%)	0.26 (1.1%)	0.20 (0.9%)	0.23 (1.0%)	0.24 (1.1%)	0.30 (1.3%)	*0.25*
Column mean	*0.23*	*0.26*	*0.22*	*0.24*	*0.23*	*0.26*	*Overall mean=0.24*

Table 7.7.
The mean errors in the distances between markers (error=1 x standard deviation of the marker distance). (The three markers of each vertebra are labelled as 1, 2 and 3.)

7.16-7.19). The trends observed for the left and right foramina were essentially the same. The mean maximum increases in the interpedicular distances ranged from 1.4mm to 3.1mm, whereas those in the transverse distances were smaller in magnitude, ranging from 0.9mm to 2.6mm.

Most of the increases in the foraminal distances occurred after the application of the 24Nm flexion moment. Addition of the 330N traction force led to little further increases. There were no major segmental variations in the magnitude of the mean increases in the foraminal distances, although the L1/2 segments tended to exhibit least increases and the L4/5 segments the greatest increases.

There were three exceptions (the left L1/2 and L3/4 foramina of specimen no.1 and the right L5/S1 of specimen no. 7) among the 70 foramina examined. In these cases, there were decreases of less than 0.9mm in the transverse foraminal distances.

7.4.5 Measurement Errors

The mean error in identifying the vertebral body corners on the photographic slides was found to be 0.37mm (table 7.6). There were no major differences in the error among different anatomical landmarks and among different vertebrae. The errors in determining sagittal rotation and translation as a result of errors in identifying of the landmarks were 0.60° and 0.76mm respectively. These values were relatively small compared to the magnitude of the movements observed.

It was shown there was very strong correlation between the two sets of measurements of the interpedicular and transverse foraminal distances ($R=0.997$; $p=0.000$) (figure 7.20). The root mean square error in the foraminal measurement was 0.30mm. It was therefore concluded that the technique provided sufficiently repeatable data.

The mean errors in the intermarker distances obtained from the different photographic slides of the seven specimens are shown in table 7.7. There were no major differences in the mean errors among different vertebrae. The mean error ranged from 0.16mm to 0.30mm and the mean coefficient of variation from 0.9% to 1.8%. The changes in the distances between markers as appeared in different slides were considered to be small, indicating that there were negligible movements outside the sagittal plane. This confirmed the assumption that the movements produced by the simulated traction loads were principally confined to the sagittal plane.

7.5 DISCUSSION

7.5.1 Intervertebral Movements Produced by Traction

The present study showed that flexion of the motion segments was consistently observed under the application of simulated traction loads. The flexion movement was generally achieved by inferior translation at the anteroinferior corner of the vertebral body and superior translation at the posteroinferior corner. Superior translations were observed at both corners in 3 out of the 35 motion segments studied (see section 7.4.1). In these cases, the superior translation was greater at the posterior corner than at the anterior corner, thereby producing the flexion movement.

The application of the dead weight generated a flexion moment of 24Nm on the motion segments. The applied traction force also produced a flexion moment on the motion segments as it was posterior to the disc centres. The observed sagittal rotation was in the direction of flexion.

The flexion movement produced was generally greater at the L4/5 and L5/S1 segments than at the upper three lumbar motion segments. This could be explained by the boundary conditions at the two ends of the spine. At the pelvic end, significant flexion moment was imposed on the spine by the dead weight and the traction force. The thoracic end was loaded by two chains, producing effectively a pinned boundary. This was free to rotate and therefore the upper motion segments would experience less flexion moment.

In addition, as explained in section 7.2, the disc centres of the L4/5 and L5/S1 motion segments are further away from the line of action of the traction force than those of the upper segments (Tracy et al, 1989). This will result in a larger amount of flexion moment at the lower segments.

The observation that the posterior corner translated superiorly and the anterior corner inferiorly was consistent with the results of the earlier studies (Colachis and Strohm, 1969; Reilly et al, 1979), which reported that there were generally increases in posterior disc heights and decreases in anterior disc heights. The L1/2 and L2/3 segments were found to exhibit less superior translation than the lower segments. The earlier studies also demonstrated the same results. The smaller changes in the translations at the upper segments were due to the lesser amount of flexion movement produced.

The magnitude of the superoinferior translations observed in the present study was similar to the values reported by Colachis and Strohm (1969) (see table 4.5). When the results of the two studies are compared, some differences would be expected. This is due to the difference in the coordinate system employed and the difference between in-vivo and in-vitro testing. The structural integrity and the passive resistance of muscles are lost in cadaveric spines, although their effects are unlikely to be large. Furthermore, the magnitude of traction force applied in the two studies is different.

It should be pointed out that Colachis and Strohm (1969) performed radiographic measurements of deforming spines and the changes in disc heights were measured by a much less accurate technique (engineering calliper). Their results were thus subjected to large errors. The use of markers and digitisation technique had significantly improved the accuracy of the technique of the present work which would provide more reliable data.

No previous study had quantified the amount of translations in the posteroanterior direction. The present work showed that the motion segments generally showed anterior translations at both the posteroinferior and anteroinferior vertebral body corners, indicating that the vertebra translated anteriorly as a whole during traction application. Unlike the other kinematic parameters, there was no major segmental variation in this parameter. The chain which simulated the reaction force at the thoracic end produced anterior shear forces of similar magnitude along the length of the spine. The above experimental results are thus expected.

Seven motion segments departed from the general trend and showed posterior translations at both vertebral body corners. It was unclear why these segments showed different behaviour. This was not likely due to the degeneration of their discs as they were not particularly degenerated. A possible explanation was that flexion of these motion segments might have rotated their endplates anteriorly so that the traction force would tend to pull the vertebra posteriorly. The effect of this posterior shear might have outweighed that of the anterior shear produced by the chain.

An interesting finding of the present study was that most of the flexion and sagittal translations produced occurred after the application of the 24Nm flexion moment which was to simulate the effects of passive hip and knee flexion. Similar observation was made by Colachis and Strohm (1969). They reported that the changes in disc heights with the adoption of the Fowler's position were larger than those obtained with traction (see table 4.5). Reilly et al (1979) also showed that posterior disc heights increased as the amount of hip flexion increased, indicating that passive hip flexion would increase the flexion of the segments and the superior translation posteriorly. The implication of the above observation is that most of the mechanical effects occur after the adoption of the Fowler's position. The additional effects produced by the traction force are small.

The large amount of spinal deformation produced by the 24Nm moment could be explained by the non-linear load-deformation characteristics of spinal motion segments (figures 2.7, 3.1 and 3.2). The spine has very low stiffness in the early toe phase of the load-deformation curve, and thus a large amount of deformation is produced when the dead weight is applied. However, after the application of this load, the segments have probably reached the linear region of the curve and exhibit a large increase in stiffness. Therefore, little further deformation is produced.

7.5.2 Changes in the Foraminal Sizes

The present study showed that traction loads produced consistent increases in the interpedicular and transverse foraminal distances. These increases were symmetrical on the left and right side of the spines as traction produced movements which were principally confined to the sagittal plane and there were insignificant rotation and lateral bending. Since both the interpedicular and transverse distances increased, there would be overall increases in the cross-sectional areas of the foramina.

The increases in the interpedicular distances were related to the superior translations produced at the posteroinferior corners of the vertebral bodies and the flexion of the segments, both of which would tend to increase the separation of the superior and inferior pedicles. In addition, the anterior and superior translations at posterior vertebral corner would increase its distance from the facets and thus the transverse foraminal distances.

Since the changes in the foraminal sizes are closely related to the movements of the segments, they generally followed the same trends as observed in the various kinematic parameters. Most of the increases in foraminal sizes occurred after the application of the 24Nm moment and the upper motion segments tended to exhibit smaller increases.

7.5.3 Experimental Validity and Reliability

Every effort was made to reproduce the in-vivo loading conditions of traction therapy. The anterior reaction force at the thoracic end was reproduced by a chain. Previous in vivo study (Lee, 1990) showed that the thoracic cage was not restrained from a small amount of rotation in the sagittal plane. No moment was thus applied at the thoracic end in this experiment. The chains effectively reproduced the pinned boundary condition. At the pelvic end, the dead weight was applied to produce the moment that would be imposed by the adoption of the Fowler's position. The reaction force at the pelvic end was not simulated in this experiment as its magnitude was small. The weight of the pelvis was almost counterbalanced by the vertical component of the traction force. In addition, the traction force was offset from the loading axis. The magnitude of the traction force and the loading rate were similar to those used in a routine clinical session. The present work was thus considered to be more realistic than the earlier cadaveric study (Twomey, 1985) which simply applied 90N force along the longitudinal axis of the spine.

The present results compared favourably with those of the study of Colachis and Strohm (1969) which was conducted on living subjects. This supported the validity of the experiment implying that the in-vivo loading conditions were successfully reproduced. However, as in any other cadaveric study, the results should be interpreted in the light of post-mortem changes in mechanical properties and the characteristics of the specimens employed.

In the present work, the specimens were generally old and showed a mild degree of degeneration as evidenced by macroscopic examination and discogram. Twomey (1985) showed that traction produced more deformation in non-degenerated motion segments than in degenerated ones. This factor should be considered in interpreting the present results. In addition, it should be noted that the sample size of the present study was small as the number of specimens which could be obtained within a reasonable period of time was limited. However, the experimental findings were highly consistent among the specimens and this allowed valid conclusions to be drawn on the mechanical effects of traction.

Vigorous assessment of the accuracy of the measurement techniques employed in the present study were conducted. The errors associated with identifying the vertebral body corners were found to be small (see section 7.4.5). Since the corners were identified in only one of the slides and their positions in the others predicted mathematically, the accumulation of errors in successive slides, which would alter the computed movement patterns, were eliminated.

The earlier motion segment study demonstrated that the photographic technique was accurate, and the errors associated with lens distortion and malalignment of the camera and slide projector were small (see section 5.3.1). The changes in the intermarker distances in successive slides were about 1%, indicating that there were very little out-of-plane movements which might influence the accuracy of the measurements. Finally, in regard to the measurement of the foraminal distances, the error involved was also found to be small (see section 7.4.5).

7.6 CONCLUSION

The present study revealed that traction would produce flexion of the motion segments leading to a loss of lordosis. This was generally achieved by superior translation of the vertebral body posteriorly and inferior translation anteriorly. The segments also tended to translate anteriorly as a whole.

Increases in the interpedicular and transverse foraminal distances were observed during the application of traction loads. These increases were associated with the intervertebral movements produced by traction.

Most of the above mechanical effects of traction appeared to take place after the application of the flexion moment which was to simulate the effects of the Fowler's position, and the effects were also generally found to be more pronounced at the lower lumbar segments.

In conclusion, the present work has developed an accurate and valid experimental technique to provide valuable data on the deformation of the spine produced by traction. The implications of these data on the mechanisms underlying traction will be fully addressed in chapter 9.

■ *Chapter 8*

Time-dependent Characteristics of Lumbar Traction

Time-dependent Characteristics of Lumbar Traction

8.1 INTRODUCTION

There is a paucity of literature on the time-dependent mechanical effects of lumbar traction (see section 4.2.4). In clinical practice, traction is often applied in an intermittent manner with periods of "hold" and "rest". A study was carried out to examine the time-dependent characteristics of intermittent traction. The data thus provided would complement those obtained in the study described in chapter 7. As demonstrated in the previous experiment, lumbar traction produced displacements principally in the sagittal plane. The present study was therefore also limited to the analysis of sagittal displacements.

More specifically, the purpose of the study was to examine the changes in

- the overall length of the whole lumbosacral spine,
- the sagittal rotation of the whole lumbosacral spine, and
- the displacements of individual lumbar vertebrae

during and after repeated applications of traction loads.

8.2 MATERIALS AND METHODS

The present study employed the same seven specimens as were used in the study described in chapter 7. In summary, they were complete osteoligamentous lumbosacral spines with moderate degree of degeneration. The T12 and sacrum of these specimens were set in moulding cups. Three markers (labelled as 1, 2 and 3; see figure 7.8) were attached to the spinous processes of each of the lumbar vertebra and

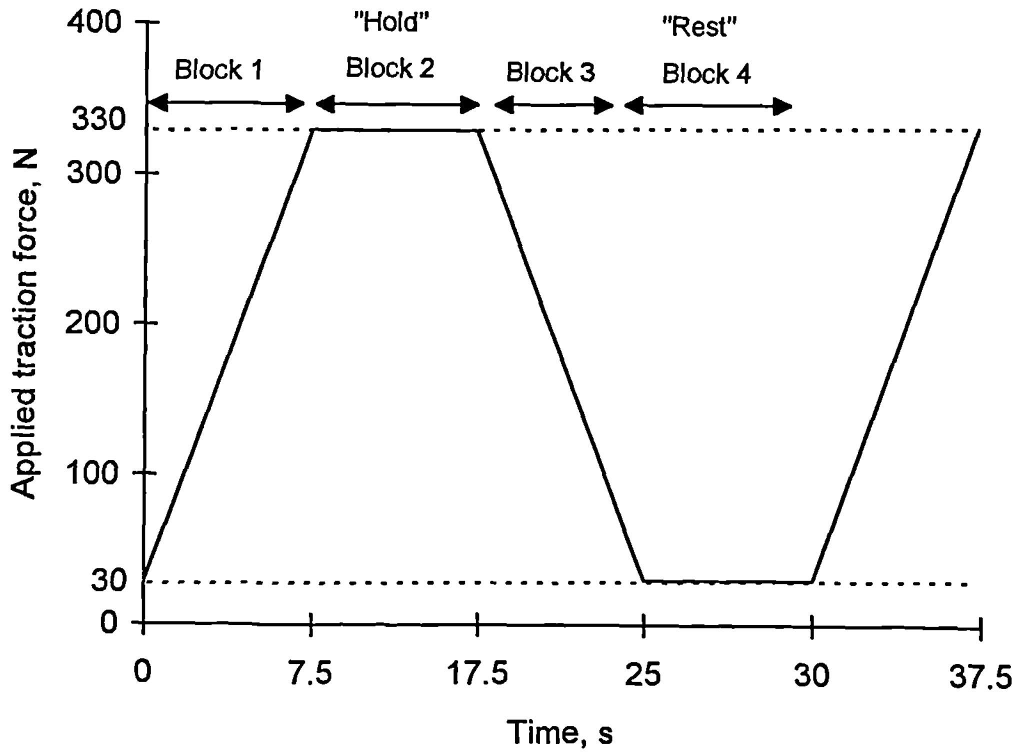


Figure 8.1
A single traction loading cycle which consisted of four blocks and lasted for 30s. The cycle was repeated 30 times during the experiment.

the inferior moulding cup. The details and preparation of the specimens are fully described in sections 7.3.1 and 7.3.2. The present experiment was conducted about 1-2 hours after they were subjected to the mechanical tests described in chapter 7.

8.2.1 The Experimental Procedure

The experimental set up was identical to that described in the previous experiment (The details of the set up are given in section 7.3.3.). The mechanical test was carried out on the Instron machine with the upper and lower moulding cups fixed to the load cell via a chain and to the cross-head respectively. The dead weight and the side chain were applied to produce the required flexion moment of 24Nm and to simulate the thoracic reaction force respectively. The specimen was arranged as shown in figure 7.6 and 7.7.

The Instron machine was set in "load control" mode. Figure 8.1 shows the sequence of loading in one cycle of traction application. The specimen was first subjected to a small traction force of 30N. The force was increased at a rate of $40Ns^{-1}$ until it reached 330N (block 1). This took 7.5s and it was then sustained for a period of 10s (block 2) (the so-called "hold" period of clinical intermittent traction). The force was then released at the same rate until it dropped back to 30N (this also took 7.5s) (block 3). The 30N force was held for 5s before the next loading cycle began (block 4) (referred to as the "rest" period clinically). The duration of each loading cycle (which consisted of the four blocks) was 30s, and the cycle was repeated 30 times. This was equivalent to a treatment duration of 15 minutes. All the above chosen parameters were similar to those used clinically (Colachis and Strohm, 1969; Hinterbuchner, 1985; Maitland, 1986; Grieve, 1988).

Clinically, intermittent traction is usually applied with a small "resting" force. Hence, in the present study, the specimen was not completely unloaded during the "rest" period and a small force of 30N was maintained. This force also ensured that the tension in the chain would not drop to zero which would affect the ability of the machine to control the load. To optimise the waveform response and minimise load overshooting, the appropriate proportional, integral and derivative gain values were set.

At the end of the mechanical testing, the traction force was completely removed. However, the experimental set-up was left undisturbed for 15 minutes so that the residual deformations of the spine after the test could be determined.

