Force measurement during spinal mobilisation

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Force Measurement during Spinal Mobilisation

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ABSTRACT

Spinal mobilisation or manipulation techniques are frequently used by physiotherapists in the treatment of musculoskeletal disorders. Despite the reliance on these techniques in clinical practice, there is little scientific evidence to substantiate their use.

A standard mobilisation couch was instrumented to enable measurement of the forces applied to the trunk during mobilisation of the lumbar spine. Six load cells were incorporated into the couch frame and linked to a personal computer to facilitate data collection. The couch allowed the assessment of the magnitude of the mobilisation force, its direction and the variation in applied load over time. The system was found to be reliable and sensitive over the range of forces applied during mobilisation.

The system was used to collect data from a sample of 30 experienced therapists to evaluate repeatability and reproducibility during the application of four grades of a posteroanterior mobilisation and an End Feel, on the third lumbar vertebra. Whilst some therapists demonstrated considerable variation in the forces applied both within one measurement session and over a two week period, others were found to be relatively consistent. The range of forces used by different therapists when performing the same technique was substantial ranging between 63 N and 347 N for a Grade IV mobilisation.

A study was carried out involving 26 young healthy subjects, to determine the characteristics of a mobilisation force applied to an asymptomatic spine. A further study was undertaken involving a clinical sample of 16 patients, aged between 47- 64 years, to evaluate the effect of age related degenerative changes of the lumbar spine on the application of these techniques. The magnitude of the mobilisation force was found to be similar for the healthy and the patient groups with median forces of 175 N and 171 N during a Grade IV procedure, respectively. However, the forces applied to the patient group exhibited a statistically significantly smaller amplitude and higher frequency of oscillation than the healthy group for the same procedure ($p < 0.01$). Such measurements are essential for the assessment of the efficacy of these techniques in clinical practice.

Published Work

The research described in this thesis has led to the following journal publications and presentations:

Papers

Harms MC, Milton AM, Cusick G, Bader DL (1995) Instrumentation of a Mobilisation Couch for Dynamic Load Measurement. *Journal of Medical Engineering and Technology* 19:4:119-122.

Harms MC, Cusick G, Bader DL (1995) Measurement of Spinal Mobilisation Forces. *Physiotherapy* 81:10:599-604

Harms MC, Cusick G, Bader DL (1995) Instrumentation of a mobilisation couch. Proceedings, Physiotherapy Research Society. 19th April 1994. Institute of Neurology, London. *Clinical Rehabilitation* 9:1: 81.

Presentations

Harms MC, Bader DL (1994) The measurement of forces applied during spinal manipulative techniques. *The Society for Back Pain Research*, Leeds General Infirmary, 11th November 1994.

Harms MC, Bader DL (1995) Variability of force levels during spinal manipulation. *The Society for Back Pain Research*. Anglo-European College of Chiropractic, Bournemouth, 27th October 1995.

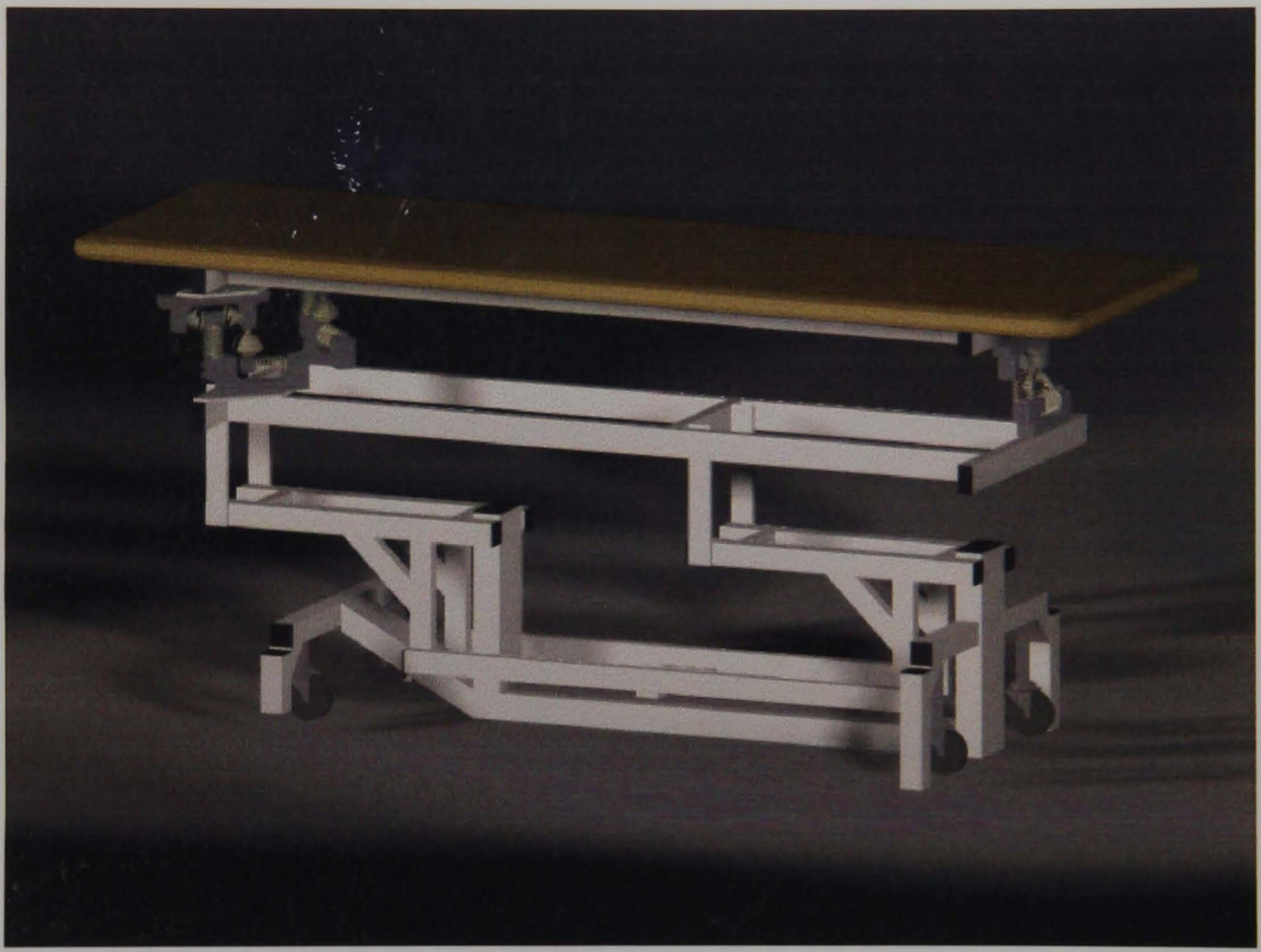
Harms MC, Cusick G, Bader DL (1994) Therapist Reliability in Spinal Mobilisation. *Chartered Society of Physiotherapy Annual Congress*, Birmingham 20-23 September.

Harms MC (1995) Spinal Manipulation: The black box. *Association of Chartered Physiotherapists in Orthopaedic Medicine*. The Middlesex Hospital, London 1st August 1995

Harms MC (1993) Measuring Mobilisation Forces: Has it all been done before? *Manipulative Association of Chartered Physiotherapists. Annual General Meeting*, Zoological Society, London, 20-21 November.

Of vertebral luxations: "one could only (adjust the vertebrae) by cutting the patient open and inserting a hand to exert pressure, 'but one might do this with a corpse but hardly with a living being' a point Hypocrates returns to from time to time with some regret"

(Hypocrates *in* Le Vay 1990)



Without a scientific basis for the assessment and measurement process, we face the future as independent practitioners unable to communicate with one another, unable to document efficacy and unable to claim scientific credibility for our profession

(Rothstein 1985)

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Glossary

Afferent:	Conveying inward, as sensory nerve
Aponeurosis:	Broad tendinous sheet, or musculotendinous junction, of dense connective tissue associated with the attachment of a muscle
Attenuation	Reduction in signal
Autonomic:	Not under voluntary control eg: regulation of blood vessel activity. Comprises of parasympathetic and sympathetic components
Axis of rotation:	Region in each bone where opposing forces cancel out. The axis remains stationary. Different forces cause different axes. Hence instantaneous axis of rotation
Coefficient of stiffness	Measure of a structure's resistance to deformation given by the ratio of force/displacement
Creep	Progressive deformation of a structure under a constant load
Cutaneous	Of the skin
Cyclical load	Dynamic load with repetitive pattern of variation
Damping	Property of object to resist speed of deformation involving energy loss
Discogram:	Roentgenography following injection of radiopaque fluid into intervertebral disc
Elasticity	The tendency of a material to return to its normal state once a deforming force has been removed

Fascia	Layer or sheet of connective tissue separating various muscles
Force couple	An association of one motion about an axis with another motion about a second axis
Hypermobility	Increased range of movement
Hypomobility	Reduced range of movement
Hysteresis	Energy loss during loading and unloading cycles
Isometric	Muscle contraction where the length of the muscle remains the same, also *Static*
Isotonic	Muscle contraction which results in shortening of the muscle
Lamella	Layer or sheet encircling the nucleus of the disc
Manipulation:	<u>Mobilisation</u> Refers to a passive joint movement defined by grade and rhythm, which the patient can prevent being performed. <u>Manipulative Thrust</u> A small amplitude rapid movement (not necessarily performed at the limit of range) which the patient cannot prevent taking place
Maximum Force	The maximum force applied during the mobilisation cycle averaged over all cycles within each 10 or 20 second recording period
Mechanoreceptor:	Variety of tactile end organs

Myelogram:	Roentgenography following injection of contrast medium into subarachnoid space
Neuritis:	Inflammation of a nerve
Nociceptor:	Free nerve ending (unspecialised) involved in chemical and mechanical nociception, all nociceptive nerves (A-delta and C category) sub-serve some other sensory function and only becomes nociceptive if the intensity of stimulation increases (Bogduk 1993), Sense organ responsive to noxious stimuli capable of eliciting pain
Osteophyte	Bony outgrowth adjacent to eroded articular cartilage
Passive Movement	Motion of a joint produced by a force external to the patient
Plastic range	Range beyond elastic limit where material suffers permanent deformation
Pre-tension:	Tension present when the spine is in a neutral position
Proprioceptor:	Sense organ responsive to movement or position
Radicular:	Relating to a nerve root, arising from disorders of the spinal nerves and roots
Radiculitis:	Inflammation of a nerve root
Referred Pain:	Pain felt at a distance from its source
Relaxation:	When the deformation of a structure is kept constant, the stress within the structure will decrease with time

Repeatability	Closeness of test results when tests performed under constant conditions: same operator, environment, during a short time interval (*BSI 1987*)
Reproducibility	Closeness of test results when tests performed under different conditions, operators, variable time interval (*BSI 1987*)
Resonant frequency	The natural frequency of the system
Rotation:	Points on a bone move in a similar direction around the centre of rotation
Sclerosis	Hypercalcification of bone
Shear force	Force which tends to cause translation
Somatic	Relating to skeleton or skeletal muscle as distinct from the viscera
Torque:	Net force causing rotation. Two unaligned forces act forming a force couple
Total force vector	Vector addition of the three component forces (Ftotal)
Trabeculae:	Supporting framework of struts within the bone
Translation:	Every point on the vertebral bone moves in the same direction
Viscoelasticity	Elasticity and viscosity in a material which make it demonstrate characteristics of creep, relaxation, load rate dependence and hysteresis

Statistical Glossary

Coefficient of variation (cv)	$\frac{sd}{X} x100\%$
Standard Error of the mean	$\frac{sd}{\sqrt{n}}$
Standard Error of the Estimate (SE)	Estimates sampling error of the estimate (see *Cohen & Holliday 1982* for expansion of formulae)
Confidence interval	SE x 1.96 (95% CI)
Regression	Method of least squares (*Cohen & Holliday 1982*): Y = a + bX
r	Pearson's product moment correlation coefficient ($r_{x.y}$)
r^2	Coefficient of determination which expresses the proportion of the variance of the dependent variable that is due to the effect of the independent variable

General Introduction

In a Health of the Nation initiative, the Government identified back pain as a strong candidate for key area status (*HMSO 1994*), but recognised that further research evidence was required before national targets could be set. Back pain is the most frequent reason for referral to physiotherapy with enormous associated costs. The annual cost of back pain management in the National Health Service in 1993 was estimated to be £481 million (Coyle & Richardson 1994). An episode of out-patient care for each patient costs between £74 to £111, with some programmes costing as much as £1052 (*Stanworth 1994*). These are in addition to the costs of post-graduate training in the use of specific techniques, calculated to be between £3,668 and £4,444 for each therapist. Aside from the frequently reported economics of back pain, the personal cost in terms of reduced capacity and human suffering should be considered. Although a large proportion of resources are allocated to its management, comparatively little is assigned to related research (*McCombe 1989, Harms 1990*).

At the beginning of this century Hugh Owen Thomas is reported to have said "I cannot find suitable cases upon which I would perform the deception known as a passive motion" (*Heyman 1930*). Yet passive techniques have a prominent rôle in current practice. 72% of patients with back pain will be treated with spinal manipulative procedures (*Meade 1990*), a form of passive mobilisation. Despite the routine use of these techniques, their effects are unknown, their efficacy remains unevaluated and there are no validated outcome measures to assess their use. The need to develop a rationale which is supported by scientific evidence has become a priority due to the increasing numbers of people who are referred for conservative management.

Previous studies have evaluated "physiotherapy" as a global treatment for back pain without sufficient control or definition of the treatment protocol. This is largely due to the lack of objective measures of the procedures used, therefore treatment can be neither quantified nor standardised. As a result, beyond the general finding that an acute episode of back pain can be resolved faster when manual therapy is prescribed, the results have been inconclusive.

The aim of this project is to develop a method of measuring the forces delivered during spinal mobilisation and to characterise a typical mobilisation force in different subject groups. The resulting system will also be used to determine the reliability of therapists in the use of common mobilisation procedures. This technique can be used to provide data which will form the basis of clinical trials of treatment efficacy, and to look at the effects of mobilisation on the spinal components.

Chapter 1: Anatomy of the Lumbar Spine

1.1 THE SPINAL CURVES

Figure 1-1
The vertebral column

The vertebral column is made up of seven cervical, twelve thoracic, five lumbar, five fused sacral and five fused coccygeal vertebrae. With the exception of the coccyx, the sections are alternately curved in the sagittal plane, convex anteriorly in the cervical and lumbar spine and posteriorly in the thoracic and sacral spine (Figure 1-1). In the thoracic and sacral regions the curves result from tapering of the vertebral bodies whereas in the lumbar and cervical spines the shape is influenced by the wedge shaped discs (*White & Panjabi 1990*).

In the lumbar spine the vertebral bodies do taper posteriorly, particularly the fifth lumbar vertebra which can be 3 mm higher anteriorly than posteriorly. Pearcy and Tibrewal (*1984*) measured the height of normal lumbar discs in standing and found them to taper between 4 and 8.5mm, increasing from L1 to L5.

Various methods of quantifying the lumbar lordosis have been described (*Branton 1984, Harms 1990*). One technique measured the perpendicular distance from a line joining the superoposterior border of L1 to the posteroinferior border of L5 to the apexes of the spinous processes and found the distance to be greatest at L3. The angle between adjacent vertebrae in the sagittal plane is greatest at L5/S1 and least at L1/L2 (*White & Panjabi 1990*).

1.2 VERTEBRAL BODIES

The lumbar vertebral body comprises of a cylinder, concave around the circumference in the vertical plane (Figure 1-2). The sides of the vertebral body are encased in a thin layer of cortical bone. The inner, cancellous bone is made up of trabeculae which are predominantly vertical in orientation to bear load. The vertical columns are interlinked by thinner horizontal trabeculae which provide the whole structure with stability (Twomey & Taylor 1994). The inter-trabecular spaces are filled with red marrow.

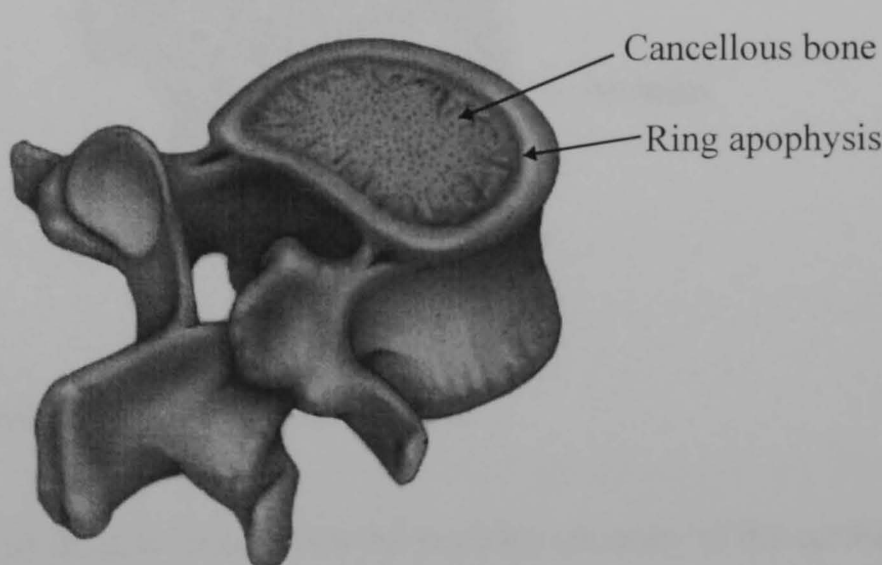


Figure 1-2: Posterolateral view of a typical lumbar vertebra

The superior and inferior surfaces of the vertebral bodies are surrounded by a compact rim of bone known as the ring apophysis. Centrally, the surfaces are continuous with the cartilage end plates which envelop the disc.

1.3 THE DISC

The discs are bound superiorly and inferiorly by hyaline cartilage end plates which attach to the vertebrae above and below. In the lumbar spine, the discs are approximately 11 mm thick (*Shirazi-Adl et al 1984*) and account for one third of the height of the lumbar spine. The disc is made up of two structures, the annulus fibrosis and the nucleus pulposus. Although these tend to be thought of as two distinct structures in young spines, the transition between the two becomes more gradual with aging reflecting changes in the collagen, water and proteoglycan content. The annulus fibrosis comprises of twelve to sixteen concentric lamellae composed mainly of collagen but with a component of elastic fibres which resist tensile forces (*Urban 1986*). They are arranged in spiralling sheets with the fibres crossing at an angle of 57-65 degrees (Figure 1-3).

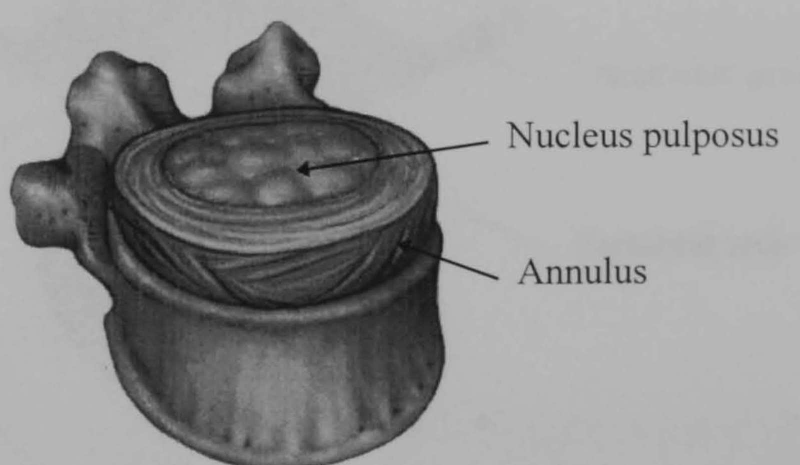


Figure 1-3: The basic structure of the intervertebral disc

The lamellae of the annulus are continuous with the lamellar structure of the cartilage end plates, both structures enclosing the nucleus pulposus. The outermost fibres of the annulus, named Sharpeys fibres, were thought to attach to the osseous tissue of the vertebral bodies and be embedded in the vertebral rim. However, recent research has shown these fibres to insert into the endplates at a significant distance from the ring apophysis (*Raushning 1993*). The nucleus is central within the disc and is made up of a fine network of fibrous strands supported by a mucoprotein gel (*White & Panjabi 1990*). Under compression, the nucleus transmits forces laterally which are resisted by tension in the annulus.

1.4 THE VERTEBRAL ARCH

The bony vertebral arch gives rise to four articular processes, two superior and two inferior, two transverse and one spinous process (Figure 1-4). The articular processes project from the lamina and the transverse processes project laterally from the junction of the pedicle and lamina. The pedicles project backwards from their attachment on the upper lateral margins of the posterior aspect of the vertebral body. They are continuous posteromedially with the laminae and together form the vertebral arch.

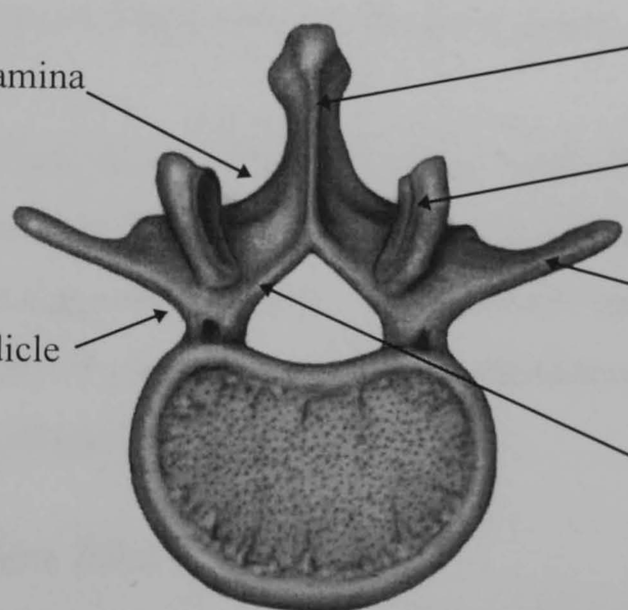


Figure 1-4: The components of the vertebral arch and the processes projecting from the pedicles and lamina

The lumbar transverse processes are long and flat, the distance between their tips increasing from L1 to L5 *(Panjabi et al 1992)*. The spinous processes are quadrangular and project horizontally from the most posterior point of the lamina. Often, the spinous process has a depression in the posterior border and on palpation may appear to be two processes. Panjabi et al (*1992*) provides detailed information on the dimensions of the vertebral components obtained from an in vitro study of 60 lumbar vertebrae.

1.5 LIGAMENTS

There are eight relevant sets of ligaments in the lumbar spine.

1.5.1 The Anterior and Posterior Longitudinal Ligaments

The longitudinal ligaments are so called because they cover the anterior and posterior surfaces of the body and disc throughout the entire vertebral column. The anterior longitudinal ligament is a broad, flat ligament attached both to the periosteum and the anterior annulus, which becomes thinner as it passes over the disc. The Posterior longitudinal ligament narrows as it passes over the vertebral body and expands over each disc where the fibres are interwoven with the outer annulus.

The superficial fibres of both the anterior and posterior longitudinal ligaments link several vertebrae, but the deep fibres link a single vertebrae with the one below. In the lumbar spine it has been suggested that fibres, conventionally thought to constitute the anterior longitudinal ligament may actually be the long tendinous attachment of the crura of the diaphragm (*Bogduk & Twomey 1991*).

1.5.2 The Annulus Fibrosus

Although the structure of the discs has been reported earlier, their role as the principle ligaments of the vertebral column warrants further description. The annulus fibrosus of the disc can be divided into a nuclear envelope portion and a ligamentous portion. The former is continuous with the cartilage end plate as previously described but the peripheral fibres are intimately interwoven with the anterior and posterior ligaments. They have the same histological basis as other tendons and ligaments (*Shirazi-Adl et al 1984*) and are subject to similar demands.

1.5.3 Ligamentum flavum

The ligamentum flavum has the highest percentage of elastic fibres of any tissue in the body. This is thought to prevent the ligament from buckling when it is in its shortened position and hence prevent it encroaching on the spinal cord (*White & Panjabi 1990*).

It is approximately 5-7 mm thick and links the lamina of one vertebra with the lamina of the vertebra below. With the laminae, it makes up the posterior border of the spinal canal and forms the anteromedial capsule of the zygapophyseal joint and the posterior boundary of the intervertebral foramen (*Schonstrom & Hansson 1991*). The ligament is stretched in flexion (*Adams et al 1980*) and is thought to prevent the zygapophyseal joint capsule being trapped in the joint during movement (*Mooney & Robertson 1976*). The ligament appears to be in two sections due to a central division, but in fact it acts as one structure (*White & Panjabi 1990*).

1.5.4 Interspinous Ligament

The Interspinous ligament is quadrilateral in shape and forms a double layer between the spines of subjacent vertebrae. This ligament was traditionally thought to limit flexion, but in vitro studies have revealed this function to be limited (*Taylor & Twomey 1987*). The posterior fibres of the interspinous ligament merge with the supraspinous ligament (*Myklebust et al 1988*).

1.5.5 Supraspinous Ligament

The Supraspinous ligament runs posterior to the posterior surface of the spinous processes attaching to their apexes, superficial to the interspinous ligaments. The supraspinous ligament generally terminates at the level of L4 (*Posner et al 1982*). There is some question as to whether the structure is a true ligament as its deep fibres are made up largely of tendinous fibres from the back muscles and superficially from the extensor aponeurosis and thoracolumbar fascia (*Bogduk & Twomey 1991)*.

1.5.6 Capsular Ligament

The zygapophyseal joint capsule attaches just beyond the articular margins of adjacent processes. The fibres are perpendicular to the plane of the facet joint (*White & Panjabi 1990*). Studies by Adams et al (*1980*) and Panjabi et al (*1982*) have shown that the role of the capsule as a ligament is substantial, controlling excessive movement and protecting the intervertebral disc from overload.

1.5.7 False Ligaments

The intertransverse, the transforaminal and the mamillo-accessory ligaments have all been termed false ligaments either because both attachments connect to the same bone or to soft tissues or because of their structure *(Bogduk & Twomey 1991)*. Their paths are closely linked to those of the nerves, particularly the transforaminal ligaments which, in traversing the intervertebral foramina, significantly reduce the space through which the nerve roots pass.

Overall, it is thought that the ligaments of the posterior vertebral arch are more important than those which lie anteriorly because they have a greater role in stabilising the bony arch and facet joints and providing a restraint to flexion.

1.6 MUSCLES

There have been many ways of dividing the posterior lumbar muscles, erector spinae and multifidus, into groups *(eg. White & Panjabi 1990, Macintosh & Bogduk 1987)*. The latter authors have chosen to divide the muscles into those between lumbar segments, those linking several segments and those which do not attach directly to the lumbar spine but span the deeper muscles from the thoracic spine to the pelvis. The posterior lumbar muscles act as a group, individual sections being recruited where necessary. Their primary functions are in postural control, initiating and controlling movement in association with gravitational forces and in controlling translation of the vertebrae. They are also responsible for producing lateral flexion and extension of the trunk in conjunction with other mechanisms utilising the thoracolumbar fascia and intra-abdominal pressure. Many of these muscles have secondary functions which are summarised in Table 1-1.

Table 1-1: Summary of main muscle groups of the lumbar spine, their attachments and additional functions

Group	Name	Main attachments	Additional Functions
Intersegmental Muscles	Intertransversari: lateralis & medialis	Link transverse, mammillary & accessory processes	Proprioceptive
	Interspinales	Small paired, flank interspinous ligament	Proprioceptive
Major Intrinsic (Poly-segmental)	Multifidus	Fan shaped layers joining spinous processes, laminae, ilium and sacrum.	Stabilises trunk against flexion during rotation. Contributes to capsule
	Erector Spinae: Lumbar Longissimus & Lumbar Iliocostalis	Link accessory & transverse processes & thoracolumbar fascia with posterior superior iliac spines.	Stabilises with multifidus.
Long Extrinsic Muscles (Poly-segmental)	Thoracic longissimus	Fascicles from ribs and transverse processes of thoracic vertebrae insert into progressively lower lumbar & sacral levels.	The majority of their length is in tendon form, the lower part is known as erector spinae aponeurosis
	Thoracic Iliocostalis	Fascicles from lower 7-8 ribs attach to ilium & sacrum.	Lower tendons contribute to the erector spinae aponeurosis.

The attachments and positions of muscles like the intertransversarii, which often lie close to joint centres, implies that they are unable to exert any appreciable force. However, the large concentration of muscle spindles within their structure suggests that their primary function could be proprioceptive, a function attributed to the majority of short small muscles of the body *(Bastide et al 1989)*.

1.6.1 Anterolateral Muscle Groups

Psoas Major extends from an attachment at the anterolateral aspect of the lumbar spine to an insertion on the lesser trochanter of the femur. Psoas Minor originates from the anterolateral aspect of T12-L1 and inserts into the iliopubic eminence on the pelvis. It is a weak flexor of the pelvis. Psoas major works in association with iliacus as a flexor of the hip joint. By reverse attachment, and with the femur fixed as in standing, Psoas major controls pelvic tilt.

1.6.2 Quadratus Lumborum

Quadratus lumborum is a wide rectangular muscle which, through a series of oblique and longitudinal fibres join the lumbar transverse processes, the ilium and the 12th Rib *(Bogduk & Twomey 1991)*. This group act to fixate the 12th rib during respiration and are lateral flexors of the lumbar spine.

1.6.3 Intermuscular aponeuroses

The lumbar intermuscular aponeurosis and the Erector Spinae aponeurosis separate sections of the Erector Spinae muscle, allowing them to move freely over the underlying components (*Macintosh & Bogduk 1987*).

1.6.4 The Thoracolumbar fascia

This comprises of three layers which arise from the anterior surface of the transverse processes (anterior layer), the apexes of the same (middle layer) and from midline (posterior layer). The three layers fuse laterally to the quadratus lumborum linking with the abdominal muscles. Little is known of the functions of the first two but there has recently been more interest in the biomechanical properties of the posterior layer. The criss-cross arrangement of fibres in the thoracolumbar fascia, combined with abdominal muscle contraction and raised intra-abdominal pressure, allow lateral tension to be converted into an extension moment of the lumbar spine (*Gracovetsky & Farfan 1986*) which reinforces the action of the posterior spinal muscles and ligaments during lifting.

1.7 THE MENISCI

Apart from the fat which fills any spaces between the capsule and synovium at the superior and inferior poles of the joint, there are other tissue inclusions within the joint. There is some controversy about the origin and constitution of these structures. They are often referred to as "meniscoid inclusions" and were originally thought to resemble the meniscus of the knee. Histologically this has now been shown as an incorrect analogy and that the synovial lined fibrofatty inclusions have a different constitution. Taylor and Twomey (*1985*) suggested that the inclusions may be pieces of articular cartilage, still attached to the capsule, which have sheared off the joint surface. This argument is supported by experimental evidence demonstrating that the repairing depression in the articular cartilage matched the shape of the inclusion (*Taylor & Twomey 1987*).

Bogduk and Engel (*1984*) identify three "meniscoid" fibrofatty structures: The first is a connective tissue rim which is a wedge shaped thickening of the capsule and acts simply as a space filler at the dorsal and ventral edges of the joint. The second is an adipose tissue pad, made up of a fold of synovium enclosing fat and blood vessels which can project into the joint cavity by approximately 2 mm. The third and largest structure is the fibro-adipose meniscoid projecting from the capsule. These enclose fat, collagen and some blood vessels and can project 5 mm into the joint cavity. In a previous study, the authors had found that all three structures were connected to the joint capsule and lined with synovial membrane (*Engel & Bogduk 1982*). Giles and Taylor (*1987*) report on the synovial folds projecting into the zygapophyseal joints. When these are traumatised by being pinched between the joint surfaces they become fibrous in nature. Giles and Taylor (*1987*) also found clearly defined nerves in these folds, thought to have vasomotor and nociceptive functions.

A review of these articles suggests that there still remains some question as to whether these structures are anatomical or pathological. However, the evidence of Engel and Bogduk (*1982*) suggests that two of the meniscus type structures have a common embryonic origin. They stem from mesenchymal cells which are present in the developing joint. In the adult, these cells have differentiated into fat, fibrous tissue and synovium which form fat pads or fibrous rims. Where they project into the joint they are subject to mechanical strains which cause the structure to become fibrous. Thus with the exception of the connective tissue rims, the inclusions are variations of the same structure. The meniscoid inclusions are thought to provide some protective function for exposed joint surfaces, and may increase the area of surface contact of the joint thereby distributing the load over a larger area. It has also been suggested that they maintain the congruity of the joint surfaces and have a lubricating function. The inclusions are thought to move in and out of the joint cavity in response to fluctuations in intrarticular pressure (*Twomey & Taylor1994*).

1.8 THE VERTEBRAL CANALS

The lumbar vertebral canal contains the conus of the spinal cord which ends between T12 and L3, and the nerve roots of the cauda equina. It is circular in cross section in the upper lumbar spine, becoming triangular or trefoil in shape towards the sacrum (*Kirkaldy-Wallis & McIvor 1976*). Average dimensions of the spinal canal are 18-20 mm in anteroposterior depth, narrowest at L3, and 24-27 mm in width increasing from L1 to L5 (*Panjabi et al 1992*). The intervertebral foramen are made up from opposing inferior and superior vertebral notches and transmit spinal nerves and vessels (Figure 1-5).

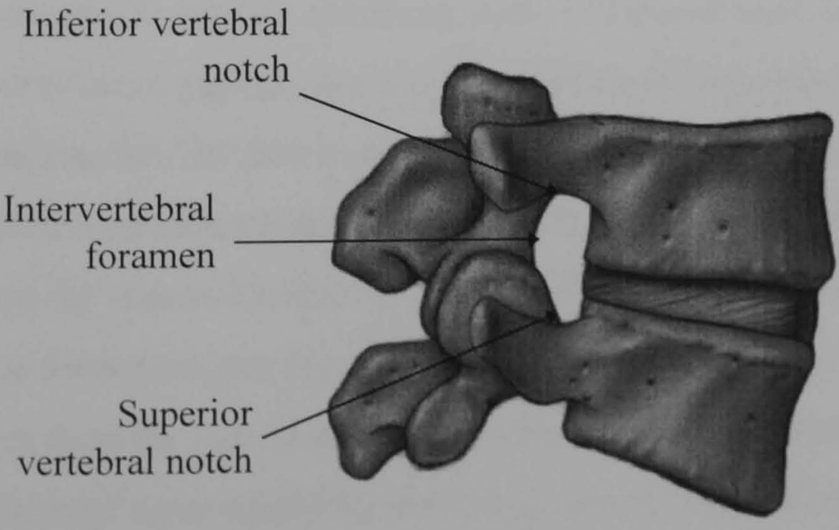


Figure 1-5: Articulated lateral view illustrating the boundaries of the intervertebral foramen

In the cervical spine the nerve roots span horizontally away from the spine, whereas in the lumbar spine the roots travel some distance vertically before spanning outwards. Reflecting this, the orientation of the intervertebral foramen changes through the lumbar spine from horizontal and lateral in the upper lumbar spine to anterior and caudal in the lower (*Stephens et al 1991*). The path that the nerve roots take from the vertebral canal, through the intervertebral foramen has been termed the radicular canal although it is not a true anatomical entity.

1.9 THE SPINAL NERVES

From the termination of the spinal cord, lumbar, sacral and coccygeal nerve roots pass down through the vertebral canal. They are encased in a dural sac and form the cauda equina. Each nerve root is composed of between 2 and 12 rootlets which emerge from the spinal cord. A pair of spinal nerve roots (one ventral and one dorsal) emerge from the dural sac encased in a dural sleeve which covers the nerve within the vertebral canal as far as the foramen.

These two nerve roots unite to form a spinal nerve which is approximately the same length as the intervertebral foramina through which it passes, although this varies between individuals (*Massey 1986*). Once through the intervertebral foramina the spinal nerves divide into ventral and dorsal rami. The dorsal rami supply tissue structures posterior to the zygapophyseal joints, including the joints themselves, the posterior vertebral muscles, the skin over the low back and glutei. The ventral rami, including the sinuvertebral nerve branches, supply structures anterior to the zygapophyseal joints including the vertebral bodies, the discs and their ligaments and joins other spinal nerves to form the lumbosacral plexus (*Taylor & Twomey 1987*). Sinuvertebral nerves also stem from the ventral rami and re-enter the vertebral canal. They contain somatic and autonomic roots supplying the disc, posterior longitudinal ligament, blood vessels and dura. These form a rich plexus of nerves which runs with the longitudinal ligaments supplying the vertebral body and related structures.

The area between the dural sac and the vertebral arch is termed the epidural region and, where space permits, is filled with connective tissue and fat. The dural sac and nerve root sleeves are thus tethered to the vertebral column. There is some debate about whether the nerve root sleeves are tethered within the intervertebral foramen and to the zygapophyseal joints. Troup (*1986*) reports that the lumbar nerve roots are relatively free allowing up to 10 mm of movement. However, Peretti and Micalef (*1989*) have shown that fibrous tissue surrounding the sleeve binds it to the lateral edge of the intervertebral foramina and that this and the tethered dural sheath prevent any movement of the spinal nerves or nerve roots.

The intervertebral foramen increase in size from L1 to L4. The L5-S1 foramina however is the smallest of the lumbar spine, yet encompasses the biggest spinal nerve. The L5 spinal nerve occupies 25-30% of the available space, whereas the nerves associated with L1 to L4 occupy between 7 and 22% (*Bogduk & Twomey 1991*).

Somatic motor and sympathetic preganglionic axons leave the spinal cord through the anterior roots. Sensory fibres, with their cell bodies in the dorsal root ganglia, enter through the dorsal roots.

1.9.1 Nerve Supply to the Disc

The suggestion that the disc is devoid of nerve endings has largely been dispelled. It is now accepted that at least the outer third and probably half of the annulus is well innervated with different types of nerve endings (*Bogduk 1987*). The greatest number are found laterally, with fewer posteriorly and least anteriorly. It is suggested that these nerve endings are sensitive both to position and painful stimuli.

1.10 BLOOD SUPPLY

The lumbar arteries arise mainly from the Aorta in pairs, one pair at each vertebral level. They divide into several branches adjacent to the intervertebral foramina Table 1-2.

Table 1-2: Summary of the main arteries surrounding the lumbar spine

Branch	Route	Supply
Lateral	Through Psoas	Abdominal wall
	With ventral and dorsal rami	Paravertebral muscles
Posterior	Under transverse processes	Back muscles, zygapophyseal joint laminae & spinous processes
Medial		
Anterior spinal canal branch	Through foramen	Network supplying vertebral bodies
Radicular branch		Nerve roots
Posterior spinal canal branch		Dural sac, laminae spinous and articular process.

The disc is largely avascular receiving nutrition from the cancellous bone of the adjacent vertebral bodies. There is some indication that this process is influenced by movement of the intervertebral joint which causes fluctuations in intradiscal pressure (*Grieve 1981*). Blood is drained from the lumbar spine from several internal and external venous plexus' via the lumbar veins which follow the arteries and nerves and drain into the inferior vena cava and through the ascending lumbar veins.

1.11 ZYGAPOPHYSEAL JOINTS

The zygapophyseal joints, synonymous with apophyseal and facet joints, are synovial joints formed between facets on the inferior articular processes of one vertebra with those of the superior articular processes of the subjacent vertebra. The superior articular facets, concave in the horizontal plane face medially (Figure 1-6) and the inferior articular processes which are convex, face laterally. However, there is some segmental variation and asymmetry in the orientation of the facets throughout the lumbar spine.

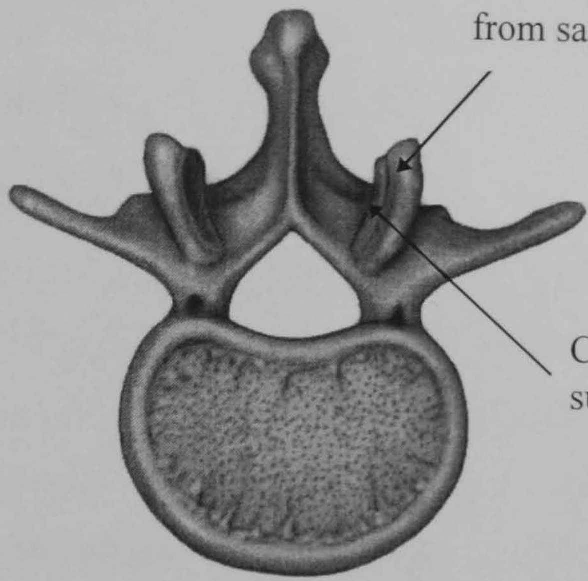


Figure 1-6: The orientation and curvature of the superior articular facets

Tulsi and Hermanis (*1993*) studied a sample of 112 lumbar spines. They demonstrated that the third lumbar vertebra (L3) exhibited the least variation in angle of inclination and curvature of the articular facet. Panjabi et al (*1993*) described the articular facet by three dimensions; its width, height and area. The width and height of L3 facet, varied between 13 and 16 mm, and the area was reported to be between 160 and 170 mm^2.

The joint is enclosed by a synovial membrane filled with joint fluid which lubricates the joint (*Hesse & Hansson 1988*). The synovial membrane, which is both vascularised and innervated (*Mooney & Robertson 1976*) attaches to the joint margins and is supported by a fibrous layer. In addition to the surrounding ligaments and tendons, the fibrous layer provides strength as a direct result of its arrangement of collagen fibres. The joint capsule is continuous with the ligamentum flavum anteriorly, and although slack posteriorly, superiorly and inferiorly, it is tensioned by the insertion of multifidus.

The majority of the joints at this level are angled between 30 and 50 degrees from the sagittal plane. The facet joints have components in both the sagittal and frontal planes, thus restricting rotational movements and anterior displacements respectively.
In the lumbar spine between 69 and 81% of the joints are curved, the remainder are flat (*Tulsi & Hermanis 1993*). In the curved joints, the anteromedial part of the superior

facet will resist anterior displacement where the upper vertebra's inferior facet will impact on the superior facet of the vertebra below. However, Stokes (*1988*) reports that the main function of the facet joint is to limit axial rotation.

1.11.1 Articular Cartilage

The joint hyaline cartilage is a specific and highly differentiated connective tissue. In the adult, the articular cartilage is approximately 2 mm thick on each facet and is attached to the underlying subchondral bone plate. It is divided into four histological zones: The tangential zone consists of three to four layers of cells lying parallel to the joint surface. Below this, the transitional zone consists of clusters of three to four cells, then the radial zone which forms the largest part of the cartilage with clusters of six to eight large cells residing mainly centrally in the joint surface. The calcified zone which is the deepest, covers the subchondral bone plate. The cells, which constitute between 1-10% of the total volume of the cartilage, are embedded in a matrix of glycosamino-glycans and collagen (*Bogduk & Twomey 1991*). The cartilage is avascular and has no nerve supply except at its periphery where it is continuous with the joint capsule. It receives nutrition from the joint fluid which is encouraged to circulate by movement. Smooth movement of the joint is dependent not only on the properties of the hyalin cartilage, but on correct lubrication. Although knowledge of the method of joint lubrication is incomplete, there is evidence that this too is promoted by movement (*Grieve 1981*).

1.11.2 The Joint Capsule

The capsule, which is lined by a synovial membrane, extends over the superior, inferior and dorsal aspects of the joint and is replaced by the ligamentum flavum over the ventral aspect. Dorsally the peripheral fibres insert 2 mm from the edge of the cartilage. Inferiorly and superiorly this distance is greater and the capsule is loose forming pockets which are filled with fat. There are also small foramina in the superior and inferior aspects of the capsule to allow the passage of fat from within the capsule (*Engel & Bogduk 1982*). The capsules are richly supplied with nerve endings which can transmit proprioceptive and nociceptive information and also contain mechanoreceptors (*Giles & Taylor 1987*).

Chapter 2: Biomechanics of Spinal Movement

2.1 THE SPINAL COLUMN

The rôle of the spinal column is complex as, whilst it bears load, it must allow movement in three planes and also protect the spinal cord. These conflicting functions are reflected by its components; the vertebral bodies are designed to bear weight, the joints guide movement and the vertebral arch provides some protection to the spinal cord. The biomechanical properties of the spine are often described with reference to the functional spinal unit (FSU) which consists of two adjacent vertebrae, the associated soft tissues, one intervertebral and two zygapophyseal joints.

Three factors contribute to the stability of the vertebral column. In weight bearing, hence under a compressive load, the configuration of the articular surfaces of the three joints offer stability (*Hsieh & Walker 1976*). Secondly, in both the loaded and unloaded positions, the FSU is stabilised by the ligaments, disc, capsule and menisci to varying degrees. Thirdly, there is some contribution from the active and passive influences of the surrounding muscles but this is dependent on the orientation of the muscle fibres. The degree of importance of the muscular contribution is still under some debate (*Taylor & Twomey 1987, White & Panjabi 1990*). However, the latter authors suggest that the major portion of mechanical stability is due to the highly developed dynamic neuromuscular control system and that without the muscles, the spine is so unstable that it cannot support the weight of the trunk.

In vivo the FSU is always subjected to some degree of loading. When standing, this is caused by the effect of gravity on the trunk above the FSU, usually in the form of a flexion moment, and the counterbalancing effect of the spinal muscles and ligaments. White and Panjabi (*1990*) report that both the ligamentum flavum and the anterior and posterior longitudinal ligaments demonstrate a degree of pretension in the neutral position. The ligamentum flavum produces a compressive stress of the intervertebral disc when the spine is in the neutral position.

Under compressive loads the fibre angle in consecutive layers of the annulus is reduced and results in tension along the annular fibres (*Pearcy & Tibrewal 1984*).

In supine lying, compressive loads of 250 N are normal in a man of 70 Kg and in sitting may increase to 1500 N (*Panjabi et al 1977*). There is little data available on spinal loading in prone lying. Whilst compression has been shown to have a significant effect on vertebral movement, generally increasing the range of motion (*Panjabi et al 1977*), it has also been shown to increase the stability of the joint (*Hsieh & Walker 1976*).

The disc, in concert with the facets of the zygapophyseal joints, carries most of the compressive load to which the trunk is subjected (*White & Panjabi 1990*). However, there is some controversy about the rôle of the zygapophyseal joints in weight bearing. Although the vertical orientation of the articular surfaces in the lumbar spine suggests that they are not designed for compressive load bearing, Adams and Hutton (*1983*) report that lordotic postures allow the facets to come into contact with the laminae below and take about one sixth of the load. Taylor and Twomey (*1987*) concur with this, suggesting that the zygapophyseal joints support approximately 20% of the axial load and increase the weight bearing base of the spinal column. Bogduk and Twomey (*1991*) report that the joints may bear anything from no load to 40%. The discrepancy in the reported contributions of the facets is likely to be the result of different testing procedures and variations in the orientation of the joint surfaces.

2.2 MOVEMENT OF THE FSU

Movements of the lumbar vertebrae are described by three translations and three rotations (Figure 2-1). Classically, the *X*, *Y* and *Z* axis of the rectangular or cartesian coordinate system are used to denote direction. In this report, the *Y* axis passes sideways through the body, the *X* axis from head to toe and the *Z* axis from front to back.

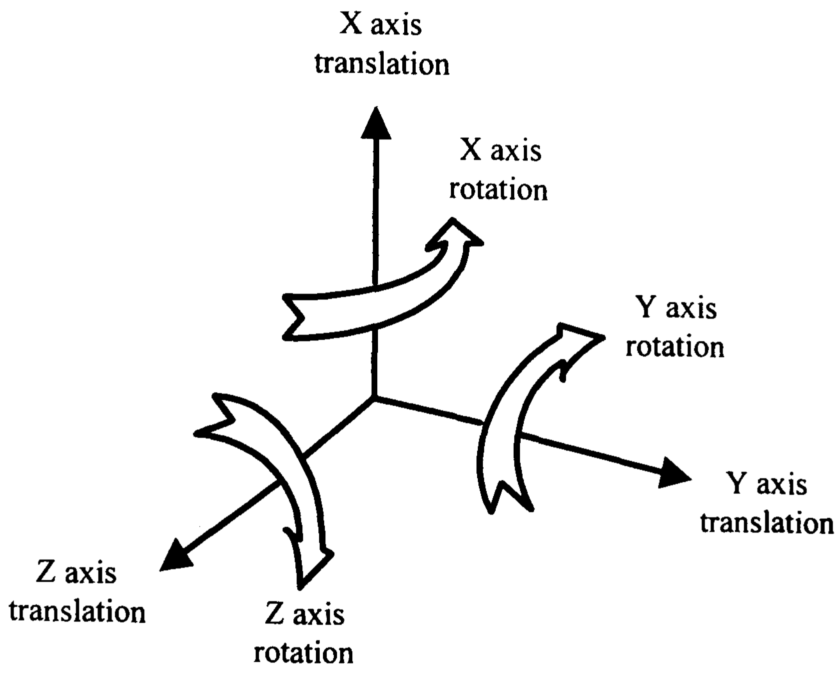


Figure 2-1: The co-ordinate system used to describe translation and rotation of the vertebrae

Clinically, movements of the lumbar spine are described as flexion, extension, lateral flexion and rotation and are composed of varying degrees of translations and rotations of individual vertebrae. Typically, movements are coupled so for example, in flexion the superior vertebra will translate 1-3 mm anteriorly, along the *Z* axis, with respect to the vertebra below while rotating 8-13 degrees about the *Y* axis (*Bogduk & Twomey 1991*).

In a study using fresh, cadaveric specimens, Panjabi et al (*1977*) applied unidirectional "physiologic" loads and measured the resulting three translations and three rotations ie: the movement in the direction of the applied force and the coupled movements. Results indicated that, for example, during extension the FSU moved between 1 and 5 degrees and was accompanied by 1 degree of both lateral and axial rotation, 1 mm of sagittal translation and 0-1 mm of lateral and vertical translation. However, in a later report on physiological movements of the lumbar spine, White and Panjabi (*1990*) suggest that although both lateral flexion and axial rotation are accompanied by rotation in the other two planes, because of sagittal symmetry of the vertebra, sagittal rotation is unlikely to be coupled by the other two rotations. This confirms the results of Pearcy (*1985*), who measured displacements in vivo using biplanar radiographic techniques.

Movements of the FSU are modified by the angle and curve of the facets. The disc permits movement in every direction by allowing one vertebra to slide forwards, backwards or sideways over another or by rocking or rotating in any direction (*Bogduk & Twomey 1991*). During movement, some of the annular fibres are under tensile stress and some will be relaxed. Which fibres are affected will be determined by the direction of movement in relation to the fibre orientation. The strength of the annulus is greatest when the force is applied in the direction of the annular fibres. Bending moments, which occur in flexion, extension and lateral flexion result in compression of one side of the disc and tension in the other. These two regions are separated by the instantaneous axis of rotation.

2.2.1 Effects of Physiological Movements on the FSU

The effects of the four physiological movements on the components of the FSU are summarised in Table 2-1. Only limited movement occurs at each lumbar segment giving the spine an inherent stability. Yet when the values are combined for all lumbar segments, the compound movement is large. Using radiographs on live subjects, Scull (*1978*) estimated that between 12 and 24 degrees of sagittal movement occurred at each segment with a total lumbar range of 60-80 degrees. This agrees with the account of White and Panjabi (*1990*) who report 12-17 degrees at each segment increasing from L1/2 to L5/S1, and Schneider (*1993*) who measured 10-22 degrees at each lower lumbar spine segment. Pearcy (*1985*) reported 13 to 16 degrees at each segment. The results are largely comparable, and the small differences are likely to be the result of differences in measuring technique. Flexion beyond the normal physiological range is prevented by apposition of the bony surfaces of the zygapophyseal joints, most usually the anteromedial aspects of the joint surfaces, and tightening of the posterior soft tissues. The zygapophyseal joint capsules are thought to provide the main ligamentous restraint (*Adams et al 1980*). Extension is limited when the joint surfaces impact, the spinous processes meet or the inferior facet comes into contact with the lamina of the vertebra below.

Table 2-1: Motion occurring at each component of the FSU during physiological spinal movement

Motion	Zygapophyseal Joint	Inter-vertebral Joint	Annulus	Nucleus
Flexion	Inferior facet of one vertebra glides upward (5-7 mm*) on superior facet of vertebra below Slight gapping allows further translation	Anterior sagittal translation. Anterior tilt	Anterior bulge. Posterior curves inward under tension	Anterior compression. Posterior deformation
Extension	Inferior facet of one vertebra glides down on superior facet of vertebra below until it contacts lamina or spinous processes impact	Posterior sagittal translation Posterior tilt	Posterior bulge. Anterior curves inward under tension	Posterior compression Anterior deformation
Lateral Flexion	Complex pattern involving gapping, approximation and gliding	Lateral translation and rotation	Bulge to side of concavity. Inward curve on side of convexity	Unilateral compression Contralateral deformation
Rotation	One approximates, one gaps. Further rotation occurs about impacted zygapophyseal joint	Rotation	Fibres in alternate lamellae under tension	Rotary deformation

White and Panjabi (*1990*) report an average of between 3 and 8 degrees of lateral flexion at each segment. These are similar to Pearcy's values (*1985*) of an average of 3 to 11 degrees of active lateral flexion between vertebrae. Lateral flexion is resisted by the soft tissues on the opposite side, especially the facet joint capsules.

There is some evidence that the joints in women are more mobile than those in men, especially in passive range (*Hesse & Hansson 1988*). The more pronounced joint laxity in women which may be a result of the size of the joints and soft tissues, which by being smaller offer less stiffness to motion.

It is still under debate whether any physiological rotation occurs in the lumbar spine. In the past it was thought that rotation would be prohibited by the angulation of the facet joints. However, several papers report that 1-3 degrees occur at each joint but further rotation causes the inferior facet of one vertebra to impact on the superior facet of the vertebra below, protecting the disc from excessive torsional shear (*White & Panjabi 1990, Pearcy 1985*) and possible damage to the subchondral bone of the facet (*Adams et al 1980*). It is possible that a few degrees of rotation may be achieved by compression of the articular cartilage within the zygapophyseal joints.

Facet asymmetry in the sagittal plane can be as great as 30 degrees (*White & Panjabi 1990*) and may allow preferential rotation in one direction. In addition, more rotation could occur in one direction at some levels and in the opposite direction at the remainder. This proposal was not substantiated in one recent study where the researchers inserted Steinmann pins into the spinous processes of living subjects and measured their angulation during rotation of the lumbar spine. Facet tropism, determined by x-ray, was not found to correlate with range of rotation (*Gunzburg et al 1991*). However, despite the use of local anaesthetic around the pin insertion sites, it is possible that the subjects movement was inhibited by knowledge of the technique and that sensory feedback from the tissues surrounding the pins may have prevented the subjects reaching their full degree of rotation at every level. The dispute about vertebral rotation is not resolved by radiographic studies. This is because of the difficulties involved in interpreting x-rays of rotary movements which may be obscured by parallax and superimposition of bony landmarks.

2.3 RESTRAINTS TO MOVEMENT

Biomechanical constraints protect the joint against excessive movement and consequent injury. Apart from isotonic muscular activity, the muscles give the spine stability by isometric contraction and by increasing the stiffness of the FSU (*White & Panjabi 1990*). This, combined with the bony configuration and ligamentous constraints, provide the main controls of excessive movement.

There is a range of movement within the so-called "neutral zone" which is virtually unresisted because of normal joint laxity. This ranges between 1.4 and 6 degrees for the main rotations in the lumbar spine (*White & Panjabi 1990*). Beyond this, the soft tissues provide increasing resistance to movement. This laxity results partially from "crimp" in the fibres of collagenous structures. When subject to tension, the collagen fibres align themselves in the direction of the applied force. Before they are ordered they offer little resistance to movement. In a ligament for example, a small degree of elongation can occur without resulting in strain within the fibres. Beyond this, stress which produces an elongation of the collagen fibres beyond 3-4% is likely to cause microscopic damage (*Bogduk and Twomey 1991*). Alteration in the arrangement of proteoglycan and water molecules within the structure and adjustments in the bonding of the collagen fibres during this elongation requires energy. Thus larger stresses are required to produce the same degree of displacement, a characteristic which allows the tissues to absorb energy.

The structural organisation of the collagenous fibres in ligaments provides resistance to tensile force but offers no opposition to compressive loading. Therefore, the resistance that they can offer depends on the orientation of their component fibres in relation to the direction of the movement. No one ligament or tissue is responsible for limiting a specific movement, but all combine in a complex pattern to allow a physiological range, while resisting excessive motion. The disc can limit extreme movement in all directions because of the pattern of orientation of collagen fibres in the annulus fibrosus. It is reported that the fibre angle between consecutive layers of 65 degrees is the optimal angle for the disc to resist all the stresses to which it is subjected (*Bogduk & Twomey 1991*).

The annulus is designed to bear compressive loads and only develops small strains within its fibres (3%) with relatively large loads (2500 N). The converse is true in torsion where small torques (12 Nm) applied to the disc cause high strains in the annular fibres (9%). Torsional damage of the annulus is prevented because the zygapophyseal joints restrict movement to only 3 degrees (*Adams & Dolan 1995, White & Panjabi 1990*).

2.4 VISCOELASTIC BEHAVIOUR OF THE FSU

Studies to determine the contribution of various components are generally performed on cadavers and involve sequential sacrifice of the structures under investigation (eg: *Posner et al 1982, Panjabi et al 1977*). Once the first structure has been sectioned, the biomechanics of the FSU are altered and the model can therefore provide only limited information. In addition, the testing rigs often confine vertebral movement to a single axis of rotation. This does not occur in vivo where the axis of rotation changes throughout the range of movement. There are few studies which examine the behaviour of the FSU in vivo. Bone, ligament, tendon and muscle are all viscoelastic tissues, which necessarily demonstrate some degree of creep, load relaxation and hysteresis. Because most studies use cadaveric material, deprived of neuromuscular influences and undergoing post-mortem changes, they can provide only limited knowledge.

The load displacement curve of a structure provides information about its stiffness, which is often expressed as the coefficient of stiffness. The load displacement curves of the disc and spinal ligaments are non-linear and are typically sigmoid in shape (*White & Panjabi 1990*). The sigmoid curve is divided into zones (Figure 2-2): the neutral and elastic zones describe the area up to the physiological limit of displacement, followed by the plastic zone up to the point of failure. This zone is also referred to as the traumatic range because of the ensuing microtrauma (*White & Panjabi 1990*). The curve reflects the small amount of force necessary to move the vertebra within its normal range. However, the rapid rise in stiffness as the joint reaches the end of physiological range means that it is protected and stabilised when acted upon by large loads.

As described above, the rearrangement of the collagen fibres during movement which requires energy, means that some energy is absorbed.

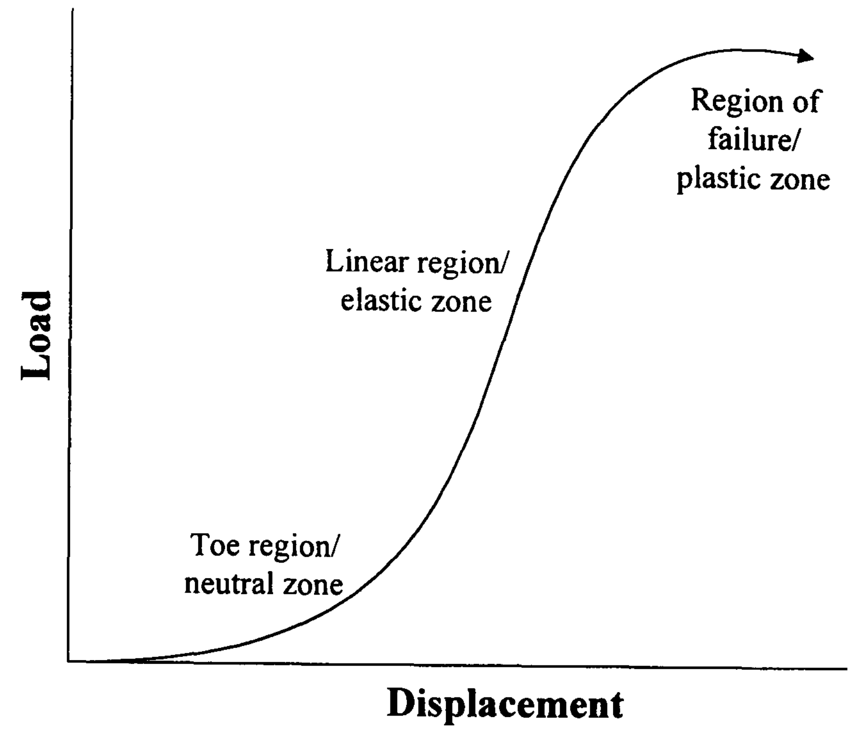


Figure 2-2: Typical shape of the load-displacement curve of the disc

The load displacement curve of viscoelastic material is dependent on the rate of loading, where an increase in rate causes the curve to become steeper. Lee and Svensson (*1993*) demonstrated that an increase in the rate of loading of the spine resulted in an increase in tissue resistance, giving smaller displacements for a given load.

The disc demonstrates viscoelastic behaviour and is subject to tensile, compressive and shear forces. The elastic properties of the annulus tend to resist deformation and return the disc to its normal shape after the deforming load has been removed. The nucleus in the healthy adult behaves as a viscous fluid (*Bogduk & Twomey 1991*) which does not change volume, but will alter its profile freely. Functionally, the disc transmits weight from one vertebra to the next and acts as a shock absorber by dissipating vertical forces horizontally. Its structure enables compressive forces applied to be translated into circumferential tension in the annulus (*Bogduk & Twomey 1991*).

Both creep and stress relaxation occur in collagenous structures and are thought to be the result of a rearrangement of collagen fibres, proteoglycans and water in response to loading within the structure. Each lumbar intervertebral disc creeps diurnally and may reduce in height by 1.5 mm during the day. This reduction in size changes the mechanical properties of the FSU and means that different structures are under greater loads at different times of the day (*Adams & Dolan 1995*) and are thus more prone to damage. The controversies discussed in Section 2.1 concerning the proportion of load taken by the zygapophyseal joints may be the result of estimates made during different stages of creep loading of the disc. At the beginning of the day when the disc has imbibed fluid during the night, load bearing through the facets may be minimal. Whereas in the evening, when the disc height is reduced, the contribution of the facets is likely to be greater. McGill and Brown (*1992*) evaluated the creep response of the lumbar spine when subject to flexion moments. In a study of healthy volunteers, they demonstrated that under normal loading, the degree of creep in the spine reached 5 degrees over 20 minutes. However, there was no indication from the data presented that the creep had reached its full extent when the experiment was terminated after a period of 20 minutes. Similarly, no conclusions could be drawn about the specific tissues responsible for this behaviour. Whether the passive influences of the soft tissues around the FSU or a gradual paying out of muscular tissue.

The biomechanical properties of the ligaments are similar to those of the disc, but are moderated by the direction of their fibres. They provide stability to the spine and help dissipate impulsive forces to protect the spinal column (*White & Panjabi 1990*). Research reports on ligaments deal mainly with their strength at failure, however this is only relevant in instances of spinal trauma. The spinal cord itself and the spinal nerves show viscoelastic properties. However, no data is available on its characteristics in shear and only limited information of what happens when it is subject to normal forces. Compression or tension in these structures can result in disturbance of the neuronal transport and vascular dynamics (*White & Panjabi 1990*).

2.4.1 Hysteresis

The energy loss as heat which occurs during the loading and unloading cycle of the spinal components is an important biomechanical consideration. Experimentally, when applied loads have caused excessive deformation in the soft tissues with an associated energy loss, there has been some residual lengthening of the loaded tissue. The spine has been shown to demonstrate hysteresis when subjected to flexion and extension in vitro (*Pope et al 1977*) and to flexion in vivo (*McGill & Brown 1992*). The disc exhibits hysteresis during walking, reducing the axial shock to the spine by its ability to absorb energy. Thus during jogging for example, the shock felt at the feet is modified by the time it reaches the head. The ability to absorb energy is thought to be greater in the lower lumbar vertebrae than those above. Most energy is absorbed during the first loading cycle, with less in subsequent cycles. This has important health implications especially in people subject to repetitive impulses or vibration. Causal links have been implicated between duration of exposure to axial vibration and spinal damage and herniated discs (*Grandjean 1986, White & Panjabi 1990*).

2.4.2 Vertebral Translation and Disc Shear

Biomechanical studies investigating the effects of forces on the spine have been almost exclusively devoted to compressive stress. However, there are a small number of reports dealing with the effects of shear. This is perhaps surprising as it is known that the spine has been shown to be more flexible in shear, than in compression and tension, and that anterior shear forces have been implicated in many studies of back injury (*Potvin et al 1991*). The angle of lordosis in supine, measured from the top of L1 to the top of the sacrum, ranges from 20 degrees to over 60 degrees with a mean of approximately 50 degrees (*Bogduk & Twomey 1991*). This angulation means that there is a tendency for one vertebra to slide forward on the one below, especially in the lower lumbar spine. The greater this tendency, the greater the posterior inclination of the facets to resist the tendency of one vertebra to slide forward on the one below (*Bogduk & Twomey 1991*).

Potvin et al (*1991*) haveestimated that the average anterior shear force on the lumbar spine, caused by the weight of the torso during normal movements, is about 400 N.

During weight lifting, forces can reach nine times this value. The authors suggest that the facets should be capable of supporting 400 to 2500 N, exclusive of that borne by the discs and ligaments. Where this limit is exceeded, the facets risk damage as no other mechanism is able to offset such excessive loads.

In standing, the posterior vertebral muscles partially brace the segment by pulling the upper vertebra down, thus compressing the facet joints. Because of the angle of pull of the paraspinal muscles, it was thought that they could only offer a small contribution. However studies have shown that the lumbar musculature may be capable of producing posterior shear forces in the region of 700 N (*Adams 1983*). The annulus does provide some resistance to shear forces. Although there is little empirical data about shear stresses in the annulus, White and Panjabi (*1990*) proposed that forces which result in vertebral shear would cause both tensile and compressive stresses in the annulus at 45 degrees to the transverse plane, which would resist further movement.

When the vertebral segment is subject to shear loading, the facets resist approximately one third of the force and the disc the remaining two thirds. During sustained loading the disc will creep and cause a significant transference of load onto the facets. In these conditions, the inferior facet of the moving vertebra impacts on the superior facet of the vertebra below, blocking further translation. Where there is articular tropism, the joints will resist shear unevenly, leading to asymmetrical loading of the facets (*Adams & Hutton 1983*). From radiographs of human volunteers, Pearcy (*1985*) and Posner et al (*1982*) estimated that physiological movements of the spine were accompanied by approximately 2.5 mm of sagittal translation. Schneider (*1993*) reported a mean of 4.6 mm of sagittal translation between full flexion and extension. Taking a different approach, Panjabi et al (*1977*) applied loads to an FSU directly. The specimens retained their ligamentous soft tissues but were devoid of muscular influences. They determined that when loads of 150 N were applied along the Z axis, 3-5 mm of Z axis translation occurred. The larger range reported may have resulted from using an FSU denuded of neuromuscular and other soft tissue influences, or from using large focused loads.

Chapter 3: Pathology

The incidence and impact of back pain on the general population is often estimated. The Clinical Standards Advisory Group (CSAG 1994) published data for the year 1993/4 which reported that in the United Kingdom at least 106 million days of work capacity were lost due to back pain during that period. In addition to the personal cost to the sufferer, back pain places demands on the health service in the region of £696 million per annum. In the USA it has been reported that between 1971 and 1981, the incidence of disability due to back pain increased at a rate fourteen times that of the population growth (*Frymoyer & Mooney 1986*). There are indications that a similar pattern is evident in the United Kingdom (*CSAG 1994*). It is possible that this is a result of increasingly sedentary lifestyles, or alternatively, the lower tolerance of illness within the community. It has been reported that 60% of the population experience low back pain at some point in their lives and that 80% of these will suffer recurrence (*National Back Pain Association 1994*).

The collection of relevant data poses many problems to the epidemiologist. In England, data collected by the Office of Population Censuses and Surveys on Health Episode Statistics is noticeably deficient. Using the International Classification of Disease Codes, in the years 1989 to 1990, between 15% and 24% of all patients were categorised "diagnosis unknown". Where cases were given a specific code, the category was often ambiguous, providing little information of either the disease process or the condition of the patient. For example, there were 364 recorded cases of Thoracic or Lumbosacral neuritis or radiculitis, and over 39,000 cases of sciatica and lumbago. Where a diagnosis could not be attempted, 2,107 cases of unspecified back disorders and 25,446 cases of "Backache, unspecified" were recorded. Neither information on the sampling frame nor on the reliability of this method of coding was made available. This demonstrates the difficulties of determining meaningful incidence statistics and of developing an accurate diagnostic classification system.

Few papers in the literature on back pain pathology report the reliability or validity of the measures used. This is surprising when 20-30% of popular diagnostic tests, including myelograms and magnetic resonance imaging, elicit positive findings in subjects with no back pain (*Hasue 1993*). In one study by Videman et al (*1990*) the reliability was assessed in cadaveric specimens. Intraobserver (ie:test-retest) reliability was "almost perfect" for detecting end plate defects using discograms (Kappa 0.92), "substantial" for annular ruptures (Kappa 0.80), "moderate" for facet degeneration (Kappa 0.57) and "fair" for osteophyte assessment when x-ray findings were compared with joint examination following dissection (Kappa 0.30). However, the reported Kappa statistics as defined by Everitt (*1989*) indicate that in some instances such as facet degeneration, agreement is only marginally better than that expected by chance. This raises important questions about the potential for diagnostic error, especially when more than one observer is involved.

3.1 PAIN SYNDROMES

Low back pain can result from either mechanical injury of the spine or from a disease process. The consequence of injury will be determined by the direction of the applied force and on the relative susceptibility of individual structures to damage. The spine is affected by many disorders which give rise to a wide range of symptoms. The following discussion is limited to those conditions where spinal manipulative therapy is considered to be effective.

Two pain syndromes are frequently reported: somatic pain or pain arising from the musculoskeletal structures; and radicular pain which results from disorders of the spinal nerves and roots. These syndromes will be discussed separately.

3.1.1 Somatic Pain

To develop a framework in which to examine the causes of somatic pain arising from the lumbar spine, those with a demonstrable nerve supply are listed below.

Structures supplied by the dorsal rami:

1. Zygapophyseal Joints
2. Muscles
3. Interspinous ligaments

Structures supplied by the ventral rami:

1. Vertebral bodies
2. Intervertebral discs
3. Longitudinal ligaments
4. Dura mater

All of these structures, with the exception of the vertebral bodies, have been shown experimentally to cause pain (*Bogduk & Twomey 1991*).

Pain occurs when nerve endings within the tissues are irritated by chemical or mechanical stimuli. Chemical irritation arises when the products associated with inflammation and cell breakdown are released following tissue damage. Although the means of mechanical irritation is unclear, it has been suggested that when collagen fibres within a given structure are subject to tension, the interspersed nerve endings may be compressed and give rise to pain (*Bogduk 1987*).

Somatic pain tends to be localised to the lumbar spine, although experimental evidence has shown that it can be referred to areas distant to the originating structure (*Bogduk & Twomey 1991*). Afferent impulses from damaged components of the lumbar spine activate neurones within the central nervous system which also receive afferent neurones from the lower limbs and other structures. This gives the impression of pain originating from all structures supplied by these neurones. Painful stimuli that are detected by cutaneous receptors can be accurately localised. However, pain arising from deeper tissues tends to be more nebulous because the nerves subserving these structures are less well organised for localising the source (*Bogduk 1993*). Thus referred pain is a reflection of central excitation and a lack of localising stimuli within the affected structure.

Diagnostic tests often involve the stimulation of the tissue thought to be causing pain. Providing the original pain is reproduced it is assumed that the structure tested is that responsible (*Mooney & Robertson 1976*). This is apparent in an interpretation by Bogduk (*1987*) on the use of provocative discography, where injection of normal saline into a subjects disc provoked pain, even when the disc appeared normal. The implicit interpretation was that the disc damage was previously undetected. Under the theory of somatic pain referral, a different explanation could be offered: If another structure supplied by the same nerve was damaged, the connecting network may be sensitised. Irritation of any structure on that network may then give rise to pain, regardless of which structure is damaged or which irritated.

3.1.2 Radicular Pain

Radicular pain occurs when the nerve root is compressed as it passes through the radicular or intervertebral canal. Radicular pain is unlikely to be confined to the lumbar spine and will be accompanied by sensory or motor disturbances in a distribution compatible with the nerve root being compressed. Sciatica is an example of radicular pain and is thought to be caused by compression of the lumbosacral nerve roots. Sciatic pain is described as "shooting" or lancinating and has been shown to follow a narrow band in the lower limb (*Bogduk 1993*).

Compression of nerves usually results in paraesthesia or numbness, not pain, which challenges nerve root compression theories. Hasue (*1993*) reports on studies which indicate that in order for compression to evoke pain, the nerve root must have suffered previous damage. However, in the presence of inflammation, the circulating chemicals may sensitise even normal nerve roots intensifying any noxious stimuli. In contrast, stimulation of the dorsal root ganglia will give rise to pain in the absence of previous damage and it appears to be more easily affected by noxious stimuli than the nerve root itself. Bogduk (*1993*) suggests that syndromes commonly attributed to nerve root compression are more likely to be the result of disturbance of the dorsal root ganglia.

Because the path of the nerve root is confined, any structure which encroaches into that space is likely to compress, tension or angulate the nerve root leading to chronic tissue irritation (*Troup 1986*). Entrapment can occur at any location within the spinal or radicular canal or intervertebral foramen (*Kirkaldy-Wallis & McIvor 1976*). The size of the spinal canal can be affected by pathological processes especially disc herniation or bulging. In some people with congenital spinal stenosis, the canal may be narrow from birth which will predispose the patient to symptoms should a space occupying lesion develop. Thickening of the ligamentum flavum has been suggested to be the most important element in spinal stenosis, where protrusion of the thickened ligament from the dorsal aspect of the canal compresses various neural structures (*Schonstrom & Hansson 1991*).

The shape and size of the intervertebral foramen is affected by different pathological processes. The predominant profile of a normal foramen is oval, but those affected by an abnormal or "barrelling" disc become auricular (*Stephens et al 1991*) while those with osteophyte encroachment from the zygapophyseal joint become teardrop in shape. A change in the dimensions of the foramen can give rise to symptoms of nerve root compression. In addition, trauma within the nerve root environment may lead to the development of adhesions, which when stretched during movement may cause irritation to the nerve root. Similarly, alterations in any of the structures forming the Radicular canals can also cause nerve root compression (*Bogduk & Twomey 1991*). These disorders are collectively referred to as nerve root entrapment syndromes.

Several structures travel through the intervertebral canal. If compression or ischaemia of the nerve root is evident it is also likely that the blood vessels will be compressed, causing venous congestion (*Tesio 1991*). Apart from blocking axonal transport, pressure on the spinal nerve will alter the blood flow and vascular dynamics of the nerve root (*Hasue 1993*). This in turn will lead to inflammation of the neural or surrounding tissues, resulting in secondary nerve root compression and ischaemia (*Jayson 1994*).

In the past, the distribution of somatic pain has been used in an attempt to identify damaged structures. However, because of significant variation between individuals and large areas of overlap in segmental distribution, this technique is neither useful nor accurate as a diagnostic indicator. Similarly, somatic pain was not thought to refer pain below the knee, therefore pain in the calf and foot was assumed to be radicular. It has now been demonstrated that although somatic pain is generally most intense in the buttock, it may be referred to the foot and the distance of referral is likely to be proportional to the pain intensity (*Mooney & Robertson 1976*). Somatic pain results in a deep ache and radicular syndromes, a sharp pain, yet because the two may coexist the symptoms may be confused. Indeed, when nerve root compression leads to radicular pain, the dural sleeve must first be compressed, resulting in local or referred somatic pain.

3.2 ZYGAPOPHYSEAL JOINT OSTEOARTHRITIS AND DEGENERATIVE DISC DISEASE

Degenerative lumbar disc disease, sometimes termed spondylosis, and osteoarthritis of the zygapophyseal joints increase with age, progressing steadily from the second decade (*Butler et al 1990, Eisenstein & Parry 1987*). It is estimated that 97% of lumbar discs will show some degree of degeneration by the age of 50 years (*Miller et al 1988*). Spinal zygapophyseal joints are identified as one of the most vulnerable sites for osteoarthritis and thus it may be suggested that almost all people over 70 years old would suffer from the condition (*Dieppe & Kirwan 1994*).

There is some debate about whether changes in the zygapophyseal joints cause degenerative changes in the disc, or whether disc degeneration alters the mechanical loading of the zygapophyseal joints, predisposing them to arthritic changes. In one study, Yang and King (*1984*) demonstrated that arthritic zygapophyseal joints were subject to greater compressive loads than their normal counterparts. Although not mentioned by the authors the increase, from between 3 - 25% of the total load to just under 50% in arthritic joints, appears to have been the result of accompanying disc degeneration. Butler et al (*1990*) found that in a sample of 68 patients with low back pain, those subjects who demonstrated osteoarthritis of the zygapophyseal joints had accompanying disc degeneration, but that disc degeneration could exist without facet involvement. However, it is conceivable that the sample used in this study demonstrated bias because they had all sought medical intervention for back pain. It is not clear whether the incidence of osteoarthritic changes, without disc degeneration, would be different in the general population.

Zygapophyseal joint problems have recently received attention as a cause of back pain. Until recently the normal anatomy of these joints was not known, and as a consequence, their importance as a cause of back pain was overlooked. In 1976 Nachemson classified arthritic changes in the "facet joint" as an irrelevant radiologic finding, not related to back pain. As recently as 1990, Videman reported on a study of 86 cadavers that facet osteoarthritis was not related to the incidence of back pain. The evidence of Videman, however, is based on an assessment procedure which had only moderate reliability (Kappa = 0.57). The history of back pain was also given by the relatives of the deceased and there was no method of determining the validity or reliability of this evidence.

The popular contention that osteoarthritis results from wear and tear of the joint is no longer held (*Ali 1978*). The trigger for Osteoarthritis is thought to be associated with chemical changes in the articular cartilage environment, although mechanical factors may also have a rôle (*Schofield & Weightman 1978*). In addition, age, gender and previous joint damage may be predisposing factors in the development of osteoarthritis (*Dieppe & Kirwan 1994*).

Osteoarthritis is a highly active process with increased cell activity in many tissue types and is thus distinguishable from changes due to the aging process (*Bland 1993*). The pathological changes associated with Osteoarthritis are summarised in Table 3-1. The disease tends to be characterised by pain, limited movement, deformity and progressive joint degeneration. However, because of the location of the spinal joints some of these symptoms may not be apparent to the examiner. For example, it is unusual to record joint effusions in the zygapophyseal joints.

Table 3-1: Summary of pathological changes occurring in intra-articular and extra-articular structures in osteoarthritis

Tissue	Pathological Change
Cartilage environment	Change in hyalin cartilage environment, increased rates of collagen synthesis and depletion of proteoglycans, chondrocyte clumps/clones form. Focal erosion of hyalin cartilage (fibrillation) denudation and regrowth, cartilage forms bumpers around joint. Grooves and ridges develop in articular surface
Subchondral Bone	Increased osteoblastic activity. Loss of compliance resulting in subchondral microfractures Callus deposited in microfractures, Synovial seepage into trabecular bone leads to pseudocysts, (more common in peripheral joints) Collapse of bone/cartilage junction with deepening of concavity of joint
Synovium	Synovial cells produce connective tissue mixture, osteophytes form on joint margins, Joint effusion
Soft tissues	Hypertrophy and stiffening of capsule, ligament, muscle and tendon

Based on: Dieppe & Kirwan (*1994*), Dieppe, Elson and Kirwan (*1993*), Bland (*1993*), Hesse & Hansson (*1988*), Mankin (*1974*)

The cartilage of the zygapophyseal joint can be preserved late in the disease process, but extensive osteophyte formation may occur before, contrary to the process occurring in large weight bearing joints (*Butler et al 1990*). The changes in the articular surfaces, stiffness in the surrounding soft tissues and the formation of osteophytes lead to a reduction in the quality and amount of joint movement. In addition, changes in the articular geometry and loss of intra-articular space can cause the joint to become unstable.

There is some evidence that the arthritic process can be halted and indeed reversed although few studies have shown the repair to be adequate. Mankin (*1974*) reports on the highly organised repair process of articular cartilage in early to middle stages of the disease process, but suggests that the repair tissue may not have the same histological structure as the original cartilage. Salter (*1981*) however, has provided some evidence of the regeneration and remodelling of articular hyalin cartilage, especially when the joint is subject to continuous movement.

Facet arthrosis syndrome, a related disorder, has been identified and compared to chondromalacia patellae in the knee joint (*Eisenstein & Parry 1987*). This condition is prevalent in young to middle aged adults and is characterised by pain on rest. The joint surfaces show full thickness cartilage necrosis, with some evidence of inflammation but no osteophytosis, differing from the normal arthritic process.

3.2.1 Pain in Osteoarthritis

Pain in arthritis has traditionally been ascribed to chemical or mechanical irritation of the nerve endings within the joint structure. However, complex intra-articular and extra-articular changes give rise to pain through a number of different mechanisms. A confusing clinical picture may arise where osteophytes around the zygapophyseal joints, for example, cause nerve root compression and result in a combination of somatic and radicular pain.

Joint swelling caused by osteophytes can reduce the size of the intervertebral foramen by 2 mm, and roughened joint margins can cause inflammation and fibrosis of the nerve root sheath, tethering the nerve within the foramen (*Sunderland 1978*). Bogduk (*1993*) suggests that there may be an additional element of central pain, similar to phantom limb pain or post herpetic neuralgia, where the neurones within the central nervous system lose their afferent input and undergo physical and physiological changes because of local joint damage.