The Panasonic Colour Video Camera Model F10CCD (Matsushita Communication Industrial Co. Ltd, Japan) was used to record the deformations of the spine during and after repeated traction loading cycles. Spirit levels were used to ensure that it was perpendicular to the plane of movement and level in both the fore-and-aft and lateral directions. The camera was fixed to a tripod and placed at a distance of 1.24m from the specimen. Its position remained stationary during the whole test procedure. The iris was set in the automatic mode and the image size made as large as possible using the zoom lens. The camera operated in the PAL mode with a sampling rate of 50Hz. The videotape used was Fuji E120 SuperHG (Fuji Photo Film Co., Ltd, Germany).

Two 500W photographic flood lamps were used to provide lighting. As described in section 7.3.3, in order to help identification of markers of different vertebrae in the video, they were painted with different colours. Fluorescent paint was used and a black cardboard placed behind the specimen to provide a good contrast.

A voltmeter was hung by the side of the specimen. This appeared in every video frame, showing the current output voltage signal from the load cell of the Instron machine. This provided a measure of the magnitude of the traction force applied in each video frame. A 0.25m ruler was placed alongside the specimen so as to provide scaling for each video frame. A small sign also appeared on the video image showing the specimen number and the test being performed.

The output signals from the load cells and the crosshead movements were recorded by an IBM compatible computer with an Amplicon analogue to digital card and the Microscope software.

Data recording and videotaping continued for 15 minutes after all the loading cycles were completed and the traction force was completely removed. This allowed an examination of the recovery of the spine after a simulated session of intermittent traction therapy.

8.2.2 Analysis of Videotape

The videotape was subsequently analysed by the Peak Motion Analysis System (version 5.0) (Peak Performance Technologies Inc., Englewood, USA). A time code, generated by the Peak system, was audio-dubbed onto the videotape. In order to allow the time code to synchronise with the video, videotape encoding should begin about one minute before the first video picture to be analysed. This meant the specimen should be videotaped at least a minute prior to the start of the mechanical test.

The first step of the analysis was to digitise the two ends of the ruler which allowed the scaling factor to be determined. The following frames were selected and stored for the frame grabber, and this would allow manual digitisation of the 18 markers which were attached to the vertebrae and the inferior moulding cup.

- at the end of block 2 of the 1st, 2nd, 3rd, 5th, 15th and 30th loading cycles (i.e. the video frame just before the traction force was released),
- immediately, 0.5 minutes, 1 minute, 15 minutes and 30 minutes after the specimens were unloaded.

These frames were identified by noting the voltmeter reading. The digitised coordinates were scaled accordingly to the magnification factor.

The repeatability of the data provided by the present experimental technique was examined by videotaping a stationary specimen with the markers attached. The

		Spinal level						
Error in determining the markers' positions, mm								
		L1	L2	L3	L4	L5	S1	*Row mean*
Marker 1	x-coordinate	0.43	0.29	0.35	0.18	0.30	0.25	*0.30*
	y-coordinate	0.37	0.22	0.38	0.30	0.48	0.37	*0.35*
Marker 2	x-coordinate	0.25	0.17	0.59	0.31	0.38	0.37	*0.34*
	y-coordinate	0.22	0.17	0.25	0.18	0.26	0.31	*0.23*
Marker 3	x-coordinate	0.44	0.23	0.18	0.23	0.40	0.28	*0.29*
	y-coordinate	0.34	0.28	0.21	0.15	0.38	0.27	*0.27*
	Column mean	*0.34*	*0.23*	*0.32*	*0.22*	*0.37*	*0.31*	
						Mean error = 0.30mm		
Error in computing the rotation of the vertebra, °								
		L1	L2	L3	L4	L5	S1	*Row mean*
Using markers 1 and 2		1.49	0.83	2.25	0.99	1.20	1.47	*1.37*
Using markers 2 and 3		1.54	0.99	1.68	0.99	2.33	1.32	*1.48*
Using markers 1 and 3		1.16	0.72	0.76	0.61	0.95	1.21	*0.90*
	Column mean	1.40	0.85	1.56	0.86	1.50	1.33	
						Mean error = 1.25°		

Table 8.1
Error in determining the markers' positions and computing the vertebral rotation from these positions. (Error = 1 x standard deviation) (The top, middle and bottom markers of each vertebra were labelled as 1, 2 and 3.)

positions of the markers were manually digitised ten times and in ten different pictures. The standard deviations of the scaled horizontal and vertical coordinates of the markers are shown in table 8.1. It was shown that with the present experimental set up, the mean error in determining the marker's positions was found to be 0.30mm. The error sizes were about the same for the horizontal and vertical coordinates of the three markers.

The rotation of the vertebra was computed from the inclination of the line joining any two markers. The mean error in this computation, as a result of error in determining the marker coordinates, was found to be 1.25° (table 8.1). It was shown that the error was significantly smaller if the movements were computed from the top and bottom markers, and in this case, the mean error was 0.90°. This was because these markers were further away from each other compared to the other pairs of markers. Hence, in the present video analysis, the computation of vertebral rotation was performed using the top and bottom markers as the reference points.

In conclusion, the data provided by the present experimental technique was generally considered to be reliable.

8.2.3 Treatment of Data

The initial position of the spine was first obtained from a video frame filmed when it was already loaded with the 24Nm flexion moment but before the application of any traction force. Deformations of the spine during and after repeated applications of the 330N traction force were determined with reference to this position.

The change in the overall length of the lumbosacral spine was examined by evaluating the distances between the three markers of the L1 vertebra and the respective ones on the sacrum. These three intermarker distances were then averaged to provide the best estimate of the change in spinal length.

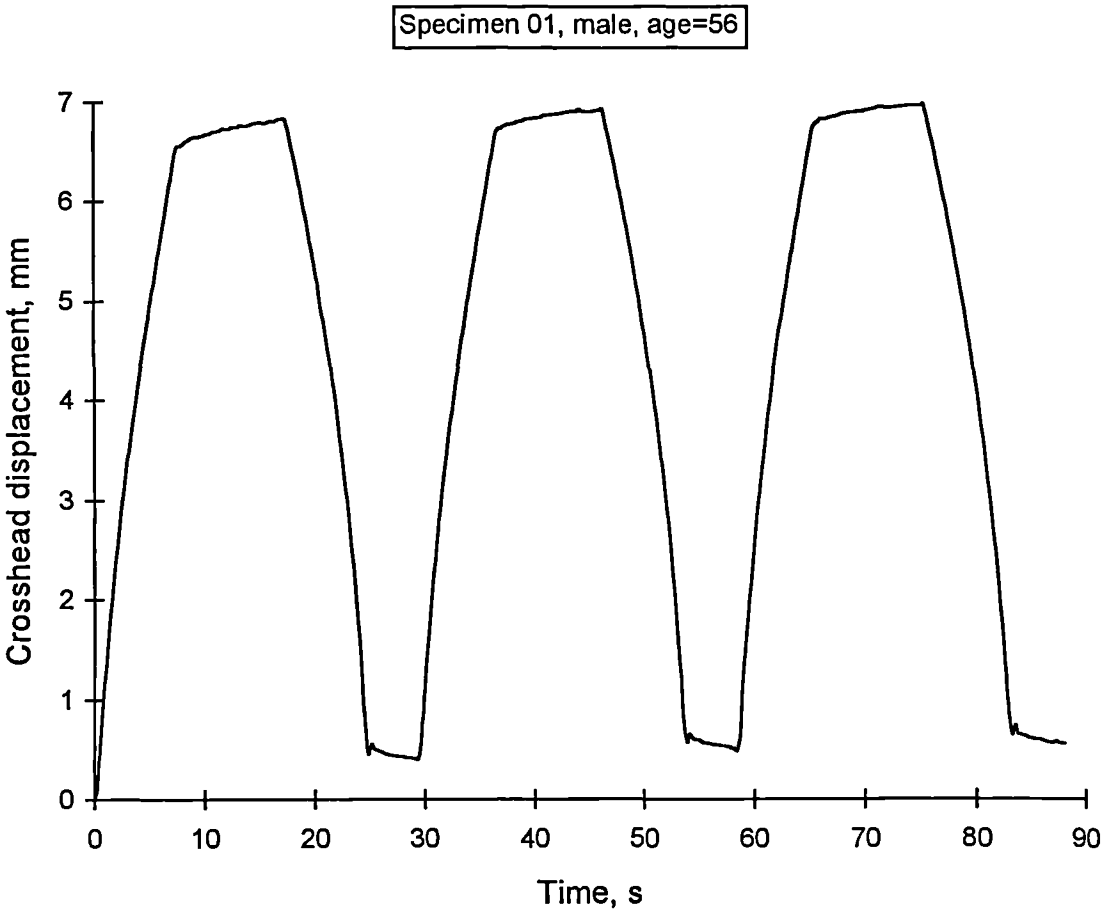


Figure 8.2
A plot of the crosshead movement against time for the first three loading cycles in a specimen (specimen no. 1, male, aged 56).

Sagittal rotation of the whole spine was computed from the difference in vertebral rotations between the L1 vertebra and the sacrum. The rotations of these two vertebrae were determined from the movements of their top and bottom markers.

The changes in the displacements of individual vertebrae with repeated traction loadings were small. It was thus important to avoid excessive accumulation of errors in the computation process which would obscure any trends produced. In computing the movement of one vertebra relative to another, the errors involved in determining the movements of the two individual vertebrae would be added together. Data analysis showed that as a result of the accumulated error, there were no consistent trends in the intervertebral movements produced by traction. These data are thus not reported here.

The displacements of each individual vertebra due to intermittent traction were assessed in absolute terms. This avoided the accumulation of errors discussed above. The positions of the anteroinferior and posteroinferior corners of the vertebral bodies were determined from those of the top and bottom markers using the method described in section A1.2. In the case of the sacrum, the chosen anatomical points were the anterosuperior and posterosuperior corners. The horizontal and vertical translations of these anatomical landmarks in space were then determined.

8.3 RESULTS

8.3.1 Deformations of the Whole Lumbosacral Spine

Figure 8.2 shows a typical plot of the crosshead movement against time for the three cycles of traction loadings of a specimen (specimen no.01, male, aged 56). The plots of all the specimens had similar characteristics. It was demonstrated that for each loading cycle, the crosshead gradually moved downward during the 10s period of sustained loading due to the creep of the specimens. Immediately upon release of the traction force (to the 30N base line), the crosshead did not return to its original

	Increase in the averaged intermarker distance, mm	
	Mean	Standard deviation
1st loading cycle	6.28	1.01
2nd loading cycle	6.50	0.95
3rd loading cycle	6.58	0.92
5th loading cycle	6.63	0.92
15th loading cycle	6.80	1.17
30th loading cycle	7.30	0.96
immediately post-traction	1.29	0.98
0.5 minutes post-traction	0.96	0.90
1 minute post-traction	1.15	0.80
5 minutes post-traction	0.70	0.87
15 minutes post-traction	0.67	0.99

Table 8.2
Mean and standard deviation of the increase in the averaged distance between the markers of the L1 vertebra and the sacrum during and after repeated cycles of traction loadings. (n=7)

	Flexion of the whole lumbosacral spine, °	
	Mean	Standard deviation
1st loading cycle	3.98	1.69
2nd loading cycle	3.63	1.33
3rd loading cycle	4.14	1.38
5th loading cycle	4.42	1.62
15th loading cycle	3.79	0.90
30th loading cycle	4.40	1.41
immediately post-traction	0.19	1.31
0.5 minutes post-traction	0.25	1.58
1 minute post-traction	0.09	1.66
5 minutes post-traction	0.46	1.52
15 minutes post-traction	0.41	1.43

Table 8.3
Mean and standard deviation of the flexion of the whole lumbosacral spine during and after repeated cycles of traction loadings. (n=7)

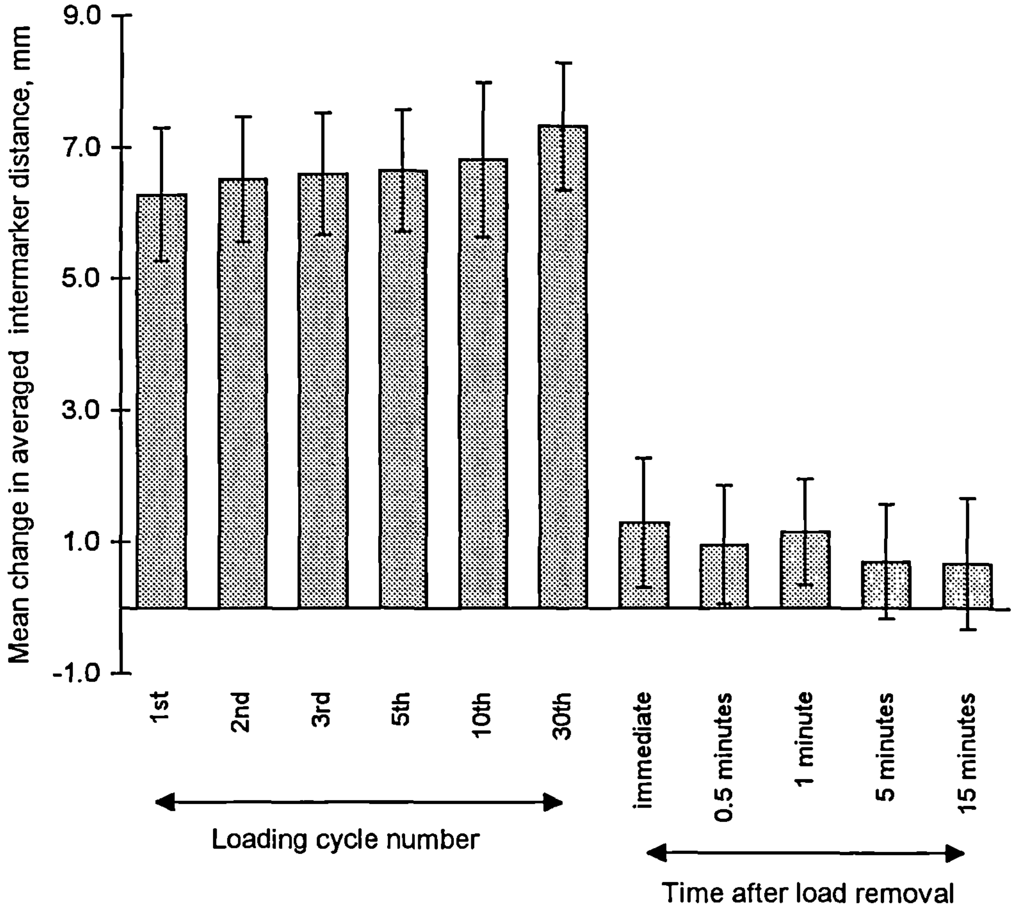


Figure 8.3
Mean change in the averaged distance between the L1 and sacral markers during and after repeated cycles of traction loadings. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

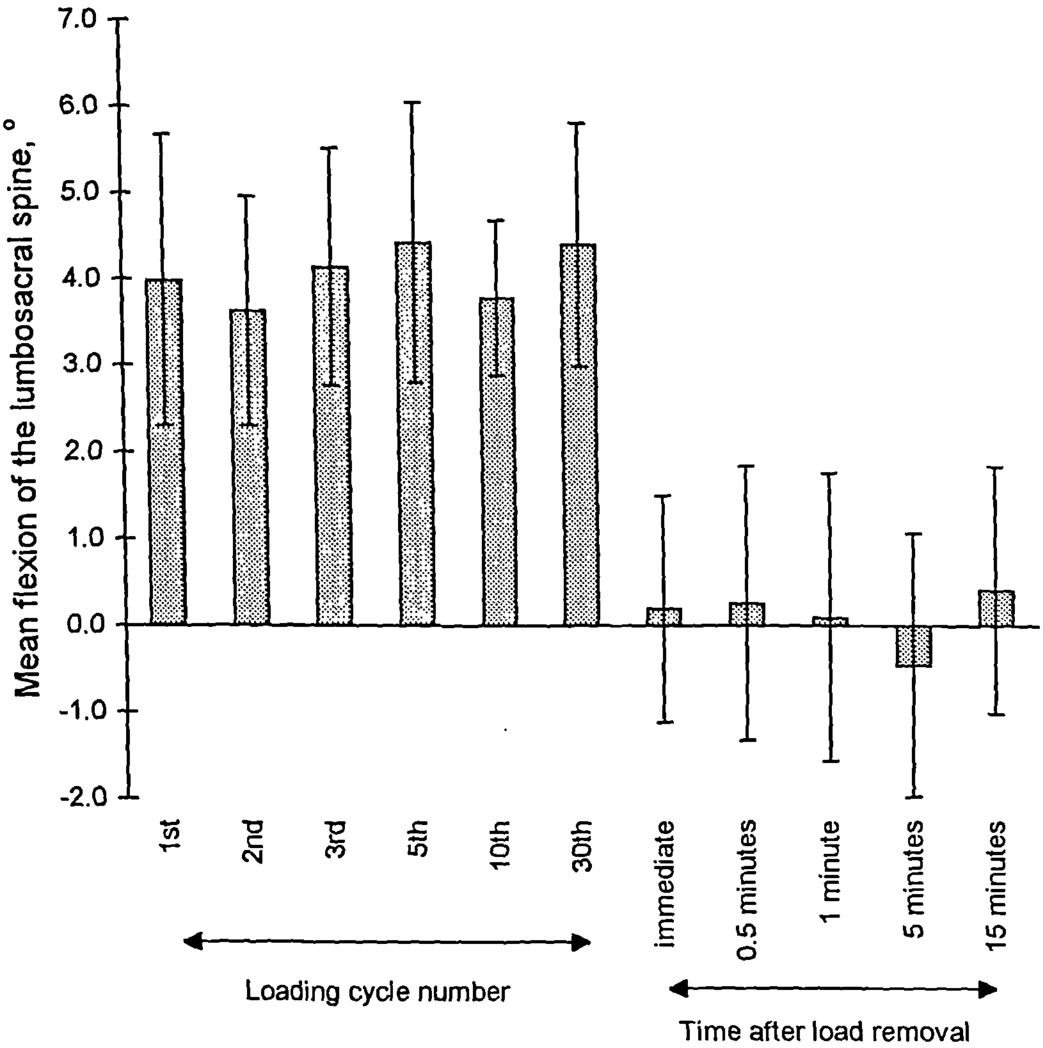


Figure 8.4
Mean flexion of the lumbosacral spine during and after repeated cycles of traction loadings. (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

position indicating that there was a residual deformation of the spine. During the 5s "rest" period, the crosshead gradually moved upwards due to the recovery of the spine. It was also observed that the crosshead moved further downwards with each loading cycle indicating an increase in residual deformation. It appeared that intermittent traction produced time-dependent effects which were typically characterised by preconditioning and creep.

It should be noted that the crosshead movement data included the deformation of the loading frame and did not truly reflected the deformation of the spine. The change in spinal length was more appropriately indicated by the averaged intermarker distance described in section 8.2.3. Figure 8.3 and table 8.2 show the increase in the distance for various loading cycles at the instant just before the traction force was released and during 15 minutes following the intermittent traction applications.

The averaged intermarker distance increased with each loading cycle and this was consistently observed in all seven specimens. The mean change increased from 6.3mm in the first loading cycle to 6.6mm in the third loading cycle, and there was an increase of 7.3mm at the 30th (the last) loading cycle. Immediately after the 30 cycles of loading, the spine did not fully return to its original length and there was a small mean increase of 1.3mm in the intermarker distance. However, the spine gradually recovered and 15 minutes after the repeated traction loadings, the mean increase in distance was only 0.7mm. In two of the specimens, there were decreases in the intermarker distances immediately after load removal, but these were small in magnitude and could have been due to error.

Figure 8.4 and table 8.3 show the flexion movement of the whole lumbosacral spines produced by the traction forces in the various cycles and during the 15 minute period following the intermittent traction applications. The observations on the flexion movement were less consistent than those on the change in the intermarker distance (i.e. the change in spinal length). The mean movements produced in the various cycles were similar. The specimens did not appear to remained flex after the intermittent

	Spinal level											
	L1		L2		L3		L4		L5		Sacrum#	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Horizontal translation at the anteroinferior corner, mm												
1st cy.	3.36	0.98	4.52	0.96	4.61	1.06	4.15	0.95	3.57	0.84	3.28	2.09
30th cy.	3.60	0.81	4.81	0.90	5.26	1.08	4.70	0.97	4.20	0.99	3.74	1.00
immed.	0.03	0.57	0.53	0.33	0.68	0.73	0.63	0.52	0.68	0.74	0.95	0.87
15 min.	-0.19	1.07	0.06	0.83	0.30	1.08	0.11	1.02	0.03	1.29	0.87	1.79
Vertical translation at the anteroinferior corner, mm												
1st cy.	-3.36	2.29	-3.77	2.02	-4.98	2.46	-4.51	3.40	-3.96	3.41	-3.81	4.26
30th cy.	-3.07	1.98	-4.28	1.52	-5.56	3.43	-5.85	2.69	-3.92	4.14	-4.34	2.33
immed.	0.57	2.19	-0.24	0.78	-2.04	2.36	-0.75	1.89	-0.45	3.31	-0.17	1.48
15 min.	0.63	1.22	0.02	1.04	-3.03	2.10	-1.29	2.65	2.31	3.89	0.89	4.26
Horizontal translation at the posteroinferior corner, mm												
1st cy.	3.38	0.97	4.57	0.96	4.63	1.05	4.25	0.95	3.83	0.92	3.55	2.33
30th cy.	3.60	0.81	4.82	0.90	5.23	1.04	4.73	0.95	4.36	0.97	4.26	1.79
immed.	0.03	0.51	0.54	0.32	0.70	0.68	0.67	0.53	0.77	0.61	1.08	0.88
15 min.	-0.17	1.07	0.08	0.84	0.23	1.22	0.15	1.05	0.21	1.08	1.19	1.95
Vertical translation at the posteroinferior corner, mm												
1st cy.	-2.57	1.35	-3.52	1.33	-4.72	1.70	-4.86	2.45	-4.88	2.17	-4.60	3.28
30th cy.	-2.42	1.41	-3.91	1.26	-5.08	2.53	-5.83	1.88	-4.84	2.83	-5.34	1.82
immed.	0.45	1.46	-0.21	0.50	-1.44	1.59	-0.65	1.29	-0.69	2.33	-0.49	1.07
15 min.	0.69	0.72	0.09	0.79	-1.80	1.51	-0.84	1.83	1.54	2.75	0.42	3.29

Table 8.4
Mean and standard deviation (SD) of the horizontal (backward/forward) and vertical (upward/downward) translations of the vertebrae at the anteroinferior and posteroinferior corners (# the anterosuperior and posterosuperior corners in the case of sacrum) of their bodies. (n=7)
(1st cy.=1st cycle of loading, 30th cy.=30th cycle of loading, immed.=immediately after the intermittent traction applications, 15min.=15 minutes after the intermittent traction applications)
(positive values denote backward and upward translations, and negative values forward and downward translations.)

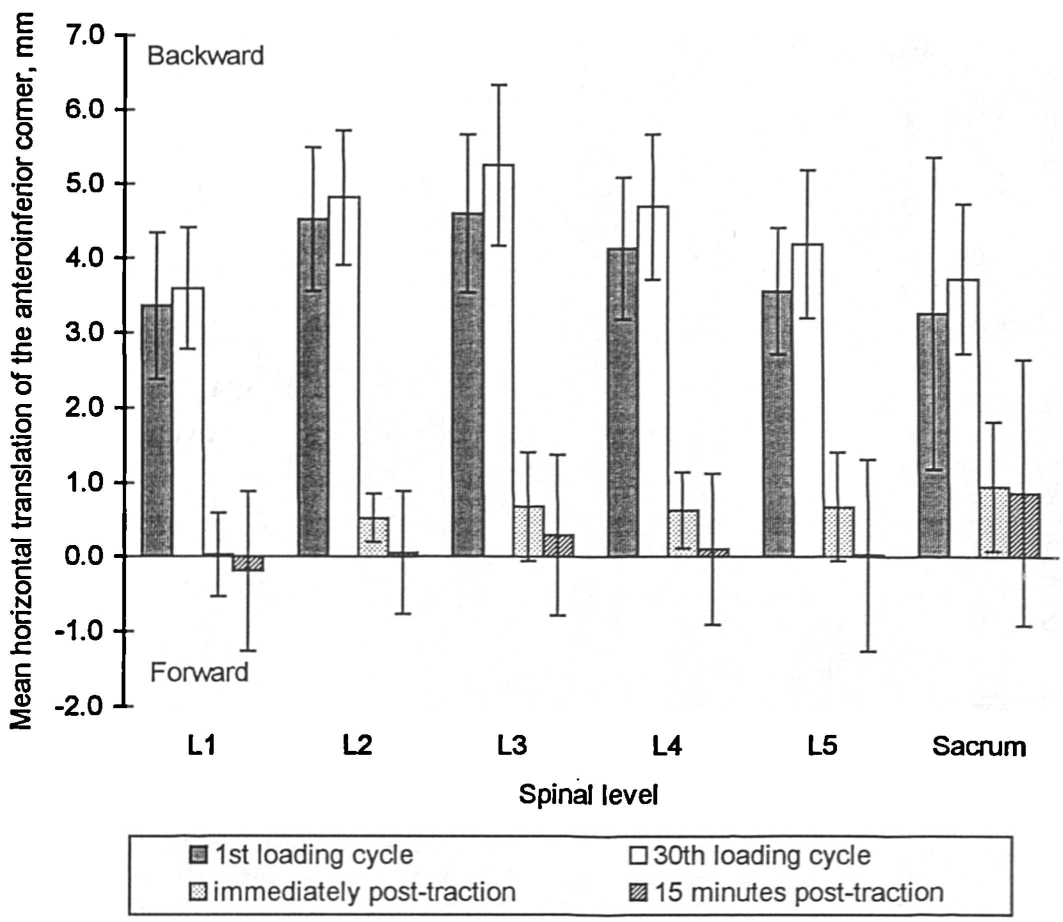


Figure 8.5
Mean horizontal translation of the anteroinferior vertebral body corner (the anterosuperior corner in the case of sacrum). (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

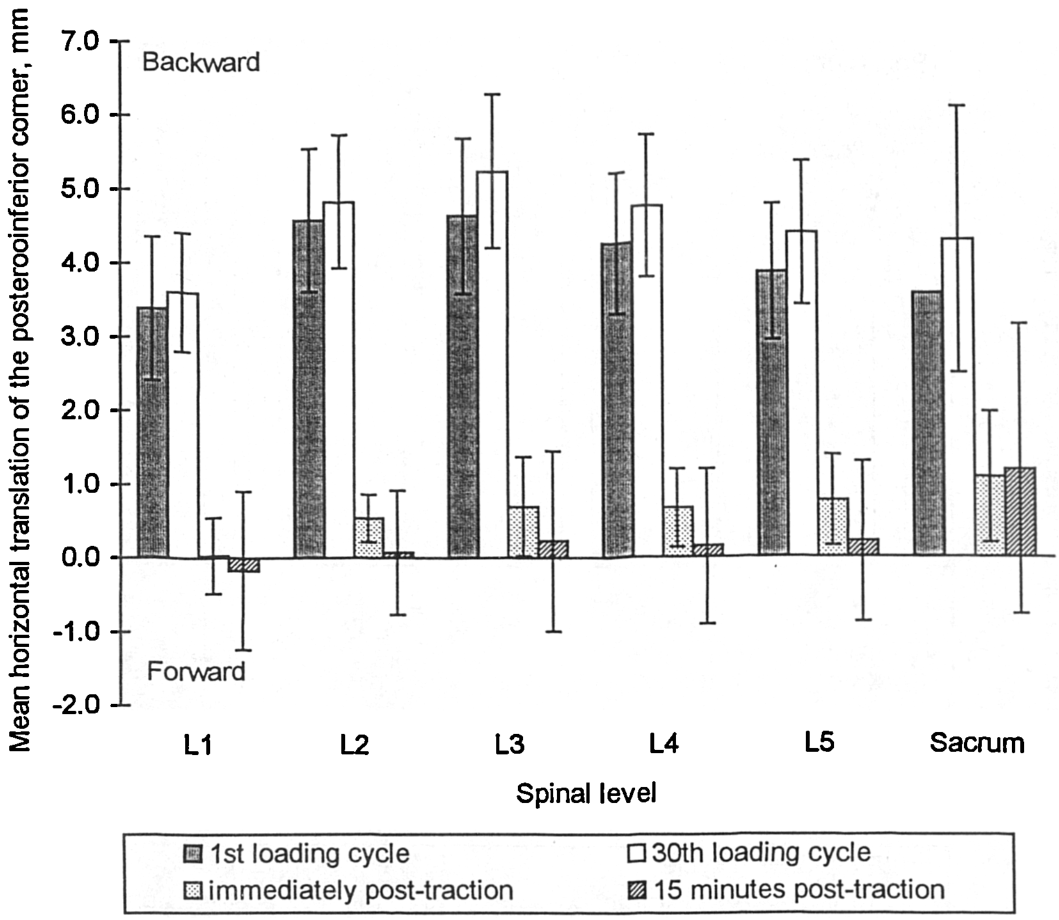


Figure 8.6
Mean horizontal translation of the posteroinferior vertebral body corner (the posterosuperior corner in the case of sacrum). (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

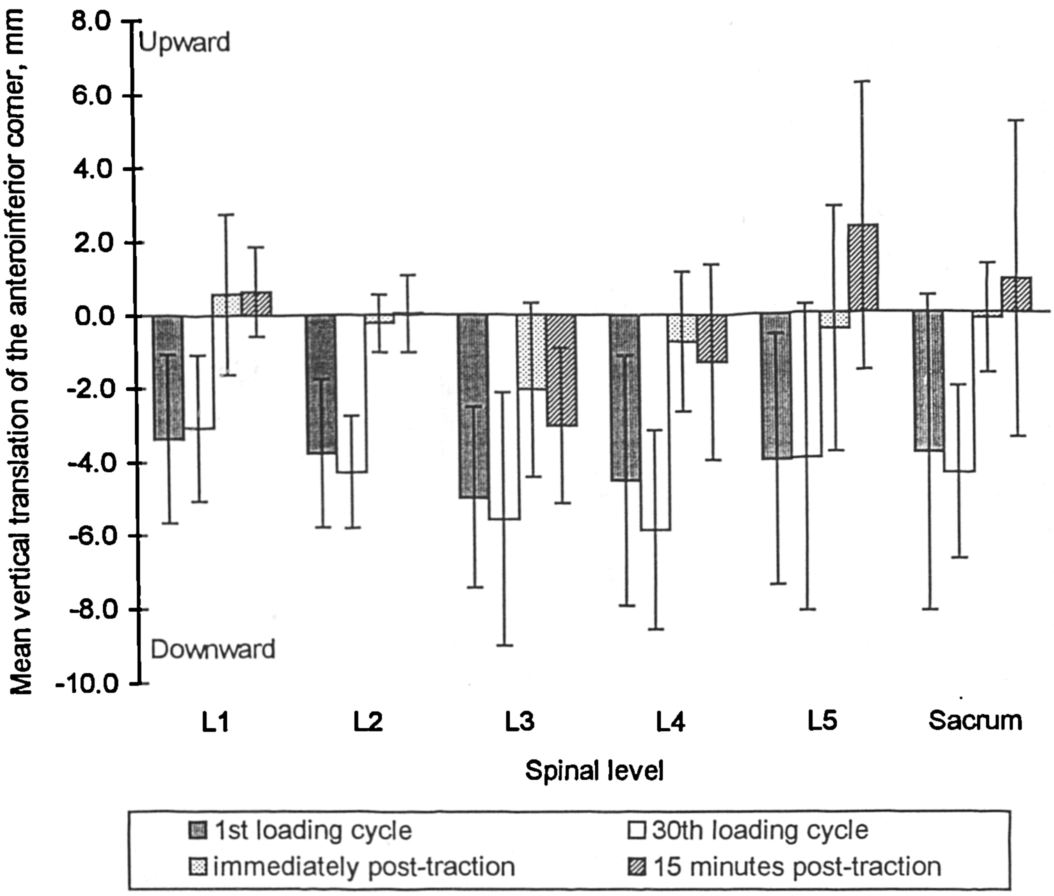


Figure 8.7
Mean vertical translation of the anteroinferior vertebral body corner (the anterosuperior corner in the case of sacrum). (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

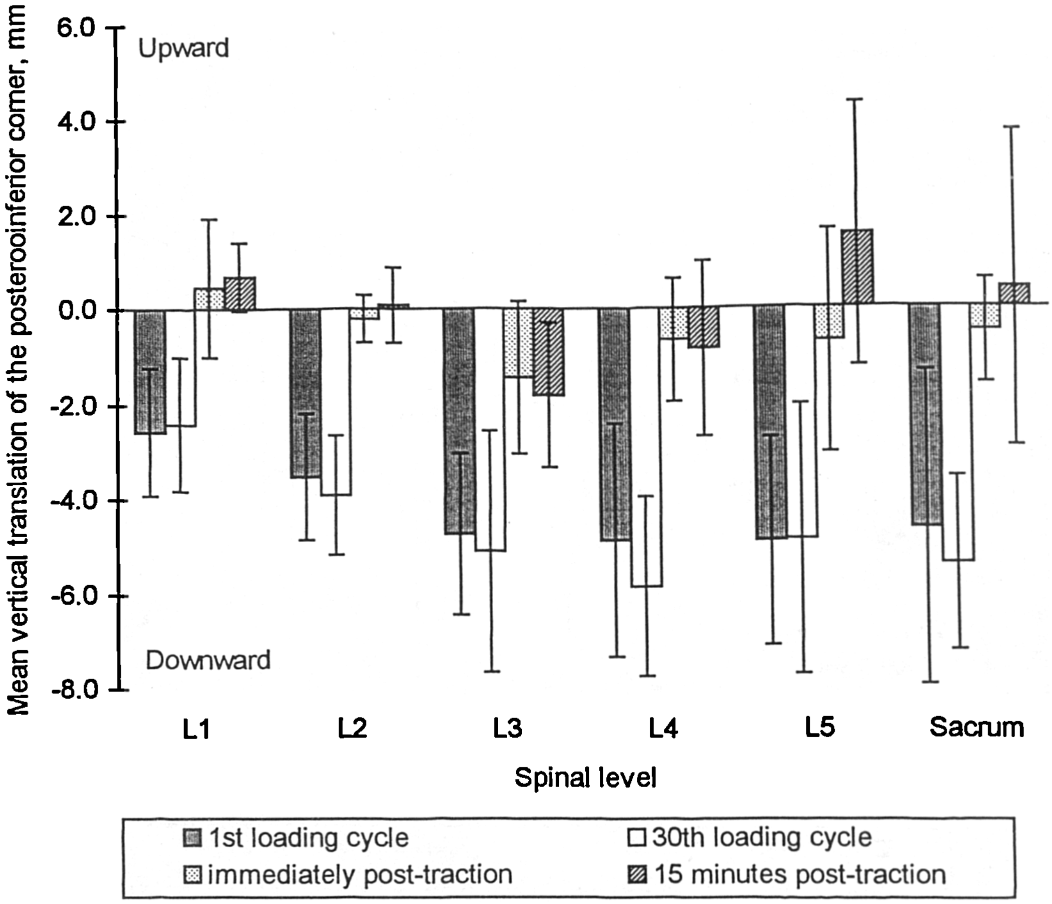


Figure 8.8
Mean vertical translation of the posteroinferior vertebral body corner (the posterosuperior corner in the case of sacrum). (n=7)
(The vertical bars denote ± 1 standard deviation for the 7 specimens.)

traction applications and there was no significant change in the sagittal rotation of the specimens during the 15 minute post-traction period.