3.3 THE INTERVERTEBRAL DISC AS A SOURCE OF PAIN

Primary disc pain occurs when the nerves in the annulus are stimulated. This can be caused by mechanical irritation as in inflammation following trauma or by chemical changes occurring as a result of degeneration (*Bogduk 1987*). If only the nucleus and inner annular fibres are damaged, the process may be painless because of the pattern of innervation of the disc. Pain is more likely to be experienced where the outer half of the annulus is affected. The anomalies seen frequently in clinical practice where the patient with severely degenerated discs does not report pain, yet the patient with a small annular tear gives a history of severe pain add credence to this theory. Twomey and Taylor (*1994*) also suggest, that as a result of trauma, annular fibres undergo metaplasia and form bulges within the annulus which may mimic disc herniation and result in primary disc pain.

3.3.1 Disc Degeneration

The intervertebral joint is a fibrocartilaginous joint and is subject to different disease processes to the zygapophyseal joint. Degeneration tends to be linked either to a history of stress and microtrauma or at the opposite extreme, disuse and stress deprivation (*Videman et al 1990*). Evidence to support the former is found in the pattern of degeneration in the lumbar spine. The lower lumbar levels which are subject to high stresses show greater degenerative changes than the upper lumbar discs (*Butler et al 1990*). Although many people exhibit disc degeneration before they are 20 years with an increasing incidence approaching middle age, only a proportion will develop related symptoms (*Miller et al 1988*). This may depend on the area of the disc that is affected.

During the degenerative process, the water content of the nucleus reduces and the structure becomes fibrotic. The division between the nucleus and annulus becomes less distinct and fissures form in the cartilage end plates (*Natarajan et al 1994*). In the annulus, degenerative changes are most noticeable postero-laterally where small radial or circumferential tears may develop, reflecting the poor metabolic supply and capacity for repair within this part of the disc (*Nachemson 1976*). In the clinical situation degeneration is classified by the degree of height loss and extent of osteophyte formation on the margins of the disc (*Miller et al 1988*). In the latter stages of degeneration where the lamellar structure has disintegrated, only the outer collagenous layer of the annulus, measuring about 1 mm thick, contains the disc fragments and prevents them encroaching on the vertebral or intervertebral foramina (*Rauschning 1993*).

Miller et al (*1988*) conducted a meta-analysis of 16 published reports on disc degeneration. They found that male discs were significantly more degenerated than female in most age groups. Butler et al (*1990*) however, found that disc degeneration affected both sexes equally. This discrepancy may arise from differences in the characteristics of the populations studied. In the former study, the use of meta-analysis techniques may have masked some information about the characteristics of the population and made inaccurate assumptions about the data.

3.3.2 Biomechanical considerations in disc degeneration

Miller et al (*1988*) demonstrate that injury and degeneration have an adverse effect on the mechanical properties of the disc, yet Taylor and Twomey (*1987*) report that moderately degenerated discs show similar mechanical behaviour to intact discs. Creep deformation occurs slowly in the normal disc, yet with degeneration the disc deforms more quickly increasing the stresses on the facets. If the annulus or end plate are injured, the disc loses its stiffness and its ability to absorb shock. It is also less able to distribute forces evenly and restore its shape after deformation. This in turn reduces the overall strength and mechanical integrity of the FSU.

Injury and degeneration which result in disc narrowing reduce the stabilising effects of the ligaments. The range of movement within the neutral zone consequently increases (*White & Panjabi 1990, Stokes 1988*). This finding led Burton and Tillotson (*1989*) to propose that subjects with low back trouble (LBT), who were likely to have early degenerative changes, would have a greater range of movement than those without. They studied a group of subjects aged between 16 and 65 years. Older subjects were excluded to reduce the risk of incorporating people with severe degeneration, where osteophytes and disc fibrosis were likely to have restablised the spine. The results, however, were contrary to those expected and subjects with LBT were found to have less movement than those without LBT. The authors conclude that the relationship between mobility and back pain is complex and is subject to a variety of influences. In future studies it would be interesting to evaluate whether subjects with LBT demonstrate protective spasm or avoid pain provoking movements, which could mask intersegmental hypermobility.

3.3.3 Problems associated with disc degeneration

Apart from the changes in shape of the intervertebral foramen caused by degeneration, the value of disc bulging which occurs on compression and lateral flexion of the spine is reported to have increased two fold when the disc is degenerated (*Bogduk & Twomey 1991, Stephens et al 1991*). In addition, where there is a loss of disc height as a result of degeneration, the ligamentum flavum has been found to gather up (*Rauschning 1993*). These factors mean that the foramen could be encroached substantially. Apart from mechanical irritation of the nerve roots, there is some evidence, although not conclusive, that the degenerated disc leaks inflammatory materials into the surrounding environment, causing chemical irritation of the nerve roots (*Hasue 1993*).

3.3.3.1 Disc bulge and herniation

Bogduk and Twomey (*1991*) report that between 5% and 30% of patients presenting with back pain will have a herniated disc. However, in order for herniation to occur the annular wall must first be damaged and the structure of the nucleus altered, reducing its cohesive properties. Herniation is most prevalent in people in their early twenties, after which the nucleus loses fluid and is unlikely to be extruded through an annular fissure. The exception is when the disc has undergone liquefaction through seepage of repair material from the vertebral body (*Twomey & Taylor 1994*)

It has been shown experimentally that the disc bulges maximally in posterolateral and lateral directions especially when subjected to compressive loads combined with a degree of lateral flexion (*White & Panjabi 1990*). Similarly, in rotation, when combined with flexion, the postero-lateral section of the disc is subject to high torsional stresses and susceptible to annular tears. This is a direct result of the distance of the annulus from the axis of rotation (*Bogduk & Twomey 1991*).

The structure of the disc can be damaged by short duration, high level loading which occurs during a sudden violent movement. It can also be damaged by long duration, low level repetitive loading which occurs when postures are sustained over a period of time, due to fatigue failure of the fibres. Once injured, the disc becomes susceptible to further injury, each insult causing damage to additional layers of the annulus, thus developing radial fissures (*Farfan 1980*). In disc herniation, a fragment of the nucleus may be displaced towards and protrude through the damaged annular wall. In addition, where the stresses are sustained, fragments of the annulus can become detached and track through the fissure, irritating either the outer layers of the annulus or the contents of the neural canal. It is also possible that a section of nuclear material, extruded through a ruptured annulus, may trap the spinal nerve within the intervertebral foramen or cause traction on the dural sleeve (*Taylor & Twomey 1987*). It is important to note that damage to the annulus may occur in conjunction with subchondral fractures of the facets of the zygapophyseal joint or capsular injuries (*Farfan 1980*).

The results of recent research suggest that in the lower lumbar spine, the discs normally bulge into the vertebral canal (*Rauschning 1993*) and that disc bulging and herniation has received unwarranted attention in the past. It is also surprisingly difficult to cause the nucleus to prolapse experimentally (*White & Panjabi 1990*). Nachemson (*1976*) reports that 95% of positive myelograms were shown to be correct at surgery, which is contrary to the evidence of White and Panjabi (*1990*) who report a significant proportion of false positive diagnoses of disc herniation in asymptomatic patients with myelography and discography. In the former study, patients may have undergone other diagnostic procedures prior to myelography which may have minimised the inclusion of subjects with healthy discs. It has recently been suggested that because of positive intradiscal pressure, any fluid injected into the nucleus will be partially expelled back into the syringe (*Twomey & Taylor 1994*). This suggests that the dye injected during discography may track through the needle hole on withdrawal, creating an appearance of herniation. Thus the diagnostician faces the problem of identifying whether an annular bulge is pathological and when positive findings on myelography are relevant.

3.4 THE LOCKED BACK

The cause of the "locked back" remains an enigma. Early theories suggested that incongruities in the surface of the articular cartilage of the zygapophyseal joints impaired movement but were unlikely to cause locking unless extraneous material was trapped between the joint surfaces (*Farfan 1980*). Taylor and Twomey (*1987*) found wedge shaped pieces of articular cartilage which had sheared off the joint surfaces, leaving areas of repairing tissue underneath. These were reported to be stronger than the synovial lined fibro-adipose pads and were also attached to the capsule, thus able to transmit traction. The articular cartilage has not been shown to be innervated, therefore pain was thought to arise from the synovial membrane or its fibrous layer (*Hesse 1988*).

This theory has been extended to include the meniscoid inclusions (see Section 1.7). The results of injury have been compared to those affecting the menisci of the knee where a portion of the tissue becomes trapped causing the joint to lock. Where a fragment is trapped, pain may arise from direct stimulation of nerve endings within the meniscus. Alternatively, a trapped fragment may transmit traction to the innervated capsule to which it is attached. However, the evidence of Engel and Bogduk (*1982*) suggests that the fibrous rims are too short to become caught between the joint surfaces; and that where the apex of a fibro-adipose structure is trapped, it is more likely to rupture leaving an intrarticular fragment rather than transmit any appreciable force to the capsule.

3.5 END PLATE FRACTURES

The importance attached to disc herniation has been superseded in recent years by the recognition of the contribution of end plate fractures. Under compressive stress, the vertebral end plates fail before the annulus, even if the disc is already damaged. This is substantiated by the results of an analysis using a finite element model of the disc (*Natarajan et al 1994*). This analysis demonstrated that the end plate was always the weak link in the body-disc-body unit. Unfortunately, it is not possible to assess how accurately the model represents the complex behaviour of the FSU, especially when many major structures were omitted from the model.

The endplates themselves are not innervated and so are unlikely to cause pain directly (*Bogduk & Twomey 1991*). Experimental evidence suggests that following injury to the end plate, the nucleus is invaded by inflammatory repair materials from the adjacent cancellous bone (*Twomey & Taylor 1994*). Exposure of the avascular nucleus to alien chemicals results in degradation, which may in turn cause annular erosion. This is thought to occur mainly along existing radial fissures, with pain and possible herniation of nuclear material.

3.6 THE SPINAL LIGAMENTS

Damage of the Intervertebral ligaments has been shown to cause pain. Disruption of ligamentous integrity can occur as a result of many of the pathologies previously discussed. However, ligaments are often the principle site of injury in spinal trauma. The interspinous ligament of normal subjects has frequently been found to be degenerated at post-mortem examination, presumed to be the result of repeated minor trauma (*Bogduk & Twomey 1991*). In incidences associated with road traffic accidents, Twomey (*1991*) reports that most injuries to the lumbar spine involved the soft tissues, including the capsule, ligamentum flavum and articular cartilage, although in one third of cases the articular processes or subchondral bone also showed signs of fracture. Unfortunately, conventional radiological assessments do not detect ligamentous injury, making differential diagnosis difficult.

3.7 THE EFFECTS OF AGING

Changes that result from the aging process may be difficult to distinguish from those of degeneration. In the future the distinction between the two may prove to be semantic. A study by Butler et al (*1990*) of a population with back pain suggested that degeneration and facet osteoarthritis were linearly related to the age of the subject. Yet in one study, 72% of an elderly sample appeared to have escaped the degenerative process, reaching old age with their discs intact (*Twomey & Taylor 1987*). Degeneration tends to result in a loss of disc height (*Miller et al 1988*), yet although the aging process leads to loss of stature, there is no loss of disc height (*Twomey & Taylor 1994*). This is partially explained by the increased concavity in the superior and inferior vertebral end plates. It is the result of a reduction in bone density, weakening the horizontal trabeculae in the vertebral bodies and causing the vertical trabeculae to buckle. The end plates subsequently collapse allowing the disc to balloon into the vertebral bodies. This increases the height of the disc centrally, whilst reducing its height at the margins.

Contrary to conventional theories only a small minority will demonstrate disc thinning (eg: *Adams & Hutton 1983*) and in these cases it is probable that the disc has suffered previous herniation or dehydration (*Twomey & Taylor 1987*). In the past it has been reported that the nucleus dehydrated with a reduction in proteoglycan concentration, but this has now been found to be minimal after adolescence (*Twomey & Taylor 1987*). There is a gradual increase in the amount and change in structure of collagen in the nucleus. This results in a loss of the distinctive boundary between the nucleus and annulus. The annulus has been found to fray and split, with loss of individual collagen and elastic fibres (*Bernick et al 1991*). However, the stiffer nucleus resulting from a change in its biochemical constitution, means that it is less likely to prolapse (*Twomey & Taylor 1987*). Age markedly influences the biomechanical properties of the disc, and indeed all ligaments, rendering them less compliant and less able to recover from deformation (*Kasra 1992*). Many of these changes result from repeated minor trauma. The formation of osteophytes around the margins of the vertebral bodies is linked to the stresses imposed from the attachment of the annulus (*Videman et al 1990*). There is a thickening of the vertebral body, and an increasing incidence of microfractures of the subchondral bone. Also, the cartilaginous end plates become calcified and replaced with bone (*Bernick et al 1991*)

The zygapophyseal joints are affected by age related changes. The subchondral bone plate becomes thinner after the age of 50 years, but there is some sclerosis in the anteromedial third of the joint, related to the stresses imposed during flexion.

Occasionally the subchondral bone plates collapse which may result in an increase in the concavity of the superior facet. Similarly, there is chondrocyte hypertrophy and cartilage fibrillation in this area, resembling the changes in the osteoarthritic joint. Articular cartilage loss is most marked around the joint margins. However, posteriorly the cartilage appears to flake as a result of shear stresses imposed by the attachment of the capsule directly to the articular cartilage, and multifidus acting through the capsule (*Taylor and Twomey 1986*). Osteophytes develop at the attachments of the capsule, particularly posteriorly, and of the ligamentum flavum. In joints with severe cartilage damage, the meniscoid inclusions have been noted to increase in size, extending further into the joint, to prevent bone contacting bone. They also show osteoarthritic changes (*Taylor & Twomey 1987*).

In adolescence, the spinal canal is oval at the level of L1, becoming triangular by L5. However, age changes it to a trefoil shape due to osteophytic outgrowths on the anterior margins of the zygapophyseal joints. This also tends to push the ligamentum flavum forwards, narrowing the spinal canal (*Taylor & Twomey 1987*).

Although it is often accepted that a loss of movement is inevitable in old age, Twomey (*1991*) remarks that it is more the result of disuse. And while degeneration and osteoarthritis may not be caused by increasing years, their prevalence grows with age. With the loss of extensibility of collagenous vertebral tissue and reduced facet joint excursion caused by degenerative changes, the range of movement tends to reduce. Fitzgerald et al (*1983*) documented a reduction in range of physiological movements of the spine with increasing age in 172 volunteers and recommend that range should be assessed in light of the subjects age. Similarly, in a cadaveric study of 200 lumbar spines, Twomey (*1979*) demonstrated some loss of movement in all physiological ranges. Yet the main loss occurred before the age of twenty. Jull (*1987*) investigated the nature of change of segmental movement in an aging painfree population. The results indicated a slight reduction in movement with age, although many of the 200 subjects remained within normal or slightly hypomobile limits. However, there were only a small number of categories of mobility into which the subject could be placed making the study insensitive to small differences. In this study, age accounted for 72% of the variance of range of movement in the lumbar spine and gender accounted for 11%, females being slightly more mobile than males.

3.8 PATHOLOGY AND TREATMENT

The diverse nature of back pain, in conjunction with the complex structure of the FSU has hindered the development of an accurate classification system for patients with back pain. This is compounded by a dearth of unequivocal diagnostic tests. The implications that this has for the treatment and provision of resources are discussed by Burton and Tillotson (*1991*).

In many cases, patients referred for conservative management of spinal pain, have underlying arthritic or degenerative changes which have been aggravated by strains or minor trauma (*Wyke 1982*). The symptoms may be further confused because the effects of the primary disease process may be modified by secondary processes like inflammation and vascular congestion. These patients can present a confusing clinical picture. In order to develop a logical protocol, treatment is directed towards groups of symptoms, or syndromes rather than addressing the disease process itself. This will be discussed further in Chapter 4.

Chapter 4: The Treatment of Low Back Pain

The self-limiting nature of back pain means that 90% of patients will recover without intervention within six weeks of onset (Waddell 1987). Indeed, a recent study of back pain referrals in primary care indicated that the majority of subjects will recover within two weeks (*Coste et al 1994*).

Only 1% to 4% of those who seek medical advice have a clearly identifiable problem suitable for surgical intervention, the remainder are prescribed some form of conservative treatment (*Dimaggio & Mooney 1987a*). In the first half of this century patients whose X-rays exhibited visible defects like fractures were treated by the surgeon and those without were referred to the physiotherapist (*Cyriax 1983*). Although over the past fifty years many methods of treatment have been added to the battery of techniques available to the physical therapist, few have been rigorously evaluated. This has resulted in a multiplicity of treatments following varied schools of thought, which change as quickly as the proponents promote their ideas.

4.1 PASSIVE SPINAL TECHNIQUES

Practised by surgeons, physicians, physiotherapists, osteopaths and chiropractors, manipulation encompasses many techniques, often lacking clear definition. The enigma that surrounds this term reflects its early use by the bonesetters. In 1930 Heyman wrote: "manipulation has been largely in the hands of the various cults or irregular practitioners. however (sic), if there were not something good in this it would not continue to exist". Then as now, skills were passed from practitioner to practitioner without being objectively quantified.

Bristow Rowley (*1926*) suggested that manipulation affected the internal derangement of joints and freed adhesions about the capsule or muscle. In 1980 similar theories were proposed and Farfan remarked "No-one has actually seen or otherwise ascertained what actually occurs with a manipulation. Evidence for the most part is circumstantial and often there is no evidence at all". Deyo (*1983*) reviewed 57 articles and found no clear rationale for the use of manipulation.

Similarly, there seems to be little to identify the type of patient most likely to benefit. In one study of 1800 patients with low back pain only 5% were selected as appropriate for manipulation. The remainder were excluded either because manipulation was contra-indicated, other forms of treatment were appropriate or because of other psycho-social criteria e.g.: drug abuse, pending litigation (*Hoehler et al 1981*). However, there was no evidence that the selected subjects would have responded differently to many of the unselected subjects.

For the purposes of this work, the definitions proposed by Maitland (*1986*) will be used, although the basic manoeuvres discussed are analogous to those of other doctrines. Maitland categorises manipulation into the two techniques of mobilisation and manipulative thrust. The former is a passive, rhythmical oscillation of the affected part which can be resisted by the patient at any time and is performed within or at the limit of joint range. This technique forms the focus of the study. The latter is a sudden thrust, which can be gentle, but cannot be prevented by the patient due to its rapid application.

Passive spinal mobilisations are widely used by physiotherapists in the assessment and treatment of patients with spinal pain, stiffness or muscle spasm resulting from the spinal disorders discussed in Chapter 3. A recent study involving eleven hospitals found that these were the treatment of choice for 72% of back pain patients seen by Physiotherapists *(Meade et al 1990)*. Few clinicians question their effectiveness, yet the supportive evidence for their value is largely anecdotal.

4.1.1 Philosophy

The "Maitland concept" centres around the patient's pain response to joint movement, to anomalies in movement and soft tissue abnormalities. During the assessment, structures are selectively stressed in an attempt to implicate those responsible (*Koury & Scarpelli 1994*). The therapist identifies the comparable sign which is usually the movement or manoeuvre which can reproduce the presenting symptom. This is repeated throughout the treatment session and at subsequent treatments to determine the effectiveness of each technique.

Many pathologies causing back pain present similar clinical pictures and conversely, a single disease can occur in many different forms. This makes it difficult to identify the individual structure responsible for the pain (*Brewerton 1986*). Theorists have developed strategies to categorise or code back pain patients including McKenzie (*1981*) and Cyriax (*1983*) and are discussed in Section 4.2. Maitland approaches the dilemma by considering the problem in two compartments (Figure 4-1). The first contains a theoretical framework and associated reasoning processes. The second contains the clinical manifestations of the disorder:

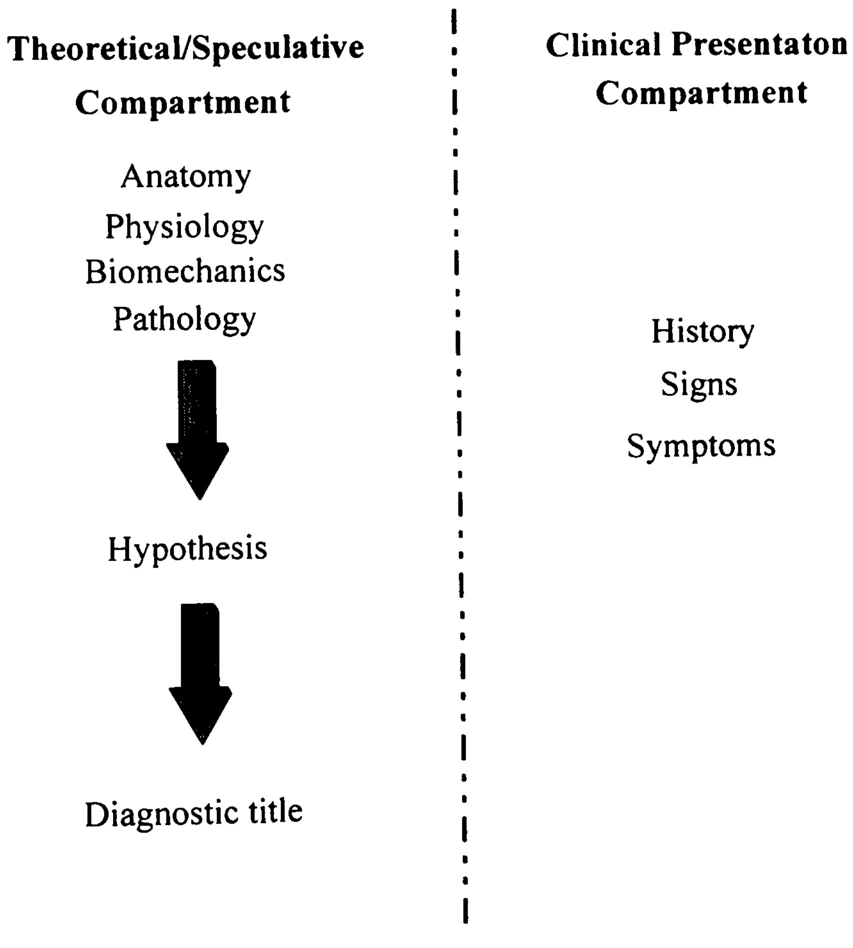


Figure 4-1: Diagrammatic representation of compartment theory under the Maitland concept

The choice of treatment is based on the history, signs and symptoms of the patient represented by the right side of Figure 4-1. The need to give a diagnostic title before beginning treatment is avoided, although a diagnosis may evolve as evidence in the two compartments is established.

Mobilisations can utilise either the physiological or the accessory range of movement of the joint. Physiological movements, as described in Section 2.2, are those which can be performed by the patient and occur during normal activity. They become mobilisations when performed passively by the therapist. Accessory movements are not under direct muscular control of the subject, but their absence is thought to result in limited physiological movement (*Maitland 1986*). They include translation, distraction and compression of the joint surfaces.

4.1.2 The Movement Diagram

Maitland (*1986*) recommends that the most significant sign found on passive movement should be recorded on a movement diagram. A basic model is shown in Figure 4-2. Many permutations of this diagram can be used. However, it has no scientific basis and conceptually is open to misinterpretation, so its value is controversial.

The slope of the line linking R1 and R2 is dependent on the nature of the resistance of the tissues to the displacing force. In a normal joint, R1 and R2 are found at the end of normal range and represent the natural resistance of the surrounding soft tissues. In a diseased joint, resistance may be encountered earlier in range and R1 and R2 are therefore detected earlier in range.

Lee and Evans (*1994*) reviewed the biomechanics of mobilisation and suggested that the point in joint range that the therapist perceives as resistance free corresponds to the toe region of the load-displacement curve (Figure 2.2). Mobilisations aimed at relieving pain are carried out within this range (*Threlkeld 1992*).

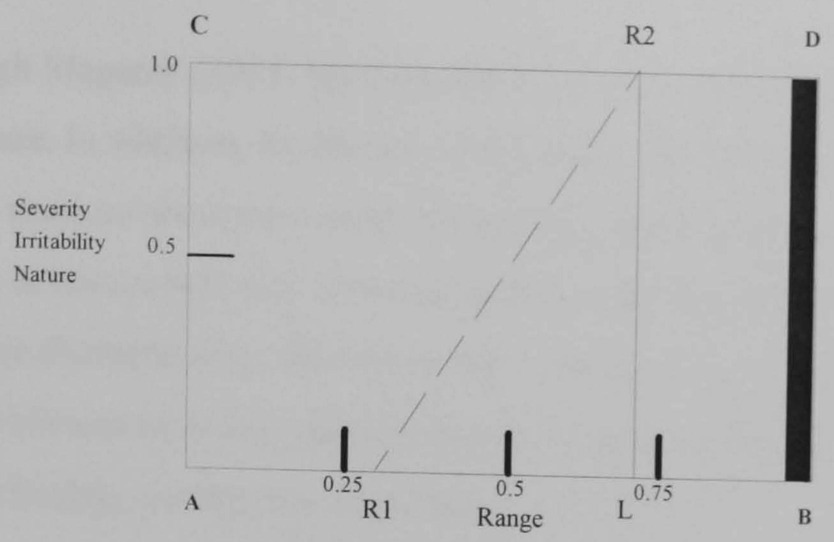


Figure 4-2: Movement diagram used as a framework for recording examination findings

Where:

A	Starting position in joint range	**AC**	Intensity of limiting factor
B	Limit of normal passive range	**C**	maximum intensity of pain that the therapist is prepared to provoke, related to the irritability of the joint or nature of the pain
AB	Range of movement of interest		
R1	point in range where resistance can first be detected	**BD**	soft tissue compliance at end of range
R2	position where resistance limits passive movement	**L**	Limit of available joint range

The point at which resistance is first felt, R1, corresponds to the beginning of the linear region of the load displacement curve, and R2, later in the linear region but well within the failure region of the soft tissues. This opposes the views of White and Panjabi (*1990*) and Bogduk and Twomey (*1991*) who suggest that microdamage to the cross links between the collagen fibrils may occur in the linear region. These authors thus suggest that therapeutic techniques should go no further than just beyond the toe region and certainly stay within the 4% elongation at which microfailure occurs. Farfan (*1980*) reports that passive motion is unlikely to stretch tissues beyond the point of elastic recoil, although Magarey (*1985*) believes that R2 is close to the plastic limit or point of structural failure. In addition, Threlkeld (*1992*) states that a low level of microfailure must occur to produce some permanent elongation. This confusion arises partly because of differences in interpretation of arbitrary points on the load displacement curve and also because no distinction has been made between normal tissue and scar tissue. Treatment should aim to destroy the scar tissue, or improve the mobility of the fibres, whilst leaving healthy connective tissue intact.

Pain behaviour is also assessed using mobilisation techniques and the results are recorded on the movement diagram. The onset of pain (P1) and the point at which pain limits movement (P2) with respect to total joint range are established by palpation of the appropriate spinal segments. Muscle spasm (S1 and S2) can also be recorded.

4.1.3 Choice of Technique

The number of passive, accessory mobilisation techniques available depends on the movements which occur at that joint, the shapes of the articular surfaces and the tension in the surrounding soft tissues. The therapist applies varying forces to palpable parts of the vertebrae. In the lumbar spine, these forces are applied:

1. Postero-anteriorly on the spinous process
2. Postero-anteriorly on the articular pillar or transverse process
3. In a transverse direction on the side of the spinous process

The technique is adapted to suit the examination findings and the movement chosen may be either the movement that relieves, or that which provokes, the symptoms. The choice of technique is governed by:

1. The symmetry of signs and symptoms
2. The structures involved
3. The pathology
4. The disorder
5. Pain, resistance and muscle spasm
6. Range of joint movement

Mobilisations can be carried out with the joint held in one of three positions: a) in mid range b) where the structures surrounding the joint offer least resistance where symptoms begin or increase or c) at the limit of available range.

4.1.4 Grading of Technique

The techniques are quantified by a subjective grading structure, which is consequently open to interpretation. It does however facilitate communication between therapists and is a useful method of recording the treatment given. The grades are selected according to the severity, nature and irritability of the joint, and are based on variations in the range and hence the force used for mobilisation. The grading structure is shown in Table 4-1.

It is generally accepted that Grade I and II are performed before resistance is detected in the joint excursion. Grades III and IV are performed in the resisted range, reaching the same point and differing only in amplitude (*Magarey 1985*). Within each grade, the notation of ++, +, - and -- are used to denote the relative force applied by the therapist to overcome resistance (*Blake 1984*). Grade IV++ would signify that the therapist was reaching the end of available joint range. The notation adds complexity to descriptions already complicated due to their subjective nature.

Table 4-1: Definitions of the Grades of mobilisation under the Maitland philosophy based on Matyas and Bach *(1985)*, Maitland *(1986)* and Jull *(1988)*

Grade	Description
I	Small amplitude movement, at the beginning of range, free of resistance
II	Large amplitude movement at any point in range but before resistance is perceived (R1)
III	Large amplitude movement, within the resisted joint range
IV	Small amplitude movement within resisted range, may reach the end of available joint range (R2)

4.1.5 Frequency of Oscillation

The irritability and severity of the patients symptoms will guide the frequency and rate of force application. When pain predominates, a lower frequency of oscillation is recommended, with smooth changes in force. Where treatment aims to increase joint range in the absence of pain, mobilisations are applied at a higher frequency or with greater force to stretch the tissues (*Blake 1984, Magarey 1985*). When a load is applied rapidly, scar tissue is thought to fail at lower loads than normal tissue (*Farfan 1980*) which suggests that stiff joints should be treated aggressively. Maitland (*1986*) advises that the rate of mobilisation can vary between one "oscillation" or cycle in two seconds i.e. 0.5 Hz to 3 Hz. This excludes the sustained stretching techniques. Blake (*1984*) suggests that the rate should be approximately 2 Hz except at the beginning of treatment with a highly irritable condition when a slower rate should be selected. However, Jull and Gibson (*1986*) found that one experienced therapist performed every technique at a rate of 1.8 Hz regardless of grade.

The known viscoelastic properties of the FSU suggest that the rate of loading during mobilisation will affect tissue response. Lee and Evans (*1992*) demonstrated that the slower the mobilisations were carried out, the greater the range that could be achieved. A force of 150 N was applied to one vertebra at two rates equivalent to a total loading period of 0.5 and 30 seconds. Maximum intervertebral displacement was found to differ by approximately 0.5 mm for the two rates, over displacements of 10 -11 mm. Although statistically significant, the confidence intervals of the mean displacements overlapped and the differences reported fell within the error of measurement of 0.7-0.8mm. Cyclic loading was also found to increase the displacement by up to 0.7mm and with sustained loading over a period of 2 minutes the vertebrae were shown to creep by 0.9 - 1.1mm. The viscoelastic response of the lumbar spine during posteroanterior mobilisation reported by Lee and Evans (*1994*) suggests that significant creep and relaxation occur within the first 30 seconds. This raises questions about the optimum duration of a set of mobilisations, currently recommended to last 20 seconds, after which the effects are assessed (*Magarey 1985*).

4.1.6 The Effects of Spinal Mobilisation

In healthy joints, mechanical stimulation from normal function maintains the physical properties of the tissues. However, when movement is inhibited either through injury or pain, the tissues deteriorate. Stress deprivation and sustained pressure have profound effects on the nutrition of the disc, articular cartilage and other soft tissues and can lead to early joint degeneration (*Bland 1993, Twomey 1991*). Immobilisation also results in soft tissue contractures, articular adhesions and protective muscle spasm (*Haldeman 1986*).

4.1.6.1 Pain

The experience of pain can be considered on four levels (*Bogduk 1993*):

- Perception of tissue or nerve damage
- Transmission (peripheral nerve)
- Transmission (spinal cord)
- Modulation (chemical, neural)

Mobilisations are thought to influence the cause of the pain as in disc prolapse, to stimulate neural mechanisms which interfere with the transmission of impulses, or may stimulate the release of modulating chemicals (*Wright 1995, Haldeman 1986, Bland 1993*). Bogduk (*1993*) suggests that physical therapy is ineffective in the treatment of radicular pain resulting from causes like impingement of the spinal nerve or disc prolapse, but may be used to treat the underlying cause.

Pain is treated using accessory movements which may be of a small or large amplitude depending on the available pain-free range. As the condition improves, the grade of mobilisation is increased to provoke some discomfort. No rationale is offered for this progression.

When pain originates in the zygapophyseal joints, treatment aims to distract the joint surfaces. This is thought to relieve pressure on a trapped synovial fringe, meniscoid inclusion or other loose body within the joint (*Zusman 1986*). There is also some evidence, although by no means conclusive, of a reduction of intrarticular pressure during passive mobilisations (*Giovanelli-Blacker et al 1985*). Changes in intrarticular pressure are likely to affect fluid egress, pain, or movement of the meniscoid inclusions within the joint.

4.1.6.2 Range of Joint Movement

Disease or injury often results in disordered movement which may be increased or, more commonly, reduced, or qualitatively affected. Where one component of the FSU is affected, there will often be secondary contracture and fibrosis of the surrounding soft tissues. Once tissue has become traumatised the repair process will lead to an inextensible scar within normal extensible tissue. Cross links in the collageneous tissues prevent normal gliding of fibres within the ligament and tendon bundles and between laminae of the disc. Mobilisations aim to influence this as discussed in Section 4.1.2.

It was not apparent whether the increase of 10-11 mm in PA mobility reported by Lee and Evans (1992), following the mechanical application of forces of 150 N, was due to the normal viscoelasticity of the tissues or to more permanent changes. In addition, the displacement measured reflected movement at the skin surface, and thus does not accurately represent a measure of vertebral movement.

The proponents of mobilisation advocate the use of powerful stretching mobilisations (Grades III and IV) alternating between accessory and physiological movements (*Blake 1984, Magarey 1985*) in the treatment of resistance. Whilst this appears logical, there are no methods of objectively quantifying these techniques or their effects.

4.1.6.3 Muscle Spasm

The type of mobilisation selected will depend on whether muscle spasm is present. If spasm limits movement and is present over part of the joint range, a sustained grade IV is recommended to stretch the capsule and muscles and achieve a reflex muscle relaxation. Where spasm is invoked by movement, a smooth mobilisation through range, or a technique performed in the range of movement which does not elicit spasm is recommended (*Maitland 1986*). Some authors suggest that the muscles are kept in a slightly shortened position to decrease neural output and reduce the likelihood of stimulating a reflex contraction (*Farfan 1980*).

Mobilisations are reported to have additional effects which are summarised in Table 4-2. However, although these phenomena have been demonstrated during gross spinal movements, the effects of the small movements occurring during spinal mobilisation have not been evaluated.

Table 4-2: Additional effects of Spinal Mobilisation. Based on Frank et al *(1984)*, Radin & Paul *(1972)*, Grieve *(1992)*, Twomey *(1992)*, Holm & Nachemson *(1983)*

Structure	**Effect**
Zygapophyseal Joint	Increases synovial fluid which reduces frictional resistance and improves synovial and articular cartilage nutrition. Shear stress stimulates cartilage formation
Intervertebral disc	Fluctuations in intradiscal pressure increase diffusion of nutrients from endplates
Connective tissues	Increased lubrication between tissue bundles Enhanced quality of repair by influencing the orientation of the fibres within the scar tissue
General effects	Changes in vascularity and tissue metabolism Psychological effects

4.1.7 Risks associated with passive movement

Passive mobilisation of unprotected joints has associated risks. Vigorous mobilisations may cause tissue microtrauma resulting in further inflammation and scar tissue (*Bland 1993*). Where joint effusions occur, the rise in intrarticular pressure can have deleterious effects including compression of the blood vessels supplying the tissues, leading to hypoxia and tissue necrosis. It is also possible that in the spine, where there are multiple articulations, adjacent joints may be subject to strain and be inappropriately mobilised.

These risks can not be minimised until mobilisation parameters, including the magnitude and rate of force application, have been objectively defined, and the boundary between healthy and traumatic stress established.

4.1.8 Clinical Trials

Because of the ambiguity between mobilisation and manipulation techniques in reports of clinical trials, it is often difficult to establish the procedure under evaluation. Many studies of manipulative techniques have also been criticised on the grounds of poor experimental design (*Deyo 1983*). Koes et al (*1991b*) evaluated 35 randomised, controlled trials on manipulation and mobilisation. Although they judged all the research to be of relatively poor quality, their own methods of meta-analysis were inappropriate. All the trials which broadly referred to manipulation or mobilisation were assessed collectively. The results were variable, with some trials sometimes favouring the manipulative approach. However, the reference treatments which were favoured by others, were variable in nature and were used under different protocols.

Jayson et al (*1981*) studied mobilisation and manipulation, as defined by Maitland, in two groups of patients. One group of 94 patients were referred from their GP and another equal group from a hospital consultant. The main difference between the two groups was the greater length of time since onset of back pain in the latter group. The two groups were further sub-divided into treatment and control groups. One experienced therapist gave all the treatments and it was left to their judgement whether passive, physiological, accessory movements, traction or manipulation were felt appropriate. All four groups improved over a four week period, with the treated GP group demonstrating the greatest improvement. By one year, when assessed by postal questionnaire, most differences between the groups had disappeared. However, one of the main assessment tools was a spirit level goniometer which was used to measure gross spinal movement. This method was assumed to be sensitive, reliable and accurate over the range of working conditions. It was also assumed that an improvement in segmental mobility of a few degrees would inevitably lead to a change in gross spinal movement. Both assumptions have to be challenged as would the value of the postal questionnaire and the fact that no assessment of pain was performed.

Coxhead (*1981*) evaluated the use of Maitland manipulation on patients with sciatic symptoms. There was no further information given on the choice of techniques which was left to the discretion of the therapist. The results favoured early intervention, but found the type of treatment to be largely irrelevant when discriminating between traction, exercises, spinal corset or manipulation. The main conclusion was that the more treatment modalities the patient received, the greater the benefit conferred. However, the complexity of the experimental design, which used four experimental treatments in sixteen permutations, carried out in eight different hospitals, limits the usefulness of the results.

A comparison of manipulation with general physiotherapy, placebo and advice given by the GP found the former two to be equal and superior to the latter two treatments in terms of pain, function and perceived effect. However, all groups were equal after twelve weeks (*Koes et al 1992*). The placebos used were de-tuned short-wave and ultrasound therapy. However, the authors did not question the possible therapeutic effects of these modalities. They did however, comment on the lack of sensitivity of their methods of assessment.

Many studies support the view that a short term benefit is conferred from spinal manipulative therapy (*Coxhead 1981, Hoehler et al 1981*). Manipulation is suggested to give short term relief from pain, shorten the duration of acute pain (*Haldeman 1986*) and hasten recovery (*Deyo 1983*). However, few have demonstrated any long term benefits (*Hoehler et al 1981, Jayson et al 1981*). This may reflect a lack of sensitivity of the measures used or poorly designed studies with compounding type II errors.

The results of clinical trials have been inconclusive. This is largely the result of the high rate of spontaneous recovery from low back pain which tends to overwhelm the results of treatment. However, the importance of a control group and sufficient follow-up to detect recurrence, in any trial to determine the efficacy of treatment should be emphasised. In addition, the lack of unambiguous definitions of the manipulative procedures used makes implementation of the results difficult and precludes useful comparative research.

4.2 ALTERNATIVE TREATMENT MODALITIES

4.2.1 Grade V: The Manipulation

The manipulative thrust may be used when improvement with mobilisation has failed. There is some confusion about the position in joint range where a manipulative thrust is applied. Maitland (*1986*) states "a manipulation is similar to a grade IV mobilisation in amplitude and position in the range; it differs only in speed". In the same text, Maitland also remarks that a "manipulation is not necessarily performed at the limit of range", and in a different chapter that it is a "forcing of a movement from the limit of range". The latter description is the one most generally adopted. The operator applies pressure until the structures to be manipulated feel taught, at which point a high velocity thrust of short amplitude is superimposed. Haldeman (*1986*) defines a manipulation as a thrust beyond the elastic region of the joint. It is characterised by a click or dull thud (*McKenzie 1981*), the cause of which leads to some speculation. Bogduk and Twomey (*1990*) report on sharp cracking sounds thought to emanate from the disc as the annular fibres fail. Similarly, during cadaveric work, an audible crack has been associated with failure of the vertebral end plate (*Farfan 1980*).

Because the crack is generally associated with clinical improvement, it is more likely that the explanation offered by Bland (*1993*) and Conway et al (*1993*) provide the answer. Bland (*1993*) describes the normal subatmospheric pressure within the joint which stabilises and maintains the proximity of the joint surfaces. Traction, which occurs during manipulation, results in a further reduction in joint fluid partial pressure causing dissolved gases to come out of solution and create bubbles within the synovial fluid, or cavitation. This is accompanied by distraction of the joint surfaces and a rise to positive intrarticular pressure accompanied by an audible crack. The bubbles remain for approximately 20 minutes before diffusing back into solution. The same phenomena has been seen in the intervertebral disc where radiographs have demonstrated air bubbles within the disc space. However, the relative importance of this sign has not been established. Manipulation is thought to have additional effects which are summarised in Table 4-3.