8.3.2 Displacements of Individual Vertebrae

Table 8.4 and figures 8.5-8.8 show the horizontal (backward/forward) and vertical (upward/downward) translations of the individual vertebrae at the anteroinferior and posteroinferior corners (the anterosuperior and posterosuperior corners in the case of sacrum) of their bodies. Only data obtained in the first and last loading cycles, and immediately and 15 minutes post-traction are presented. This is because the changes in the translations with the individual loading cycles and with a short period of time were too small to be of any significance.

It was shown that the two vertebral body corners translated backwards by about 3-5mm during the application of the traction force (figures 8.5-8.6). These backward translations had a general tendency to increase in the 30th cycle compared to those in the 1st cycle. The increases at both vertebral corners and at the different vertebrae were similar in magnitude.

The two corners of all the vertebrae translated in the downward direction by about 2-6mm during the application of the traction force (figures 8.7-8.8). In the 30th loading cycle, the downward translations at the corners of most vertebrae tended to increase slightly, except the L1 and L5 vertebrae which remained more or less the same in position in the vertical direction.

It was revealed that immediately after the intermittent traction applications, very small amount of horizontal and vertical translations were retained at the two vertebral body corners (figures 8.5-8.8). This implied that the vertebral bodies had returned more or less to their original positions in both the horizontal and vertical directions. There were no further significant changes in the vertebral body positions 15 minutes post-traction.

8.4 DISCUSSION

An attempt was made to simulate a clinical traction treatment session when the force was applied intermittently with both "hold" and "rest" periods. The experimental set-up employed was identical to that described in chapter 7 which was designed to reproduce the in vivo loading conditions. The validity of the set-up has been discussed in the section 7.5.3.

The experimental findings of chapter 7 revealed that most of the deformations of the spine took place after the application of the 24Nm flexion moment which simulated passive hip and knee flexion. The additional deformations produced by the traction force as observed in this study were thus expected to be small.

The video technique had error sizes of 0.3mm and 0.9° (using makers 1 and 3 for computation) and was sufficiently accurate to trace the movements of the markers. However, the technique was found to be unable to reveal the small time-dependent changes in the intervertebral movements. As discussed in section 8.2.3, the computation of these parameters involved a number of steps leading to accumulated errors which were excessive. The present study was thus limited to the changes in the deformations of the whole spine and the absolute displacements of the individual vertebrae.

The most consistent finding of the present experiment was that the intermarker distance gradually increased with each loading, reaching a maximum at the 30th cycle of loading. This implied that the spine lengthened gradually, but the additional amount of lengthening with each cycle of loading was small. There was a difference of 1.0mm in the mean intermarker distances between the first and last loading cycles. This small gain in spinal length was probably due to the combined effects of creep and preconditioning as a result of repeated applications of the traction force.

Flexion of the whole lumbosacral spine did not appear to increase with increasing number of loading cycles. The mean flexion in the different cycles ranged from 3.6° to 4.4°. The change in flexion with each loading cycle was close to the error size reported in table 8.1 and thus was unlikely to be revealed by the present experimental technique.

The backward translations at the anteroinferior and posteroinferior corners of the vertebral bodies were produced by flexion of the spine. The translations in the 30th loading cycle tended to be slightly larger than those in the 1st cycle. This implied that there was a very small increase in spinal flexion although this was not revealed by direct computation of the movement.

The downward translations at the two vertebral corners also tended to increase slightly by the 30th loading cycle. Any changes at the L1 and L5 vertebrae were probably too small to be revealed by the present technique. It appeared that the increase in the distance between the markers of L1 and the sacrum, or the overall length of the spine was the combined result of flexion and downward movements of the vertebrae.

It was shown that the mechanical effects of repeated applications of traction force were temporary. The spine did not appear to remain flexed (i.e. the lordosis was regained) and the individual vertebrae returned more or less to their original positions immediately after the intermittent traction application. Small amount of residual effects was only observed in the overall length of the spine. The distance between L1 and the sacrum remained increased by 1.3mm immediately post-traction (table 8.2), but this was equivalent to less than 0.3mm per motion segment. The distance almost returned to its original value 15 minutes post traction (table 8.2).

8.5 CONCLUSION

The present study showed that repeated application of traction loads led to a gradual increase in the overall length of the spine. There were also slight increases in the downward and backward translations of the vertebrae observable in the 30th cycles of loading. These changes implied increased flexion of the spine. It was believed that the increase in spinal length was attributable to the separation of the vertebrae and the flexion of the spine.

With each loading cycle, the change in the deformation of the spine was generally small. The mechanical effects of traction were temporary and very little deformations were retained afterwards. Immediately after the repeated traction force applications, the various movement parameters indicated that the individual vertebrae returned more or less to their original positions. There was a significant residual gain only in the overall length of the spine, but this became negligible 15 minutes post-traction.

■ *PART IV*

DISCUSSION AND CONCLUSION

■ *Chapter 9*

General Discussion

General Discussion

9.1 THE MECHANICAL EFFECTS OF MOBILISATION AND TRACTION

The present work details of a series of experiments which examined the mechanical effects of two very common manual therapy techniques, namely, posteroanterior mobilisation and traction. The mechanics of these procedure is much more complicated than their names might imply.

Posteroanterior mobilisation is not simply an application of a discrete posteroanterior force to the lumbar spine but it generates a complex loading pattern. The biomechanical model proposed by Lee (1990) suggests that the technique represents a three-point bending of the lumbar spine (section 4.1.4). It produces posterior shear at the motion segments above the vertebra being mobilised and anterior shear at those below. An extension moment is also produced at all segments.

Likewise, during lumbar traction, the lumbar spine is not simply subjected to tension. The technique produces flexion moment, tension and anterior shear at the motion segments. The forceplate study described in section 7.2 showed that the adoption of the Fowler's position generated a significant flexion moment on the spine. The traction force also produces a flexion moment as its line of action is posterior to the centres of rotation of the lumbar segments (section 7.2). In addition, the motion segments are subjected to anterior shear as a result of the reaction force generated at the thoracic cage.

As a result of the complex loading patterns, mobilisation and traction produce a combination of different intervertebral movements. The radiographic study (chapter

6) clearly demonstrated that the motion segments underwent extension during posteroanterior mobilisation. The upper lumbar segments were found to translate posteriorly and the L5/S1 anteriorly. These data strongly supported the three-point bending model proposed by Lee (1990) and refute the general belief that posteroanterior mobilisation only produces anterior "gliding" of one vertebra upon another.

During traction, the motion segments were found to exhibit complex motion patterns which are thoroughly reported in chapter 7. In addition to an increase in the overall length of the spine, the motion segments were found to flex and undergo anterior translation generally.

The complex movement patterns produced by mobilisation and traction had not been reported elsewhere in the literature. Previous studies on mobilisation had quantified the vertical displacements of the vertebrae using only skin measurement techniques (Thompson, 1983; Lee, 1990; Lee and Svensoon, 1990; Lee and Evans, 1991 and 1992). In the case of traction, Colachis and Strohm (1969) measured only the changes in anterior and posterior disc heights. The present work has fully quantified all the sagittal plane movement components (rotation and translation) and will therefore significantly contribute to the current understanding of the mechanical effects of these therapeutic procedures.

Clinically, mobilisation and traction are applied in both an oscillatory or sustained manner, and thus their mechanical effects on the lumbar spine are time-dependent. These had been examined by previous works (Twomey, 1985; Lee, 1990; Lee and Evans, 1991 and 1992; Lee, 1994) and the present study. Mobilisation was shown to exhibit preconditioning and creep effects (section 4.1.4) which tended to decrease with repeated loadings or with time. The time-dependent effects of traction were demonstrated in the cadaveric study described in chapter 8. It was shown that repeated traction loadings produced gradual lengthening of the spine. Twomey (1985) also showed that sustained traction elicited the creep response.

The time-dependent mechanical effects of lumbar traction were shown to be temporary. Twomey (1985) showed that the residual deformation obtained were almost completely lost 30 minutes after cessation of loading. A similar observation was made in the present work in which traction was applied with repeated cycles of "hold" and "rest". About 50% of the increase in the length of the spine was lost 15 minutes post-traction. The vertebrae were also shown to return to their original positions indicating that the initial lordosis of the spine was regained. Although the above effects had only been experimentally demonstrated to be temporary in traction, it is expected that the mechanical effects of mobilisation or other manual therapy techniques are also short-lived.

9.2 CLINICAL IMPLICATIONS

9.2.1 Clinical Manual Examination

The cadaveric (chapter 5) and in-vivo (chapter 6) studies had both demonstrated that the intervertebral movements produced by posteroanterior mobilisation were small in magnitude. The motion segments exhibit a few degree of extension movements and the mean translation in the posteroanterior direction was less than 1mm.

The above finding has cast doubt on the reliability of clinical manual examination using mobilisation techniques. It is extremely unlikely that the therapist could feel such a small amount of movement. Since the vertebra rotates and translates at the same time, it is not clear what movement components the therapist is palpating. It is therefore not surprising to find poor reliability in clinical manual examination (see section 4.1.3).

The use of the "movement diagram" as a tool for communication between therapist is even more unreliable. As pointed out in section 4.1.4 (figure 4.4), the diagram is actually a reflection of the load-deformation curve of the motion segment.

In this case, the therapist has to feel both the resistance of the tissue and the movement produced. Hence, the task is even more complicated than feeling the movements only. It appears that the movement diagram may only be determined by instrumentation like those developed by previous authors (Thompson, 1983; Lee, 1990; Lee and Svensoon, 1990; Lee and Evans, 1991 and 1992).

In addition, both the cadaveric and radiographic studies (chapters 5 and 6) showed that there were wide variations in the amount of intervertebral movements produced by mobilisation. These findings further complicate the clinical diagnosis of segmental hypermobility or hypomobility, even if segmental mobility could be determined accurately.

Although the magnitude of extension produced by mobilisation in each motion segment is small, the total amount of extension in the whole lumbar spine will be significant, producing a noticeable increase in the lordosis or a "sagging" of the spine. The radiographic study (chapter 6) has provided experimental data to support this argument. It was found that the spinous processes of the upper segments moved caudally and those of the lower segments cranially indicating that the lordosis of the spine had increased and brought the two ends of the spine closer together. The spinous processes were also found to be displaced in the downward or ventral direction during mobilisation, thus indicating sagging of the entire spine.

The mean downward displacement of the spinous process was found to range from 9mm to 12mm. This is about ten times larger than the segmental movements produced and thus far easier to be detected manually. It is thus probable that what the therapist is palpating is the downward movement of the spinous process rather than the corresponding segmental translation.

In view of the apparent unreliability of clinical manual examination, various authors had attempted to quantify mobilisation using objective instrumentation (Thompson, 1983; Lee, 1990; Lee and Svensson, 1990; Lee and Evans, 1992) (section 4.1.4). They generally used displacement transducers resting on the skin overlying the spinous processes. Lee (1990) reported that during mobilisation of L4, the mean downward displacement of the L3 spinous process relative to the L4 process was 11.1mm and that of L5 relative to 4 was 12.3mm.

However, the radiographic study (chapter 6) showed that the relative displacements of the spinous process of L3 and L5 (relative to L4) were only 0.9mm and 1.3mm respectively. Data obtained from previous studies were about ten times larger in magnitude than those in the present work. This indicated that surface measurements were subjected to large errors due to deformation of skin and soft tissues overlying the spinous process. In addition, in the work of Thompson (1985) and Lee (1990), the mobilisation force applicator was fitted with a soft pad to reduce the discomfort of the patient. The deformation of the pad would also contribute to the error in surface measurements. Hence, segmental mobility or relative spinous process displacement cannot be reliably determined by non-invasive surface techniques, but only by radiographic techniques, as in the present study. It appears that clinical instrumentation using skin displacement transducers is only useful for quantifying the absolute displacements of the spinous processes in space.

Clearly, clinical manual examination using posteroanterior mobilisation force should be interpreted as a passive test of the stiffness of the lumbar spine in three-point bending. The therapist is actually attempting to feel the change in the bending stiffness which may be affected by clinical pathology and/or a local change in segmental mobility. The downward displacement of the spinous process may be a good clinical measure of the stiffness of the spine in three-point bending. As pointed out earlier, this may be manually perceived or measured by instrumentation with more accuracy than segmental movements.

The cadaveric study described in chapter 5 provides information on how posteroanterior stiffness may be altered by pathology. It was shown that the intervertebral disc was the principal structure resisting the mobilisation loads. Degeneration and injuries of the disc will likely lead to decreases in posteroanterior stiffness. Injuries of the soft tissues and the zygapophyseal joints are unlikely to produced major decreases in stiffness. On the other hand, increases in stiffness may be produced, for example, by formation of scar tissues in the discs during the healing process. Such stiffening may also be caused by scarring of other tissues which render them capable of better resisting the applied mobilisation load.

Clinical manual examination should not be limited to detection of abnormal segmental mobility. The pain response of the patient is particularly important if such mobility cannot be determined with accuracy in routine clinical practice. Previous investigations showed that the level of lesions could be identified accurately based on the pain response (Phillips and Twomey, 1993; Maher and Adams, 1994) (section 4.1.3). It should be noted that pain is dependent on the strains produced in the innervated tissues. The present study showed that the soft tissue elements and the zygapophyseal joints did not appear to resist the mobilisation movements to any great extent. However, they may have a significant role in eliciting the pain response. During mobilisation, significant strains are produced in these well-innervated structures. For instance, it was shown that simulated mobilisation loads produced a strain of up to 24% in the anterior longitudinal ligament (see section 5.4.2). Although injuries of this ligament may not change the posteroanterior stiffness of the segment, the tissue stretch may produce pain during mobilisation. In addition, a decrease of segmental stiffness as a result of injuries of other structures such as the disc, may stretch the soft tissues excessively and produce pain.

In summary, clinical manual examination of segmental mobility is unreliable. It is believed that the therapist is actually palpating the stiffness of the spine in three-point bending or the downward displacement of the spinous process which could also be reliably measured by instrumentation. The present work has also provided information on how changes in posteroanterior stiffness or pain may be brought about by pathology.

9.2.2 Mobilisation Treatment

Earlier studies showed that posteroanterior mobilisation produced creep and preconditioning effects (Lee, 1990; Lee and Evans, 1991 and 1992). This explains the improvement in the range of movement which is often observed immediately after mobilisation treatment. As pointed out earlier, these time-dependent effects are temporary. Therefore, they do not fully explain any permanent improvement in joint mobility. Other mechanisms may operate at the same time. Grover (1982) proposed that the mobilisation force applied may be sufficient to produce failure or damage to scar tissues in the disc or soft connective tissues and thus cause the increase in segmental mobility as discussed above.