Table 4-3: Additional effects of Spinal Manipulation. Based on Herzog (*1993*), Haldeman (*1986*), McKenzie (*1981*), Taylor & Twomey (*1987*), Farfan (*1980*), Cyriax (*1983*)

Structure	Effect
Zygapophyseal Joints	Reduction of locking or subluxation, realignment
Meniscoid inclusion or torn cartilage	Freed by distracting joint surfaces
Soft tissues	Adhesions torn, reflex muscle relaxation
Nerve root	Freed from entrapment
Disc	Loose material tracking through annular fissure induced back to central disc
General	Restore spinal motion, reduction in stiffness, stimulation of release of anti-inflamatory agents

Many of the proposed mechanisms remain unsubstantiated and terms like re-alignment have become outmoded (*Grieve 1992, Twomey 1992*). Lee and Latimer (*1993*) used a simulator to apply a load of 200 N to the spinous process, whilst measuring the displacement of the same vertebrae, to evaluate changes in posterior-anterior stiffness following a manipulative procedure. The authors used normal subjects, and measured total spinal displacement rather than relative intervertebral displacement. It is perhaps unsurprising that they found manipulation to have no effect.

A damming indictment of manipulative techniques was given by Gibson et al (*1985*) who compared osteopathic manipulation with short-wave diathermy (SWD) and a placebo treatment of de-tuned SWD. Neither treatment group improved beyond the placebo and for some variables the placebo group demonstrated the greatest improvement. These results led the author to suggest that what is important about treatment is regular contact with the therapist and the placebo effect when harmless treatments are applied with conviction.

4.2.2 Cyriax

The Cyriax philosophy suggests that the majority of spinal problems are due to intrarticular derangements of the disc (*Cyriax 1983*). This may take the form of either an annular cartilaginous fragment which Cyriax considers a manipulative lesion, or a herniation of the nucleus pulposus treated by traction or epidural injection. Treatment techniques are not localised to a specific joint, but Cyriax states that the "strain is borne by the blocked joint". A manipulation consists of a combination of manual traction, a passive movement and then over-pressure applied by the therapist (*Hutson 1990*).

4.2.3 McKenzie

The early McKenzie philosophy was based on principles advocated by Cyriax. There is an underlying assumption that 95% of patients with lumbar back pain have disc problems (*McKenzie 1981*). McKenzie favours the approach that by altering their movement or position, the patient can treat their pain without therapist intervention. It remains only to identify which movement or posture is appropriate for that patient and a customised set of progressive exercises are developed for that patient based on flexion, extension and sidegliding, combined rotation and sideflexion. Additional mobilisation or manipulation is advocated only when the previous approaches have failed. McKenzie identifies three different syndromes causing mechanical pain:

1. Postural syndrome resulting from the application of stress during prolonged loading of structures. This does not involve a pathology and is treated by postural correction

2. Dysfunction syndrome resulting from adaptive shortening of structures following trauma or lack of movement into full range. Pain is caused when these tissues are stressed. Treatment is based on stretching

3. Derangement syndrome resulting from damage to the disc from a radial fissure, annular bulge or nuclear protrusion. The size of the derangement is estimated from the distance of pain radiation

This approach relies on categorisation of patients into the three groups with no alternatives for patients who cannot be clearly categorised. A study of 49 therapists who categorised 363 patients with low back pain demonstrated poor intertherapist reliability, even when specialist training had been received (*Riddle & Rothstein 1993*). Research evidence is scarce although it has been reported that the McKenzie approach is more effective than traction and back school in reducing pain and increasing function (*Dimaggio 1987b*).

4.2.4 Exercises

Exercise programmes generally aim to strengthen the surrounding musculature to protect and stabilise the spine. Programmes tend to be classified by the components of spinal movement, flexion or extension regimes for example, but there is little evidence of a more effective programme. Koes et al (*1991a*) evaluated sixteen randomised, controlled trials carried out mainly by doctors, evaluating exercise therapy. From a possible score of 100 for methodological quality, only four scored over 50 (range 24-61). The lack of scientific scrutiny in these trials indicates that any conclusions about treatment efficacy is potentially inaccurate. Recently, the importance of exercise programmes aimed at restoring function and improving back fitness has been recognised (Frost et al 1995). These are now recommended to form an integral part of the rehabilitation of patients with low back pain (CSAG 1994).

4.2.5 Mobilisation of the Nervous System

The techniques advocated by Butler (*1991*), termed adverse neural tension (ANT), extend the concepts proposed by Maitland to include assessment and treatment of the neurological components. It is recognised that the nervous system as a whole is capable of a large adaptive range. For example, it will accommodate joint movement without suffering damage as a result of tension or compression. However, injury in one part which results in tethering and scar tissue will have both localised and remote effects. The methods of assessment and treatment are aimed at identifying abnormal tension in the nervous system caused by injury. Techniques like the slump test are used to stress the nerves over several joints in an attempt to mobilise the neural tissues. These techniques have been used by the Osteopathic profession for some years and they have recently gained popularity in other disciplines.

4.2.6 Combined Movements

The techniques advocated by Edwards (*1992*) are an extension of Maitlands concepts. The complexity of the movement patterns of the FSU are recognised. The subject is positioned in combinations of the main axial movements of the spine, thus the spine may be supported in extension and side flexion if this combination is found to be limited in range, whilst passive accessory techniques are applied. The efficacy of these techniques has not been addressed.

4.2.7 Kaltenborne

Kaltenborne suggests an educational approach, teaching patients about back pain, prophylaxis and treatment. Automobilisation exercises are taught which the patient can continue at home with the aid of an instruction sheet. The exercises aim to reduce pain by increasing the proprioceptive input and to stretch tight tissues. Strengthening exercises and postural awareness are also included.

4.2.8 Palliative Treatment

Rest as a form of treatment for back pain is now largely outmoded, mainly due to the adverse effects of periods of immobilisation. Short periods of bed rest are occasionally advocated during the acute phase. Corsets reduce spinal mobility during periods of activity, provide abdominal support and postural correction. They are only recommended for short term use because of the deleterious effects they have on the trunk musculature.

4.2.9 Counter-stimulation

Transcutaneous nerve stimulation and acupuncture are thought to modulate pain impulses by blocking or confusing peripheral nerve transmission. This occurs either by saturating the axons or by creating background noise which confuses the transmission of nociceptive stimuli (*Bogduk 1993*). Research has yet to show a clear benefit for these types of treatment. Other treatment modalities like ultrasound, ice packs, analgesic, muscle relaxant or anti-inflammatory drugs are largely palliative and fall outside the scope of this work.

4.3 DISCUSSION

In 1992, Grieve wrote of mobilisations "we don't really know what we are doing, nor do we know enough about what we are doing this to". Research in this field has been held back by the view that those who attempt to evaluate small groups of techniques will miss the "whole" concept or picture (*Maitland 1986*). Whilst this may be an appropriate criticism of studies which evaluate one technique on a heterogeneous group of patients, it is not true of all studies which partition a section of theory or practice for evaluation. Indeed, this would seem the most appropriate method of study.

A literature search for trends in treatment effectiveness highlights the lack of comparable data. The protocols are often inadequately reported and encompass a diversity of approaches, indicative of the absence of a clearly superior method. Reference treatments are inconsistent, inclusion criteria and outcome measures are not standardised, similarly the subject groups lack clear diagnostic categories. The parameters of all these need closer definition before any valid comparisons can be made. If the effects of mobilisations are to be determined, it is essential that the forces are defined accurately and are assessed in the light of the mechanical properties of relevant biological tissues.

Chapter 5: Therapist Reliability

The use of mobilising techniques relies on the skill of the therapist and their perception of the quality and quantity of joint movement. To select the appropriate treatment, the therapist must first establish the nature of tissue resistance as the joint is moved, the amount of movement of the vertebral body and the behaviour of the pain. However, because of the subjective nature of this assessment, the results are susceptible to differences in interpretation.

The assessment of spinal joints require the following skills:

- Knowledge of what constitutes normal range of movement and its variation, taking into account age, gender, and body type of the individual

- Knowledge of what constitutes normal tissue resistance and its variation

- Perceptual skill in the detection of resistance to movement and other abnormalities

- Motor skill in technique application

The assessment is a complex task and is made more difficult by a lack of objective measurements. Because there are no commonly agreed normative values for the healthy population, the therapist develops an experiential model of what can be considered normal. However, in the clinical situation, the therapist may rarely feel normal movement and memory for this model may be distorted by years of feeling abnormal spinal motion. The results of a study by Gonella et al (1982) substantiate this theory. Although not noted by the authors, the five therapists who subjectively assessed joint movement in normal, young subjects reported a decrease in mobility at the majority of vertebral levels tested.

Over the past decade there have been some attempts to evaluate the reliability and validity of various assessment and treatment techniques, predominantly undertaken by researchers in Australia. However, much of this work has been done by students of manipulative therapy and the majority, which is summarised by Matyas and Bach (1985), remains unpublished.

5.1 THE ASSESSMENT OF RELIABILITY

5.1.1 Repeatability

The consistency with which a therapist can repeat a given technique under the same experimental conditions, also known as test-retest reliability, is important both in the initial assessment of spinal movement and for detecting changes during and following treatment. Training, level of expertise and even fatigue of the operator will all influence this consistency.

5.1.2 Reproducibility

The ability of the therapist to repeat a procedure either at a different time, or under different experimental conditions is termed reproducibility. This term can also be applied to the consistency in technique application between different therapists, which is often referred to as intertherapist reliability.

The parameters given by Maitland's movement diagram described in Section 4.1.2, have been used to evaluate reliability and are discussed below. However, the limitations in the measurement techniques used in these studies, which will be discussed further in Chapter 6, suggest that the results should be interpreted with caution.

5.2 THE PARAMETERS USED TO ASSESS RELIABILITY

5.2.1 Resistance

The ability of the therapist to detect the position in joint range where resistance can first be detected (R1) and the position where it limits passive movement (R2) have been evaluated. Carty (*1986*) evaluated therapist precision in locating R1 and R2 in the joints of asymptomatic subjects. He found test-retest reliability coefficients of 0.23 for R1 and 0.59 for R2 with intertherapist coefficients of 0.20 and 0.53 respectively. This indicates that the limit of joint excursion can be detected with greater reliability than the position of onset of resistance. Although these coefficients are statistically significant, clinically, this reliability is poor.

The study was criticised by Matyas and Bach (*1985*) who highlighted the limited variation resulting from the use of a normal sample and suggested that this would cause the reliability to appear lower than it might otherwise be. However, Matyas and Bach (*1985*) reported on a series of studies using patients and found that reliability coefficients were generally low (Table 5-1) supporting Carty's results.

Reports of poor reliability led Evans et al (*1988*) to examine the therapist's ability to detect changes in resistance. Using a mechanical model of the neck, he found that therapists could perceive relatively small changes in resistance when applying small forces, but their ability was influenced by the gradient of the increase in resistance and the position of the onset of resistance within the range of movement. Maher and Adams (1996) used a series of springs to evaluate therapists ability to discriminate between stiffness stimuli. They demonstrated that therapists were able to discriminate between relatively small differences in stiffness characteristics. However, the stiffness of the springs used of 12-14 N/mm, was lower than that generally associated with mobilisation procedures. It seems unlikely that therapists would be able to detect small changes in resistance with precision whilst they are applying the relatively large forces associated with treatment.

As shown in Figure 2.2, the load displacement curve of the motion segment is non-linear and the rate of increase in resistance will vary depending not only on the rate of load application, but on which components (eg. ligament, capsule or disc) are subject to loading at each point in the range of movement. In addition, the results of many disease processes will have unpredictable effects on joint resistance. The clinical task of distinguishing between normal anatomical resistance and pathological resistance is therefore not adequately simulated in the task reported by Evans.

5.2.2 Pain

The onset of pain (P1) and the point at which pain limits movement (P2) have also been used to evaluate therapist reliability. Matyas and Bach (*1985*) report the results of a series of trials in which students training in manipulation assessed therapist reliability in recording the point in range where the patient reported the onset of pain (P1). They found a test-retest reliability coefficient of 0.73 and intertherapist reliability coefficients of between 0.48 and 0.62. The coefficient used was not specified. Overall, the results of these studies indicate that the assessment of pain characteristics is more reliable than the assessment of resistance. This may be because whilst the detection of resistance requires the perceptual skills of the therapist, the report of pain during the manoeuvre is made directly by the patient.

5.2.3 Tissue Compliance

The assessment of tissue compliance during joint movement is closely linked to the detection of resistance. Research in this area has examined the therapists ability to rate joint movement into one of several categories. Jull evaluated the reliability of one experienced therapist who, using a PA accessory technique graded the lumbar spine joints of 20 subjects on a five point rating scale on two successive days (*Jull & Bullock 1987b*). The scale categories ranged from "hypermobile" to "very stiff". The resulting Pearson's product moment correlation coefficient was 0.98 indicating good reliability. When the results of this therapist were compared with one other experienced therapist the intertherapist correlation coefficient was 0.94. However, the use of only two experienced therapists, the minimum requirement for a study of intertherapist reliability, and the restriction of the rating scale to five broad categories and use of an inappropriate statistical test, may have made the reliability appear artificially high.

In an earlier study, Jull (*1982*) repeated 41 tests of manual examination of the lumbar spine of one subject on three successive days. Again, a 5 point rating scale was used. She reports a reliability coefficient of $r = 0.10$ (significant at 95% level). Jull then repeated this procedure with a second experienced therapist, but this time using three subjects, to estimate intertherapist reliability. She reports that the results "correlated highly" with a coefficient of $r = 0.35$ (significant at 95% level). Whilst this description is statistically correct, the r values of 0.10 and 0.35 cannot be considered reliable in clinical terms. In addition, Pearson's product moment coefficient, which appears to have been used in these studies, is not recommended for use in reliability studies for reasons discussed in Section 5.3.

A further study reported by Matyas and Bach (*1985*) involved three therapists selecting the appropriate level for treatment from five spinal levels of 12 patients. To achieve this, the therapists needed to discriminate between abnormal and normal movement. The average test-retest agreement rate was 47.2% and the intertherapist agreement 26.4% which was little better than a model assuming random agreement which would have yielded 20% agreement. A further study reported by these authors where the therapist was required to make a decision as to which factor (pain, muscle spasm or resistance) was the main cause of limitation to movement, demonstrated that there was only 66.6% agreement between therapists. This figure was only moderately better than the expected random agreement rate of 51.8%.

A study of the ability of chiropractors to detect the presence or absence of movement between lumbar spine segments used an articulated spinal model (*Harvey & Byfield 1991*). The experimenter could fixate each segment independently. However, as each segment was fixed via attachments in the centre of the vertebral body, no translatory glide was available at any segment which would have changed the sensation of normal joint movement regardless of any further fixation. This, in addition to the questionable accuracy of the simulation, may explain the low levels of reliability reported (Kappa 0.34 - 0.36).

5.2.4 Movement diagrams

The accuracy with which a therapist can transpose an assessment of joint characteristics onto paper, and hence the usefulness of the movement diagram has been questioned. Matyas and Bach (*1985*) commented on the difficulty of assessing tissue compliance and producing the associated movement diagrams. Within this paper it was demonstrated that therapists could repeat the force necessary to reach P1 with greater reliability than they could record comparable findings on a visual analogue scale, a simplified form of movement diagram.

Lee and Evans (*1990*) used a mechanical spinal mobiliser to produce load-displacement curves of L4 and adjacent vertebrae, which they equated with the movement diagram. The data produced was reported to be highly repeatable which suggests that human error accounts for the unreliability found in other studies. There has as yet been little published work on the therapist's ability to record the R1 R2 curve. In a brief account, Young (*1986*) reports the results of a study where 540 movement diagram curves were classified according to curve shape. Young concludes that the reliability amongst therapists is poor, when recording the detection and observation of the behaviour of resistance. No data is given to substantiate this finding.

Stoelwinder et al (*1986*) argues that the movement diagram is not purported to be a precise measurement tool, and is used only to facilitate communication between therapists. Yet if it is neither reliable nor accurate then it would appear to have little use even for purposes of communication. However, as a teaching aid it is possible that there are advantages in providing a conceptual model for the nature of joint characteristics.

5.2.5 Force

There have been few assessments of the therapists ability to apply consistent forces during spinal mobilisation. Jull and Gibson (*1986*) found that an experienced therapist could reliably apply a force of consistent magnitude to "displace a vertebral segment to its end of range" (r = 0.93). The mean force applied was 215 N (sd 25 N). The conclusions of this study were based on the results of one therapist, used in many of the trials by this author, over a relatively short time period and may indicate a good memory for motor ability rather than reliability of technique given the perceived feedback.

Gonella et al (*1982*) suggests that the experienced therapist tends to develop idiosyncratic behaviours as a result of their experience and thus may not represent the norm. Gonella's results, which involved assessment of tissue compliance, supported this hypothesis where the best agreement and least variation was found between the two least experienced therapists. Using a myometer, which will be described further in Section 6.1, Grieve and Shirley (*1987*) measured the forces applied to the transducer by two therapists on eleven occasions. Although no analysis is given, the authors reported that the therapists found it difficult to repeatedly apply a consistent force.

Matyas and Bach (*1985*) measured the average peak forces used during mobilisation in a sample of eight experienced therapists using a force platform. They report an intertherapist coefficient of 0.16-0.25 and a test-retest coefficient of 0.22-0.42 which are far lower than the 0.93 reported by Jull. In one set, the forces used by seven post-graduate students of manual therapy, ranged from less than 10 N to over 87 N.

Hardy and Napier (*1991*) used a pressure sensitive platform to measure the differences between the forces applied by therapists using Maitland Mobilisations. They found significant variability both in a test-retest design and between different therapists. Aside from the limitations of their equipment which will be discussed in Section 6.1, they also used inappropriate tests of significance to analyse their results.

Table 5-1: Results of previous studies of the reliability of passive accessory intervertebral movements

Parameter	Principal Author	Statistic	Reliability	
			Test-Retest	Inter-Therapist
Resistance	Carty 1986	Kappa	0.23-0.59	0.2 - 0.53
		r (ICC)	0.09-0.46	0.25-0.38
Pain	Matyas 1985	r (ICC)	0.73	0.48-0.62
	Maher 1994	ICC % agreement		0.67-0.72 31-43
Force Applied:				
Linked to resistance	Jull 1986	Pearson's r		0.93
	Matyas 1985	r (ICC)	0.42	0.16
	Matyas 1985	r (ICC)	0.22	0.25
Linked to pain	Matyas 1985	r (ICC)	0.83	0.75
Tissue Compliance	Jull 1982	Pearsons r	0.10	0.35
	Jull 1987b	Pearsons r	0.98	0.94
Identify Abnormal Level	Harvey 1991	Kappa		0.34
	Maher 1994	ICC % agreement		0.03-0.37 21-29
	Matyas 1985	% agreement	31-47	25-26

Many of the studies assessed reliability by measuring parameters recorded by the therapist on a movement diagram or visual analogue scale. However, no account is given of how much variance can be attributed to recording errors using this method. A more accurate assessment of reliability may be made if the forces used during each grade of mobilisation are assessed, which contain implicit information on parameters like R1 and R2.

5.2.6 Rate

Only one study has been highlighted in the literature which addressed the ability of the therapist to apply a load at a constant rate (*Jull & Gibson 1986*). The rate of mobilisation of one examiner was assessed, whilst examining and grading postero-anterior motion at L4 on 28 subjects. The results showed that the therapist consistently applied loads at a rate of 1.8 Hz.

5.3 STATISTICAL APPROACH

Discrepancies between results can largely be accounted for by the variety of statistical analyses used. Traditionally, the Pearson product moment correlation coefficient has been used as a measure of reliability (*eg: Dillard et al 1991*). However, attention to the derivation of this coefficient demonstrates that it is not appropriate to assess agreement between two sets of scores. Whilst it can be used to express association or covariance between two variables, the coefficient will assume a high value when two sets of measures change together uniformly with no respect to the scaling of the scores. The analysis only takes random error into account and perfect agreement can be shown even if there is substantial systematic bias. Thus one therapist may apply twice the force of a second therapist, yet the correlation will be 1.0, or perfect. Use of r also assumes that one variable is error free. This is not appropriate to reliability studies where both sets of measurements are estimates of a hypothetical true value and will contain some degree of error.

More recently, the Intraclass Correlation Coefficient (ICC or R) has been recommended for reliability studies (*Eliasziw et al 1994, Shrout & Fleiss 1979*). In its simplest form, the ICC expresses the ratio of true score variance to true score variance plus error variance:

$$\text{Thus: ICC} = \frac{\text{True score variance}}{\text{True score variance} + \text{error variance}}$$

The smaller the error variance, the closer the ratio is to unity and the nearer the ICC is to 1.0. However, several factors can influence the result of this calculation. If sufficient samples are taken and the true score variance approximates 0, then even where the two sets of scores are reliable, minimal error will result in a small reliability coefficient. Conversely, provided the variance of the data, or true score variance is sufficiently large, any testing procedure will give the appearance of reliability.

Another approach to reliability has been to test the data obtained, from the two replications, for significance using an analysis of variance or a t-test. This approach is perhaps the most misleading. One would not expect two sets of data which have been obtained under the same conditions, using the same technique to differ significantly. However, failure to reject a null hypothesis between two repeated measurements does not imply that there is agreement between the scores. Ironically, the greater the measurement error, the greater the variance within the sample, the less likely the result is to be significant.

Recently, Generalisability theory has been developed to allow many sources of error to be evaluated within the same statistical model (*Morrow 1989*). This involves an analysis of variance which not only takes into account both random and systematic bias, it also takes error from all the main effects into account. It then partitions the variance into its component parts. One can then determine whether the therapists, the measuring procedure or the subject for example, provided the greatest proportion of variance. Again, it is dependent on the variance of the sample and only provides proportions for comparison rather than absolute values in the unit of measurement.

5.4 DISCUSSION

The reliability with which mobilisation techniques are performed will affect the assessment of tissue compliance. If different forces are used, at different rates, the apparent joint compliance will change. Magarey (*1985*) points out that as the degree of pressure exerted by different therapists varies, so will the range of joint movement found. Conversely, inaccurate assessment of tissue compliance may lead to the choice of inappropriate grades of mobilisation. These factors often frustrate attempts to standardise and quantify the techniques.

There is no standard set of mobilisation procedures and the therapist is encouraged to modify the basic technique to suit the examination findings (*Maitland 1986*). Maitland (*1986*) states that often therapists do not realise how gently a technique can be applied nor how strongly in spite of the patients discomfort. Yet until the grades of mobilisation can be reliably characterised and reproduced, it is not possible to establish the importance of the grading structure nor evaluate the differences in treatment outcome due to different grades of mobilisation.

The majority of research on reliability has been carried out by a small number of people who have used controversial experimental designs. This, in conjunction with the wide range of inappropriate reliability coefficients used, makes it difficult to draw confident conclusions. This is reflected in Table 5-1 which shows a general lack of consensus between studies. In addition, the summary statistics reported do not allow the data to be usefully interpreted in appropriate units of measurement.

As a result of these criticisms, the method recommended by Bland and Altman (*1983, 1986*) and Chinn (*1991*) was used in this study although it is not yet widely adopted in the literature. This technique provides an estimate of the error between sets of data. It is based on a simple parametric model, which can be usefully interpreted in the units of measurement used. It also allows the estimation of both bias and error, the two essential components in the assessment of reliability. This technique will be discussed in greater detail in Chapter 9.

Chapter 6: Force Measurement

Previous attempts at measuring mobilisation and manipulation forces have been limited both in experimental and equipment design. However, three reported methods have been identified whose principles include:

- A rig which simulates a section of the spine, through which the loads applied by the therapist can be measured

- Direct measurement by interposing a measuring device between the therapist and the subject

- Indirect measurement of forces applied by the therapist using a force platform

The potential advantages and disadvantages of these methods are discussed below.

6.1 SPINAL SIMULATION

Evans et al (*1988*) constructed a model of the neck, with a plunger shaped to represent a spinous process. The model allowed direct measurement of the loads applied by the therapist and enabled close control of the experimental conditions. When a postero-anterior (PA) force was applied, the plunger depressed by up to 5mm. Its excursion could be controlled by electronically varying the resistance against the force applied by the therapist. Evans monitored various parameters including the ability of the therapist to detect the onset of resistance. They also assessed whether the resistance gradient and position of resistance in the range of movement influenced its detection. Whilst the conditions could be kept equitable for every test there were obvious limitations to this approach, not least the question of accuracy of the simulation. It can be assumed with some confidence that the sensory feedback given to the therapist would be very different to that experienced during mobilisation in vivo. The small forces recorded which in some instances were less than 0.02 N are not within the normal range of forces used for mobilisation. It is probable that when the therapist is applying small loads, their sensitivity to changes in resistance may be very different to that when applying normal mobilisation forces of up to 200 - 300 N.

In an earlier study, Grieve and Shirley (*1987*) used an "inverted myometer" which consisted of a small displacement transducer composed of a diaphragm attached to a potentiometer, mounted in a measuring head and connected to a digital display. The diaphragm was covered with a selection of sponges to represent superficial and deep tissues (Figure 6-1). The therapist performed the technique directly onto the sponges covering the head of the myometer.

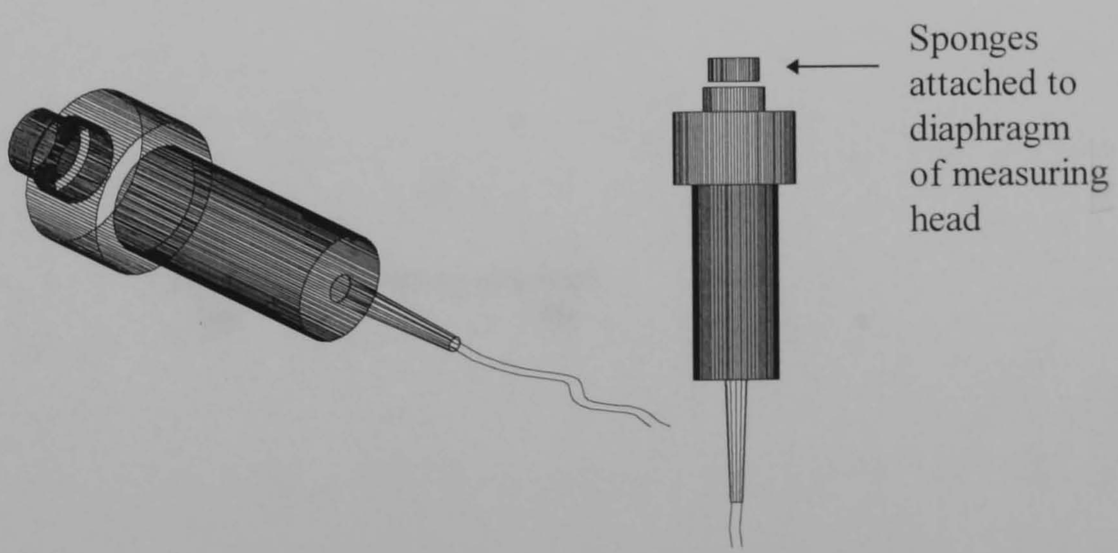


Figure 6-1: The measuring head of the Myometer used by Grieve and Shirley (*1987*)

The study was undertaken to measure forces applied by the therapist during PA oscillatory mobilisation of the model vertebra. The authors reported some success with this technique, although they acknowledged that it did not truly simulate the response of living tissue. A similar device was developed by Hardy and Napier (*1991*) who used a block of rubber to represent human tissues, mounted on a force sensitive platform. Neither study justified their chosen material as being representative in terms of the stiffness of the vertebral segment or on criteria used to decide whether their form of simulation was acceptable. The forces measured demonstrated considerable variation and tended to underestimate those measured in other studies (Table 6-1). This questions the conditions established in the simulated tasks.

6.2 MEASUREMENT OF FORCES APPLIED TO THE PATIENT

Figure 6-2 shows five potential sites where transducers could be located to allow the measurement of forces applied during spinal mobilisation.

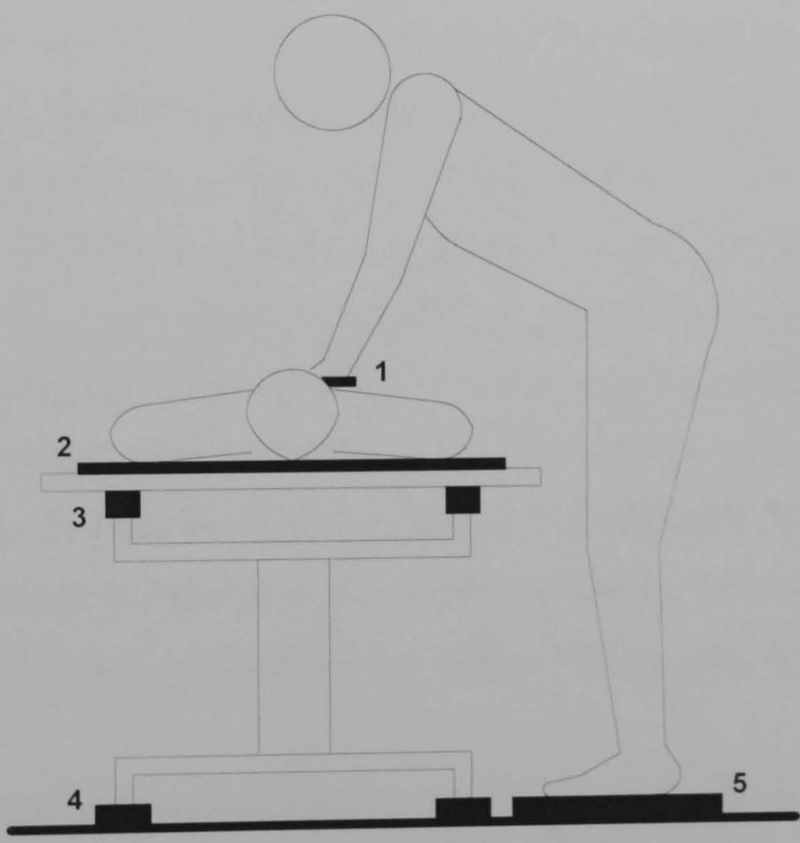


Figure 6-2: Potential transducer locations for measurement of spinal mobilisation forces

1. Direct measurement under the hands of the therapist
2. Force plate under the model
3. Instrumentation of the mobilisation couch
4. Couch standing on transducers
5. Indirect measurement using force platform

With the exception of location 2, which was considered to be too uncomfortable for the model, the merits of each location are discussed below.

6.2.1 Direct Measurement

The Interposition of a measuring device at the point of load application was considered (location 1 in Figure 6-2). Three transducers used in previous applications were assessed for their suitability and are discussed below.

6.2.1.1 Capacitive Transducers

Jull (*1988*) and Jull and Gibson (*1986*) placed a capacitive transducer, originally designed to measure vertical forces transmitted during the gait cycle, under the hands of the therapist. A capacitive transducer uses a compliant material sandwiched between two electrodes. With force, the material is compressed bringing the electrodes closer together which results in a change in capacitance. However, these types of transducer tend to exhibit hysteresis and time related drift, which affects their reliability (*Grant 1985*). The transducers used by Jull (*1988*) needed replacement on several occasions due to breakage, which also indicates that they may not be sufficiently robust for this application. In addition, the heat generated at the interface between the therapist's hand and the soft tissues is likely to affect the performance of the capacitive transducer.

A similar flexible pressure mat was employed by Herzog et al (*1993*) to measure forces during chiropractic manipulation. They claimed that the palpation of bony points of the spine was not hindered by the transducer. However, it should be emphasised that the use of mobilisation and manipulative techniques relies on palpatory feedback and any measuring device which is interposed between the therapist and patient is likely to interfere with the therapists perception of the quality and quantity of joint movement. Whilst the application of mobilising techniques remains highly subjective any interference due to measurement technique should be minimised.

6.2.1.2 Magneto-Resistor Transducers

Tappin et al (*1980*) developed a transducer to measure the horizontal shear components of force under the foot of the subject when wearing different types of footwear. The transducer, which was attached to the sole of the foot, consisted of a magneto-resistor mounted on one stainless steel disc with a magnet mounted on a second. The two discs were cemented together with a layer of silicone rubber. The resulting signal from the magneto-resistor was proportional to the displacement of the magnet. Movement was limited to one direction by a groove and ridge on opposing faces of the transducer discs (*Tappin et al 1980, Pollard et al 1983*).

It would appear that this system is prone to some conflicting influences. The silicone rubber would deform with both tangential loads and vertical loads, the influence of which could not be isolated. In addition there would be some energy loss in overcoming the friction caused by contact of the sides of the ridge with the restraining grove. This friction would increase, the more perpendicular the applied force was to the grove. In addition, although the top surface of the transducer was attached to the sole of the foot, the lower disc did not appear to be fixed and the accuracy of measurement would therefore rely upon the friction between the lower disc's surface and that of the footwear under evaluation.

6.2.1.3 Pneumatic and Electropneumatic Transducers

Pneumatic pressure transducers operate on the principle of displacement of air within a flexible pneumatic chamber. They have often been used to evaluate the effects of externally applied pressure on orthotic performance (*Chase et al 1989*). Pressure can be measured directly with the Oxford pressure monitor or the Denne Gauge which measure the displacement of air within one or more capsules; or over a larger area with the Texas Interface Pressure Evaluator (*Grant 1985, Palmieri et al 1980*). These transducers tend to supply limited information about the forces applied and due to their flexibility, transducers like the Texas monitor are prone to damage.

6.2.2 Indirect Measurement

Matyas and Bach (*1985*) and Petty (1995) report on a series of studies in which the therapist stood on a force platform during manipulation (location 5 in Figure 6-2). In order to calculate the applied forces, the former authors made assumptions about the behaviour of the acceleration of the therapist's body during the procedures. In one method, the therapist was required to "hold" their position for 0.5 to 1.0 second whilst measurements were made. Acceleration was thus assumed to be virtually zero. It is difficult to judge how accurately the therapist could hold their position whilst maintaining a load which would usually be applied dynamically. In addition, during dynamic measurement, no comparison was made of the loads applied at the hand and those at the feet of the therapist. Because the therapist accelerates their torso towards the subject, applying a load against this inertia, the loads seen at the spine are not totally represented by those measured under the feet of the therapist. It is also likely that the loads would have been modified by resonance and inertia of the therapists body.

6.2.3 Further limitations of these approaches

Force application during mobilisation will undoubtedly comprise of components along vertical, longitudinal and horizontal axes. Therefore, any measuring technique should be able to monitor forces along these three axes of movement. Many of the methods described above only provide information on vertical forces or forces applied perpendicular to the transducer.

Thus, in order to measure mobilisation forces, a single device made up of separate transducers would be required to measure forces in three orthogonal planes. This would result in a transducer in excess of 10 mm in depth and approximately 10 mm in diameter. When location 1 in Figure 6-2 is considered, this bulky device would be likely to cause significant distortion of the patient/operator interface and with a relatively rigid construction, would tend to cause discomfort if interposed between patient and therapist. To test this hypothesis, a feasibility study was undertaken to determine whether transducers based on this model could be used. Five therapists were required to apply the techniques with a rigid metal disc, 10 mm in circumference and with a depth of

2 mm. over the subject's spinous process. The therapists reported a loss of sensitivity caused by the disc and the subjects reported discomfort on higher grades of mobilisation.

The alternative of using a force platform was also considered. However, in the studies discussed above, a force platform only provides information on vertical forces and has been criticised because the therapist must alter their normal stance to avoid leaning against the plinth (*Stoelwinder 1986*). In fact, each of these methods causes some disturbance to normal operator practice which explains, in part, the wide variation in forces recorded during similar mobilisation techniques (Table 6-1).

Table 6-1: Forces recorded during spinal mobilisation using different measuring techniques

Study	Mean force (N) Grade of Mobilisation				Comment
	I	II	III	IV	
Grieve & Shirley 1987	7 25	13 37	27 63	28 64	thumb pisiform
Napier & Hardy 1991	1-30	1-38	4-51	8-122	Mean force at each grade
Jull 1986				215	One therapist, End feel
Petty 1995				93	One therapist
Matyas & Bach 1985		2-7		89-329	Intertherapist range
M.Lee 1989	200				Force applied mechanically
R.Lee 1990	150				Force applied by simulator
Evans 1988	< 0.2				Neck simulation

6.2.4 Instrumentation of the Mobilisation Couch

The weight of the subject and any additional forces are transmitted through the supporting couch so methods of measuring these forces by instrumenting the couch were considered. Wheatley et al (*1982*) described a low profile load transducer specifically designed to be placed under the legs of a hospital bed. Barbenel (*1990*) used this system to measure the magnitude and direction of the movement occurring during sleep estimated from the displacement of the centre of mass of the subject. The method was developed to allow identification of patients who were at high risk from developing pressure sores due to lack of nocturnal movement. However, this system gave only limited information on force magnitude or direction.

A method of measuring the displacement of the centre of mass of the body as a result of mass movements caused by the cardiac cycle has been reported (*Manley 1973*). Ballistocardiography initially utilised a technique of suspending a table on long wires whose movement was resisted by stiff springs. More recent designs have used a lightweight table supported on air bearings with guide fins to limit movement to the head-foot plane and monitored by accelerometers. Air bearings have the advantage of being able to support extremely large loads and remain relatively friction free. Although this method provided a sensitive measure of displacement, it was limited to measuring movement along the horizontal axes.

Triano and Schultz (*1990*) instrumented a manipulation table to measure the peak force exerted during a chiropractic manipulation procedure although limited information was made available on the technique used. To enable an accurate evaluation of mobilisation forces, various methods of instrumenting the mobilisation couch were considered. However, it was considered important that the limitations recognised in previous approaches were addressed.

6.3 OBJECTIVES

The main objectives of the programme of research were:

- To design, develop and build a system capable of measuring the characteristics of the forces applied by a therapist during spinal mobilisation on a range of subject groups, and to calibrate and characterise the behaviour of the system

- To use this system to evaluate the reproducibility and repeatability of the forces applied by a therapist and to establish the variation in the characteristics of the mobilisation force performed by different experienced therapists

- To define the characteristics of the mobilisation force used on a normal lumbar spine and to examine the effects of age related changes in the lumbar spine

Chapter 7: Development of an Instrumented Mobilisation Couch

7.1 DESIGN SPECIFICATION

The measuring system had to allow determination of several parameters without disrupting the mode of operation of the therapist or causing any deformation or interference at the patient/therapist interface. These included:-

1. the magnitude of the applied force (F_{total})
2. the component forces (F_x F_y F_z)
3. the amplitude of the mobilisation force
4. the temporal variation in force application

The measuring system was also required to:

1. Support a static load of 150 to 180 kg
2. Remain sensitive to mobilisation forces estimated to range between 6.9 N to 300 N
3. Exhibit natural frequencies outside the range of mobilisation frequencies of 1 - 3 Hz
4. Minimise the separation between the point of force application and the plane of measurement

It was originally intended to mount one unit comprising of three orthogonal transducers, corresponding to *X*, *Y* and *Z* axes, under each leg (location 4 in Figure 6.2). This would enable calculation of the component forces and the total applied force. An alternative arrangement involved mounting the whole couch on a rigid plinth, which would rest on one centrally positioned unit instrumented with three force transducers. However, both these approaches had several perceived disadvantages: Firstly, the large static load resulting from the weight of the mobilisation couch and the subject was likely to mask the relatively small mobilisation forces. Secondly, the inertia of the couch, coupled with the separation of the load cells from the point of force application, would have caused unwanted attenuation

7.2 THE INSTRUMENTED COUCH: VERSION I

To overcome these disadvantages, a standard mobilisation couch (Akron Therapy Products Ltd, Ipswich, UK) was modified to allow the couch top to be removed from the tubular steel frame. The couch top was mounted on four bending beam load cells, each one of which conformed to the specification detailed in Appendix 1. In this arrangement the load cells were not required to support the legs and frame of the bed, reducing the effective dead load by approximately 65 kg. It also minimised the distance between the plane of measurement and force application (Figure 7-1).