Clinically, mobilisation is also believed to relieve pain. As illustrated in section 5.4.2 and discussed earlier, mobilisation loads produce significant strains in the ligaments and joint capsules of the motion segment. Therefore, oscillatory application of mobilisation force may stimulate the mechanoreceptors and close the pain gate at the spinal cord level (Melzack and Wall, 1965; Wyke, 1976) (section 4.1.1.2). The resulting reduction in pain may also lead to an improvement in the range of movement.

Since mobilisation loads were shown to be primarily resisted by the disc, it is suggested that major injuries of the ligaments and the zygapophyseal joints do not present as contraindications to therapy. Such injuries will not lead to excessive vertebral movements which may compromise the neural tissues. However, as pointed out earlier, the strains produced in these injured tissues may elicit pain which cannot be tolerated by the patient. In this case, mobilisation treatment should be avoided. On the other hand, when there are major decreases in posteroanterior stiffness due to disc injuries, mobilisation should be applied with caution.

The radiographic study (chapter 6) revealed that posteroanterior mobilisation produced movements at all the lumbar motion segments and there were no major differences in the magnitude of the movements among the various segments. This implies that the mechanical and perhaps the therapeutic effects are not limited to the segments which are adjacent to the mobilised vertebra. Therefore, in clinical practice, when a symptomatic segment is too tender for palpation, it may be possible to induce therapeutic effects on that segment by mobilising a distant segment.

9.2.3 Traction Treatment

The clinical rationale for the use of traction therapy is based on its mechanical effects on the intervertebral disc and the zygapophyseal joints, and possibly on the relief of nerve root compression (see section 4.2.1). These hypotheses are critically evaluated based on the experimental data obtained in the present work.

The present study showed that traction loads produced flexion of the motion segments. This implies a loss of lordosis which, in isolation, tends to raise the intradiscal pressure (Andersson et al, 1974). Beattie et al (1994) also found that the nuclear pulposus moved posteriorly when the spine was in a less extended position in supine lying. On the other hand, traction also produces longitudinal distraction of the spine which tends to reduce the pressure in the disc. The overall effect will thus be dependent on the relative contributions of the two mechanisms. Andersson et al (1983) found that no significant changes in intradiscal pressure were observed during traction therapy, indicating that the effects of the two mechanisms may cancel each other. Hence, the observation does not support the belief that traction can "suck back" a posterior disc protrusion by reducing the intradiscal pressure (Cyriax, 1978; Cailliet, 1981).

It appears that traction reduces a disc lesion by other mechanisms. The present study showed that the posteroinferior vertebral body corner translated superiorly during traction. This implies that there will be an increase in the tension of the posterior annular fibres and the posterior longitudinal ligament. This may prevent excessive posterior movement of the disc materials and reduce a posterior disc bulge. However, the mechanism is likely to operate only if the annular and ligamentous tissues are intact and if the tension is sufficiently strong. In addition, this will not reduce a prolapse which has extended beyond the annular and ligamentous boundaries.

Under the application of the traction loads, an overall increase in the foraminal cross-sectional area was observed and this was associated with superior and anterior translations of the posterior vertebral body. Traction therapy may therefore be clinically useful when the foraminal size has been compromised by osteophyte formation, thinning of the disc or excessive posterior translation of the superior vertebra.

The flexion movement and the superior translation at the posteroinferior vertebral body corners imply that the zygapophyseal joints will be separated and the joint capsules stretched. This will also stretch the posterior longitudinal ligament as explained earlier, and other tissues posterior to the disc. These tissues are well-innervated and as in the case of mobilisation, oscillatory application of traction forces will stimulate the mechanoreceptors of these tissues which may lead to a reduction of pain by closing the pain gate (Melzack and Wall, 1965; Wyke, 1976). Furthermore, the associated separation of the zygapophyseal joints may release an entrapped fold of capsule or synovial membrane leading to a reduction in symptoms (Twomey, 1985).

The present study showed that during traction, the magnitude of flexion and superior translation at the posterior vertebral corner was larger at the lower motion segments than at the upper ones. This suggests that the mechanical and therapeutic effects of traction will be more pronounced at these segments.

Another interesting finding of the present work was that most of the mechanical effects of traction took place after the application of the flexion moment which simulated the effects of adopting the Fowler's position. It may appear that the application of the subsequent traction force has little additional therapeutic value. However, it should be noted that the posterior soft tissues are slack when the spine is in the neutral position (section 2.4.2). The posterior tissues are simply unfolded in the initial stage of flexion of the motion segments as when the Fowler's position is being adopted. Reduction of disc prolapse and stimulation of mechanoreceptors are possible only in the later stage when the tissues are sufficiently tightened by the subsequent application of the traction force. The Fowler's position is clinically important in that it reduces the traction force required by first unfolding the tissues.

As discussed earlier, the mechanical effects of traction are temporary. This was demonstrated experimentally in isolated cadaveric spines in the present study. In the clinical situation, as the patient gets up from the plinth after traction therapy, it is likely that the residual effects of traction will be even smaller and more short-lived due to the effects of body weight. The temporary nature of the mechanical effects suggests that the therapeutic mechanisms by which traction relieves pain and regains mobility takes place during therapy. However, these therapeutic effects may not be short-lived. The effects achieved by mechanisms, such as reduction of a disc bulge, stretching of mechanoreceptors and enlargement of the foraminal size, may persist after treatment.

9.2.4 Mobilisation versus Traction

Clinically, there is generally a lack of consensus on the choice of manual therapy techniques. The present study reveals that there are some fundamental differences in the mechanical effects of mobilisation and traction. An understanding of these differences may help the clinicians decide the relative appropriateness of the two techniques in a particular clinical situation.

Traction produces flexion and elongation of the spine, and thus enlarges the foraminal size. However, the spine is extended and shortened in the case of mobilisation, and the technique will therefore narrow the interverebral foramina. It is thus believed that mobilisation is not a preferred choice of treatment when the intervertebral formaina are pathologically narrowed, and in this case, traction will be more appropriate. In addition, the flexion movement produced by traction will stretch the posterior soft tissues to a greater extent compared to the extension movement produced by mobilisation. Patients with acute injuries of these tissues may therefore find traction more painful and less tolerable than mobilisation treatment. It may also be argued that traction may be more useful in regaining the flexion movement and mobilisation the extension movement.

There are differences in the nature of the shear force produced by traction and mobilisation. During traction, all the vertebrae are subjected to anterior shear of similar magnitude. This suggests that traction therapy should be avoided if patients have anterior translational instability, such as spondylolisthesis and spondylolysis. However, in the case of mobilisation, only the motion segments below the mobilised vertebra are subjected to anterior shear. Patient may therefore still be able to tolerate the treatment if the unstable motion segment is above the mobilised vertebra. In fact, the posterior shear force produced at this segment may help relieve pain . Clinically, spondylolisthesis and spondylolysis are most common at the inferior motion segments (Taylor and Twomey, 1994). In the case of L5/S1, relief of symptoms may be achieved by mobilising the sacrum, but mobilisation of L5 should be avoided.

9.3 EXPERIMENTAL MEASUREMENTS OF SPINAL DEFORMATIONS

An experimental technique was developed in the present study to measure the intervertebral movements produced by mobilisation and traction. Vigorous error checks were performed and the technique was found to be accurate and reliable when applied in both cadaveric and radiographic studies.

In the cadaveric studies (chapters 5 and 7), markers were employed to enhance the accuracy. Segmental rotation and translations were computed from the the change in positions of chosen anatomical landmarks which were in turn determined from the movements of the markers. Since only two markers are required for this computation, the three markers used in the present work allowed for repeated computations to provide an averaged value which would further enhance accuracy.

In the radiographic study (chapter 6), the experimental technique was modified since markers could not be attached. Intervertebral movements were computed directly from the movements of the anatomical landmarks. However, these landmarks could not be readily identified on radiographs. The sides of the vertebral bodies were thus traced by four tangent lines. The four intersections of these lines, which served as reference points for computing the movements, were more well-defined than the images of the vertebral corners and would therefore introduce less error. Since repeated computations of movements were allowed by having more than two reference points, together with repeated digitisation and tracing, the errors involved were reduced to a minimum.

Several other measures were adopted in the radiographic study in an attempt to improve accuracy. These included rigid fixation of the x-ray cassette, the use of reference markers in the films (to eliminate error due to different positioning of the cassette among different exposures), accurate and proper alignment of the x-ray beam using a special device, and the use of a long film-focus distance so as to minimise the obliquity of the incident x-ray beam. These precautions are generally not taken in routine x-ray examinations. It is suggested that future radiographic studies should take these measures into consideration if the errors in measurements are to be reduced to a minimum, and in particular, if the magnitude of the movements being measured is small.

In the present study, a technique was also developed to measure the deformation of the intervertebral foraminal size. Pins were used to define points around the foramen so that its size could be determined by a digital calliper. The technique was shown to be sufficiently reliable to reveal the changes produced. It appears that it is superior to other techniques such as radiographic measurement. Determinations of foraminal size by lateral radiographs are unlikely to be reliable as the left and right foraminal images were superimposed on the films. This technique also suffers from errors in locating relevant landmarks around the foramen which has rounded boundaries. Previous authors had also attempted to measure foraminal size by filling the foramen with silicone moulds and determining the areas of their cut sections (Stephens et al, 1991). However, pilot studies showed that this is unlikely to reveal small changes in size as the moulds have irregular shapes and it is technically difficult to produce the same cut sections with different moulds of the same foramen.

In chapter 8, the time-dependent effects of traction were examined by video techniques which provide dynamic measurements of spinal movements. Errors associated with digitisation in video analysis were significantly larger those observed in still photography where the images are much more magnified and not subjected to errors due to instability or lack of definition of the video pictures. While the technique was shown to be sufficiently reliable to detect the changes in the overall spinal length and the absolute displacements of the vertebrae in space, it was unable to reveal the small changes in the magnitude of the intervertebral movements.

It is suggested that future studies should employ techniques with more precision if small changes in intervertebral movements are to be determined. This may be achieved by using displacement transducers which are rigidly attached to the vertebrae. These devices may also be attached to reference points which are rigid extensions of the vertebrae. These points can be made as far apart as possible so as to further minimise angle calculation errors.

Previous authors developed magnetic tracking device to provide dynamic measurement of movements (An et al, 1988). The device generates a electromagnetic field and detects the position of a sensor in the field, and was shown to be reliable. Therefore, the small time dependent changes in interverebral movements may also be evaluated by rigidly attaching the sensors to the vertebrae. However, if the Instron machine is used for the mechanical testing, such a tracking device will not be appropriate as the large metal frame will distort the electromagnetic field and affect the measurements.

9.4 SUGGESTIONS FOR FUTURE STUDIES

The present study has thoroughly examined the biomechanical basis of mobilisation and traction techniques. However, much further research work still needs to be done if manual therapy techniques are to be further developed rationally.

As it has been demonstrated that clinical manual examination of segmental mobility is unreliable, studies should be carried out to establish other parameters which can be quantified easily with sufficient reliability in the clinical setting. Studies are required to examine the clinical validity and reliability of detecting the spinous process displacement as a measure of the stiffness of the spine in three-point bending during mobilisation.

Mobilisation has been shown to elicit creep and preconditioning effects which may lead to temporary improvement in spinal mobility. Investigations are needed to study other mechanisms that may possibly bring about permanent therapeutic effects. For instance, histological studies examining the effects of mobilisation on the composition of normal soft tissues and scars will be clinically useful.

It should be noted that the above explanation of the likely mechanisms of traction are hypothetical (section 9.2.3), although the present experimental results have provided some indirect evidence against which the hypotheses may be tested. These hypotheses need to be examined directly. For instance, neurophysiological studies are required to examine how the mechanoreceptors are deformed and stimulated during traction.

The present work has only examined the effects of mobilisation and traction when they are performed in the most common fashion. Clinically, the force of posteroanterior mobilisation may sometimes be applied at different inclinations, at different points of contact and in different lying positions. The mechanics of the procedure in these situations needs to be studied. The present mechanical analysis of traction is also limited to the situation when the Fowler's position is adopted. Future works should be extended to the analysis of traction in other positions such as prone lying and supine without flexion of the hips and knees.

A reliable experimental technique was developed in the present work to measure spinal deformations produced by mobilisation and traction. It is considered that similar technique may also be used to examine the biomechanical basis of other spinal manual therapeutic procedures.

The present work was largely limited to the examination of the mechanical effects of mobilisation and traction on the osteoligamentous structures. Their effects on spinal muscles is another issue that needs to be addressed. In addition, most of the previous works and the present study were carried out in cadaveric spines and normal living subjects. Future studies should be focused on patients with non-specific and well-defined spinal pathologies. The effects of pain and pathology on the mechanical behaviour of mobilisation, traction and other techniques remain to be tested.

■ *Chapter 10*

Conclusion

Conclusion

The deformations of the lumbar spine during mobilisation and treatment were examined in the present work. Both clinical procedures were shown to produce complex loading and movement patterns. An attempt was also made to examine the mechanisms underlying these techniques.

Radiographic study revealed that under the application of posteroanterior mobilisation loads, all the motion segments extended. The upper segments also translated posteriorly and the lower ones anteriorly. These movement patterns were the result of three-point bending of the spine during mobilisation. It was demonstrated that the spinous processes were displaced in the downward or ventral direction. They also moved caudually for the upper segments and cranially for the lower ones, indicating that the increase in lordosis due to extension of the spine had brought the two ends of the spine together.

The anatomical basis of posteroanterior mobilisation was thoroughly examined in a cadaveric study. It was shown that the intervertebral disc was the principal structure resisting the mobilisation loads. Although the soft tissues and the zygapophyseal joints did not appear to resist these loads to any great extent, large strains were produced in these pain-sensitive tissues. The above findings imply that disc injuries will have a significant effect on posteroanterior stiffness. Although injuries of soft tissues and the zygapophyseal joints are unlikely to change the stiffness, they may elicit a painful response during mobilisation examination.

Both the radiographic and cadaveric works mentioned above showed that the intervertebral movements produced by posteroanterior mobilisation were small in magnitude. This casts doubt on the reliability of clinical manual examination of segmental mobility as the therapists are unlikely to be able to "feel" these movements. Posteroanterior mobilisation provides a passive test of the stiffness of the entire spine in three-point bending. It should not be interpreted as a direct test of local segmental stiffness.

The complex loading and movement patterns of lumbar traction were examined in a cadaveric study. Simulated traction loads were found to produce flexion of the motion segments. This was because the line of action of the traction force was posterior to the centres of rotation of the discs. The adoption of the Fowler's position also produced a flexion moment on the spine. In addition, the motion segments were found to translate anteriorly as a result of the anterior shear imposed on them. The changes in the foraminal sizes induced by traction were also determined in this cadaveric study. The interpedicular and transverse foraminal distances were found to increase. Finally, it was shown that most of the mechanical effects of traction occurred after the application of the flexion moment which simulated the effects of the Fowler's position.

The observed deformations of the spine produced by traction has provided evidence against which its therapeutic hypotheses may be tested. Stretching of the posterior ligaments and joint capsules is suggested as the segments are flexed and the posterior vertebral corner translated superiorly. The increase in tension of the ligaments may reduce a disc bulge and the stretching of the innervated tissues may stimulate the mechanoreceptors leading to a closure of the pain gate. Traction is also believed to be useful in enlarging the size of a foramen which has been compromised by pathology.

The effects of repeated applications of traction force were also investigated in the present work. Residual increase in the overall length of the spine was observed immediately after the repeated loadings but most of this increase was lost 15 minutes afterwards. The vertebrae were also found to return more or less to their original positions. The above findings suggested that the time dependent mechanical effects of traction are temporary. However, the therapeutic effects achieved during therapy may continue to operate after the treatment.

Clearly, the scientific evidence reported in this thesis has added distinctly to the body of knowledge on the biomechanical basis of mobilisation and traction which are commonly employed in the practice of spinal manual therapy. The work has challenged some traditional beliefs of the clinicians. Palpation of segmental stiffness is highly improbable. It is also unlikely that traction may "suck back" a disc prolapse by reducing the intradiscal pressure.

It is hoped that the present work will stimulate further research. More studies are required to examine the effects of mobilisation and traction on tissue histology and the neuromuscular system, the effects on patients with back pain or spinal pathology and the biomechanics of other manual therapy techniques. The knowledge base of therapy must not be based on a model of tradition or authority but sound scientific principles. Research must be continued if the treatment techniques are to be able to stand up under the scrutiny of the scientific community and if the theories of practice are to be developed rationally.

Bibliography

Bibliography

Adams MA, Dolan P and Hutton WC (1988) The lumbar spine in backward bending. Spine, 13, 1019-1026.

Adams MA and Hutton WC (1983) The mechanical function of the lumbar apophyseal joints. Spine, 8, 327-330.

Adams MA and Hutton WC (1985) Gradual disc prolapse. Spine, 10, 524-531.

Adams MA, Hutton WC and Stott JRR (1980) The resistance to flexion of the lumbar intervertebral joint. Spine, 5, 245-253.

Allbrook D (1957) Movements of the lumbar spinal column. J Bone Joint Surg, 39B, 339-345.

Anderson JE (1985) Grant's atlas of anatomy, 8th edition, William and Wilkins Company, Baltimore.

Andersson GBJ, Ortengren R, Nachemson A and Elfstrom G (1974) Lumbar disc pressure and myoelectric back muscle activity during sitting. Part 1. Studies on experimental chair. Scand J Rehab Med, 6, 104-114.

Anderson GBJ, Schultz AB and Nachemson AL (1983) Intervertebral disc pressures during traction. Scand J Rehab Med Suppl, 9, 88-91.

Andersson GBJ, Schultz AB, Nathan A and Irstam L (1981) Roentgenographic measurement of lumbar intervertebral disc height. Spine, 6, 154-158.

Barbenel JC, Evans JH and Finlay JB (1973) Stress-strain-time relations for soft connective tissues, in "Perspectives in Biomedical Engineering", ed RM Kenedi, Macmillan, London, pp 165-172.

Beattie PF, Brooks WM, Rothstein JM, Sibbitt WL, Robergs RA, MacLean T and Hart BL (1994) Effect of lordosis on the position of the nucleus pulposus in supine subjects. A study using magnetic resonance imaging. Spine, 19, 2096-2102.

Berkson MH, Nachemson A and Schultz AB (1979) Mechanical properties of human lumbar spine motion segment. Part II: Responses in compression and shear; influence of gross morphology. J Biomech Eng, 101, 53-57.

Boden SD and Wiesel SW (1990) Lumbosacral segmental motion in normal individuals. Have we been measuring instability properly? Spine, 15, 571-576.

Bogduk N (1983) The innervation of the lumbar spine. Spine, 8, 286-293.

Bogduk N (1994) Anatomy and function of the lumbar back muscles and their fascia, in "Physcial Therapy of the Low Back", eds. LT Twomey and JR Taylor, 2nd edition, Churchill Livingstone, New York, pp 111-147.

Bogduk N and Twomey LT (1991) Clinical anatomy of the lumbar spine, 2nd edition, Churchill Livingstone, Melbourne.

Bogduk N, Tynan W and Wilson AS (1981) The nerve supply to the human lumbar intervertebral discs. J Anat, 132, 39-56.

Brown T, Hanson R and Yorra A (1957) Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs. J Bone Joint Surg, 39A, 1135-1164.

Burns ML and Kaleps I (1980) Analysis of load-deflection behaviour of intervertebral discs under axial compression using exact parametric solutions of Kelvin-solid models. J Biomech, 13, 959-964.

Burns ML, Kaleps I and Kazarian LE (1984) Analysis of compressive creep behaviour of the intervertebral unit subjected to a uniform axial loading using exact parametric solution equations of Kelvin-solid models. Part I. Human intervertebral joints. J Biomech, 17, 113-130.

Cailliet R (1981) Low back pain syndrome, 3rd edition, FA Davis, Philadelphia.

Chaffin DB and Andersson GB (1991) Occupational biomechanics. 2nd edition, Wiley Interscience, New York.

Chazal J, Tanguy A, Bourges M, Gaurel G, Escande G, Guillot M and Vanneuville G (1985) Biomechanical properties of spinal ligaments and a histological study of the supraspinal ligament in traction. J Biomech, 3, 167-176.

Christian GF, Stanton GJ, Sissons D, How HY, Jamison J, Alder B, Fullerton M and Funder JW (1988) Immunoreactive ACTH, β-endorphin and cortisol levels in plasma following spinal manipulative therapy. Spine, 13, 1411-1417.

Colachis SC and Strohm BR (1969) Effects of intermittent traction on separation of lumbar vertebrae. Arch Phys Med Rehabil, 50, 251-258.

Corrigan B and Maitland GD (1983) Practical orthopaedic medicine, Butterworths, London.

Cossette JW, Farfan HF, Robertson GH and Wells RV (1971) The instantaneous center of rotation of the third lumbar intervertebral joint. J Biomech, 4, 149-153.

Coxhead CE, Inskip H, Meade TW, North WRS and Troup JDG (1981) Multicentre trial of physiotherapy in the management of sciatic symptoms. Lancet, i, 1065-1068.

Cyriax J (1978) Textbook of orthopaedic medicine, vol.1 diagnosis of soft tissue lesions, 7th edition, Bailliere Tindal, London.

Daly CH (1966) The biomechanical characteristics of human skin, PhD thesis, University of Strathclyde, Glasgow.

Difabio RP (1986) Clinical assessment of manipulation and mobilisation of the lumbar spine. A critical review of the literature. Phys Ther, 66, 51-54.

Dimnet J (1980) The improvement in the results of kinematics of in vivo joints. J Biomech, 13, 653-661.

Dimnet J, Carret JP, Gonon G and Fischer LP (1976) A technique for joint centre analysis using a stored program calculator. J Biomech, 9, 771-778.

Doran DML and Newell DJ (1975) Manipulation in the treatment of low back pain - a multicentre study. Br Med J, 2, 161-164.

Dvorak J, Panjabi MM, Chang DG, Theiler R and Grob D (1991a) Functional radiographic diagnosis of the lumbar spine. Flexion-extension and lateral bending. Spine, 16, 562-571.

Dvorak J, Panjabi MM, Novotny JE, Chang DG and Grob D (1991b) Clinical validation of functional flexion-extension roentgenograms. Spine, 16, 943-950.

Dupuis PR, YH Ken, Cassidy JD and Kirkaldy-Willis WH (1985) Radiologic diagnosis of degenerative lumbar spinal instability. Spine, 10, 262-276.

Edwards WT, Hayes WC, Posner I, White AA and Mann RW (1987) Variation of lumbar spine stiffness with load. J Biomech Eng, 109, 35-42.

Evans JH (1973) Structure and function of soft connective tissue, PhD thesis, University of Strathclyde, Glasgow.

Farfan HF (1973) Mechanical disorders of the low back. Lea and Febiger, Philadelphia.

Farfan HF (1975) Muscular mechanism of the lumbar spine and the position of power and efficiency. Orthop Clin North Am, 6, 135-143.

Farfan HF (1980) The scientific basis of manipulative procedures. Clin Rheum Dis, 6, 159-178.

Farrell JP and Twomey LT (1983) Acute low back pain: comparison of two conservative treatment approaches, in "Proceedings of the International Conference on Manipulative Therapy", ed KD Bower, Perth, pp 162-179.

Figar S and Krausova L (1975) Measurement of degree of resistance in vertebral segments. Rehabilitacia Supp, 10, 60-62.

Frank C, Akeson WH, Woo SLY, Amiel D and Coutts RD (1984) Physiology and therapeutic value of passive joint motion. Clin Orthop, 185, 113-125.

Frymoyer JW (1988) Back pain and sciatica. New England J Med, 318, 291-300.

Frymoyer JW, Pope MH, Clements JH, Wilder DG, MacPherson B and Ashikaga T (1983) Risk factors in low back pain: an epidemiological survey. J Bone Joint Surg, 65A, 213-218.

Galante JO (1967) Tensile properties of the human lumbar annular fibrosus. Acta Orthop Scand Suppl, 100, 1-91.

Gertzbein SD, Chan KW and Tile M (1981) A new technique for the determination of the centre of rotation of the lumbar disc. Orthop Trans, 5, 477.

Gertzbein SD, Holtby R, Tile M, Kapasouri A, Chan KW and Cruickshank B (1984) Determination of a locus of instantaneous centers of rotation of the lumbar disc by Moire fringes. A new technique. Spine, 9, 409-413.

Gibson T, Grahame R, Harkness J, Woo P, Blagrave P and Hills R (1985) Controlled comparison of short wave diathermy treatment with osteopathic treatment in non-specific low back pain. Lancet, i, 1258-1261.

Glover JR, Morris JG and Khosla T (1974) Back pain: a randomized clinical trial of rotational manipulation of the trunk. Br J Ind Med, 31, 59-64.

Goel VK, Fromknecht SJ, Nishiyama K, Weinstein J and Liu YK (1985) The role of lumbar spinal elements in flexion. Spine, 10, 516-523.

Gonnella C, Paris SV and Kutner M (1982) Reliability in evaluating passive intervertebral motion. Phys Ther, 62, 436-444.

Gower WE and Pedrini V (1969) Age-related variations in the protein polysaccharides from human nucleus pulposus, annulus fibrosus and costal cartilage. J Bone Joint Surg, 51A, 1154-1162.

Grabias SL and Mankin HJ (1980) Pain in the low back. Bull Rheum Dis, 30, 1040-1045.

Gregersen GG and Lucas DB (1967) An in vivo study of the axial rotation of the human thoracolumbar spine. J Bone Joint Surg, 49A, 247-262.

Grieve GP (1988) Common vertebral joint problems, 2nd edition, Churchill Livingstone, Edinburgh.

Grover M (1982) Proposed mechanical effects of manipulative therapy, in "Proceedings of the Manipulative Therapists' Association of Australia Symposium: Towards A Better Understanding of Spinal Pain", Brisbane, pp 158-169.

Gunzburg R, Hutton W and Fraser R (1991) Axial rotation of the lumbar spine and the effec of flexion. An in vitro and in vivo biomechanical study. Spine, 16, 22-28.

Gupta RC and Ramarao SV (1978) Epidurography in reduction of lumbar disc prolapse by traction. Arch Phys Med Rehabil, 59, 322-327.

Hadler NM, Curtis P, Gillings DB and Stinnett S (1987) A benefit of spinal manipulation as adjunctive therapy for acute low back pain: a stratified controlled trial. Spine, 12, 703-706.

Hamilton WJ (1976) Textbook of human anatomy, 2nd edition, Macmillan and Co. Ltd, London.

Hayes MA, Howard TC, Gruel CR and Kopta JA (1989) Roentgenographic evaluation of lumbar spine flexion-extension in asymptomatic individuals. Spine, 14, 327-331.

Helliwell P, Moll J and Wright V (1992) Measurement of spinal movement and function, in "The Lumbar Spine and Back Pain", ed MIV Jayson, 4th edition, Churchill Livingstone, Edinburgh, pp 173-205.

Heylings DJA (1978) Supraspinous and interspinous ligaments of the human lumbar spine. J Anat, 125, 127-131.

Hickey DS and Hukins DWL (1980) Relation between the structure of the annulus fibrosus and the function and failure of the intervertebral disc. Spine, 5, 100-116.

Hindle RJ, Pearcy MJ, Cross AT and Miller DHT (1990) Three-dimensional kinematics of the human back. Clin Biomech, 5, 218-228.

Hinterbuchner C (1985) Traction, in "Manipulation, Traction and Massage", ed JV Basmajian, 3rd edition, Williams and Wilkins, Baltimore, pp 172-200.

Hirsch C and Galante J (1967) Laboratory conditions for tensile tests in annulus fibrosus from human intervertebral discs. Acta Orthop Scand, 38, 148-162.

Hirsch C and Nachemson (1957) A new observation on the mechanical behaviour of lumbar discs. Acta Orthop Scand, 23, 254-283.

Hirsch C, Ingelmark BF and Miller M (1963) The anatomical basis for low back pain. Acta Orthop Scand, 33, 1-17.

Hood CJ, Hart DL, Smith HG and Davis H (1981) Comparison of electromygraphic activity in normal lumbar sacrospinalis musculature during continuous and intermittent pelvic traction. J Orthop Sports Phys Ther, 2, 137-141.

Hood LB and Chrisman D (1968) Intermittent pelvic traction in the treatment of ruptured intervertebral disk. Phys Ther, 48, 21-30.

Horton GW (1958) Further observations on the elastic mechanism of the intervertebral disc. J Bone Joint Surg, 40B, 552-557.

Jayson MIV, Sims-Williams H, Young S, Baddeley H and Collins E (1981) Mobilisation and manipulation for low back pain. Spine, 6, 409-416.

Johnstone B, Urban JPG, Roberts S and Menage J (1992) The fluid content of the human intervertebral disc. Comparison between fluid content and swelling pressure profiles of discs removed at surgery and those taken postmortem. Spine, 17, 412-416.

Judovich B and Nobel G (1957) Traction therapy: a study of resistance forces. Am J Surg, 93, 108-114.

Jull G (1987) Manual examination - prove it or lose it, in "Proceedings of the Fifth Biennial Conference of the Manipulative Therapists' Association of Australia", Melbourne, pp 351-358.

Jull G and Bullock M (1987) A motion profile of the lumbar spine in an ageing population assessed by manual examination. Physiother Practice, 3, 70-81.

Kazarian LE (1975) Creep characteristics of the human spinal column. Orthop Clin North Am, 6, 3-18.

Keller TS, Holm SH, Hansson TH and Spengler DM (1990) The dependence of intervertebral disc mechanical properties on physiologic conditions. Spine, 15, 751-761.

Keller TS, Spengler DM and Hansson TH (1987) Mechanical behaviour of the lumbar spine. Part I. Creep analysis during static compressive loading. J Orthop Res, 5, 467-477.

Kelsey JL, Mundt DJ and Golden AL (1992) Epidemiology of low back pain, in "The Lumbar Spine and Back Pain", ed MIV Jayson, 4th edition, Churchill Livingstone, Edinburgh, pp 537-549.

King AI, Yamashita T, Ozaktay AC and Cavanaugh JM (1990) A morphological study of the lumbar facet joint capsule and its innervation, in "Proceedings of the 6th International Conference on Biomedical Engineering", Singapore, pp 32-35.

Koes BW, Assendelft WJJ, van der Heijden GJMG, Bouter LM and Knipschild PG (1991) Spinal manipulation and mobilisation for back and neck pain: a blinded review. Br Med J, 303, 1298-1303.

Lee M and Svensson N (1990) Measurement of stiffness during spinal simulated spinal physiotherapy. Clin Phys Physiol Meas, 11, 201-207.

Lee M, Esler MA, Mildren J and Herbert R (1993) Effect of extensor muscle activation on the response to lumbar posteroanterior forces. Clin Biomech, 8, 115-119.

Lee RYW (1990) Biomechanics of spinal posteroanterior mobilisation, MPhil thesis, Hong Kong Polytechnic, Hong Kong.

Lee RYW and Evans JH (1991) Biomechanics of spinal posteroanterior mobilisation, in "Proceedings of the Seventh Biennial Conference of the Manipulative Therapists' Association of Australia", Sydney, pp 59-64.

Lee RYW and Evans JH (1992) Load-displacement-time characteristics of the spine under posteroanterior mobilisation. Aust J Physiother, 38, 115-123.

Lee RYW and Evans JH (1994) Towards a better understanding of posteroanterior mobilisation. Physiotherapy, 80, 68-73.

Letchuman R and Deusinger RH (1993) Comparison of sacrospinalis myoelectric activity and pain levels in patients undergoing static and intermittent lumbar traction. Spine, 18, 1361-1365.

Lin HS, Liu YK and Adams KH (1978) Mechanical response of the lumbar intervertebral joint under physiological (complex) loading. J Bone Joint Surg, 60A, 41-55.

Liu YK, Ray G and Hirsch C (1975) The resistance of the lumbar spine to direct shear. Orthop Clin North Am, 6, 33-49.

Lorenz M, Patwardhan A and Vanderby R (1983) Load-bearing characteristics of lumbar facets in normal and surgically altered spinal segments. Spine, 8, 122-130.

Lovett RW (1905) The mechanism of the normal spine and its relation to scoliosis. Boston Med Surg J, 153, 349-358.

Macintosh JE and Bogduk N (1987) The morphology of the lumbar erector spinae. Spine, 7, 658-668.

Macintosh JE, Valencia F, Bogduk N and Munro RR (1986) The morphology of the human lumbar multifudus. Clin Biomech, 1, 196-204.

Maher C and Adams R (1994) Reliability of pain and stiffness assessments in clinical manual lumbar spine examination. Phys Ther, 74, 801-811.

Maitland GD (1986) Vertebral Manipulation, 5th edition, Baltimore, London.

Markolf KL (1972) Deformation of the thoracolumbar intervertebral joints in response to external loads. J Bone Joint Surg, 54A, 511-533.

Masturzo A (1955) Vertebral traction for treatment of sciatica. Rheumatism, 11, 62-67.

Mathews JA (1968) Dynamic discography: a study of lumbar traction. Ann Phys Med, 9, 275-279.

Mathews JA and Hickling J (1975) Lumbar traction: a double-blind controlled study for sciatica. Rheumatol Rehabil, 14, 222-225.

Mathews W, Morkel M and Mathews J (1988) Manipulation and traction for lumbago and sciatica: physiotherapeutic techniques used in two controlled trials. Physiother Practice, 4, 201-206.

Matyas TA and Bach TM (1985) The reliability of selected techniques in clinical arthrometrics. Aust J Physiother, 31, 175-200.

Mennell JM (1960) Back pain: diagnosis and treatment using manipulative techniques. Little, Brown and Co, Boston.