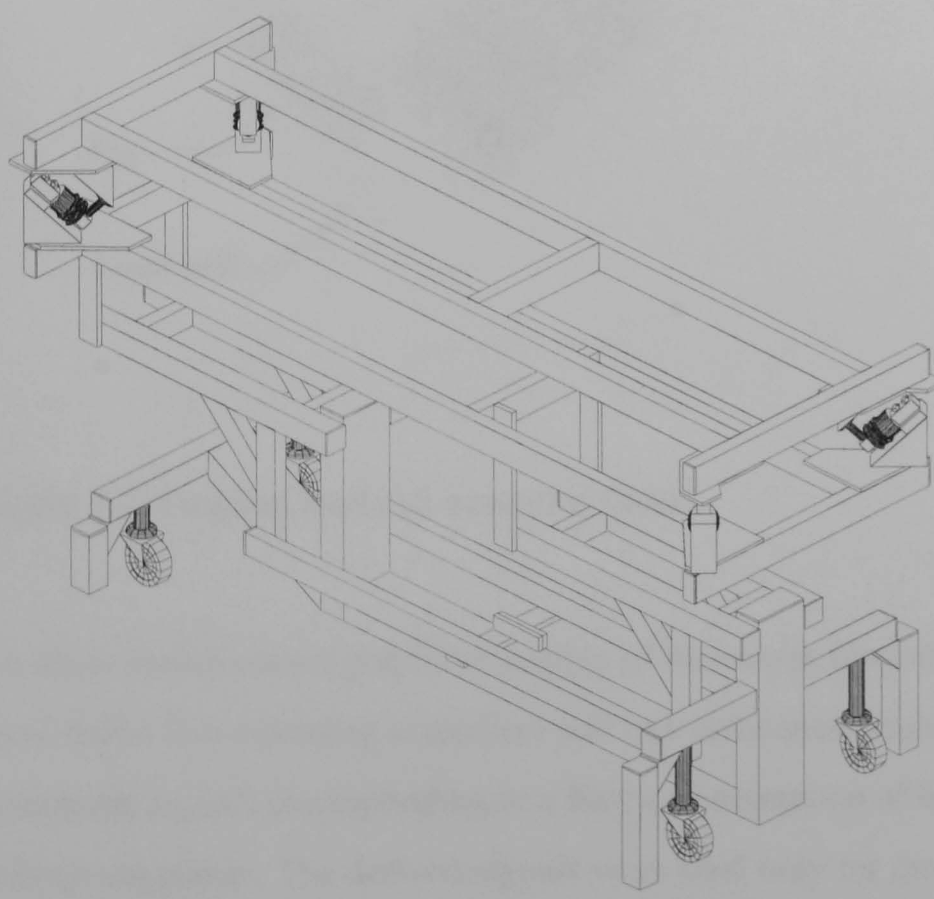


Figure 7-1: The Instrumented Mobilisation Couch: Version I

Two load cells were positioned at the head and two at the foot of the couch and were inclined at forty five degrees to the three orthogonal planes (Figure 7-2). The primary (sensing) axis of the load cell was therefore at an angle of forty five degrees to *X*, *Y* and *Z* planes. Pins projected from the load bearing surface of the load cells and corresponding plates were screwed to the underside of the couch top perpendicular to the pin axis. The pins and plates were polished to allow them to slide freely over each other. Thus, any force applied to the couch top would be transmitted through the pins and measured by the load cells.

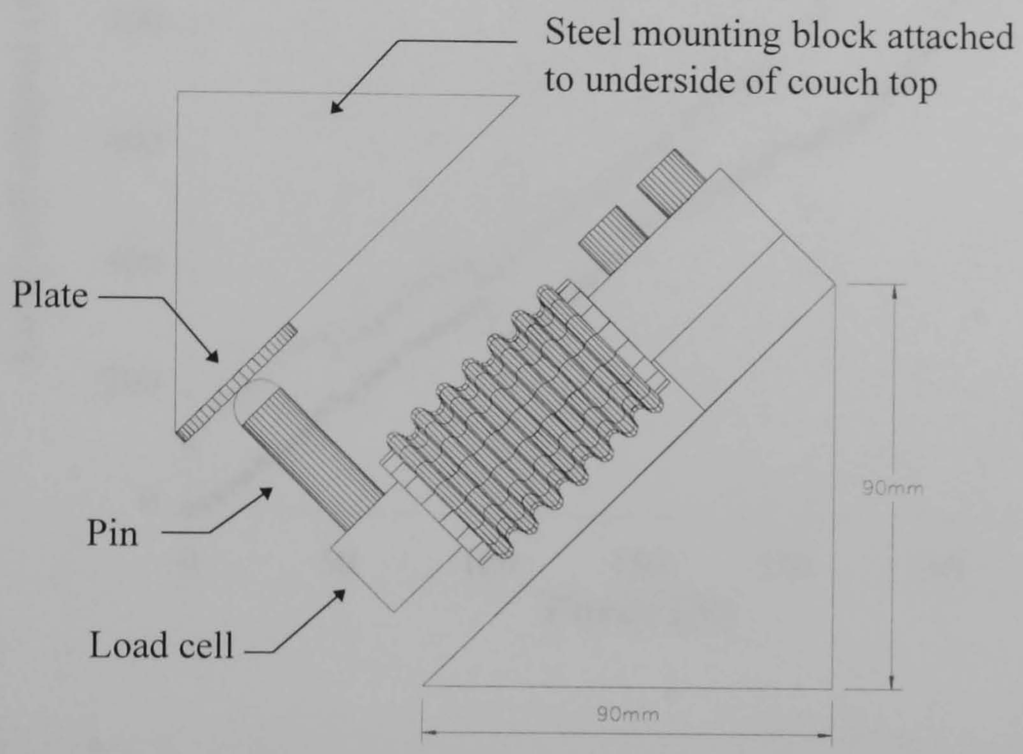


Figure 7-2: Original load cell mounting block

To allow simultaneous graphical display of the output from the load cells, the signals were fed to five summing amplifiers and two difference amplifiers which provided three composite signals corresponding to a first approximation of the forces in the three orthogonal planes. The derived signals were used only for graphical display purposes. The amplified raw load cell signals were collected for further data analysis.

Initial calibration procedures demonstrated acceptable accuracy in terms of linearity and sensitivity. However, it soon became apparent that the top was unstable and if loaded asymmetrically would slide over the load cells until the couch top came into contact with part of the frame. This was demonstrated in the data by a sudden decrease in load cell output at specific applied forces as shown at 100 N in Figure 7-3.

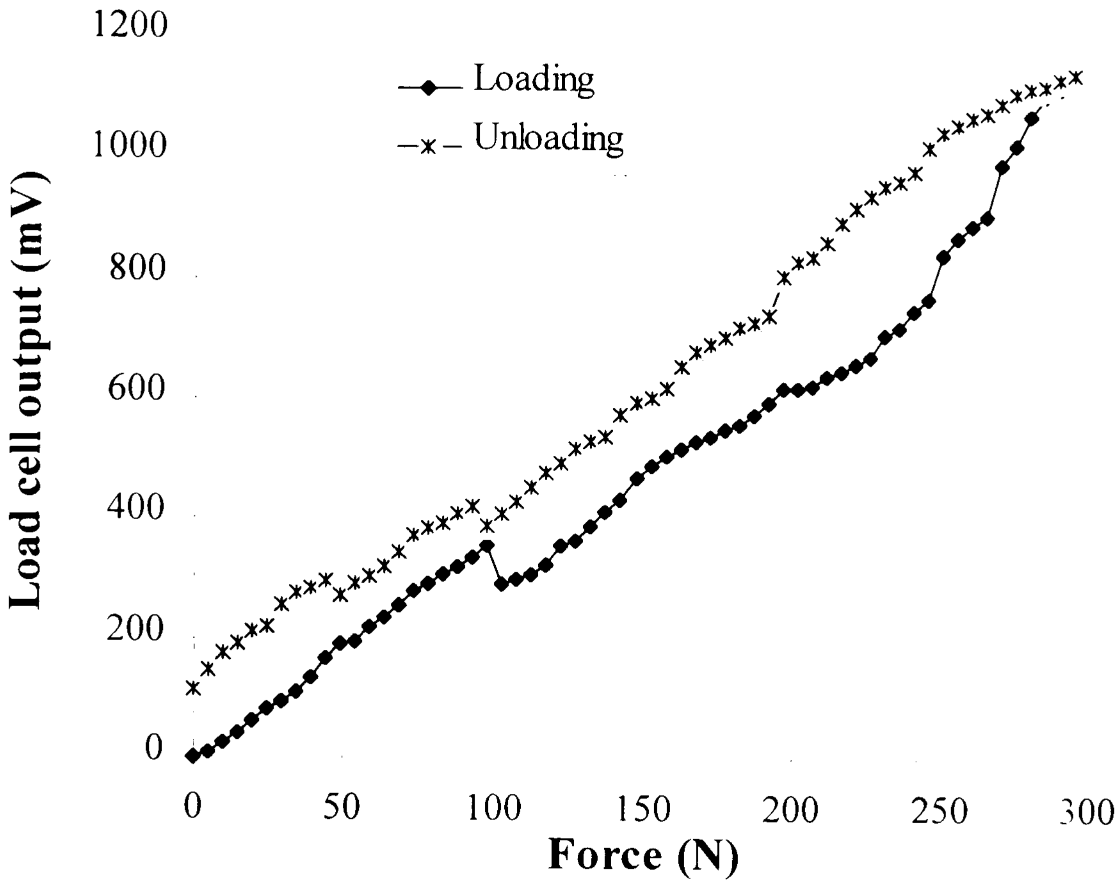


Figure 7-3: Typical response of the couch, version I, under vertical loading

In an attempt to rectify this problem, the load cells were rotated so that the pins projected outwards. However, similar limitations were demonstrated.

7.3 THE INSTRUMENTED COUCH: VERSION II

It was clearly important to increase the stability of the couch top. Therefore, neoprene mounts were interposed between the load cells and their corresponding plates (Figure 7-4). These mounts prevented untoward movement of the couch top and ensured the load cells returned the plinth to its original resting position between loading cycles. The mounts chosen were stiff in compression, but allowed some lateral deformation in shear with stiffness coefficients of 87 N/mm and 29 N mm respectively.

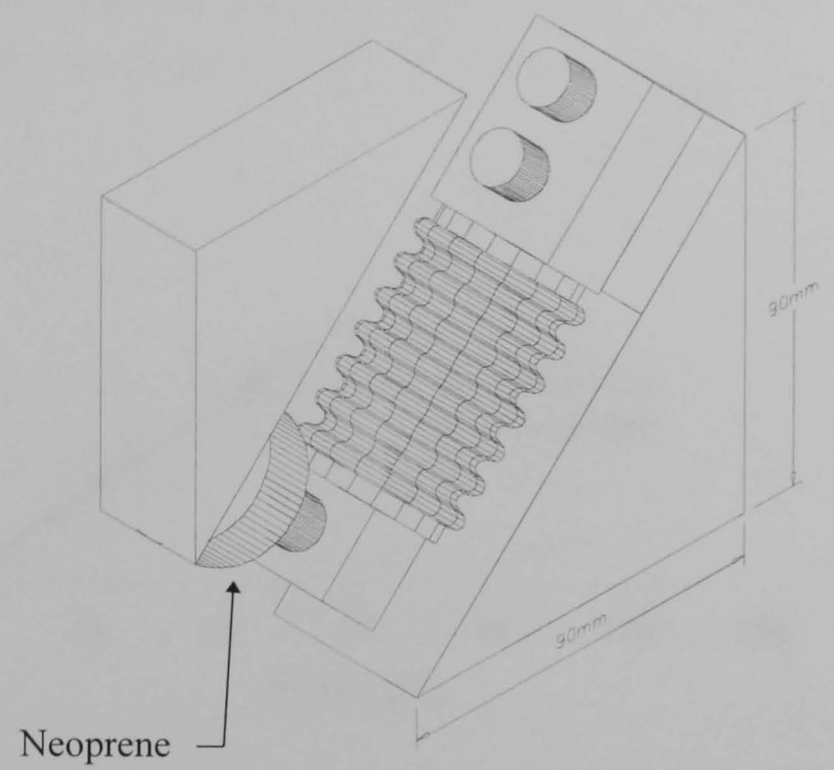


Figure 7-4: One load cell with neoprene mount

Equations were derived to resolve the component forces from the signals generated by the load cells. It had been hypothesised that providing the point of load application was known, in relation to *X, Y* and *Z* axes, the moments about each load cell could be calculated and the resultant forces summed to give F_x, F_y and F_z.

Simple calibration procedures about single axes demonstrated excellent results. Forces applied normal to each axis in turn demonstrated linearity with a very small error component (Figure 7-5). Indeed, where the vertical load response was assessed over the extended range of 10 to 300 N the maximum coefficient of variation was estimated to be less than 1%. Hysteresis and drift were negligible with a coefficient of variation of less than 0.2%. And when a sample of five therapists performed a set of Grade IV postero-anterior mobilisations, the range of values demonstrated that the load cells were recording forces within their working range of 0 to 500 N, reported by the manufacturers

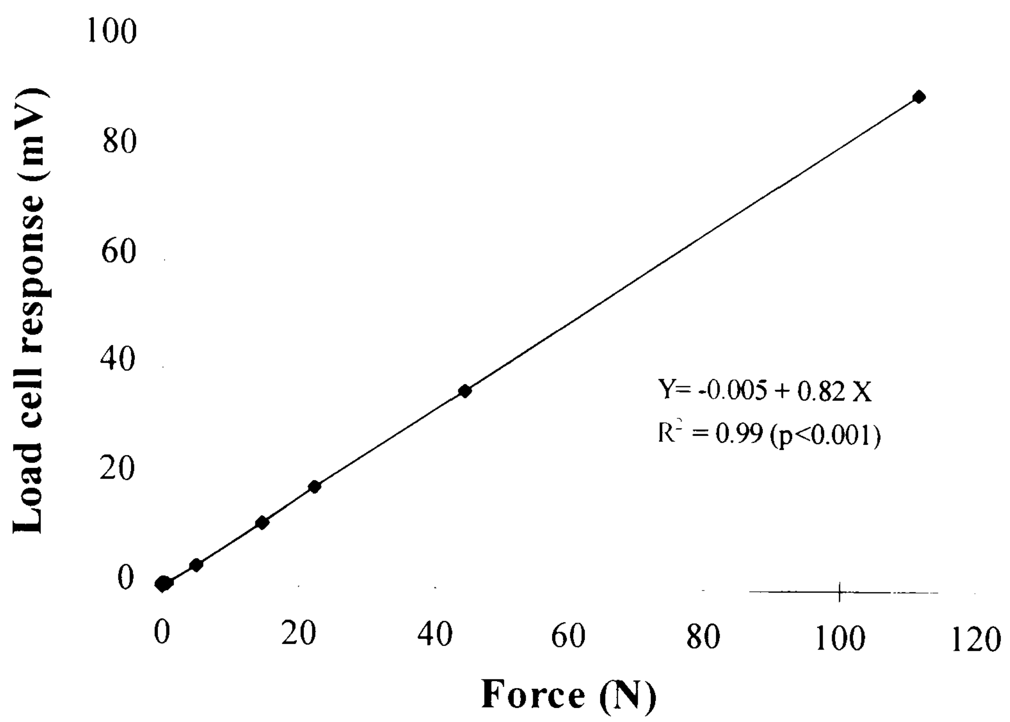


Figure 7-5: Typical response of the couch, version II, under vertical loading

Sensitivity was determined by calculating the minimal signal detected above the noise inherent in the system. This was found in response to forces of 0.19 N in the vertical plane, 0.40 N in the longitudinal plane and 0.73 N in the horizontal plane.

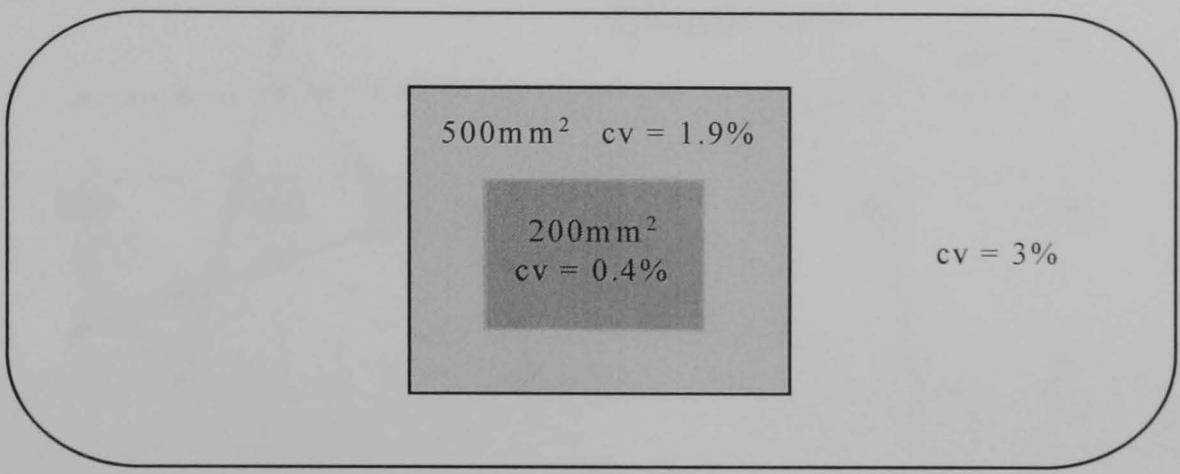
A 3 *x* 5 grid was marked out on the couch surface. Forces of 5 N, 25 N and 111 N were placed at each of the 15 positions. The response in mV was recorded for each position, a summary of which is given in (Table 7-1).

Table 7-1: The effect of changing the position of vertical force application over the couch surface

Force (N)	Mean Response (mV)		
	$200mm^2$	$500mm^2$	Whole couch surface
5	26 (0.0)	25.7 (0.5)	24.8 (0.7)
25	113.8 (0.4)	114.3 (1.0)	115.8 (2.5)
111	520.4 (1.9)	520.9 (4.9)	526.9 (10.6)

standard deviation shown in parenthesis

The coefficients of variation in the response calculated for the three areas are illustrated in Figure 7-6. The coefficients were similar for the three applied forces.



cv = coefficient of variation

Figure 7-6: The variability associated with changing the position of vertical force application over the couch surface

It is clear that the measurement error resulting from moving the point of vertical force application over the couch surface was small, shown by the coefficients of variation included in Figure 7-6. However, it was important to assess whether this vertical force, applied over the couch surface, produced any measurements of the two horizontal components. A test procedure was established in which a 10 N vertical force was applied at eleven positions at 50 mm increments across the width of the couch.

The resulting outputs in the three orthogonal axes are shown in Figure 7-7. The vertical force measurement demonstrated that F_z remained relatively constant with a mean response of 7.8mV (SD 0.4) regardless of the position of the applied force. However, although there was no force applied along the horizontal axes, the results suggest that there was a substantial component along both *X* and *Y* axes which varied depending on the position that the force was applied across the couch top.

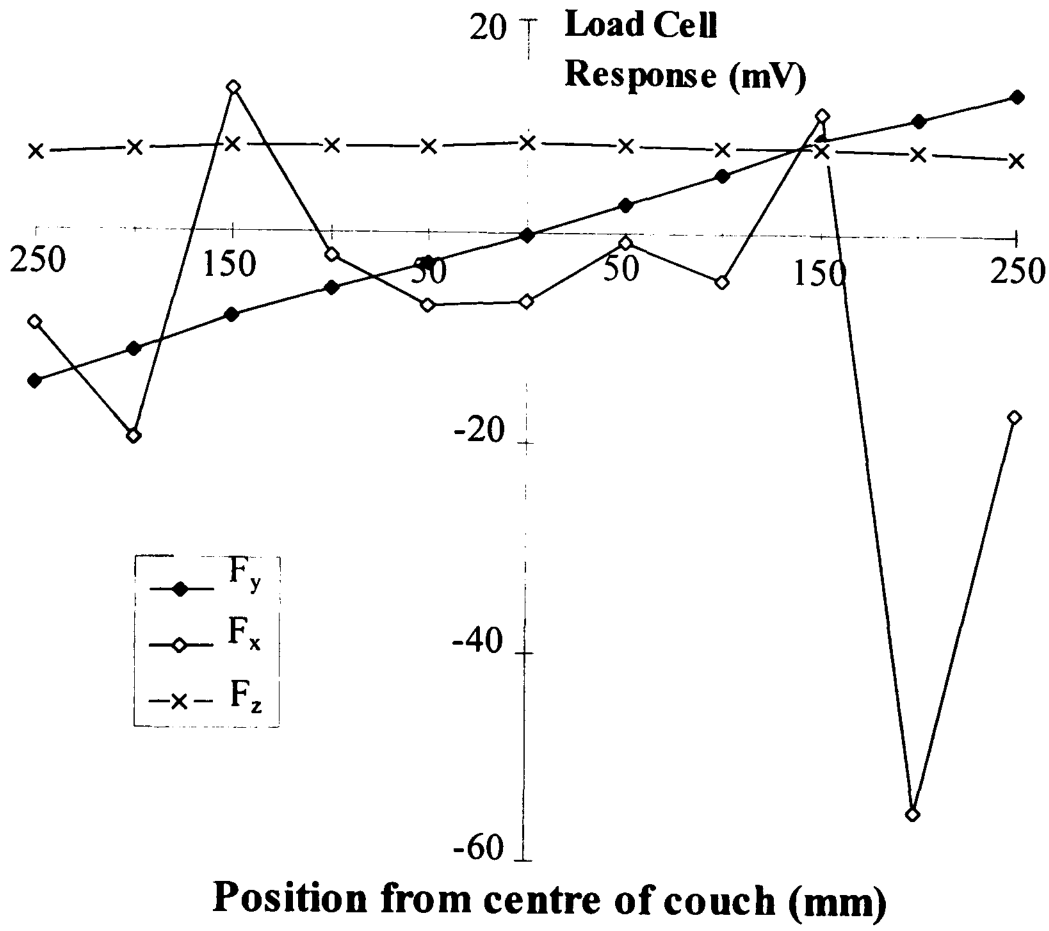


Figure 7-7: The effect of position of vertical force application across the couch width on the measurement of the three component forces

Initially, because there was a linear relationship between the point of force application across the width of the couch and the response in F_y, it had been intended to incorporate a correction factor, derived from the calibration data, into the formulae. However, further results demonstrated an unstable force couple between F_x and F_y. This is illustrated by the unpredictable nature of F_x in Figure 7-7.

When forces were applied along two or three axes simultaneously to simulate a mobilisation force, two further problems became apparent:

- The couch had six degrees of freedom yet the measuring system only allowed the measurement of four components. Therefore the appropriate derived equations were indeterminate

- The neoprene mounts were transmitting shear forces, although these forces were not registered by the load cells

These limitations led to unpredictable behaviour of the couch under varying loading conditions. The component forces could not be resolved, and the data could not be interpreted with confidence.

7.4 THE INSTRUMENTED COUCH: VERSION III

The problems identified in the previous section led to a re-evaluation of the couch design and a new approach to the calculation of forces. The new design involved mounting the couch top on six bending beam, strain gauge load cells (Model No: SHBXM, Revere Transducers Europe) The specification for one load cell is included in Appendix 1. Three load cells with vertical sensing axes were positioned in reference positions *A, B* and *C,* shown in Figure 7-8. The stability of a three point support system is greater than with four supports and ensured that the three load cells in the system cumulatively recorded the vertical load.

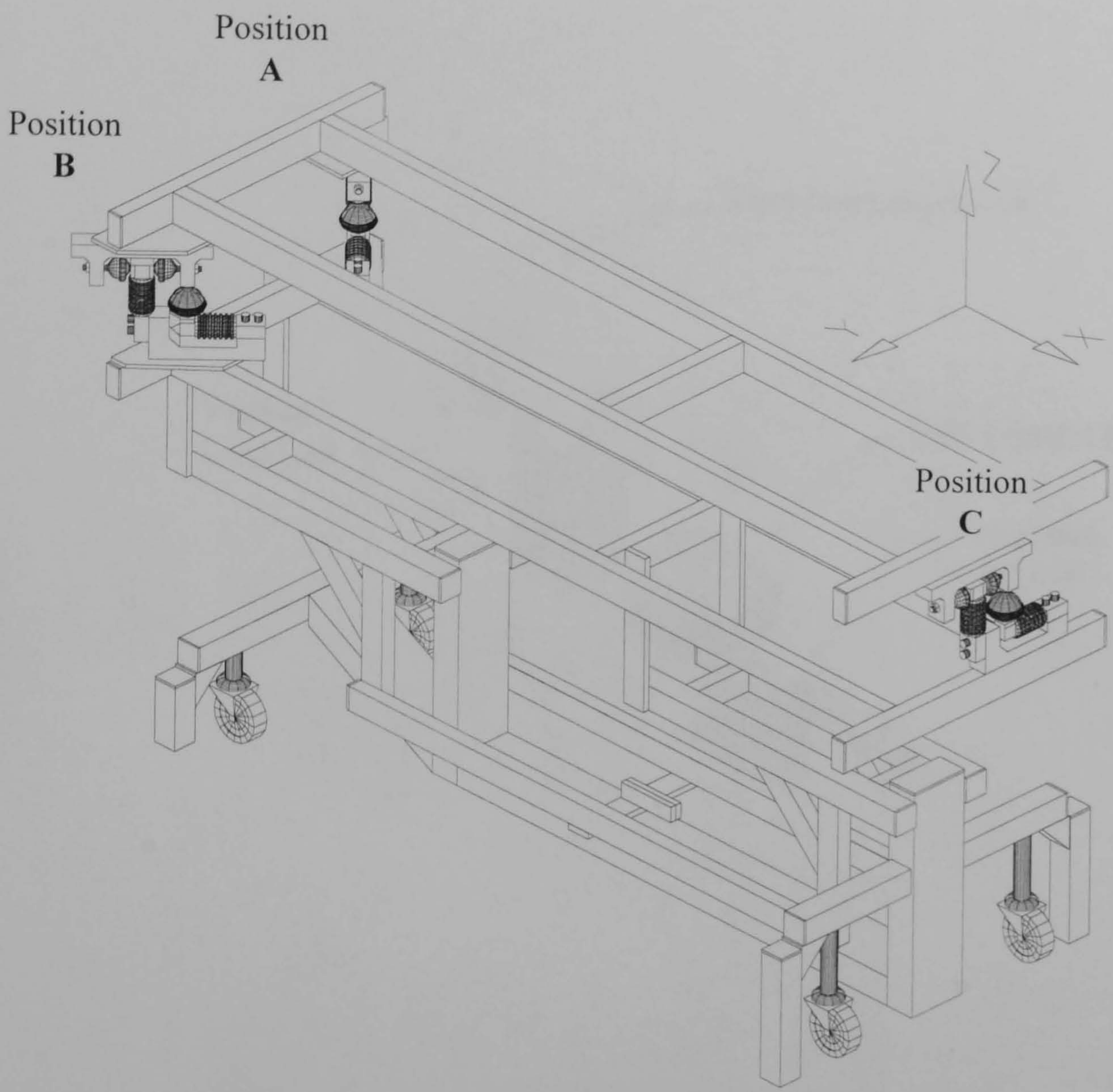


Figure 7-8: The final version of the Instrumented Mobilisation Couch

Each vertical sensing load cell was mounted in a unit with a second load cell positioned to sense horizontal forces (Figure 7-9). These units were positioned within the couch frame in a configuration which allowed calculation of forces directed along the *X*, *Y* and *Z* axes. The units at reference positions A and B were orientated at 45° to both horizontal axes of the couch and unit C was positioned parallel to the *Y* axis.

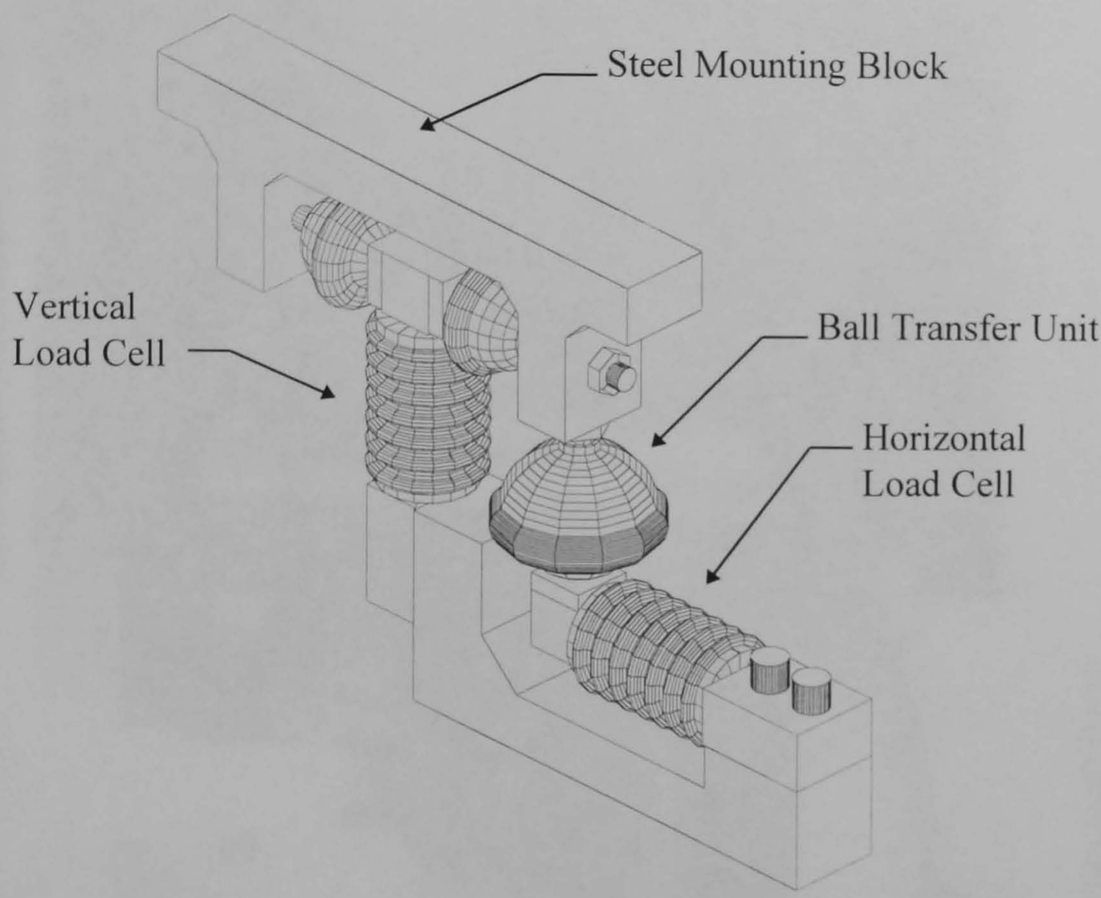


Figure 7-9: One unit comprising of a vertical and a horizontal load cell in a mounting block

The engineering drawings generated during the design of the measuring system are included in Appendix 2. Figure 7-10 shows one unit comprising of two load cells mounted on the frame of the couch. In addition, a stand-off bar, illustrated in Figure 7-11 was attached to the lower frame to allow the therapist to lean against the side of the couch to gain stability, without affecting the forces measured

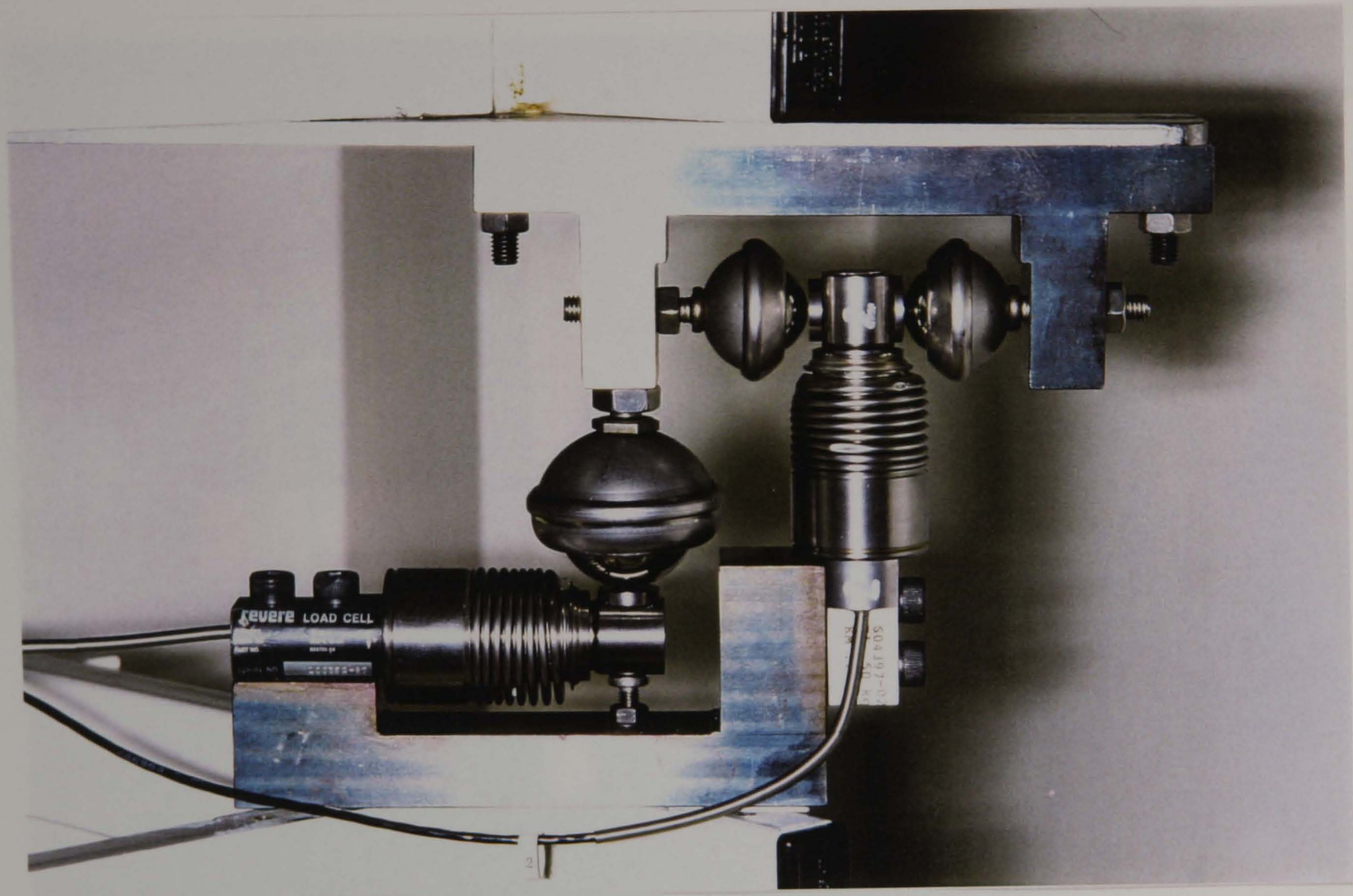


Figure 7-10: One load cell unit mounted on the couch frame

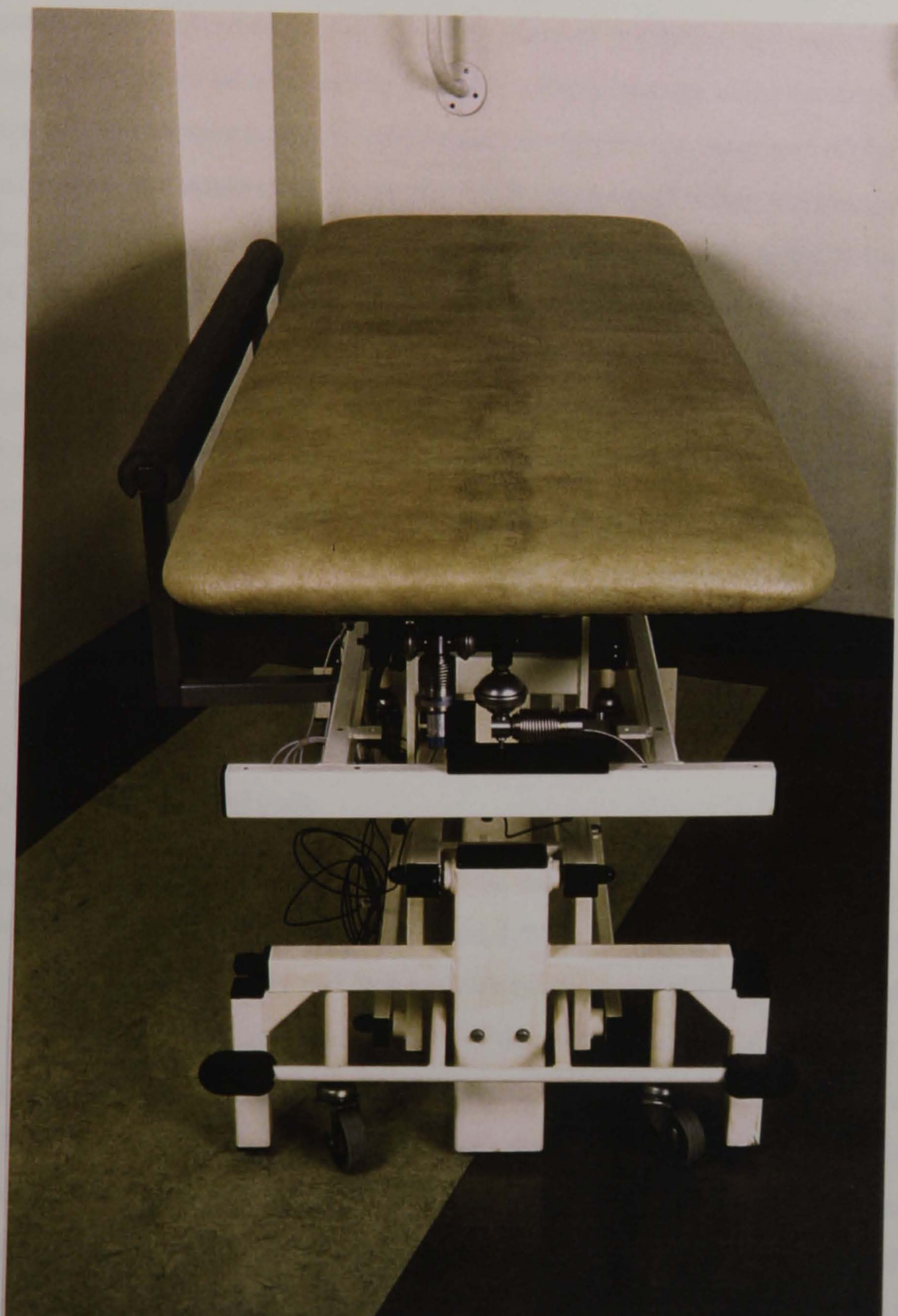


Figure 7-11: The Stand-off bar attached to the front of the couch

7.4.1 Load Cells

Each bi-directional load cell had a nominal capacity of 500 N although the manufacturers reported that each would be accurate to150% of this capacity, enabling forces of 750 N to be measured by each cell without damage to the components. The three vertical sensing load cells could therefore support the static load of the couch top and subject, estimated to be 300 N and 700 N respectively, while remaining sensitive to the superimposed mobilisation forces of up to 300 N. According to the manufacturers, the arm of the load cells deflected by less than 0.3mm under full scale load. The blocks into which the load cells were mounted incorporated mechanical stops to prevent their damage due to an accidental overload. The four strain gauges contained in each load cell produced an electrical signal proportional to their deformation and hence, the applied force. Each load cell was connected to a differential amplifier within a control box mounted on the couch frame. The amplified signals were collected for data analysis (Figure 7-12).

7.4.2 Data Analysis

The load cell signals were fed into a personal computer using a data acquisition package (Labtech Notebook, Massachusetts, Boston, USA). This provided the interface between the instrumentation system and data analysis facilities and allowed graphical display of the load cell outputs in real time, whilst raw data was collected for subsequent analysis. The software was configured to allow data collection from six load cell channels in addition to elapsed time. Sampling rate (Hz), duration and number of stages could be determined for each application. Data was stored in ASCII format. For data reduction and analysis, the raw data was translated and read into Excel (Microsoft, Reading, UK), in which calculations were performed to obtain F_x, F_y and F_z using the signals from each of the six load cells.

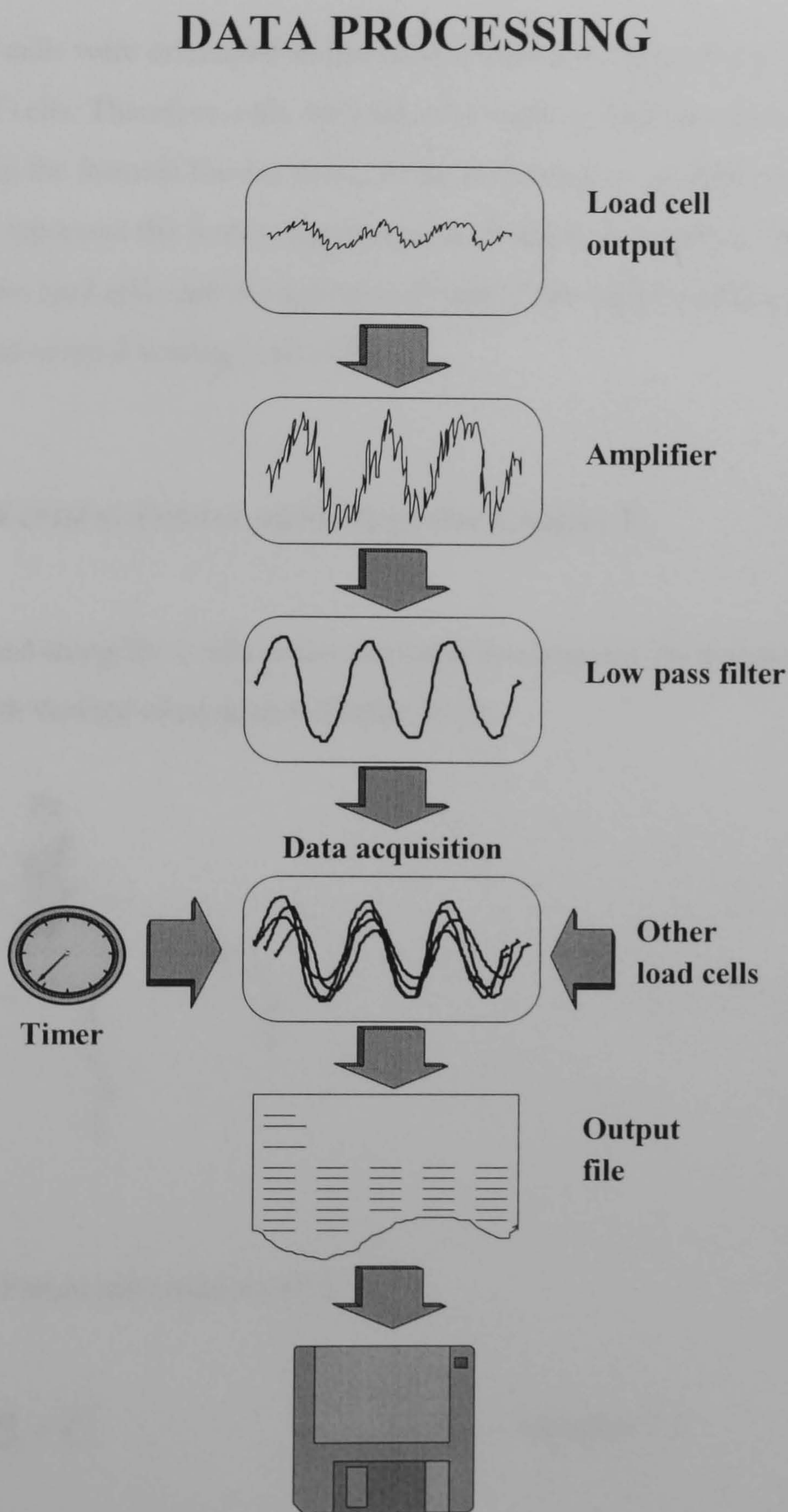


Figure 7-12: Data acquisition process from load cell output to storage in a data file

7.4.3 Determination of Forces

The six load cells were orientated so that each component force would be measured by one group of cells. Therefore, only the load cells reacting against a given force are represented in the formula for that force. In the following set of figures and equations, F_z, F_x and F_y represent the force components, A, B and C the load cell blocks, each containing two load cells and the notations 'h' and 'v' are used to differentiate between horizontal and vertical sensing load cells.