Mooney V and Robertson J (1976) The facet syndrome. Clin Orthop Rel Res, 115, 149-156.

McCollam RL and Benson CJ (1993) Effects of posteroanterior mobilisation of lumbar extension and flexion. J Manual Manipulative Ther, 1, 134-141.

McGill SM (1988) Estimation of force and extensor moment contributions of the disc and ligaments at L4/5. Spine, 13, 1395-1402.

McGill SM and Norman RW (1986) Partitioning of the L4-L5 dynamic moment into disc, ligamentous and muscular components during lifting. Spine, 11, 666-678.

McGlashen KM, Miller JJA, Schultz AB and Andersson GBJ (1987) Load displacement behaviour of the human lumbosacral joint. J Orthop Res, 5, 488-498.

McNally DS and Adams MA (1992) Internal intervertebral disc mechanics as revealed by stress profilometry. Spine, 17, 66-73.

Melzack R and Wall PD (1965) Pain mechanisms: a new theory. Science, 150, 971-979.

Miles M and Sullivan WE (1961) Lateral bending at the lumbar and lumbosacral joints. Anat Rec, 139, 387-398.

Mimura M, Panjabi MM, Oxland TR, Crisco JJ, Yamamoto I and Vasavada A (1994) Disc degeneration affects the multidirectional flexibility of the lumbar spine. Spine, 19, 1371-1380.

Nachemson AL (1960) Lumbar intradiskal pressure. Acta Orthop Scand Suppl, 43.

Nachemson AL (1975) Towards a better understanding of low back pain: a review of the mechanics of the disc. Rheumatol Rehab, 14, 129-143.

Nachemson AL (1976) The lumbar spine. An orthopaedic challenge. Spine, 1, 59-71.

Nachemson AL (1992) Lumbar mechanics as revealed by lumbar intradiscal pressure measurements, in "The Lumbar Spine and Back Pain", ed MIV Jayson, 4th edition, Churchill Livingstone, Edinburgh, pp 157-171.

Nachemson AL and Elfstrom G (1970) Intravital dynamic pressure measurements in lumbar disc. A study of common movements, maeuvers and exercises. Scand J Rehabil Med Suppl, 1, 1-40.

Nachemson AL and Evans JH (1968) Some mechanical properties of the third human lumbar interlaminar ligament (ligament flavum). J Biomech, 1, 211-220.

Nachemson AL and Morris JM (1964) In vivo measurements of intradiscal pressure. J Bone Joint Surg, 46A, 1077-1092.

Nachemson AL, Schultz AB and Berkson MH (1979) Mechanical properties of human lumbar spine motion segments. Influences of age, sex, disc level and degeneration. Spine, 4, 1-8.

Nwuga VC (1982) Relative therapeutic efficacy of vertebral manipulation and conventional treatment in back pain management. Am J Phys Med, 61, 273-278.

Office of Home Economics (1985) Back pain. Studies of current health problems series no. 78. Office of Home Economics Publications, London.

Onel D, Tuzlaci M, Sari H and Demir K (1989) Computed tomographic investigation of the effect of traction on lumbar disc herniations. Spine, 14, 82-90.

Ogston NG, King GJ, Gertzbein SD, Tile M, Kapasouri A and Rubenstein JD (1986) Centrode patterns in the lumbar spine. Baseline studies in normal subjects. Spine, 11, 591-595.

Ottenbacher K and Difabio RP (1985) Efficacy of spinal manipulation/mobilisation therapy. A meta analysis. Spine, 10, 833-837.

Oxland TR, Crisco JJ, Panjabi MM and Yamamoto I (1992) The effect of injury on rotational coupling at the lumbosacral joint. A biomechanical investigation. Spine, 17, 74-80.

Pal B, Mangion P, Hossain MA and Diffey BL (1986) A controlled trial of continuous lumbar traction in the treatment of back pain and sciatica. Br J Rheumatol, 25, 181-183.

Panjabi MM (1979) Centres and angles of rotation of body joints: a study of errors and optimization. J Biomech, 12, 911-920.

Panjabi MM, Chang D and Dvorak J (1992) An analysis of errors in kinematic parameters associated with in vivo functional radiographs. Spine, 17, 200-205.

Panjabi MM, Goel VK and Walter SD (1982) Errors in kinematic parameters of a planar joint: guidelines for optimal experimental designs. J Biomech, 15, 537-544.

Panjabi MM, Krag M and Chung TQ (1984) Effects of disc injury on the mechanical behaviour of the human spine. Spine, 9, 707-713.

Panjabi MM, Krag M, Summers D and Videman T (1985) Biomechanical time-tolerance of fresh cadaveric human spine specimens. J Orthop Res, 3, 292-300.

Panjabi MM, Krag M, White AA and Southwick WO (1977) Effects of preload on load-displacement curves of the lumbar spine. Orthop Clin North Am, 8, 181-192.

Panjabi MM, Oxland T, Takata K, Goel V, Duranceau J and Krag M (1993) Articular facets of the human spine. Quantitative three-dimensional anatomy. Spine, 10, 1298-1310.

Panjabi MM, Goel VK and Takata K (1982) Physiological strains in the lumbar spinal ligaments. Spine, 7, 192-203.

Panjabi MM. Takata K and Goel VK (1983) Kinematics of the intervertebral foramen. Spine, 8, 348-357.

Panjabi MM, Yamamoto I, Oxland T and Crisco J (1989) How does posture affect coupling in the lumbar spine? Spine, 14, 1002-1011.

Paris SV (1979) Mobilisation of the spine. Phys Ther, 59, 988-995.

Pearcy MJ (1979) A biomechanical analysis of interbody fusion of the lumbar spine. PhD thesis, University of Strathclyde, Glasgow.

Pearcy MJ (1985) Stereoradiography of lumbar spine motion. Acta Orthop Scand Suppl 212, 1-45,

Pearcy MJ (1986) Measurement of back pain and spinal mobility. Clin Biomech, 1, 44-51.

Pearcy MJ and Bogduk N (1988) Instantaneous axes of rotation of the lumbar intervertebral joints. Spine, 13, 1033-1041.

Pearcy MJ and Hindle RJ (1989) New method for the non-invasive three-dimensional measurement of human back movement. Clin Biomech, 4, 73-79.

Pearcy MJ, Portek I and Shepherd J (1984) Three-dimensional x-ray analysis of normal movement in the lumbar spine, Spine, 9, 294-297.

Pearcy MJ, Portek I and Shepherd J (1985) The effect of low back pain on lumbar spinal movements measured by three dimensional x-ray analysis. Spine, 10, 150-153.

Pearcy MJ and Tibrewal SB (1984a) Lumbar intervertebral disc and ligament deformations measured in vivo. Clin Orthop Rel Res, 191, 281-286.

Pearcy MJ and Tibrewal SB (1984b) Axial rotation and lateral bending in the normal lumbar spine measured by three-dimensional radiography. Spine, 9, 582-587.

Pearcy MJ and Whittle MW (1982) Movements of the lumbar spine measured by three-dimensional x ray analysis. J Biomed Eng, 4, 107-112.

Perey O (1957) Fracture of the vertebral end-plate in the lumbar spine. Acta Orthop Scand Suppl, 25, 1-101.

Phillips DR and Twomey LT (1993) Comparison of manual diagnosis with a diagnosis established by a uni-level lumbar spinal block procedure, in "Integrating Approaches. Proceedings of the Eighth Biennial Conference of the Manipulative Therapists' Association of Australia", ed KP Singer, Perth, pp 55-61.

Pope MH, Bevins T, Wilder DG and Frymoyer JW (1985) The relationship between antropometric, postural, muscular and mobility characteristics of males, ages 18-55. Spine, 10, 644-648.

Pope MH, Hanley EN, Materi RE, Wilder DG and Frymoyer JW (1977) Measurement of intervertebral disc space height. Spine, 2, 282-286.

Pope MH, Wilder DG, Matteri RE and Frymoyer JW (1977) Experimental measurements of vertebral motion under load. Orthop Clin North Am, 8, 155-167.

Pritzker KPH (1977) Ageing and degeneration in the lumbar intervertebral discs. Orthop Clin North Am, 8, 65-77.

Putto E and Tallroth K (1990) Extension-flexion radiographs for motion studies of the lumbar spine. A comparison of two methods. Spine, 1990, 107-110.

Rab GT and Chao EY (1977) Verification of roentgenographic landmarks in the lumbar spine. Spine, 2, 287-293.

Reilly JP, Gersten JW and Clinkingbeard JR (1979) Effect of pelvic-femoral position on vertebral separation produced by lumbar traction. Phys Ther, 59, 282-286.

Rockoff SD, Sweet E and Bleustein J (1969) The relative contribution of trabecular and cortical bone to the strength of human lumbar vertebrae. Calcif Tissue Res, 3, 163-175.

Rolander SD (1966) Motion of the lumbar spine with special reference to the stabilising effect of posterior fusion. Acta Orthop Scand Suppl, 90, 1-144.

Romanes GJ (1981) Cunningham's textbook of anatomy, 10th edition, Oxford Medical Publication, Oxford.

Saunders HD (1979) Lumbar Traction. J Orthop Sports Phys Ther, 1, 36-45.

Schultz AB and Andersson GBJ (1981) Analysis of loads on the lumbar spine. Spine, 6, 76-82.

Schultz AB, Belytschko TB and Andriacchi TP (1973) Analog studies of forces in the human spine: mechanical properties and motion segment behaviour. J Biomech, 6, 373-383.

Schultz AB, Warwick DN, Berkson MH and Nachemson AL (1979) Mechanical properties of human lumbar spine motion segments. Part II. Responses in flexion, extension, lateral bending and torsion. J Biomech Eng, 101, 46-52.

Seldin ED and Hirsch C (1966) Factors affecting the determination of the physical properties of femoral cortical bone. Acta Orthop Scand, 37, 29-48.

Seligman JV, Gertzbein SD, Tile M and Kapasouri (1984) Computer analysis of spinal segment motion in degenerative disc disease with and without axial loading. Spine, 9, 566-573.

Shaffer WO, Spratt KF, Weinstein J, Lehmann TR and Goel V (1990) The consistency and accuracy of roentgenograms for measuring sagittal translation in the lumbar vertebral motion segment. An experimental model. Spine, 15, 741-750.

Shambaugh P (1987) Changes in electrical activity in muscles resulting from chiropractic adjustment: a pilot study. J Manipulative Physiol Ther, 10, 300-304.

Shirazi-Adl A, Ahmed AM and Shrivastava SC (1986) A finite element study of a lumbar motion segment subjected to pure sagittal plane movements. J Biomech, 19, 331-350.

Sims-Williams H, Jayson MIV, Young SMS, Baddeley H and Collins E (1978) Controlled clinical trial of mobilisation and manipulation for patients with low back pain in general practice. Br Med J, 2, 1338-1340.

Sims-Williams H, Jayson MIV, Young SMS, Baddeley H and Collins E (1979) Controlled clinical trials of mobilisation and manipulation for low back pain: hospital patients. Br Med J, 2, 1318-1320.

Stephens MM, Evans JH and O'Brien JP (1991) Lumbar intervertebral foramens. An in vitro study of their shape in relation to intervertebral disc pathology. Spine, 16, 525-529.

Stokes IAF and Frymoyer JW (1987) Segmental motion and instability. Spine, 12, 688-691.

Stokes IAF, Medlicott PA and Wilder DG (1980) Measurement of movement in painful intervertebral joints. Med Biol Eng Comput, 18, 694-700.

Stokes IAF, Wilder DG, Frymoyer JW and Pope MM (1981) Assessment of patients with low back pain by biplanar radiographic measurement of intervertebral motion. Spine, 6, 233-240.

Sunderland S (1974) Meningeal-neural relations in the intervertebral foramen. J Neurosurg, 40, 756-763.

Svensson HO and Andersson GBJ (1982) Low back pain in forty to forty-seven year old men. Frequency of occurrence and impact on medical services. Scand J Rehab Med, 14, 47-53.

Svensson HO and Andersson GBJ (1983) Low back pain in forty to forty-seven year old men. Work history and work environment factor. Spine, 8, 272-276.

Tanz S (1953) Motion of the lumbar spine. A roentgenologic study. Am J Roentgenol, 69, 399-412.

Taylor JR and Twomey LT (1986) Age changes in lumbar zygapophyseal joints: observations on structure and function. Spine, 11, 739-745.

Taylor JR and Twomey LT (1994) The lumbar spine from infancy to old age, in "Physical Therapy of the Low Back", eds. LT Twomey and JR Taylor, 2nd edition, Churchill Livingstone, New York, pp1-56.

Thompson R (1983) Measurements of relative intervertebral displacements in the lumbar spine during application of a PAIVM. Postgraduate diploma dissertation, Lincoln Institute of Health Sciences, Melbourne.

Tkaczuk H (1968) Tensile properties of human longitudinal ligaments. Acta Orthop Scand Suppl, 115, 1-69.

Trott P, Evans D and Frick R (1989) Accuracy of manual palpation skills: the ability of manipulative therapists to detect changes in force and displacement on a palpation simulator, in " Proceedings of Sixth Biennial Conference of the Manipulative Therapists' Association of Australia", Adelaide, pp 207-214.

Twomey LT (1985) Sustained lumbar traction - an experimental study of long spine segments. Spine, 10, 146-149.

Twomey LT and Taylor JR (1982) Flexion creep deformation and hysteresis in the lumbar vertebral column. Spine, 7, 116-122.

Twomey LT and Taylor JR (1983) Sagittal movements of the human lumbar vertebral column: a quantitative study of the role of the posterior vertebral elements. Arch Phys Med Rehabil, 64, 322-325.

Twomey LT and Taylor JR (1987) Age changes in lumbar vertebrae and intervertebral discs. Clin Orthop Rel Res, 224, 97-104.

van der Heijden GJMG, Beurskens AJHM, Koes BW, Assendelft WJJ, de Vet HCW and Bouter LM (1995) The efficacy of traction for back and neck pain. A systematic, blinded review of randomized clinical trial methods. Phys Ther, 75, 93-104.

Vanharanta H, Sachs BL and Spivey MA (1987) The relationship of pain provocation to lumbar disc deterioration as seen by CT/discography. Spine, 12, 295-298.

Viidik A (1968) A rheological model for uncalcified parallel-fibered collagenous tissue. J Biomech, 1, 3.

Vernon-Roberts B and Pirie CJ (1977) Degenerative changes in the intervertebral discs of the lumbar spine and their sequelae. Rheumatol Rehab, 16, 13-21.

Waters RL and Morris JM (1973) An in-vitro study of normal and scoliotic interspinous ligaments. J Biomech, 6, 343-348.

Watson MJ, Burnett M and Dickens W (1989) Experiment in recording passive spinal movement. Physiother, 75, 747-749.

Weatherell VF (1987) Comparison of electromyographic activity in normal lumbar sacrospinalis musculature during static pelvic traction in two different positions. J Orthop Sports Phys Ther, 8, 382-390.

Weber H (1973) Traction therapy in sciatica due to disc prolapse. J Oslo City Hosp, 23, 167-173.

Weber H, Ljunggren AE and Walker L (1984) Traction therapy in patients with herniated lumbar intervertebral discs. J Oslo City Hosp, 34, 61-70.

White AA and Panjabi MM (1990) Clinical biomechanics of the spine, 2nd edition, JB Lippincott Company, Philadelphia.

Williams PL, Warwick R, Dyson M and Bannister LH (1989) Gray's anatomy, 37th edition, Churchill Livingstone, Edinburgh.

Winter DA (1990) Biomechanics and motor control of human movement. 2nd edition, Wiley Interscience, New York.

Woo SLY, Gomez MA and Akeson WH (1985) Mechanical behaviours of soft tissues: measurements, modifications, injuries and treatment, in "The Biomechanics of Trauma", eds. AM Nahum and J Melvin, Appleton-Century-Crofts, New York, pp109-133.

Wyke BD (1976) Neurological aspects of low back pain, in "The Lumbar Spine and Back Pain", ed MIV Jayson, Sector Publishing, London, pp 189-256.

Wyke BD (1985) Articular neurology and manipulative therapy, in "Aspects of Manipulative Therapy", eds. EF Glasgow, LT Twomey, ER Scull and AM Kleynhans, 2nd edition, Churchill Livingstone, Melbourne, pp 72-80.

Yang KH and King AI (1984) Mechanism of facet load transmission as a hypothesis for low back pain. Spine, 9, 557-565.

Yates DAH (1972) Indications and contraindications for spinal traction. Physiother, 58, 55-57.

Zusman (1986) Spinal manipulative therapy: a review of some proposed mechanisms, and a new hypothesis. Aust J Physiother, 32, 89-99.

Zylbergold RS and Piper MC (1981) Lumbar disc disease: comparative analysis of physical therapy treatments. Arch Phys Med Rehab, 62, 176-179.