7.4.3.1 Vertical Forces applied to the Couch: F_z

Forces directed along the *Z* axis were calculated by summing the signals from the three load cells with vertical sensing axes (Figure 7-13).

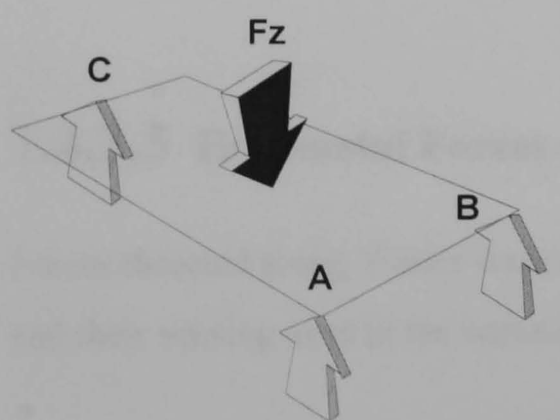


Figure 7-13: Forces and reactions (F_z)

$$F_z = A_v + B_v + C_v$$ equation 7.1

7.4.3.2 Horizontal Forces applied along the length of the Couch: F_x

Forces directed along X axes would be resisted by two of the load cells with their sensing axes in the horizontal plane (Figure 7-14). The load cells in positions A and B were orientated at 45° to the horizontal axes of the couch.

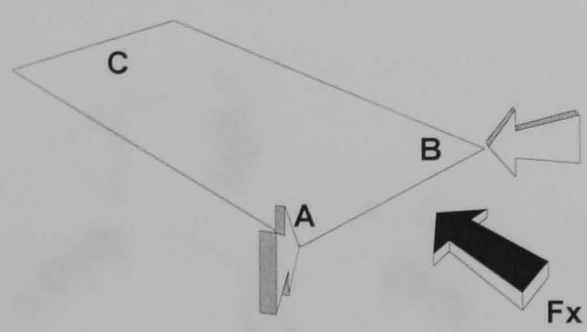


Figure 7-14: Forces and reactions (F_x)

$$F_x = \frac{A_h}{\sqrt{2}} + \frac{B_h}{\sqrt{2}}$$ equation 7.2

7.4.3.3 Horizontal Forces applied across the width of the Couch: F_y

Forces directed along Y axes were derived from functions of the three load cells which had their sensing axes in the horizontal plane (Figure 7-15).

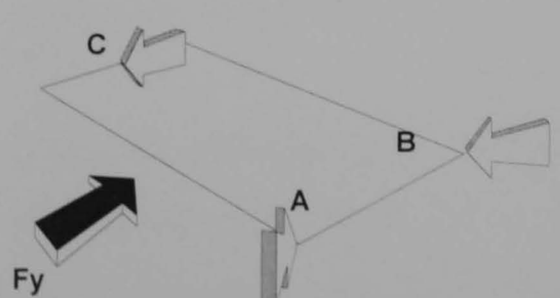


Figure 7-15: Forces and reactions (F_y)

$$F_y = \frac{A_h}{\sqrt{2}} + \frac{B_h}{\sqrt{2}} + C_h$$ equation 7.3

7.4.3.4 The Resultant Force: F_{total}

The total force applied (F_{total}) could be calculated by summing the individual components of the overall force vector (Figure 7-16).

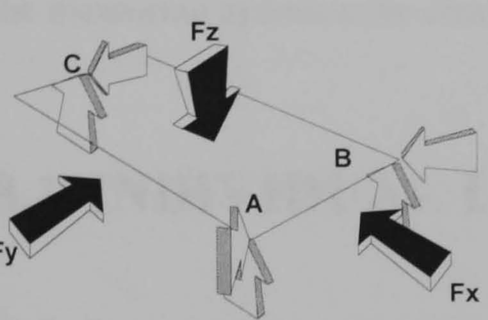


Figure 7-16: Forces and reactions (F_{total})

$$F_{Total} = \sqrt{F_x^2 + F_y^2 + F_z^2} \qquad \text{equation 7.4}$$

Chapter 8: Performance Data

Before the system could be used to collect subject data, it was necessary to establish its performance under specified conditions. The following calibration procedures allowed the measuring system to be characterised.

8.1 INDIVIDUAL LOAD CELL CALIBRATION

8.1.1 Assessment of linearity and scaling differences

Once the load cells had been mounted on the couch frame, they were calibrated individually before the couch top was re-positioned. Forces were applied to the vertical sensing load cells by suspending weights underneath them. First order pulley systems were used to redirect the forces to allow calibration of the horizontal force sensing load cells. To minimise the scaling differences between the load cells, the offsets and gains were adjusted via potentiometers on the amplifier boards. Further calibration of individual load cells allowed correction factors to be calculated to rectify any persistent scaling inequalities. Calibration of the vertical sensing load cells was confined to gravitational forces in one direction. The three horizontal sensing load cells were calibrated bi-directionally and a summary of the data is presented in Table 8-1.

In seven out of nine cases, adjustment of the gain on the amplifier board obviated the need for correction factors. However, a scaling factor was needed to correct the output from load cell B_h, which was included in subsequent calculations. This factor was calculated from the mean of the positive and negative values.

Table 8-1: Calibration data from the six individual load cells

Load Cell	Sense	R^2 *	mV/N	Correction factors
A_v	-ve	1.00	8.1	none
B_v	-ve	1.00	8.1	none
C_v	+ve	1.00	8.0	none
A_h	+ve	1.00	6.9	F_x: none
	-ve	0.99	6.8	F_y: none
B_h	+ve	1.00	9.2	F_x: 0.745 (corrected to A_h)
	-ve	1.00	9.2	F_y: 0.750 (corrected to C_h)
C_h	+ve	1.00	6.9	none
	-ve	1.00	6.9	none

* Coefficient of determination of linear model

A typical calibration graph of one load cell is shown in Figure 8-1. Each procedure was repeated 10 times and the mean result used in subsequent calculations.

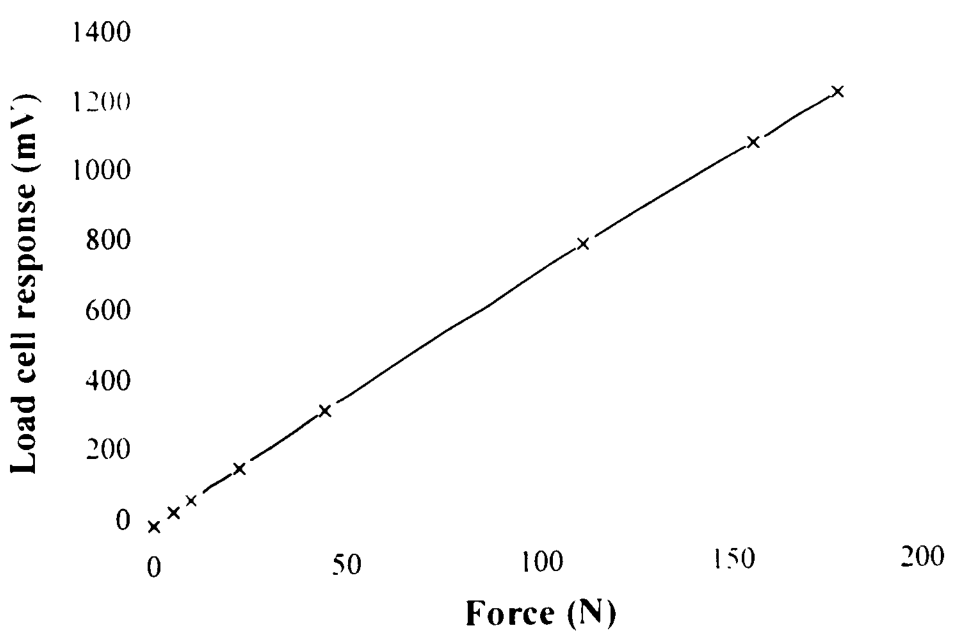


Figure 8-1: A typical calibration line illustrating the response of the load cell (B_Z)

8.1.2 Position of force application on load cell

A 20 mm disc of ground and polished stainless steel was mounted on the free end of the load cell (Figure 8-2). When the couch top was positioned over the three load cells, it was noted that in one instance, the point of force application was not central to the disc due to machining error or mechanical misalignment. This effectively lengthened the load arm of the cell. Therefore a calibration procedure was undertaken to determine the effect of changing the point of force application along the disc.

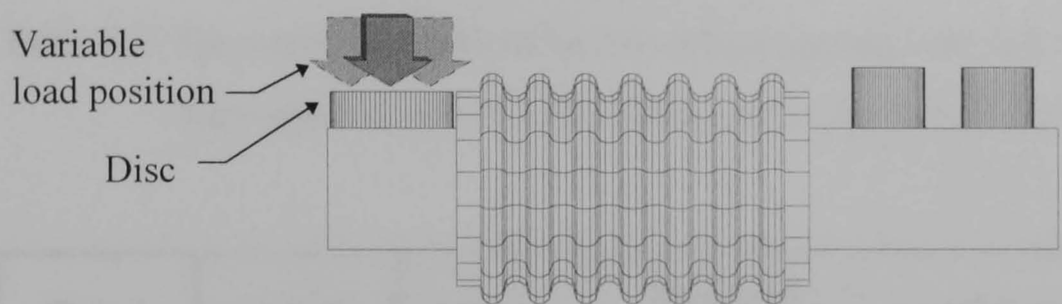


Figure 8-2: The three positions on the load cell disc used to determine the effect of the point of loading on response

A spreader bar was constructed to allow a weight to hang, free of the frame of the couch (Figure 8-3). A stylus was located in the centre of the bar and enabled the force to be applied accurately (+/- 0.25mm) in three positions along the disc.

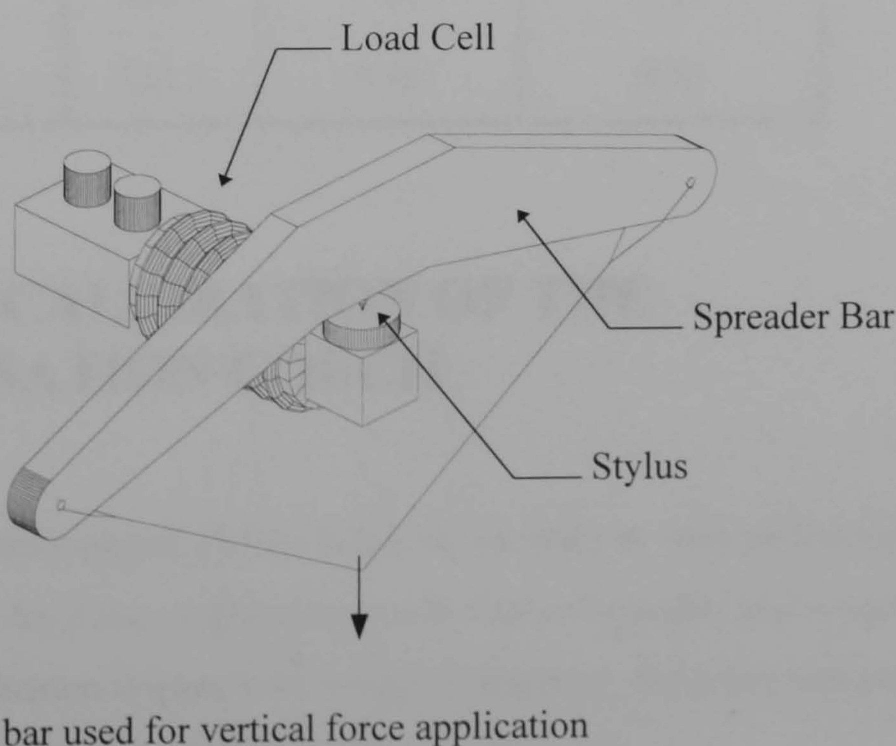


Figure 8-3: Spreader bar used for vertical force application

The effect of changing the point of loading, from a central position on the disc, to 5mm proximal or distal to this point, was examined. The procedure was repeated on three occasions for each vertical sensing load cell, at two different forces. The results, summarised in Table 8-2, indicate variability not exceeding 1.3%. Indeed there was no systematic increase in load cell output as the force was moved distally, indicating that the results would not be affected by the mechanical misalignment noted.

Table 8-2: The variation exhibited by the vertical sensing load cells when the point of force application was varied along their longitudinal axis

Force (N)	Load cell	Mean (mV)	Standard deviation (mV)	Coefficient of variation (%)
	A_V	79.5	0.78	0.99
9.81	B_V	77.7	0.95	1.22
	C_V	78.9	1.02	1.30
	A_V	358.7	1.07	0.30
44.5	B_V	361.9	1.00	0.28
	C_V	355.3	0.45	0.13

8.2 STATIC CALIBRATION OF THE MOBILISATION COUCH

The couch top was then replaced and the following procedures were performed. Forces were applied directly for Z axis calibration and via first order pulley and weight systems for X and Y axes calibration (Figure 8-4). Where appropriate, the couch was preloaded with a 690N static load to represent the weight of a subject.

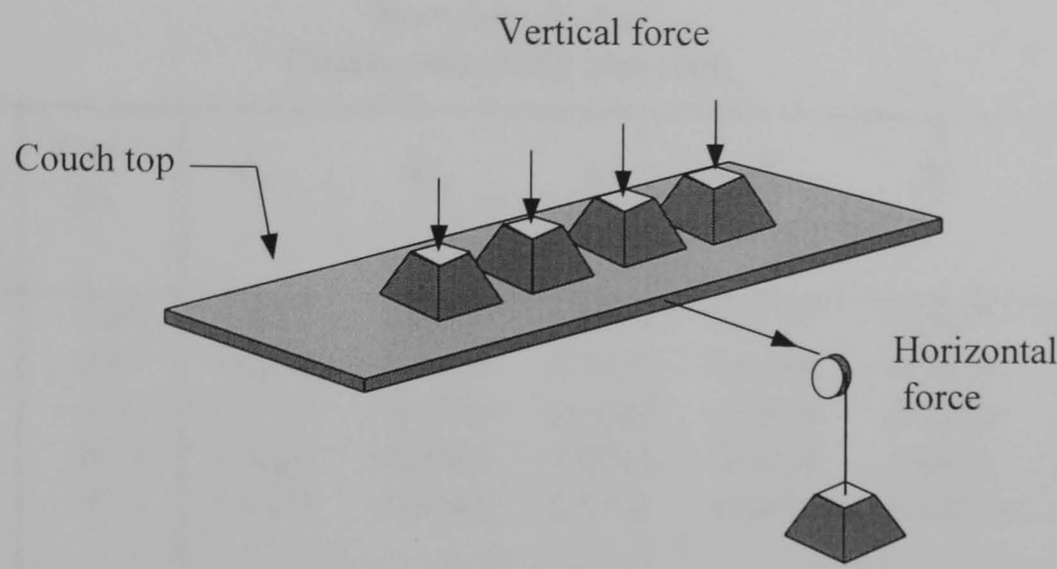


Figure 8-4: Schematic of the method of applying vertical and horizontal calibrating forces

The baseline activity of the load cells was calculated by sampling the load cells at 20 Hz with the subject or static load in position, for 10 seconds, prior to each measurement set (Table 8-3). This data was used to initialise each set of measurements to allow the effects of the static load to be removed from the calculations (Table 8-5).

Because of the extent of the data collected during calibration procedures, the raw data is not included. Therefore, a series of analysis tasks are illustrated below to show an example of data manipulation (Table 8-3 - Table 8-8). At each stage the data were checked manually to remove erroneous values which may have skewed the summary data.

Table 8-3: Baseline data used to calculate load cell means, which were used to initialise data file before further analysis

Raw data (volts) Couch with static load only							
Sample number	**Time (s)**	C_v	C_h	A_v	A_h	B_h	B_v
1	0.00	3.6523	-0.0537	2.5635	0.0977	-0.0879	2.5537
2	0.05	3.6670	-0.0635	2.5732	0.0879	-0.0879	2.5537
3	0.10	3.6719	-0.0586	2.5684	0.0879	-0.0830	2.5537
4	0.15	3.6670	-0.0586	2.5781	0.0830	-0.0830	2.5537
5	0.20	3.6670	-0.0684	2.5732	0.0830	-0.0830	2.5488
.	.	.	.	.	.	.	.
200	10.00	3.667	-0.0439	2.5732	0.0879	-0.0830	2.5537
	Mean	3.666	-0.0586	2.5698	0.0886	-0.0850	2.5542

Table 8-4: Raw data before initialisation, during vertical force application to couch surface

Raw data (volts) Vertical calibration force added							
Sample number	**Time (s)**	C_v	C_h	A_v	A_h	B_h	B_v
201	27.00	3.9893	-0.0537	2.7881	0.0977	-0.1074	2.6855
202	27.05	3.9893	-0.0732	2.7979	0.0977	-0.0977	2.6611
203	27.10	3.9893	-0.0488	2.7979	0.1025	-0.1025	2.6465
204	27.15	3.9893	-0.0537	2.7881	0.0977	-0.1025	2.6563
205	27.20	3.9893	-0.0488	2.7979	0.1025	-0.0977	2.6709
.	.	.	.	.	.	.	.
400	37.00	3.9893	-0.0537	2.7979	0.1025	-0.1025	2.6660

Table 8-5: Data from Table 8-4 initialised using load cell means and converted from volts to millivolts

Data in rows 201-400 initialised using means from rows 1-200 (Table 8-3)							
Sample number	**Time (s)**	C_v	C_h	A_v	A_h	B_h	B_v
201	27.00	323	5	218	9	22	131
202	27.05	323	-15	228	9	33	107
203	27.10	323	10	228	14	17	92
204	27.15	323	5	218	9	17	102
205	27.20	323	10	228	14	13	117
.	.	.	.	.	.	.	.
400	37.00	323	5	228	14	17	112

Table 8-6: Load cell output from Table 8-5 resolved into component forces i.e.:

$$F_z = A_v + B_v + C_v$$

Individual load cell signals combined in equations				
Sample number	**Time (s)**	F_z	F_y (mV)	F_x
201	27.00	672	10	18
202	27.05	658	-14	13
203	27.10	643	9	19
204	27.15	643	8	16
205	27.20	668	7	16
.	.	.	.	.
400	37.00	663	4	19
	Mean	662	10	16

Table 8-7: Ten repetitions of each calibration procedure were performed, the data from the ten replications was then combined

Combined with data from other files (Fz used in example)						
Force (N)	**File 1**	**File 2**	**File 3**	**File 4**	**File n..**	**File 10**
0.00	0	-2	0	-3	·	2
9.81	76	73	75	72	·	74
14.72	115	117	119	113	·	116
36.98	304	295	299	294	·	294
81.42	662	663	658	657	·	657
192.67	1564	1566	1569	1565	·	1563
303.91	2467	2467	2465	2465	·	2461

Table 8-8: Grand mean from 10 replications

Force (N)	**Fz (mV)**
0.00	-1
9.81	74
14.72	116
36.98	297
81.42	660
192.67	1565
303.91	2465

This data was used in combination with other data files to generate summary statistics, conversion factors, regression equations and measures of dispersion. The examples shown above illustrate a small proportion of the data collected. Over two hundred data files were generated and analysed to characterise the couch.

8.2.1 Linearity of the Measurement System

The couch was loaded incrementally with forces of between 0 - 304 N, considered to be of similar magnitude to mobilisation forces. The effects of loading the couch along the *Z* axis are shown in Figure 8-5.

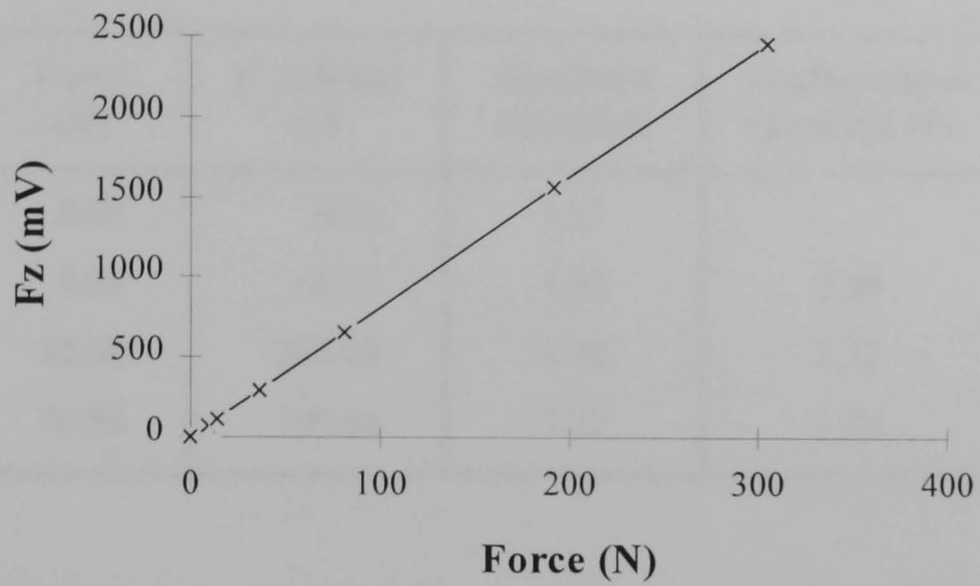


Figure 8-5: Typical response of the couch, version III, under vertical loading

Summary data are given in Table 8-9 to show the variation over ten repetitions.

Table 8-9: F_z calculated in response to vertical loading along the *Z* axis

Force (N)	F_z (Mean) mV	Standard deviation	Coefficient of variation (%)
0.00	-4.32	3.75	-
9.81	75.70	3.07	4.05
14.71	115.48	4.21	3.64
36.98	296.39	4.48	1.51
81.42	658.73	3.86	0.59
192.67	1562.07	6.84	0.44
303.91	2464.41	4.31	0.17

Similar procedures were used to calibrate *X* and *Y* axes responses and are summarised in Table 8-10 and Table 8-11. However, because the magnitude of the force directed along the horizontal axes during mobilisation was anticipated to be smaller than that directed along the vertical axis, a smaller horizontal force range was used for calibration.

Table 8-10: F_y calculated in response to horizontal loading along the *Y* axis

Force (N)	F_y (Mean) mV	Standard deviation	Coefficient of variation (%)
0.00	0.26	1.57	-
9.81	66.30	4.30	6.49
32.30	206.08	4.78	2.32
76.50	509.84	7.37	1.45

Table 8-11: F_x calculated in response to horizontal loading along the *X* axis

Force (N)	F_x (Mean) mV	Standard deviation	Coefficient of variation (%)
0.00	0.63	4.13	-
9.81	50.77	4.59	9.04
44.5	235.35	6.29	2.67
111.25	660.92	13.39	2.03

Table 8-9- Table 8-11 demonstrate a smaller coefficient of variation in the response of the measurement system for forces applied vertically, than for those applied along the horizontal axes. This suggested that vertical forces could be measured with greater accuracy than forces directed along the *X* and *Y* axes, however, the coefficients reported all demonstrated an acceptable degree of variation.

Linear models were employed to examine the relationship between force and system response, and to provide conversion factors for each axis, given by the slope of the regression line. Because the data was initialised for the static load before each set of readings, the regression line was forced through the origin as theoretically there should not have been an offset. The regression equations are given in Table 8-12. The linear model was an excellent approximation of the response of the couch, in all three directions over the force ranges considered to be appropriate for mobilisation procedures. In addition, the effects of different static loads were found to have no effect on load cell response.

Table 8-12: Correlation between applied force and response during single axis loading

Statistical Parameter	**Axis**		
	Z	***Y***	***X***
Linear Model*	$y = 8.10x$	$y = 5.79x$	$y = 5.46x$
Coefficient of determination (r^2)	0.999	0.999	0.997
p	< 0.001	< 0.001	< 0.001
Conversion factor (mV/N)	8.10	5.79	5.46

*The regression equations do not contain a constant factor because the models were forced through the origin

8.2.2 Hysteresis of the Measurement System

Hysteresis was evaluated by comparing the results of loading and unloading cycles in each of the three axes. Figure 8-6 allows a comparison of the degree of hysteresis under similar forces along each axis. The *Z* axis response has been truncated to allow comparison of the three axes under similar load ranges.

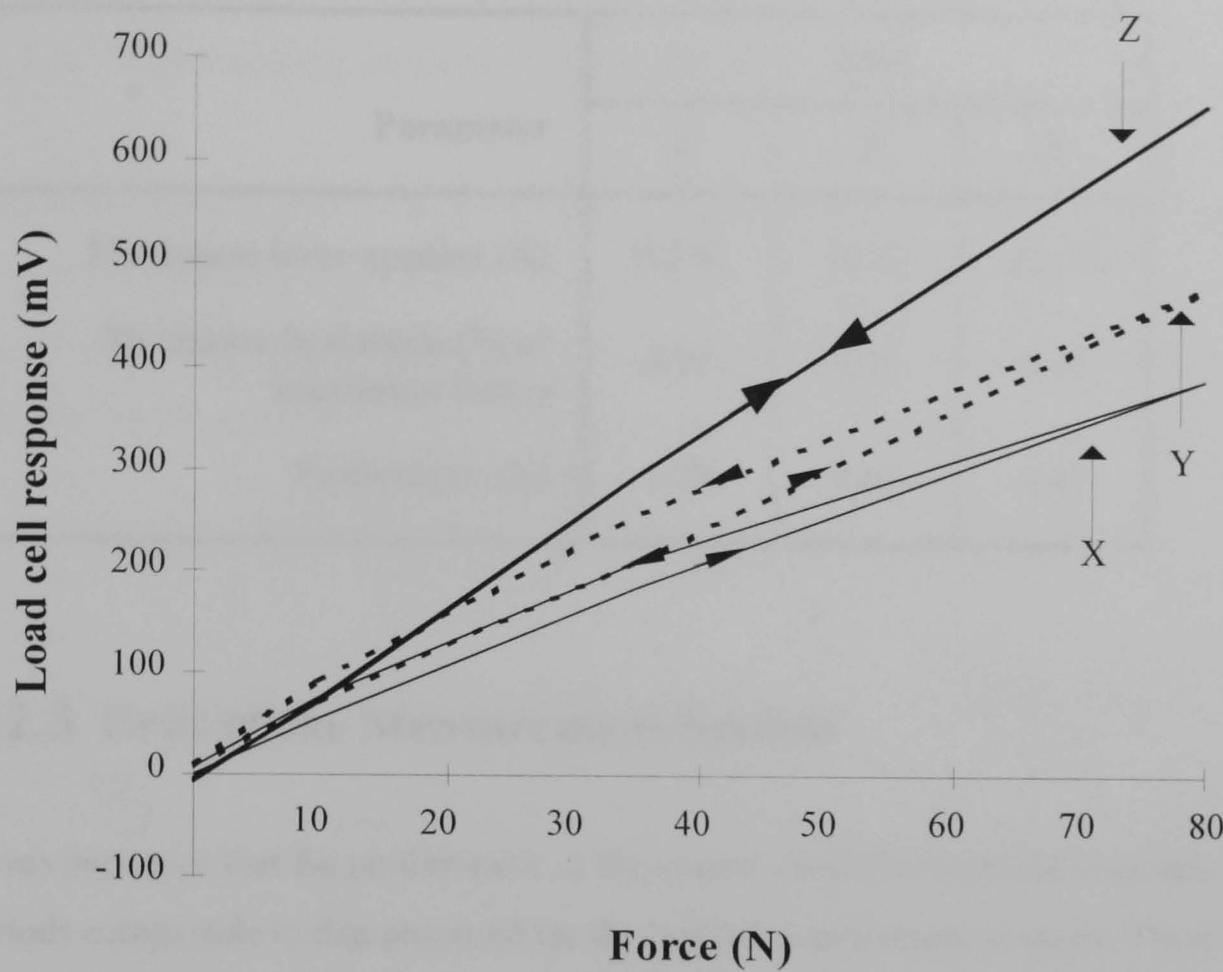


Figure 8-6: The effect of hysteresis under single axis loading

It was clear that the couch demonstrated negligible hysteresis under vertical loading up to values of 300 N. However, there was some recorded hysteresis under horizontal loading conditions (Table 8-13). This hysteresis was thought to be due primarily to the method of applying the calibrating force. The pulley system in use was not devoid of friction and may also have introduced some movement artefact, which registered as variation in the data.

However, because the measurement of mobilisation forces is primarily concerned with peak forces, the effects of hysteresis are unlikely to affect the results significantly. It is worthy of note that over ten repetitive loading/unloading cycles, there was no systematic change in load cell response, suggesting that repeated loading had no effect.

Table 8-13: Hysteresis expressed as a percentage of the maximum force applied

Parameter	Axis		
	Z	*Y*	*X*
Maximum force applied (N)	303.91	76.52	81.42
Maximum hysteresis (% of maximum force)	0.11	9.21	6.55
Equivalent (N)	0.29	6.67	4.41

8.2.3 Drift of the Measurement System

It was important that the performance of the system should be assessed over time periods comparable to that proposed for the typical measurement sessions. Therefore, the temporal response of the system to a static body weight was assessed. Figure 8-7 depicts the effects of sustained vertical loading of approximately 980 N, representing a large body weight. Drift of the load cells, expressed as the difference in recorded force over time as a percentage of that applied, was found to be minimal at 0.07%.

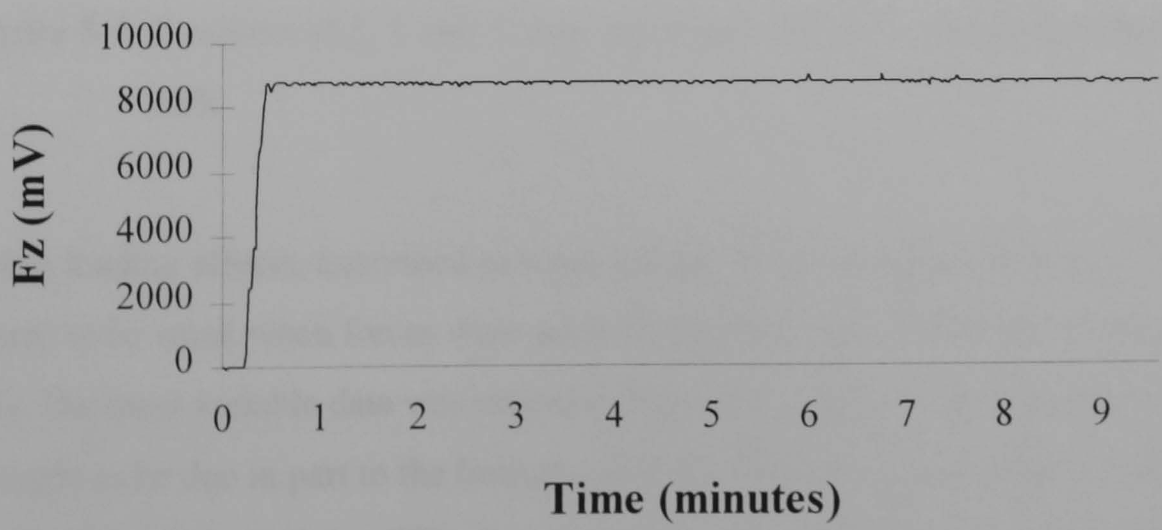


Figure 8-7: Response of the measurement system to a sustained vertical force of 980 N

8.2.4 Interaction between Load Cells

To determine whether there was any coupling between the *X, Y* and *Z* axes, the effects of loading in one axis were measured by assessing the signals in the two remaining axes. Forces were applied both at the centre of each axis and at the ends of the couch, as illustrated in Figure 8-8.

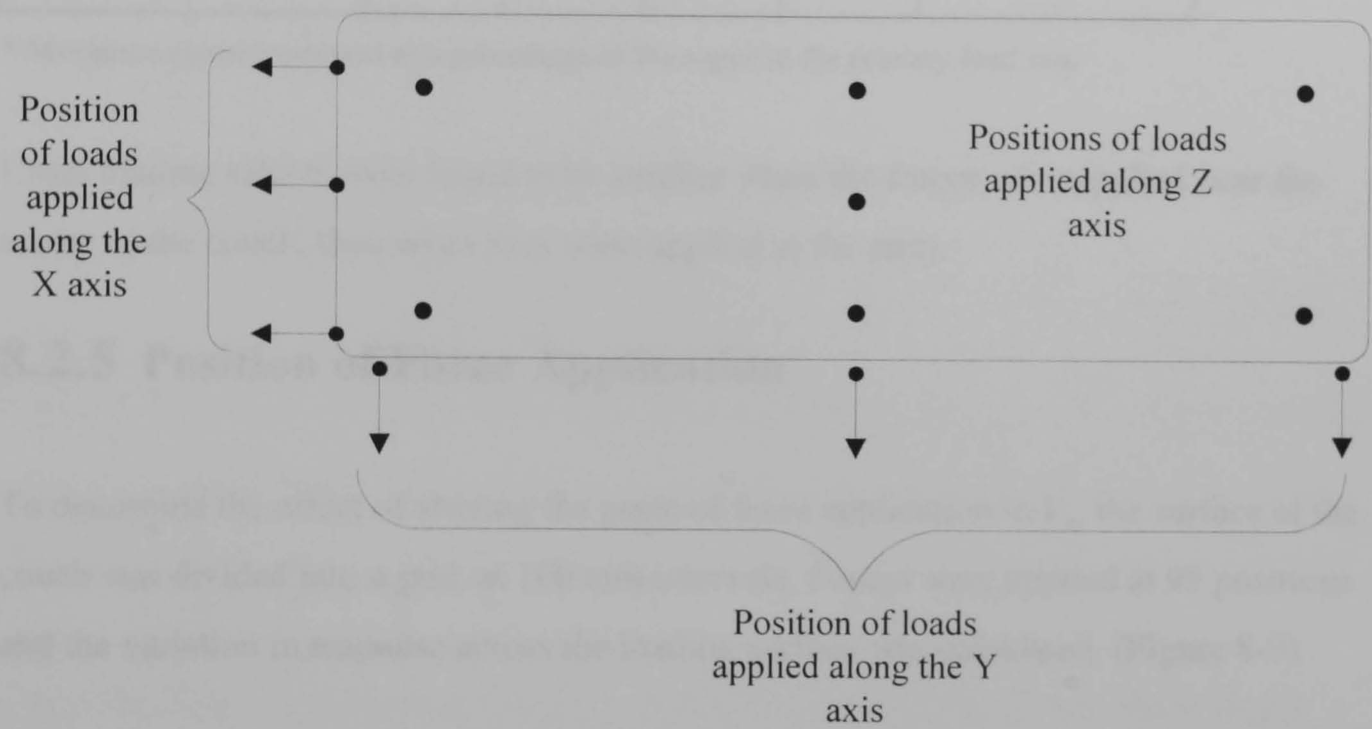


Figure 8-8: Positions of *Z, Y* and *X* axes loading to assess the interaction between load cells

Cross loading effects, expressed as a percentage of the signal in the primary axis, were found to be small when forces were applied normal to the axes of the couch (Table 8-14). The most variable data was obtained from the response in the *Y* axis, but this was thought to be due in part to the limitations of the method of applying the calibrating force. It demonstrated the difficulty of applying a force normal to one axis and perpendicular to two axes, simultaneously

Table 8-14: Cross loading effects under single axis loading

	Load Axis					
	Z		*Y*		*X*	
	Centre	Ends	Centre	Ends	Centre	Ends
Coupling in *X* (%)*	0.5	2.1	1.4	5.0		
Coupling in *Y* (%)*	1.2	4.6			4.4	6.5
Coupling in *Z* (%)*			1.4	1.5	2.5	3.6

* Maximum signal expressed as a percentage of the signal in the primary load axis

Cross loading effects were found to be smaller when the forces were applied near the centre of the couch, than when they were applied at the ends.

8.2.5 Position of Force Application

To determine the effect of altering the point of force application in F_z, the surface of the couch was divided into a grid, at 100 mm intervals. Forces were applied at 95 positions and the variation in response across the loading surface was calculated (Figure 8-9)

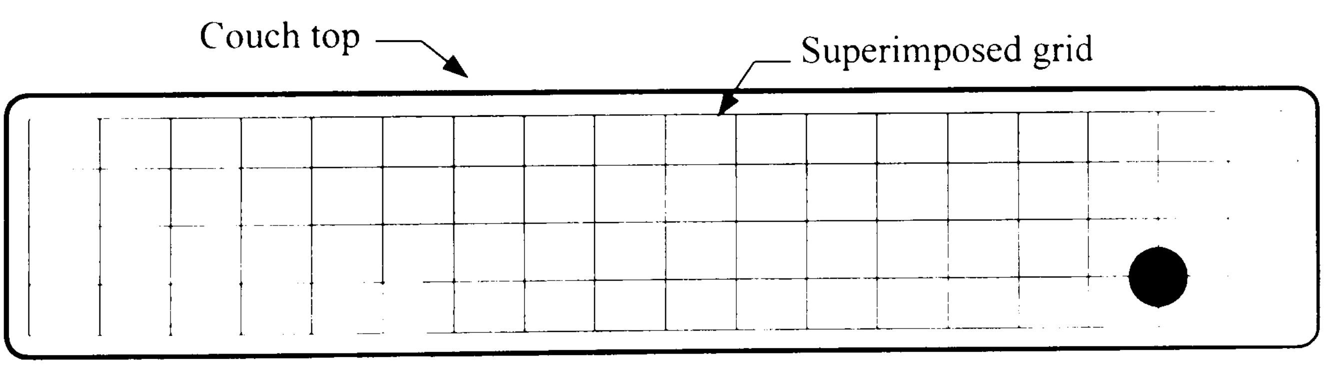


Figure 8-9: Schematic of 95 positions of force application for F_z

To determine the effect of changing the point of force application along the *Y* and *X* axes, forces were applied at seven positions along the length and three positions across the width of the couch, respectively (Figure 8-10).

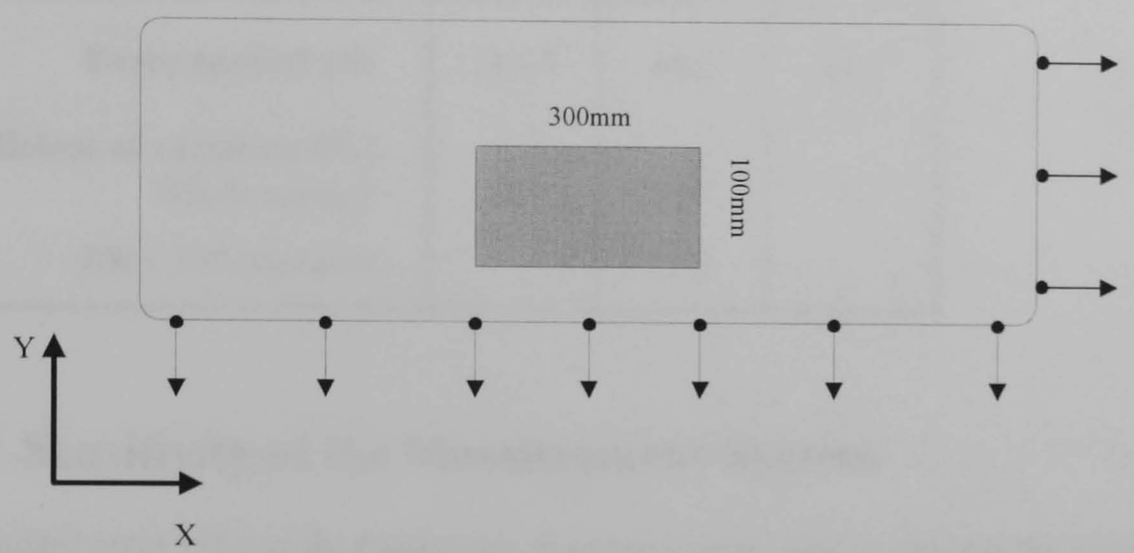


Figure 8-10: Schematic of 7 positions of force application for F_y and 3 positions of force application for F_x

There was little variation in output with changes in *Z* and *X* axes positions, but some degree of variation when the force was applied in different positions along the length of the couch (Table 8-15). However, this was considered to be an extreme condition as in practice, mobilisation forces would be applied in a more confined area. The size of this area was estimated in a pilot study using normal subjects. Ten therapists positioned two subjects of extreme body types on the couch. The position of force application to L3 in relation to the couch was found consistently within an area 300mm x 100mm. When the variability of signal in response to positional loading within this area was calculated, it was considered to be within acceptable limits of 2%.

Table 8-15: Coefficient of variation in response due to changes in the position of force application

Parameter	Axis		
	Z	*Y*	*X*
Force applied (N)	111.3	44.5	44.5
Coefficient of variation (%) Whole surface	0.9	11.2	1.7
300 x 100 mm area		2.1	

8.2.6 Sensitivity of the Measurement System

The standard error (SE) of the regression equations were used to express the error and therefore the sensitivity of measurement in each axis. The SE is a measure of the amount of error in the prediction in mV for a given force. Therefore, by establishing the 95% confidence interval for the slope of the regression line, the sensitivity can be expressed as the minimal signal detectable over the noise inherent in the system at any given value of applied force.

Table 8-16: The sensitivity of the couch to forces applied along the three orthogonal axes

Statistical Parameter	Axis		
	Z	*Y*	*X*
Linear Model	$y = 8.10x$	$y = 5.79x$	$y = 5.46x$
Standard Error (mV)	3.21	4.96	5.23
95% Confidence interval (mV)	6.29	9.72	10.25
Equivalent Force (N)	0.78	1.68	1.88

The sensitivity for Fz given in Table 8-16 is well within the requirements necessary to record the smallest peak mobilisation forces, reported to be between 6.9 to 24.5 N (*Grieve & Shirley 1987, Matyas & Bach 1985*).

8.2.7 Combined Loading

The effect of applying a force along two axes simultaneously was evaluated to determine whether the signals from each individual axis were independent (Table 8-17).

Table 8-17: The effect of applying a force along two axes simultaneously

Parameter	Combined Axes			
	Z and *Y*		*Z* and *X*	
Applied Force (N)	111.3	44.5	111.3	44.5
Response (mV)	897.0	266.0	909.0	235.0
Equivalent Force (N)	110.6	45.9	112.1	43.0

When the load cell readings were substituted into the formulae described in Section 7.4.3, the differences between the equivalent force and the actual force applied, summarised in Table 8-17, were within the 95% confidence intervals reported in Table 8-16. This indicated that the responses of each axis were independent when forces were applied along two axes simultaneously.

8.3 DYNAMIC CALIBRATION OF THE MOBILISATION COUCH

8.3.1 Resonance of the couch

The resonance of the system was determined experimentally by delivering an impulse to the centre of the couch. Three different weights, 2.5 N, 9.8 N and 24.5 N, were dropped from a distance of 500 mm above the couch, their path was intercepted as they rebounded from the couch surface, to prevent any further excitation from the weight making contact again with the surface. A typical response to a force of 24.5 N of an individual load cells is shown in Figure 8-11.

The load cells were sampled at 200 Hz to allow the effects of frequencies up to 100 Hz to be established. This was considered important because it had been previously noted that erroneous values occurred randomly in the data. The magnitude of these values suggested that averaging procedures would cause the data to become skewed. These components were likely to be high frequency spikes which could be eradicated with appropriate filtering techniques.

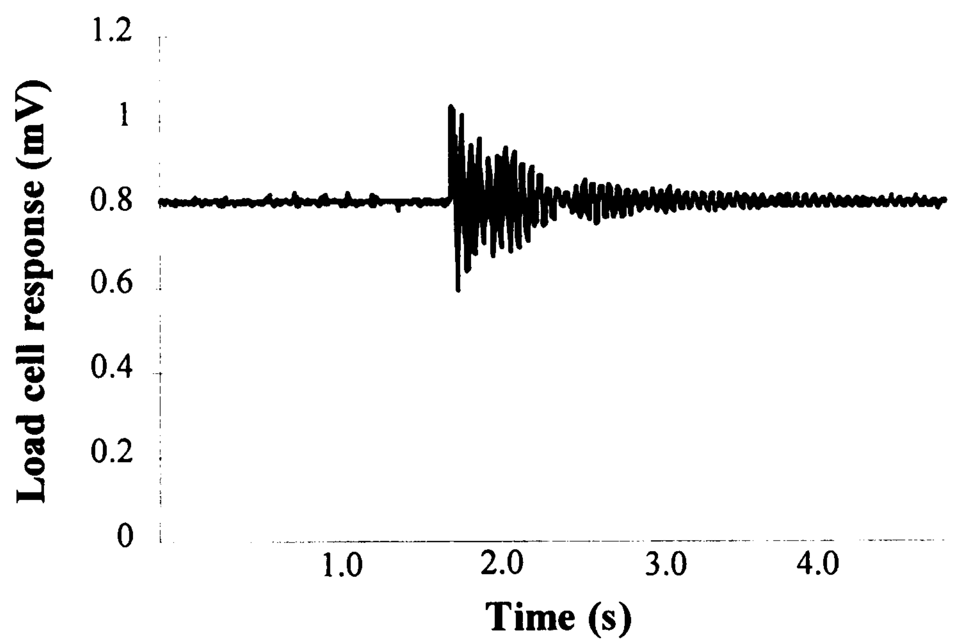


Figure 8-11: Typical response of a single load cell to an impulse (A_v)

Fourier transformation of the signals from the six load cells demonstrated that the main resonant frequencies of the system were between 8 and 18 Hz (Figure 8-12). These were significantly higher than the frequencies associated with mobilisations of 2 Hz and could therefore be filtered from the signal without loss of relevant data at around 2 Hz. A selection of vertical and horizontal load cell responses are shown in Figure 8-12. Although there was some variation in response, the main resonant frequencies were above the frequency of mobilisation.

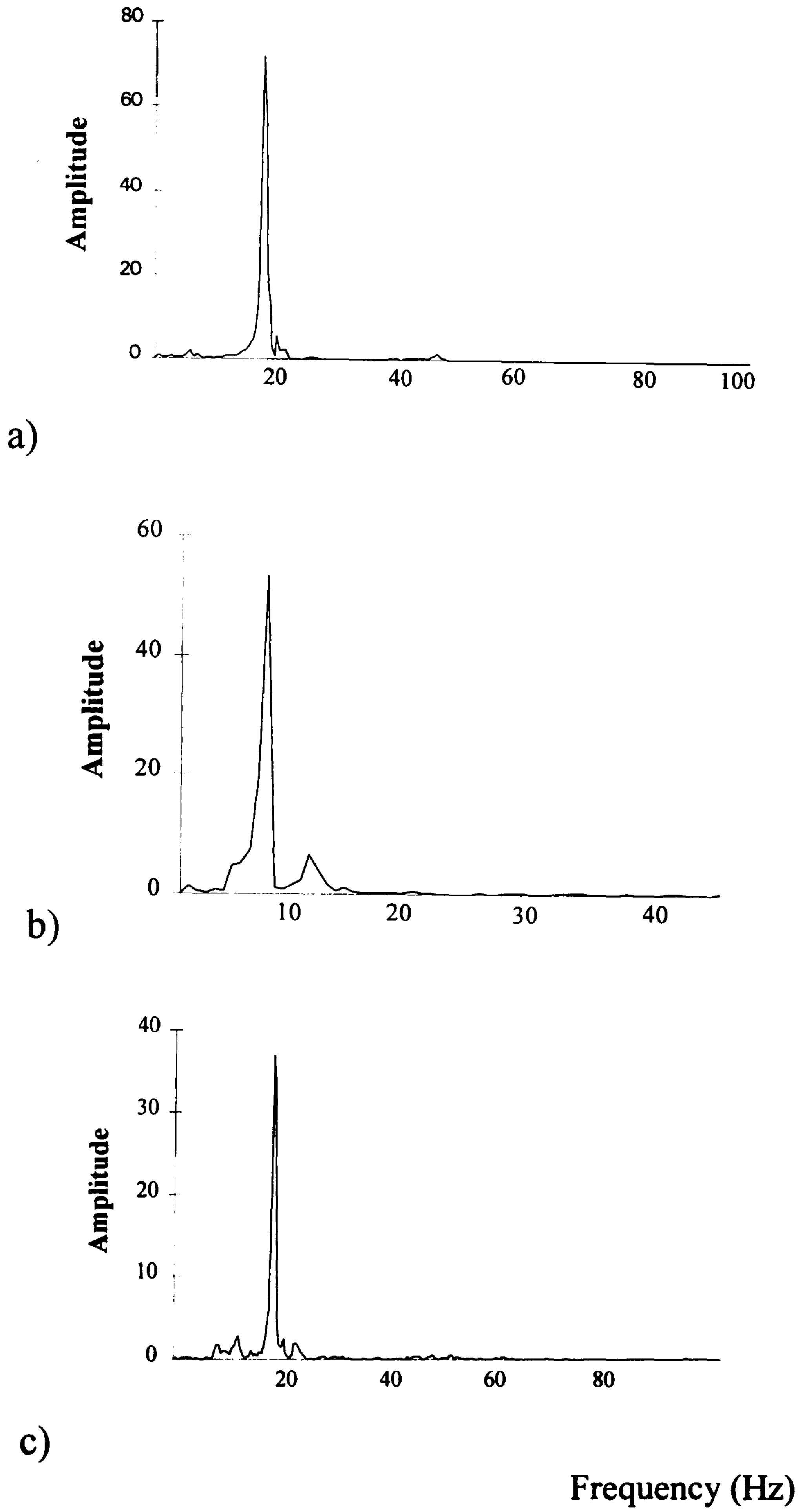


Figure 8-12: Fourier transform of impulse data for three typical load cell responses to forces of 24.5 N: Power Spectrum for load cell a) A_v b) B_v c) C_v

8.3.2 Method of Data Filtration

Because there was a significant amount of noise associated with each signal (Figure 8-13), methods of filtering the data using suitable software were considered. A facility available in Labtech Notebook allowed the data to be filtered during the sampling period. However, despite the use of a dedicated personal computer with a 286 processor and data card, the machine was not capable of processing the volume of data necessary for these calculations. It also meant that the data acquisition time would have to be reduced to approximately 5 seconds which was insufficient for the purposes of the study.

The possibility of collecting the data at a high sampling rate and subsequently filtering the data files was also considered. In this situation, the sampling rate had to be at least double the frequency of the highest frequency to be filtered. Thus to filter noise associated with mains voltage of 50 Hz, a sampling rate exceeding 100 Hz was required. However, this also meant that the capacity of the hardware was exceeded.

As a result, a low pass filter in the form of a capacitor, was incorporated into the circuit board of each signal amplifier within the control box on the couch frame. This meant that only frequencies below 8 Hz were transmitted. Whilst information occurring at higher frequencies was no longer present following this processing stage, it permitted relevant data to be sampled at a lower frequency over longer time periods. Following the addition of capacitors to the circuit boards, a modified calibration procedure was performed to assess whether the disturbance caused by removal and replacement of the circuit boards had resulted in changes in the offset and gains of the load cells. The results demonstrated that these had not been affected.

Figure 8-13 to Figure 8-15 show pairs of force-time curves before and after the low pass filtering process was applied. Figure 8-13 a and b were recorded when a therapist applied a force to the third lumbar vertebra of a normal subject to reach the end of the range of movement, to determine the characteristics of joint excursion.

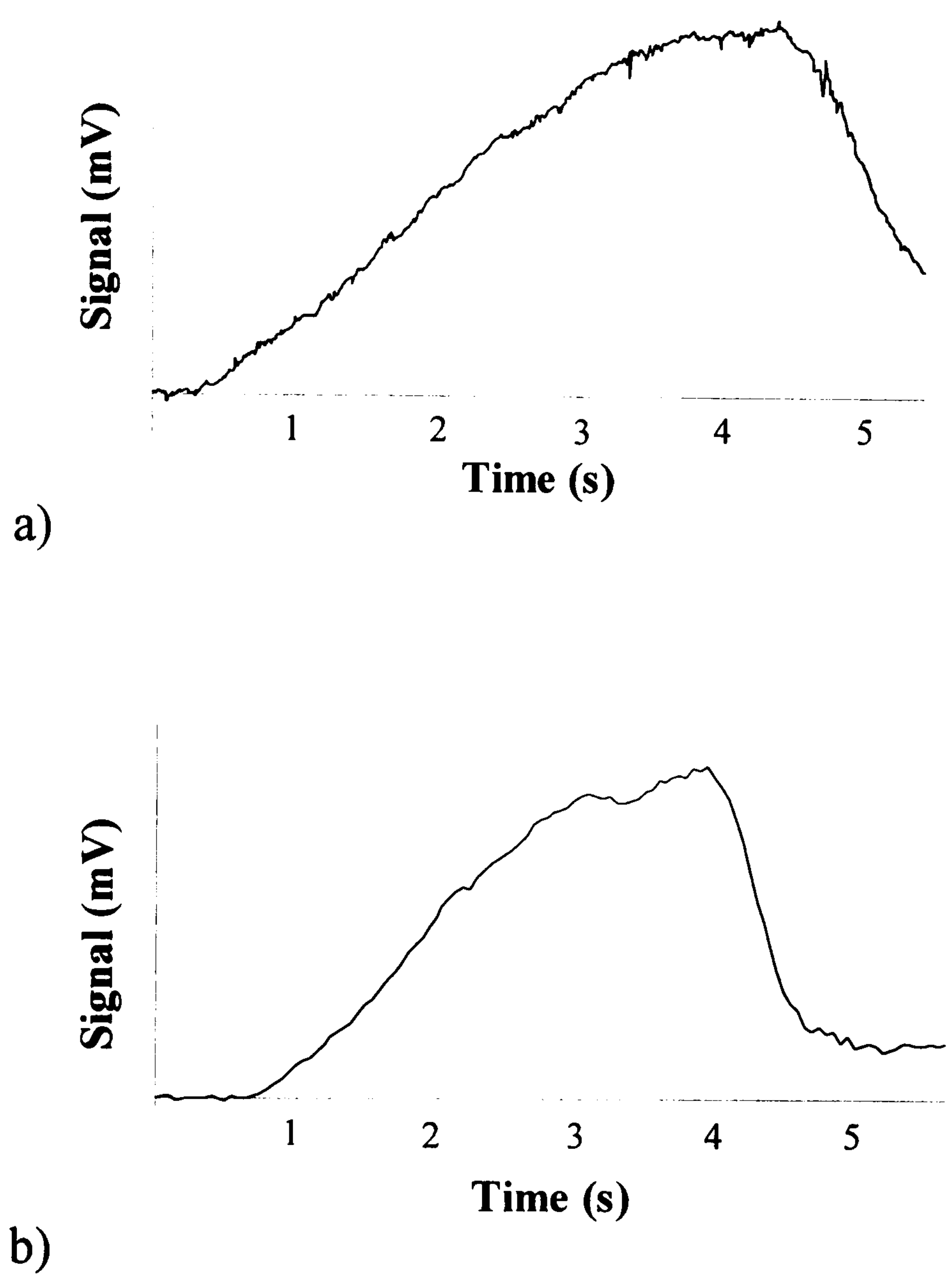
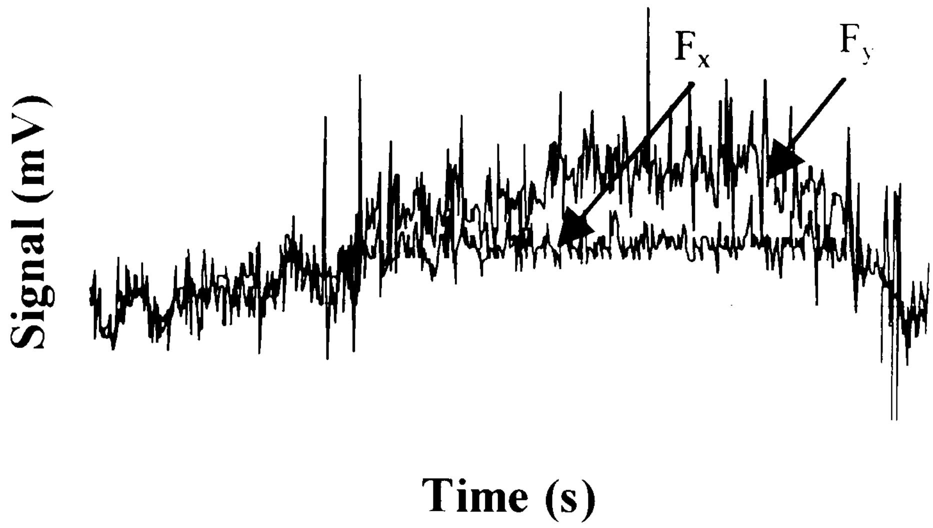


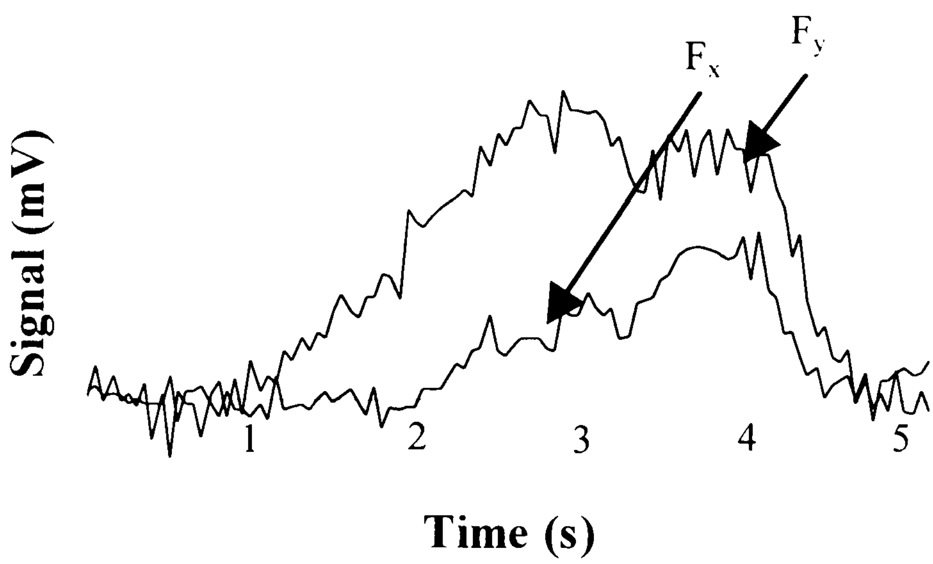
Figure 8-13: Signal recorded during mobilisation procedure (F_z)

a) before filtering process b) after filtering process

The signals recorded for F_x and F_y demonstrated greater variation than those recorded for F_z (Figure 8-14). Although the addition of low pass filters reduced the noise considerably, the data were still affected by erroneous values. This is reflected by the relative sizes of the confidence intervals reported in Table 8-16.



a)

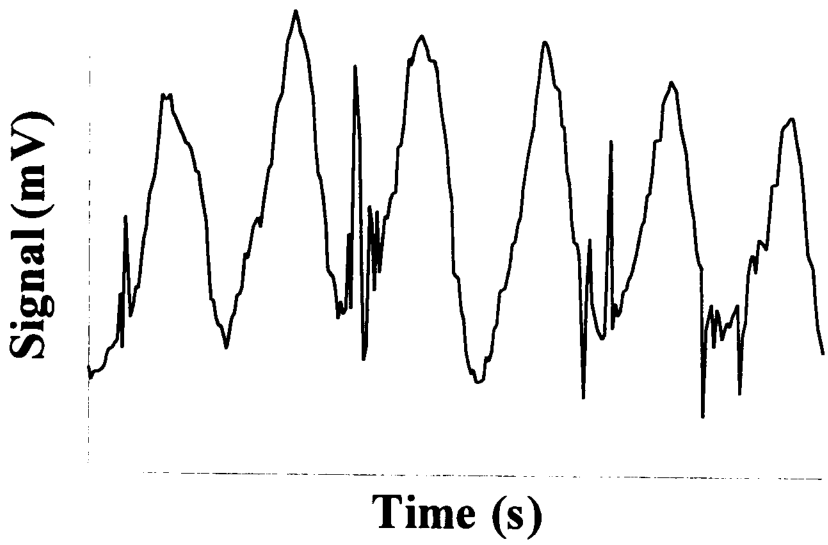


b)

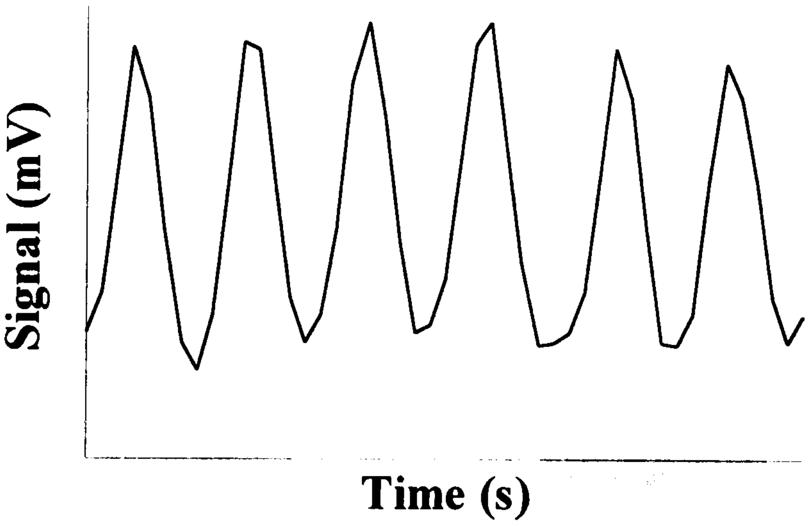
Figure 8-14: Signal recorded during mobilisation procedure (F_x and F_y)

a) before filtering process b) after filtering process

The effect of filtering the load cell signals collected during the application of a varying mobilisation force was also examined. The raw signal recorded for F_z is shown in Figure 8-15a and a similar signal after filtration in Figure 8-15b.



a)



b)

Figure 8-15: Signal recorded during a Grade II Mobilisation procedure (F_z)
a) before filtering process b) after filtering process

8.3.3 Resonance of the body

Whilst the characteristics of the couch can be determined experimentally, it is more difficult to assess the complex dynamic influence of the human body. Whilst it is neither practical nor ethical to deliver an impact force to the couch with the body interposed, some inferences can be made from the known characteristics of the body.

Where the frequency of the applied force falls within the natural frequencies of the body, resonance may occur. The majority of work on body resonance has been conducted with the body either sitting or standing, and under axial vibration. In these conditions trunk resonance is reported to be between 4-6 Hz (*Grandjean 1986*). Unloaded, the axial resonant frequency of the lumbar spine is approximately 1,200 Hz. When a mass of 40 kg representing the weight of the torso is added, resonance occurs at a lower frequency of 23-32 Hz (*Kasra et al 1992*). It seems reasonable to assume that the prone spine will fall somewhere between these two values. Kasra et al (*1992*) provides some support for this assumption. Although designed to examine conditions of axial vibration on lumbar segments, their report provides information about the resonant frequency of the lumbar segment during lateral loading, and indicates that lateral resonance of the spine exceeds 20-30 Hz.

The separation of the point of force application and the plane of measurement coupled with the large masses of the subject and couch top, may cause attenuation of the recorded force. However, it can be argued that the rigidity of the mobilisation table produced a system which was relatively stiff in compression. Coupled with the small accelerations of the couch top and subject occurring during mobilisation and the slow change in force over time, signal attenuation was likely to be minimal.

The forces may also be subject to some damping, tending to reduce their amplitude. However, Lee (*1989*) suggests that the path of force transmission is through the rib cage and pelvis which are comparatively stiff in compression. Grandjean (*1986*) also reports that frequencies are not heavily damped by tissues of the body until they exceed 30Hz. This supports the evidence of Kasra et al (*1992*) who found that when the spine was not subject to preload, a similar situation to when the subject lies prone, the damping effect was negligible.

8.4 DISCUSSION

The development of an instrumented couch to facilitate the measurement of forces applied during spinal manipulative procedures has been described. This is an important development in an area where there are few objective measures or scientific definitions of treatment practices.

Calibration procedures have demonstrated that the system exhibits characteristics which are suitable for the measurement of spinal manipulative techniques. Providing the limitations of the measurement system are accepted, which may have been overestimated due to limitations of the calibration procedures themselves, the design has potential for use in many other situations where the force delivered to a subject needsto be quantified. It is particularly appropriate for use in the assessment of manual therapy techniques.

Chapter 9: Study One: Repeatability and Reproducibility

The instrumented mobilisation couch was used to evaluate the reliability of the therapist when performing the mobilisation techniques described in Section 4.1. The study design took into account the ability of the therapist to remain consistent both within one treatment session and from week to week, as would be appropriate in the clinical situation when the patient attends for a series of appointments.

9.1 OBJECTIVES

The objectives of Study One were:

- To evaluate the consistency with which one therapist could repeat a procedure within one measurement session and again at a session two weeks later

- To establish the variation in the characteristics of the mobilisation force performed by different experienced therapists

- To determine the effect of performing one grade of mobilisation outside the normal grade sequence

9.2 PILOT STUDY

A pilot study involving 10 therapists and two models was carried out to establish the feasibility of the experimental protocol. It highlighted several methodological issues which were addressed before the main study was undertaken and are described below.

9.2.1 Differences in interpretation of Grade I mobilisation

Discussion between the therapists during the pilot study revealed a disparity in interpretation of a Grade I mobilisation. Some therapists applied a minimal force sufficient to "bend a fly's legs" in line with the common idiom (*Maitland 1986*), which was only liable to depress the overlying soft tissues. Other therapists performed a Grade I mobilisation at the beginning of joint range. This definition was adopted in the final study.

9.2.2 Couch Height

The couch height at which the therapist operated was measured for a sample of therapists, using an unmodified couch. The relationship between therapist height and preferred couch height is illustrated in Figure 9-1. The inclusion of load cells in the frame raised the couch top by 0.2 m (Figure 9-2), making it too high for many of the therapists, even at the lowest setting of 0.65 m. Using the regression equation predicted by the linear model, only therapists who were taller than 1.75 m could achieve their preferred height on the adapted couch. To compensate, the therapist stood on a platform 0.2m high, and the couch was raised until it was at the preferred height for that therapist.

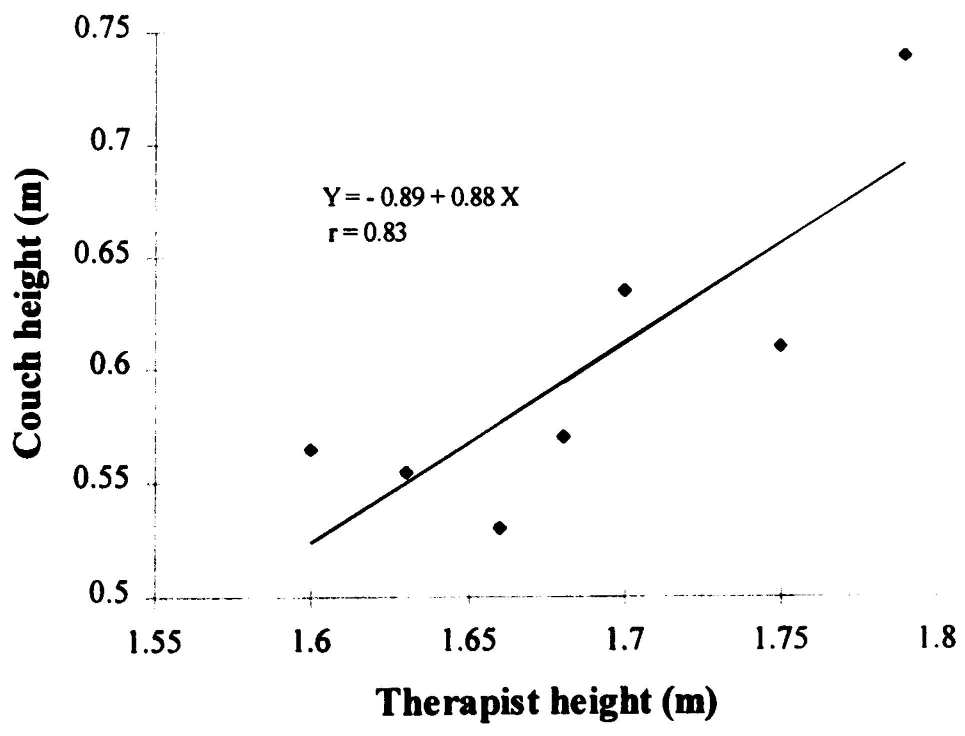


Figure 9-1: Relationship between the height of the therapist and their preferred couch height

Figure 9-2: The instrumented couch illustrating the increase in height resulting from the inclusion of load cells within the frame

9.2.3 Stand-off Bar

In normal clinical practice the therapist gains stability from leaning against the edge of the couch. As this manoeuvre would be recorded as a variable force by the load cells, a stand-off bar was attached to the front of the couch which allowed the therapist to adopt their usual supported posture, without influencing the measurement of the forces applied to the lumbar spine. In the pilot study, several therapists had commented that the bar, illustrated in Figure 7-11, which was square section tubular steel was uncomfortable to lean against. It was therefore covered in a circular sheath of high density foam which rectified the problem of discomfort.

9.3 EXPERIMENTAL DESIGN

An evaluation of the consistency with which one therapist can apply a force over two repetitions requires an experimental model with 30 degrees of freedom (*Chinn 1990*). Therefore, thirty therapists were employed who repeated the test procedure twice within the same measurement session (Table 9-1). In a subsequent study, time was introduced as a variable, with each therapist repeating the procedure approximately two weeks after the first measurement session, under similar experimental conditions. A comparable experimental model was used (Table 9-1).

Table 9-1: Experimental design and degrees of freedom used in the assessment of repeatability and reproducibility

Parameter	Study 1 (Repeatability)	Study 2 (Reproducibility)
Therapists (n)	30	30
Sessions (s)	1	2
Repetitions (r)	2	1
Degrees of freedom	30 i.e.: n.x (r - 1)	30 i.e.: n.x (s - 1)

9.3.1 Therapists

It has been reported that with experience, therapists become more consistent in the application of a specific technique *(Jull 1988)*. Therefore, only therapists with a minimum of three years postgraduate experience, employed in a Senior post, who used mobilisation techniques regularly, participated in the study. The majority were either members, eligible for membership or involved in the training which would lead to membership of the Manipulative Association of Chartered Physiotherapists.

9.3.2 Model

In order to compare all the therapists under similar circumstances and minimise the sources of variation, one model was used. Because it was not known whether intensive periods of spinal mobilisation would cause soft tissue fatigue in the model, the period of data collection was spread over several weeks. The model was subjected to a maximum of two sessions each week, to allow any effects of mobilisation to dissipate. The pilot study had involved two models, neither of whom reported any discomfort following the test procedures.

9.4 PROTOCOL

Two techniques were included in the evaluation:

1. Postero-anterior central spinal mobilisation

- Grade I
- Grade II
- Grade III
- Grade IV

2. Displacement of joint through full range of translatory movement

- End Feel

9.4.1 Standardisation

The experimental environment had been designed to avoid disrupting the technique or behaviour of the therapist during the testing procedure. Each therapist was allowed to use thumb or pisiform techniques according to preference. In the later technique, the hypothena eminence of one hand contacts the spinous process of the appropriate vertebra and is reinforced by the second hand (*Maher & Adams 1996*). The following conditions were standardised:

- Time of day
- Vertebral level
- Order of technique performance
- Verbal instructions given to therapist
- Data collection protocol

The third lumbar vertebra (L3) was selected because of its central position in the lumbar lordosis which suggests that anterior translation, and hence the direction of force application would be mainly along the vertical axis. (Figure 9-3). It is also subject to fewer anomalies than the other lumbar vertebrae (*Tulsi & Hermanis 1993*), is central in the lumbar curve and is relatively symmetrical in structure (*Twomey & Taylor 1987*).

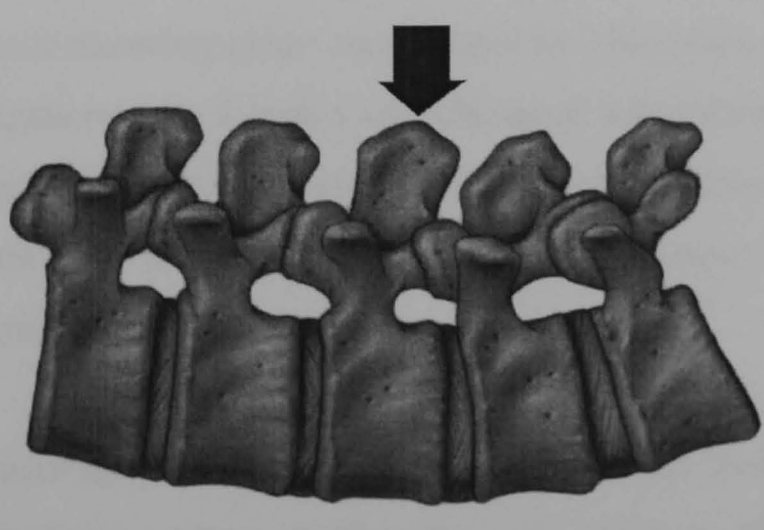


Figure 9-3: The direction of an anteroposterior mobilisation on the spinous process of L3

Therapists were allowed a period of familiarisation at each grade of mobilisation, usually lasting approximately five seconds, until they indicated that they were ready for the recording to begin. This allowed the therapist to reach their normal rhythm whilst preconditioning the vertebrae. Therapists were screened from the computer VDU to avoid any influence resulting from visual feedback on either the magnitude of the forces or the rate of application. No feedback was provided by the model or the assessor.

9.5 DATA COLLECTION

The data collection protocol was based on the following considerations:

- Minimisation of the time period in which the spine of the model was mobilised
- Collection of sufficient samples to reconstruct a mobilisation force waveform without loss of information especially at the peaks and the troughs
- Standardisation of the format of the output data file
- Limitations of the computer processor/memory

The load cells were sampled at 40 Hz for periods of 10 seconds during the application of each technique. Data were collected over 12 phases of 10 seconds at the first measurement session (termed A) and 7 phases of 10 seconds at the second measurement session, B. Details of the procedures are given in Figure 9-4. The software was configured so that each recording phase was initiated by a key press on the PC by the assessor, at the instigation of the therapist and terminated automatically after 10 seconds. This ensured that a standard number of data samples, equivalent to 400 samples for each load cell, were recorded for each 10 second phase. This was important for subsequent analysis procedures.

Data collection for each therapist was preceded by recording the baseline activity of the load cells with the model in position, to allow the static load to be offset in subsequent calculations.

Study 1

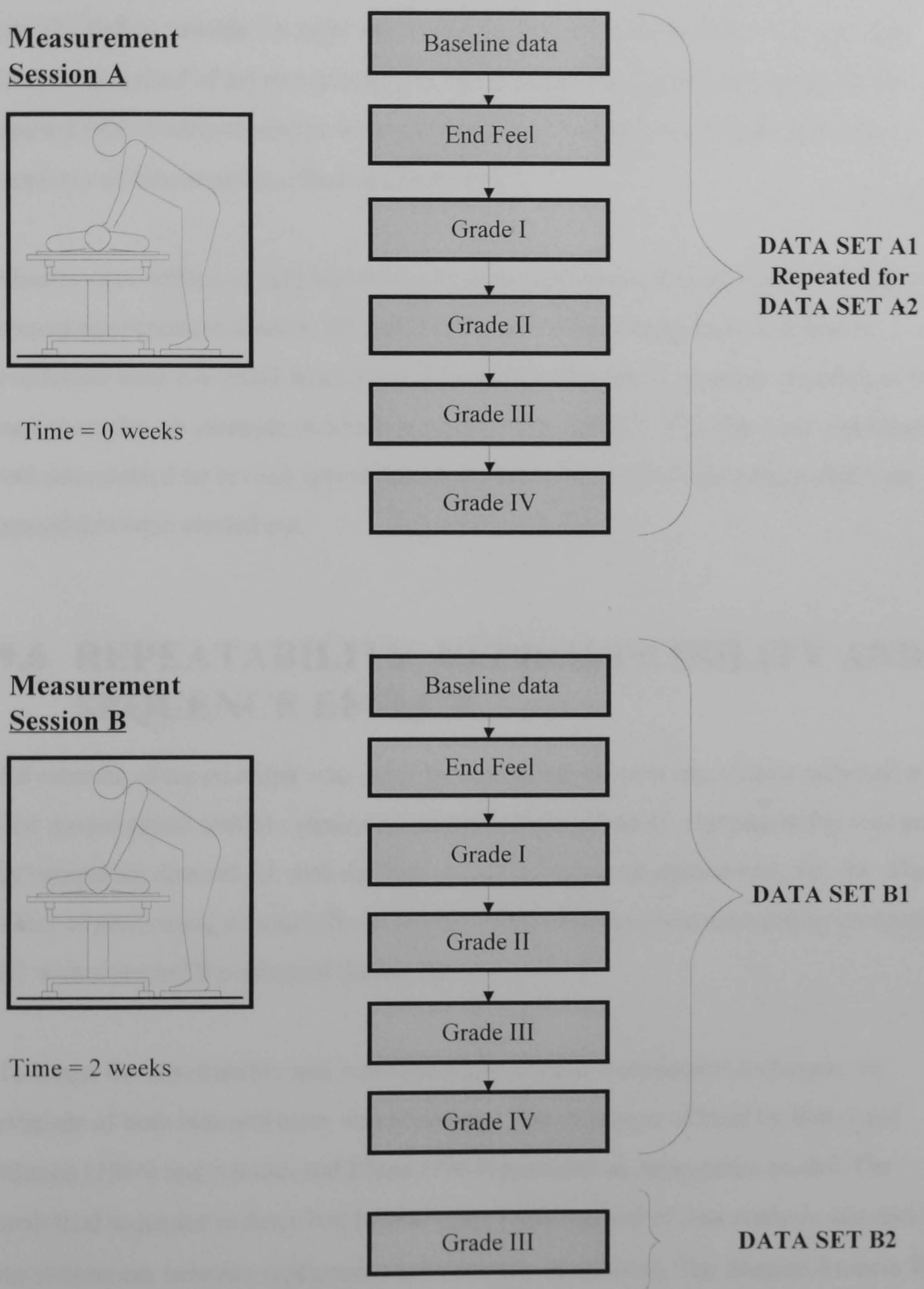


Figure 9-4: Flow diagram of the experimental protocol and sequence of data collection

9.5.1 Preliminary analysis

Data files acquired through the software package, Labtech Notebook, were translated into the format suitable for input into Excel, as discussed in Section 7.4.2. The data format comprised of seven columns corresponding to the six load cell channels and elapsed time. Fourteen macros were compiled within Excel, which performed the sequence of functions described in Figure 9-5.

Macros were written to perform similar functions on data collected during A2 and at the second measurement session, B1 and B2. Values of force magnitude and rate of oscillation were estimated from the graphs and entered onto a separate spreadsheet for each therapist, an example of which is given in Appendix 3. The data were condensed and summarised on several spreadsheets (Appendix 4), where appropriate statistical procedures were carried out.

9.6 REPEATABILITY, REPRODUCIBILITY AND SEQUENCE EFFECT

An estimate of repeatability was made by comparing the two sets of data collected at the first measurement session, phases A1 and A2. An estimate of reproducibility was made by comparing data set A1 with the first measurement set of session two, viz. B1. The effect of performing a Grade III out of the normal sequence was assessed by comparing B2 with a Grade III performed during B1.

To assess the repeatability and reproducibility of each mobilisation technique, an estimate of both bias and error was necessary. The technique offered by Bland and Altman (*1986*) and Altman and Bland (*1983*) provided an appropriate model. The analytical sequence is described below. Use of this method of data analysis assumes that the differences between replications are normally distributed. The Shapiro-Francia W' test of non-Normality was used to determine whether the data were compatible with a null hypothesis of Normality (*Altman 1991*).