■ *Appendix I*

Computation of the Movements of the Motion Segments

A1.1 TRANSFORMATION OF COORDINATE SYSTEM

A1.2 DETERMINATION OF THE COORDINATES OF ANATOMICAL LANDMARKS

A1.3 COMPUTATION OF SAGITTAL ROTATION

A1.4 COMPUTATION OF THE SAGITTAL TRANSLATIONS OF THE MOTION SEGMENT

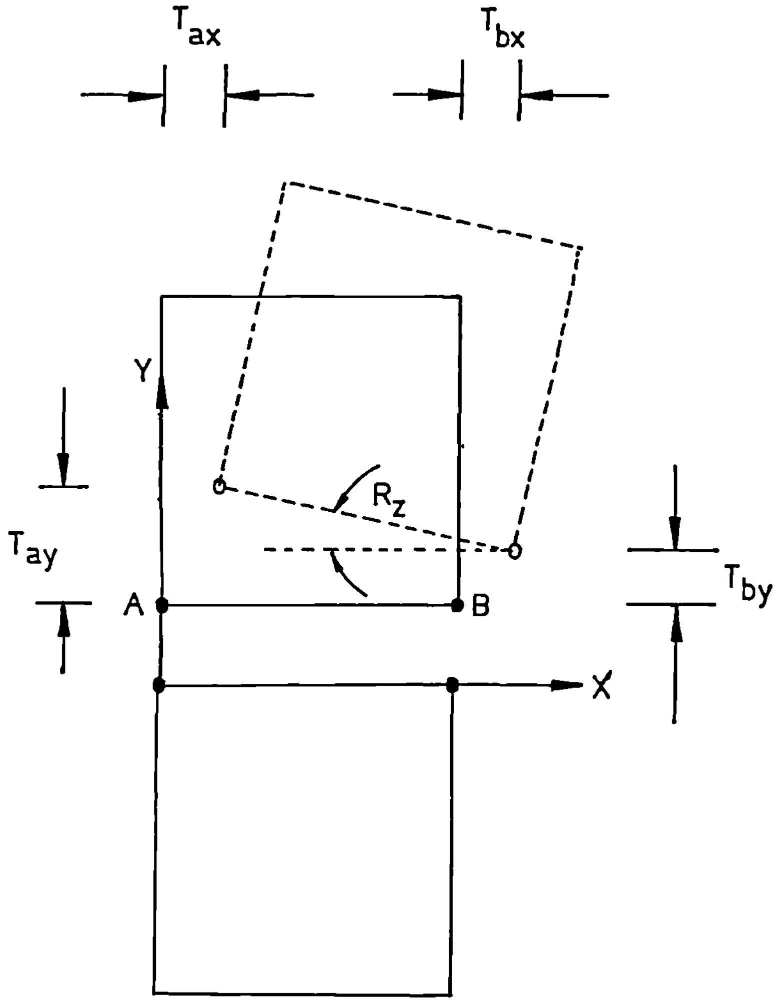


Figure A1
The Cartesian coordinate system used to describe movements of the motion segment. Segmental movements are described in terms of sagittal rotation (R_z), and x- and y-translations at the anteroinferior (A) and posteroinferior (B) corners of the superior vertebral body of the segment (T_{ax} and T_{ay}, T_{bx} and T_{by}).

Computation of the Movements of the Motion Segments

The mathematical procedures involved in the determination of the movements of the motion segment are presented in this Appendix. The techniques were employed in the various experiments of the present work.

A Cartesian coordinate system is established to describe the segmental movements. The origin is located at the anterosuperior corner of the inferior vertebral body of the motion segment (figure A1). The X axis of the system is directed posteriorly and formed by the superior endplate of the inferior vertebra, that is, the line joining the anterosuperior and the posterosuperior corners of the inferior vertebral body. The Y axis is directed superiorly perpendicular to the X axis.

The coordinate system moves with the inferior vertebra. The inferior vertebra is considered to be rigidly fixed while the superior one moves relative to it. Sagittal movements of the superior vertebra are described in terms of five movement parameters, namely,

- rotation of the superior vertebra about the mediolateral (Z) axis (the cross-product of the X and Y axes) on the sagittal plane (R_z),
- translations of the superior vertebra at the anteroinferior and posteroinferior corners of the superior vertebra (A and B) along the anteroposterior (X) and longitudinal (Y) axes (T_{ax}, T_{ay}, T_{bx} and T_{by}) (figure A1).

These motion parameters are determined from the movements of anatomical landmarks. The mathematical computation involves the transformation of coordinate

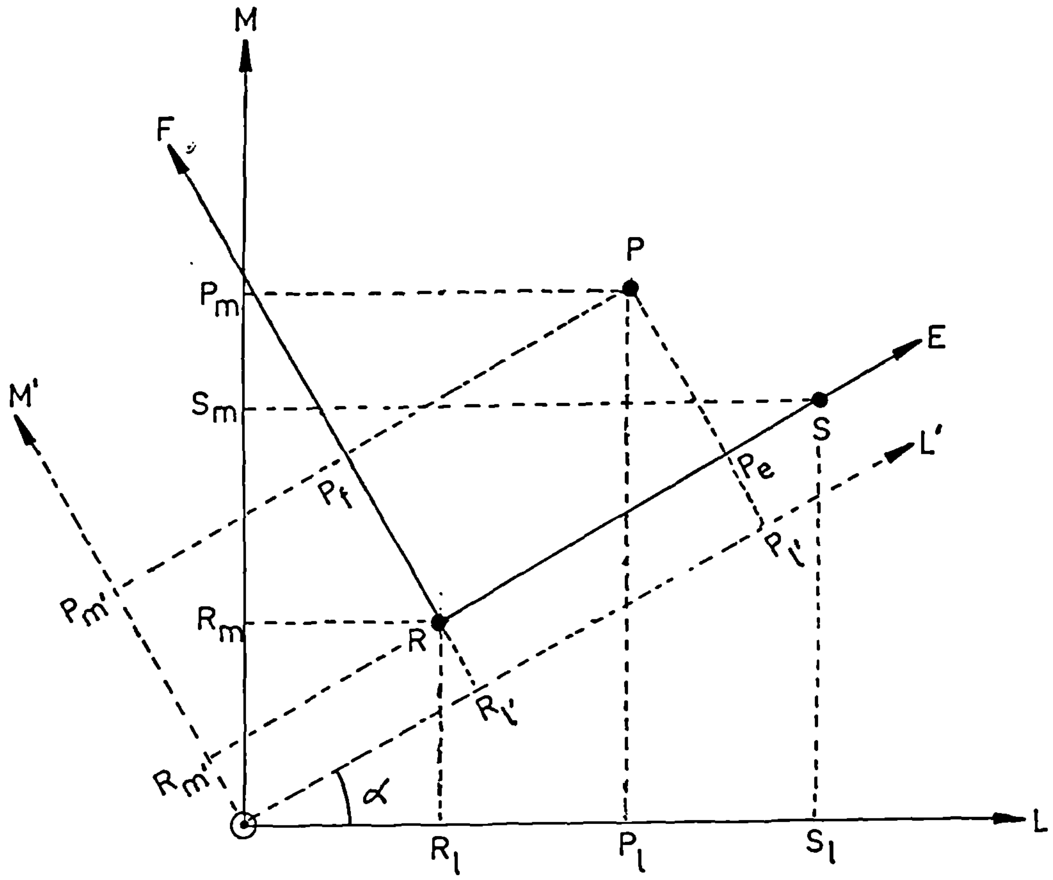


Figure A2
The transformation of the coordinates of a point P from a LM coordinate system to a new EF coordinate system which is established by the reference points R and S.

systems and the determination of the coordinates of anatomical landmarks which are also described in this appendix.

A1.1 TRANSFORMATION OF COORDINATE SYSTEM

During the process of computing the segmental movements, the coordinates of the markers or the anatomical landmarks are required to be transformed from one coordinate system to another, for instance, from a LM coordinate system to a new EF coordinate system (figure A2). In order to perform these, the projected inclination of the E axis of the new coordinate system relative to the L axis of the LM system, α, is first calculated as follows.

$$\alpha = \tan^{-1}\left(\frac{S_m - R_m}{S_l - R_l}\right) \qquad (A1.1)$$

where

(R_l, R_m) and (S_l, S_m) are respectively the l- and m-coordinates of two reference points R and S which form the E axis of the new coordinate system with the origin at R(figure A2)

The transformation of the coordinates of a given point P involves

1. rotation of the LM system by α, that is,

   $P_{l'} = (P_l \cos\alpha + P_m \sin\alpha)$

   $P_{m'} = (- P_l \sin\alpha + P_m \cos\alpha)$

where

(P_l, P_m) and $(P_{l'}, P_{m'})$ are the coordinates of the point P with respect to the LM system and the rotated LM system (the L'M' system) respectively

and,

2. translation of the origin of the L'M' system to the origin of the EF system which is located at R, that is,

   $P_e = P_{l'} - R_{l'}$

$P_f = P_{m'} - R_{m'}$

where

(P_e, P_f) are the coordinates of the point P with respect to the XY system, and

$(R_{l'}, R_{m'})$ are the coordinates of R with repect to the L'M' system.

In matrix representation, the transformation may be summarised as

$$\begin{bmatrix} P_e \\ P_f \end{bmatrix} = \begin{bmatrix} \cos\alpha & \sin\alpha \\ -\sin\alpha & \cos\alpha \end{bmatrix} \begin{bmatrix} P_l - R_l \\ P_m - R_m \end{bmatrix} \qquad (A1.2)$$

where

(R_l, R_m) are the coordinates of R with respect to the LM system.

A1.2 DETERMINATION OF THE COORDINATES OF ANATOMICAL LANDMARKS

The computation of segmental movements requires a knowledge of the coordinates of anatomical landmarks in different films or radiographs. These coordinates may be determined directly by digitisation of their locations on the films, or computed mathematically from the positions of the markers attached or any other chosen reference points.

Direct measurement of the position of anatomical landmark is subjected to errors (see section 5.3.1). Radiographic landmark such as the most posterior point of the spinous process is rounded, ill-defined and difficult to be identified consistently. The mathematical technique requires the identification of the landmark P in one film only and allows one to determine the new positions in other films from the positions of two chosen reference points. The reference points may be either markers which are rigidly attached to the vertebra or other anatomical landmarks which could be readily identified and located with accuracy. The mathematical technqiue avoids the problem of inconsistency in identifying an anatomical landmark in different films which will lead to significant error in the computation of the segmental movements. The error involved in determining the coordinates of the landmark in all films is consistent and will not add to each other.

The mathematical procedures involved in the determination of the coordinates of anatomical landmarks are described as follows. Assuming that the vertebra is a rigid body, there is a fixed relationship between the position of a given anatomical landmark P and any two markers or reference points (R and S). The coordinates of the landmark (P_e, P_f) with respect to a moving coordinate system (the EF system) established by the moving points R and S with R as the origin will remain constant in all films. The coordinates (P_e, P_f) thus represent the fixed relationship between P, R and S.

The coordinates (P_e, P_f) are determined using data from one of the photographic or radiographic films which shows the points P, R and S. Suppose this is the first film when the spine is unloaded and the global coordinate system is the LM system. The angle between the E and L axes before loading, α_o, is computed by equation A1.1, giving

$$\alpha_o = \tan^{-1}\left(\frac{S_{om} - R_{om}}{S_{ol} - R_{ol}}\right)$$

where

(R_{ol}, R_{om}) and (S_{ol}, S_{om}) are initial coordinates of the reference points R and S. Equation A1.2 is then used to compute (P_e, P_f).

$$\begin{bmatrix} P_e \\ P_f \end{bmatrix} = \begin{bmatrix} \cos\alpha_o & \sin\alpha_o \\ -\sin\alpha_o & \cos\alpha_o \end{bmatrix} \begin{bmatrix} P_{ol} - R_{ol} \\ P_{om} - R_{om} \end{bmatrix} \qquad (A1.3)$$

(In the experiment described in chapter 5, the anatomical landmarks were hidden inside the moulding cups. The coordinates (P_e, P_f) could not be found from any of the films and was thus determined after the experiment by direct measurements using digital callipers.)

The new position of the landmark in another photographic or radiographic film, for instance, in the one after loading, may be calculated from the new position of the EF coordinate system. The new angle between the moved E axis and the L axis, α_f, is given by

$$\alpha_f = \tan^{-1}\left(\frac{S_{fm} - R_{fm}}{S_{fl} - R_{fl}}\right)$$

where

(R_{fl}, R_{fm}) and (S_{fl}, S_{fm}) are new coordinates of the reference points R and S.

The new coordinates of the landmark (P_{fl}, P_{fm}) in the second film are then determined by transforming the coordinates (P_e, P_f) from the moved EF system back to the LM system using the new position of R and the new α_f angle as the control parameters. Mathematically,

$$\begin{bmatrix} P_{fl} \\ P_{fm} \end{bmatrix} = \begin{bmatrix} \cos(-\alpha_f) & \sin(-\alpha_f) \\ -\sin(-\alpha_f) & \cos(-\alpha_f) \end{bmatrix} \begin{bmatrix} P_e \\ P_f \end{bmatrix} + \begin{bmatrix} R_{fl} \\ R_{fm} \end{bmatrix}$$

or,

$$\begin{bmatrix} P_{fl} \\ P_{fm} \end{bmatrix} = \begin{bmatrix} \cos\alpha_f & -\sin\alpha_f \\ \sin\alpha_f & \cos\alpha_f \end{bmatrix} \begin{bmatrix} P_e \\ P_f \end{bmatrix} + \begin{bmatrix} R_{fl} \\ R_{fm} \end{bmatrix} \qquad \textit{(A1.4)}$$

A1.3 COMPUTATION OF SAGITTAL ROTATION

Sagittal rotation (R_z) is a motion parameter which is independent of the coordinate system employed. It can be computed directly from the movements of any two reference points (either the markers or the anatomical landmarks) in any coordinate system.

Mathematically, it is given by the change in inclination in the line joining the two reference points (P_{1x}, P_{1y}) and (P_{2x}, P_{2y}), that is,

$$R_z = \tan^{-1}\left(\frac{P_{f2y} - P_{f1y}}{P_{f2x} - P_{f1x}}\right) - \tan^{-1}\left(\frac{P_{o2y} - P_{o1y}}{P_{o2x} - P_{o1x}}\right) \qquad (A1.5)$$

where the subscripts o and f denotes the initial and final positions of the points respectively.

A1.4 COMPUTATION OF THE SAGITTAL TRANSLATIONS OF THE MOTION SEGMENT

The x and y translations of the superior vertebra are different for different points of the vertebra. The translations at the anteroinferior (A) and posteroinferior (B) corners of the vertebra were determined in the present work (figure A1). These points are chosen because their movements are clinically relevant. Their superoinferior (y) and anteroposterior (x) translations represent the changes in disc heights and shear displacements respectively.

The translational motion parameters are also dependent on the coordinate system employed. The coordinates of the anteroinferior and posteroinferior corners before (A_{ox}, A_{oy}, B_{ox}, B_{oy}) and after (A_{fx}, A_{fy}, B_{fx}, B_{fy}) loading with respect to the XY reference coordinate system are thus first determined. The technique described in section A1.1 (equation A1.2) is employed to perform these coordinate transformations. The x- and y-translations at these points are then given by

$$\begin{bmatrix} T_{ax} \\ T_{ay} \end{bmatrix} = \begin{bmatrix} A_{fx} \\ A_{fy} \end{bmatrix} - \begin{bmatrix} A_{ox} \\ A_{oy} \end{bmatrix} \qquad (A1.6)$$

$$\begin{bmatrix} T_{bx} \\ T_{by} \end{bmatrix} = \begin{bmatrix} B_{fx} \\ B_{fy} \end{bmatrix} - \begin{bmatrix} B_{ox} \\ B_{oy} \end{bmatrix} \qquad (A1.7)$$

■ *Appendix II*

Computer Programs

Computer Programs

The following computer programs were specially written for the data analysis required in the various experiments of this thesis:

- PAMOB.PAS (chapter 5)
- PAXRAY.PAS (chapter 6)
- TRACTION.PAS (chapter 7)

(The data obtained in chapter 8 were analysed by the commercial software "Peak Performance Motion Analysis system version 5.0" developed by the Peak Performance Technologies Inc., Englewood, USA.)

The computer programs are saved in the floppy disc attached. They can be run in the Turbo Pascal Version 5.0 Compiler Environment or in DOS (version 3.0 or above) by executing the relevant *.EXE file. In general, the programs contain the following procedures:

1. Data input
2. Correction for magnification
3. Computing intermarker distances (for the programs PAMOB.PAS and TRACTION.PAS)
4. Transformation of coordinate system
5. Determination of the coordinates of the anatomical landmarks
6. Computation of segmental rotation and translation (and displacement of spinous process for the program PAXRAY.PAS)
7. Display and saving of data

Menus are provided in the programs to facilitate their execution.