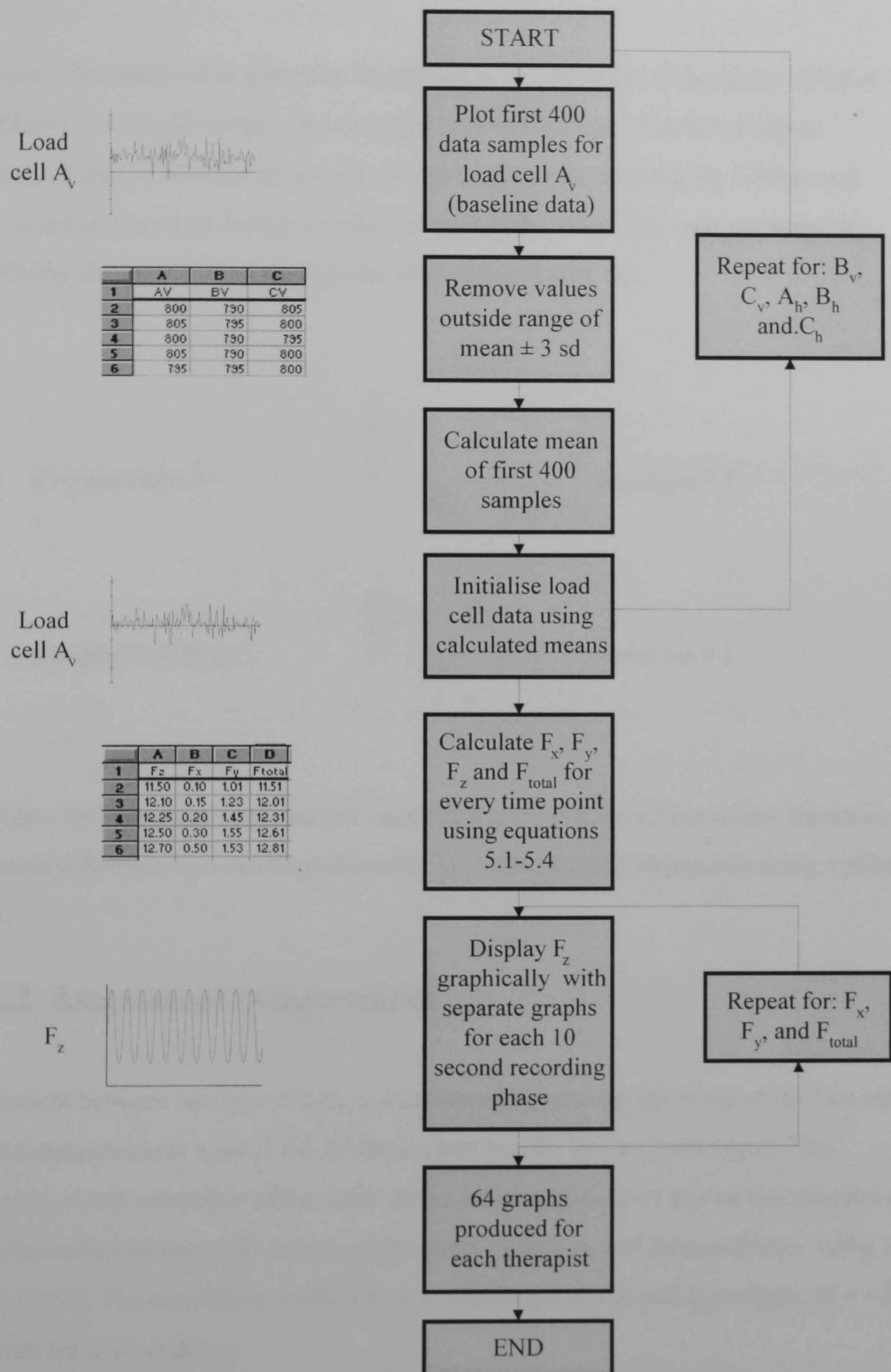


Figure 9-5: Sequence of events performed to prepare data set A1 for analysis

9.6.1 Bias

The data were analysed to determine the presence, or otherwise, of any order effect or condition in the model which systematically altered with time. The data analysis protocol evaluated whether the measured forces changed systematically during each measurement session or during the whole period of the study. Bias was estimated by calculating the mean difference between the two repetitions (*d*).

thus: $d\,(\text{repeatability}) = \dfrac{\sum_{i=1}^{30} A2_i - A1_i}{30}$ equation 9.1

$d\,(\text{reproducibility}) = \dfrac{\sum_{i=1}^{30} B1_i - A1_i}{30}$ equation 9.2

The same therapists and experimental conditions were used on all occasions, therefore the mean difference was examined formally by testing the null hypothesis using a paired t-test.

9.6.2 Assessment of agreement

Agreement between two sets of data was examined by plotting the mean of the first and second measurements against the difference between the two measurements. This allowed a visual estimation of the value of the measurement error and an examination of the relationship between this value and the mean of each pair of measurements, using a linear model. The correlation coefficient was tested against the null hypothesis of $r = 0$ to check for independence.

The standard deviation of the differences (s) was calculated. Provided the differences were normally distributed, which was determined by calculating the Shapiro-Wilks W', probability theory dictates that 95 per cent of these differences will lie between $d \pm 1.96s$. These boundaries define the limits of agreement as described in BS5497(*1987*) and can be used to describe the error in a set of measurements.

9.7 RESULTS

Unless stated, all results are based on the magnitude of the total force vector (F_{total}), as calculated in equation 7.4. The data for force magnitude is included in Appendix 4, Tables A4-1 to A4-8. The corresponding data for frequency of oscillation is shown in Table A4-9.

9.7.1 Characteristics of the Model and Therapist Group

One male physiotherapist aged 26 years, of average build (64 kg, 1.70m) in good general health, acted as the model for all measurement sessions. Schrobers method for measuring spinal movement was used to provide an indication of spinal mobility of the model. Using this method, which is described in Section 10.2.2, the lumbar spine range of movement of the model was measured as 76mm. The characteristics of the therapists employed in the study are summarised in Table 9-2.

9.8 ASSESSMENT OF REPEATABILITY AND REPRODUCIBILITY

The ability of each therapist to replicate a given force under similar conditions was evaluated. The two sets of data collected within the first measurement session were used to assess repeatability. The first set of data collected at the first measurement session (A1) and the first collected at the second session (B1) were used to assess reproducibility. For each therapist, session B was an average of 2 weeks after session A. Table 9-3 summarises the results of the paired t-tests used to assess bias between replications.

Table 9-2: Physical characteristics and experience of the therapists participating in the study

Therapist Number	Qualified(yrs)	Height (m)	Weight (kg)	Out-Patient experience (yrs)	Use of mobilising techniques
1	10	1.63	60	6.0	Daily
2	6	1.68	63	4.3	Daily
3	21	1.68	60	11.7	Monthly
4	4	1.67	58	1.6	Daily
5	5	1.65	65	2.8	Daily
6	4	1.60	58	2.3	Daily
7	11	1.70	62	8.8	Daily
8	6	1.75	70	4.0	Weekly
9	4	1.88	83	2.4	Daily
10	10	1.58	47	8.5	Daily
11	6	1.75	71	0.9	Daily
12	12	1.70	57	5.4	Daily
13	5	1.68	59	3.2	Daily
14	8	1.65	80	3.0	Weekly
15	25	1.58	58	4.5	Weekly
16	10	1.60	57	8.0	Daily
17	9	1.73	67	7.0	Weekly
18	4	1.60	60	0.4	Daily
19	9	1.68	64	4.5	Daily
20	8	1.58	55	4.2	Daily
21	8	1.60	53	4.0	Daily
22	7	1.68	64	3.0	Daily
23	35	1.63	70	30.0	Weekly
24	7	1.65	60	4.0	Daily
25	7	1.63	60	1.2	Daily
26	19	1.70	73	5.0	Daily
27	5	1.75	62	1.3	Daily
28	8	1.70	86	5.5	Weekly
29	5	1.68	64	1.0	Daily
30	3	1.63	57	1.0	Daily
Mean	9.4	1.67	63.4	5.0	
S.D	7	0.07	8.6	5.4	
Range	3 - 35	1.58 -1.88	47 - 86	0.4 - 30	

Table 9-3: Data to illustrate assessment of bias between sets of repeated measurements comparing A1 with A2, and A1 with B1

Grade of Mobilisation	A1 mean (N)	A2 mean (N)	B1 mean (N)	*d* repeatability (N)	*d* reproducibility (N)
End Feel	198	203	201	5 (0 - 10) ns	9 (-3 - 12) **ns**
Grade I	35	42	34	7 (5 - 9) **	-1 (-9 - 7) ns
Grade II	50	58	52	9 (7 - 11) **	2 (-7 - 11) ns
Grade III	137	135	132	-2 (-7 - 3) ns	-5 (-20 - 10) ns
Grade IV	158	155	156	-3 (-8 - 2) ns	-3 (-17 - 11) ns

95% Confidence interval in parenthesis,

** = $p < 0.01$, ns = not significant at 5% level

The 95% confidence intervals around the mean differences are given by:

$$SEdiff = \sqrt{\frac{s^2}{n}} \qquad \text{equation 9.3}$$

$$95\% \text{ Confidence interval} = \pm\ t \times SEdiff \qquad \text{equation 9.4}$$

where:

SEdiff = Standard error of the mean difference

s = standard deviation of the differences between two sets of measurements

n = number of pairs of observations (n = 30)

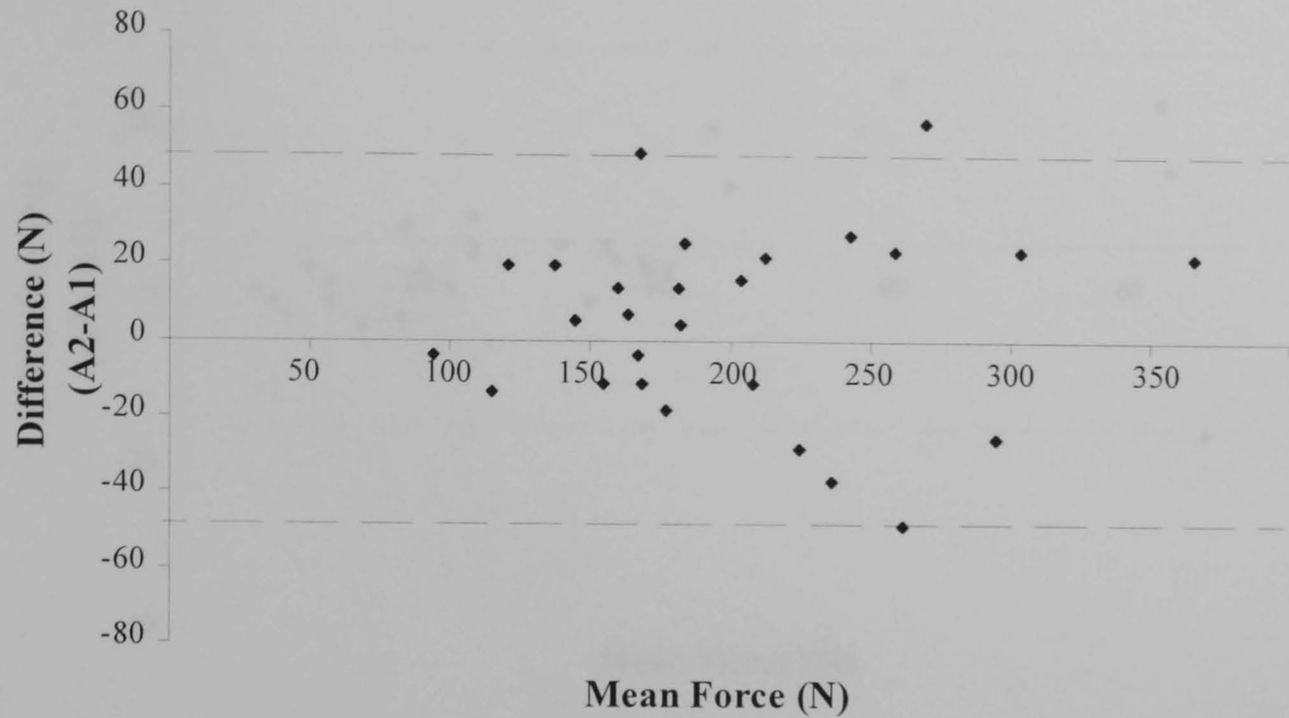
t = is the appropriate value on the t distribution with (n-1) degrees of freedom (df =29, p = 0.05, t = 2.045)

The results demonstrate that there was no significant bias between A1 and A2 for the higher grades of mobilisation and for the End Feel. The analysis indicates a significant increase at the 1 per cent level in the mean force used for a Grade I and Grade II mobilisation on the second measurement (A2). The bias, present in the comparison of the lower grades of mobilisation may have been the result of soft tissues changes in the FSU, induced by the first set of mobilisations. The results suggest that relatively small forces used during Grades I and II may have been sensitive to this change, where the higher grades were not. This indicates that it is more appropriate to use the higher grades of mobilisation to assess reliability because they are less prone to systematic error.

The assessment of bias between the measurements from the two sessions A1 and B1 exhibited no statistically significant differences at the 5 per cent level between repetitions. This suggests that any changes in the soft tissues that had occurred during the first set of mobilisations had returned to their baseline condition before the second measurement session.

For every grade of mobilisation, the relationship between the mean of the two measurements for each therapist and the difference between them was assessed. Separate comparisons were undertaken for repeatability and reproducibility. Figure 9-6 to Figure 9-10 summarise the assessment of repeatability and reproducibility. They illustrate the difference between the two replications plotted against the average measurement, for each grade of mobilisation. The limits of agreement were calculated as $d \pm 1.96s$ and are shown as horizontal lines. Mean force is calculated as:

$$\text{Mean force} = \left(\frac{A1 + A2}{2}\right) \qquad \text{equation 9.5}$$



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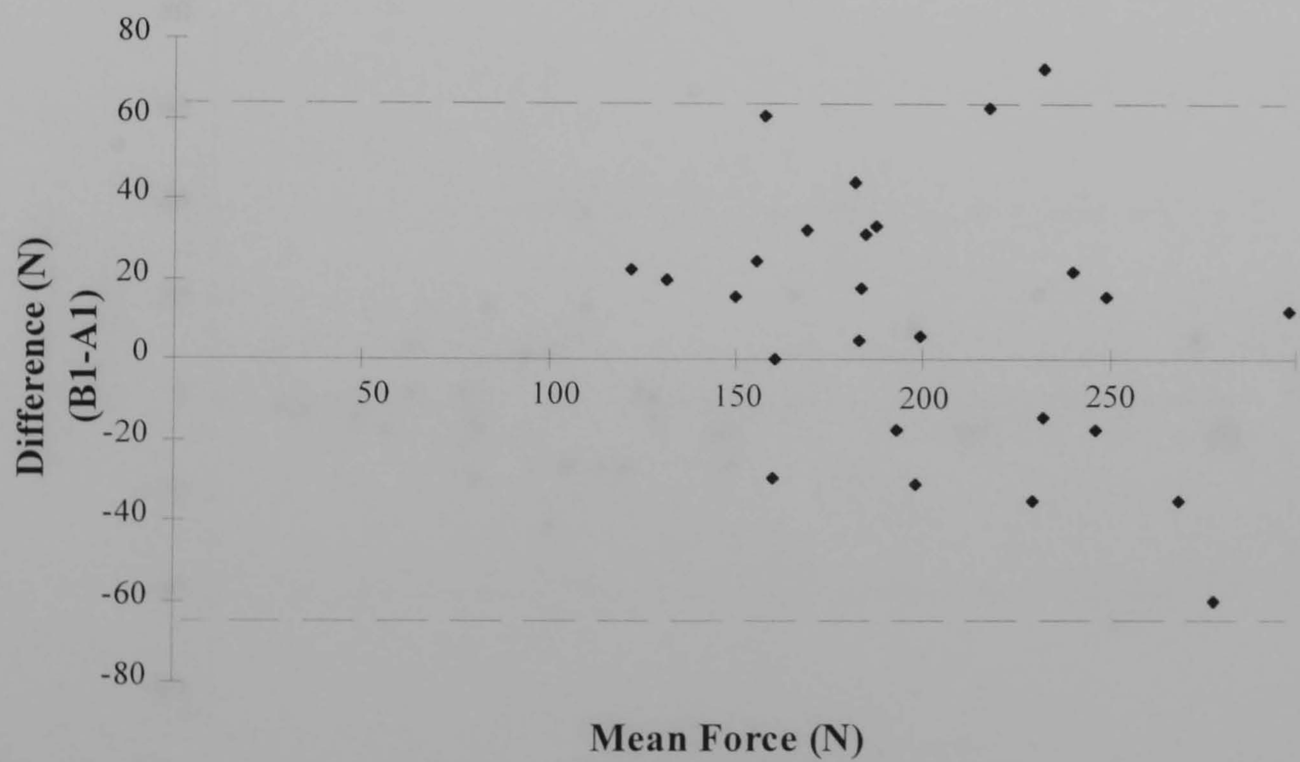
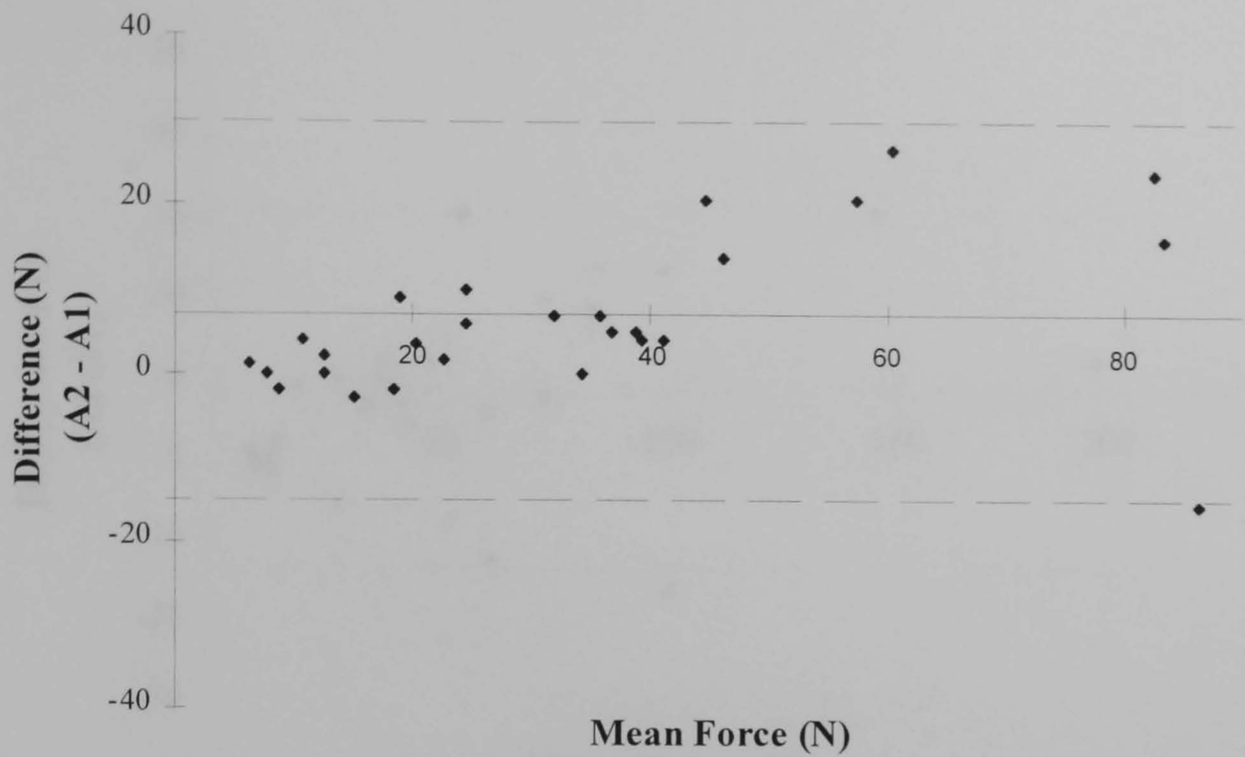
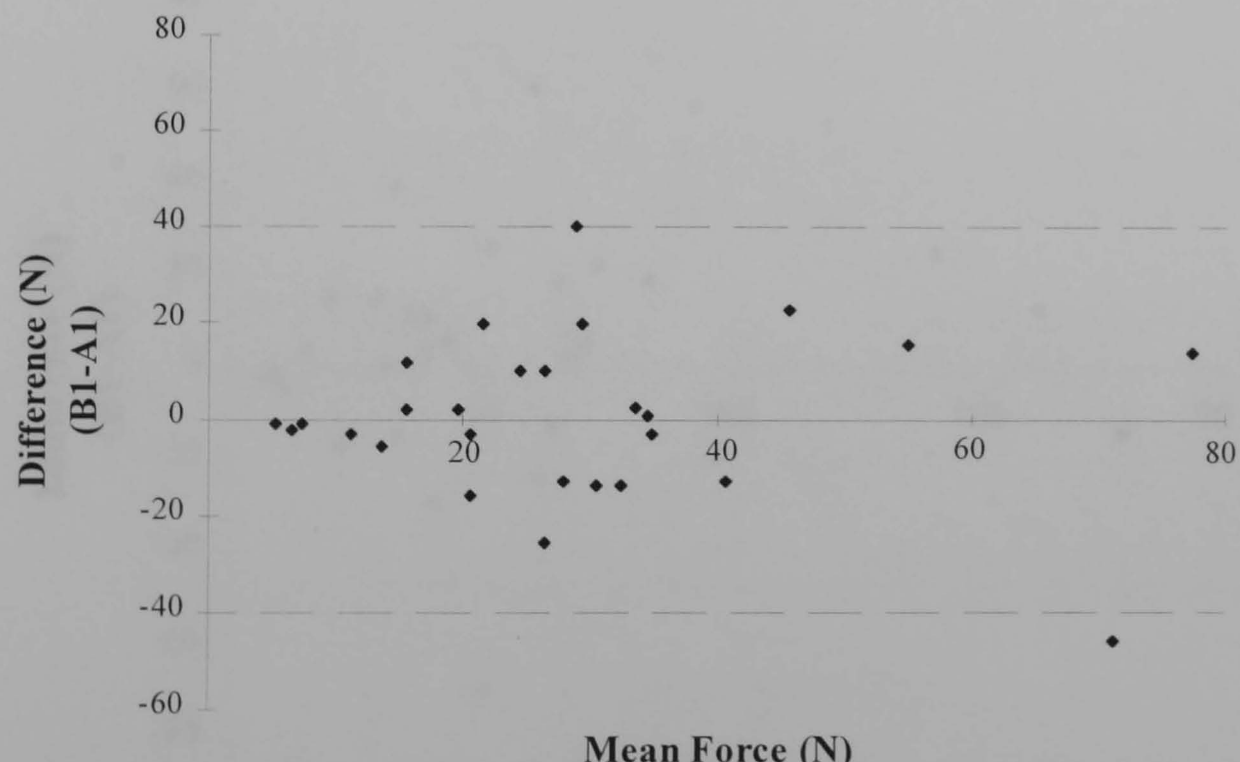


b)

Figure 9-6: Comparison of data collected during measurement sessions A and B to show the difference between two replications plotted against the average measurement for a) repeatability and b) reproducibility of forces applied during End Feel

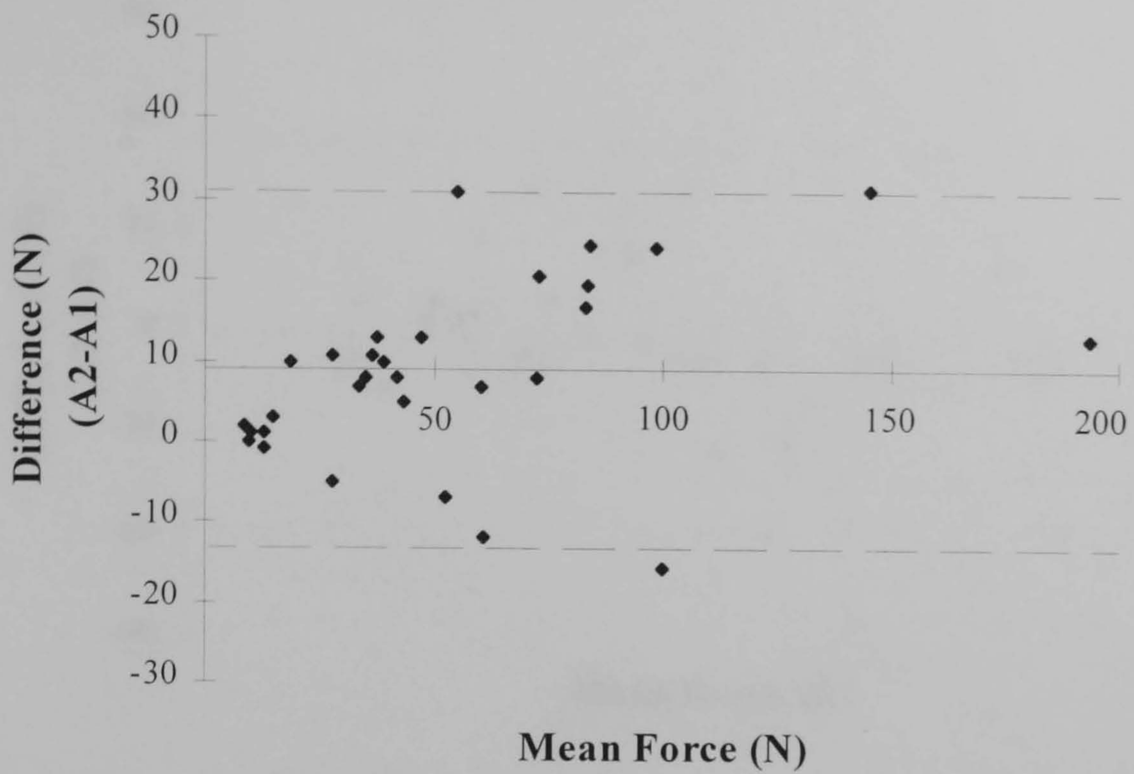


a)

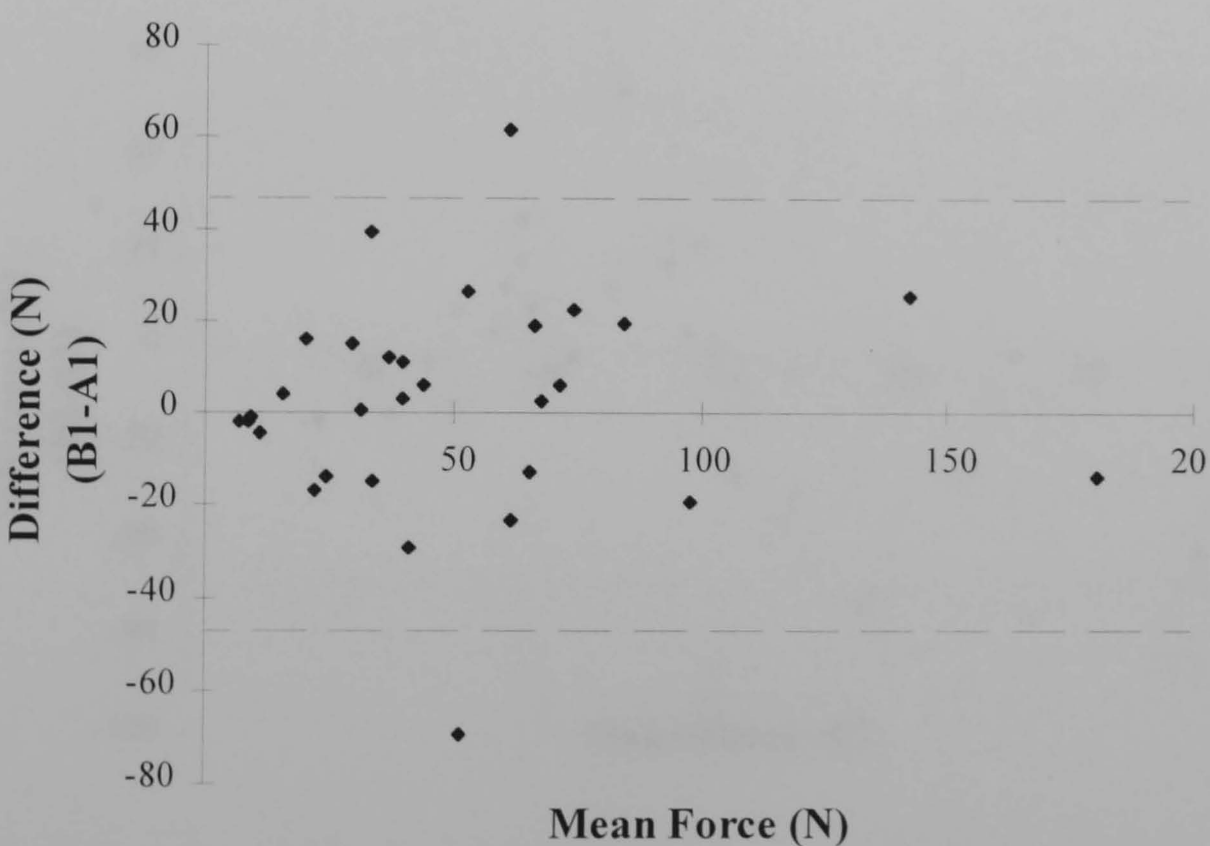


b)

Figure 9-7: Comparison of data collected during measurement sessions A and B to show the difference between two replications plotted against the average measurement for a) repeatability and b) reproducibility of forces applied during Grade I

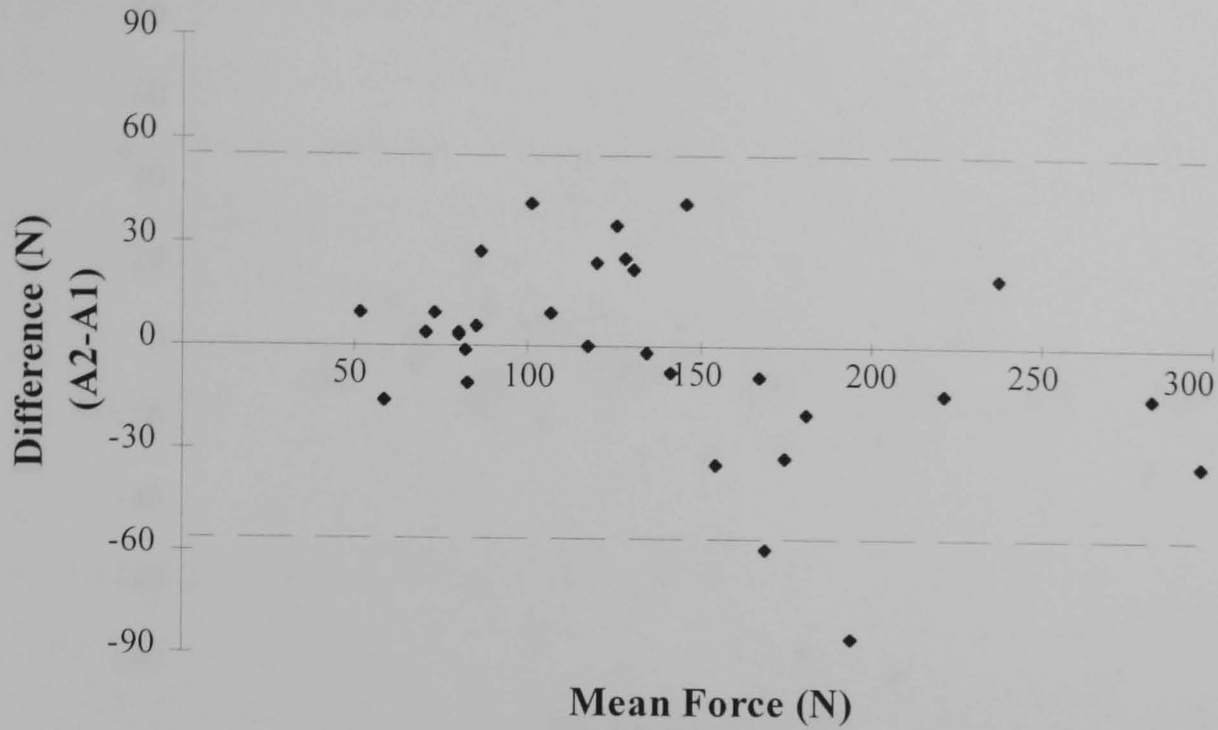


a)

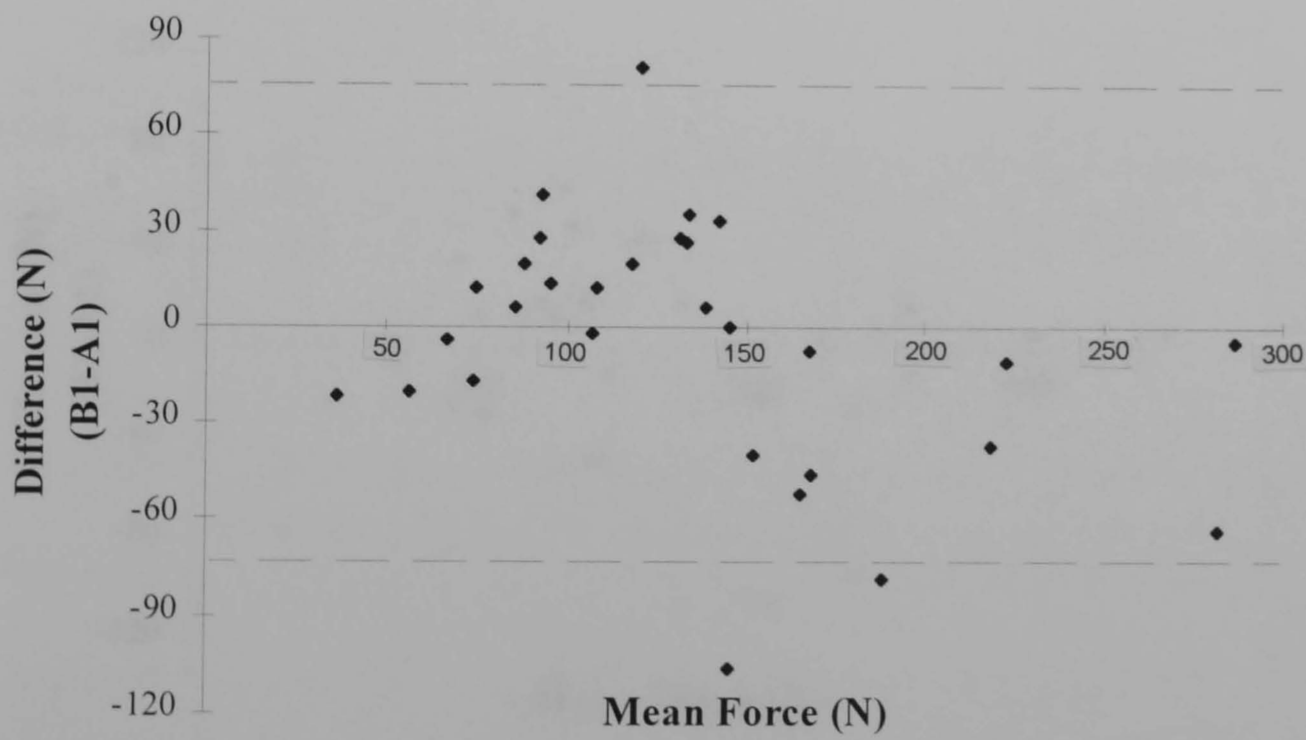


b)

Figure 9-8: Comparison of data collected during measurement sessions A and B to show the difference between two replications plotted against the average measurement for a) repeatability and b) reproducibility of forces applied during Grade II

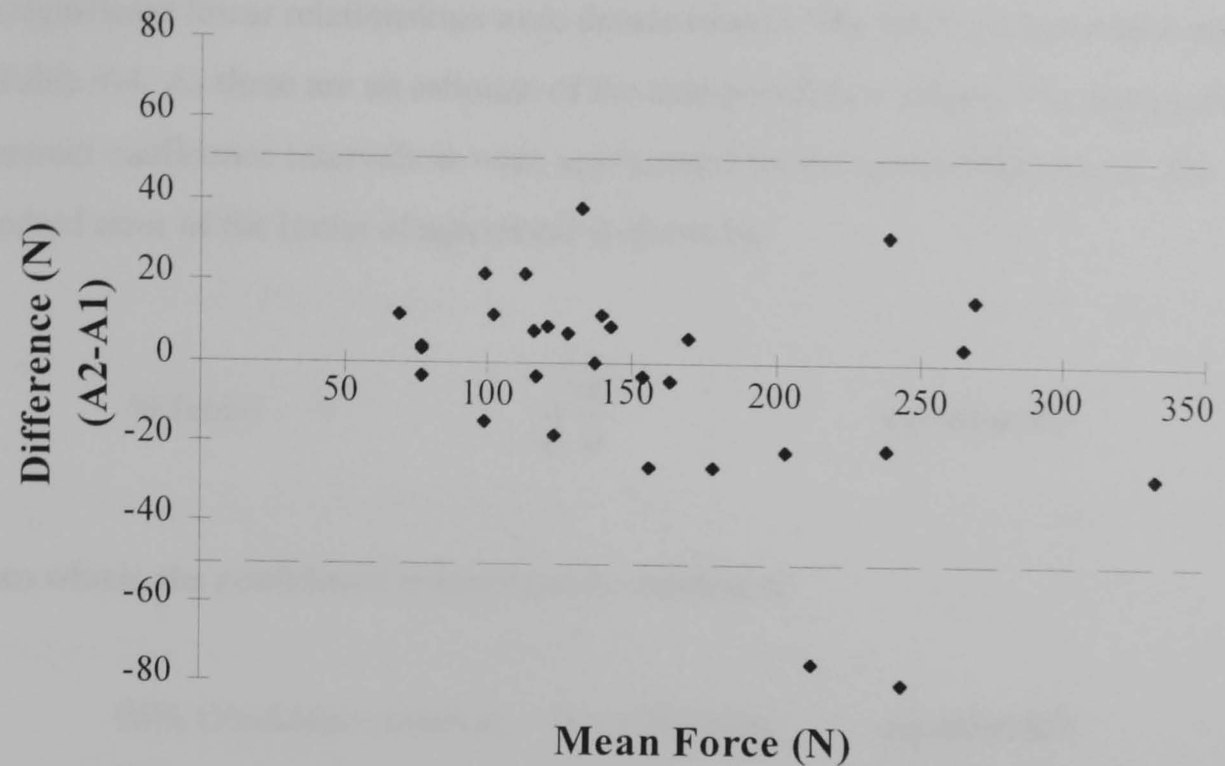


a)

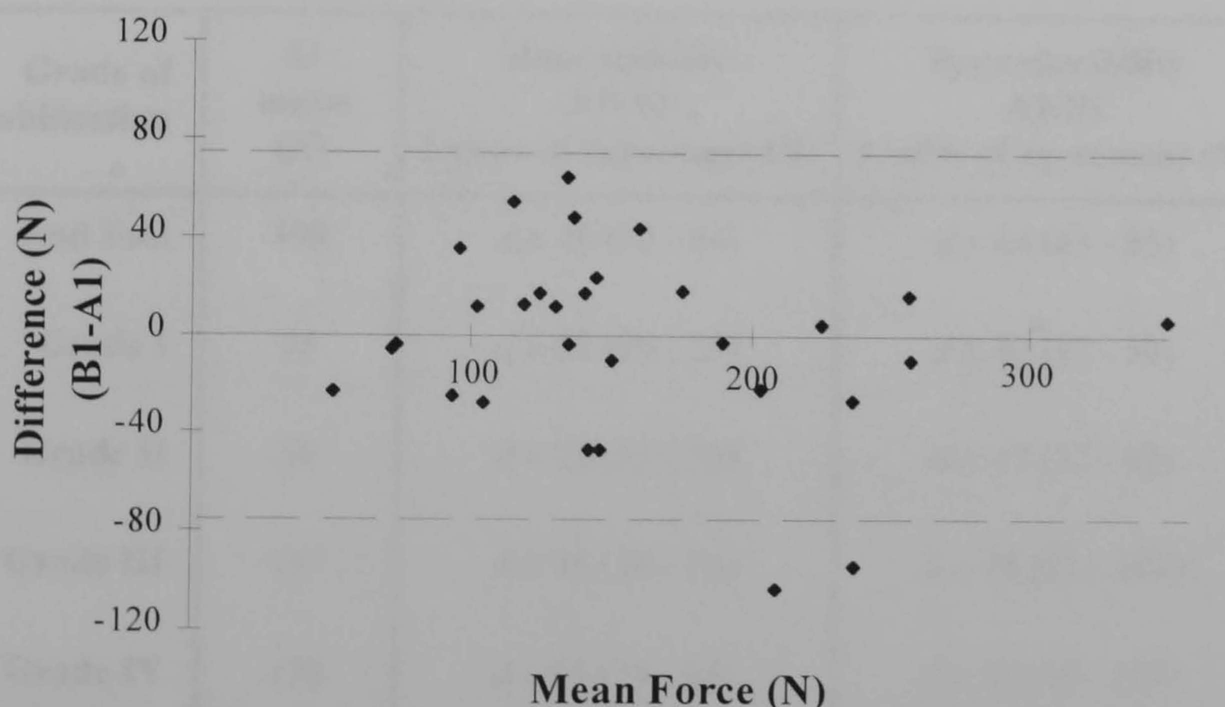


b)

Figure 9-9: Comparison of data collected during measurement sessions A and B to show the difference between two replications plotted against the average measurement for a) repeatability and b) reproducibility of forces applied during Grade III



a)



b)

Figure 9-10: Comparison of data collected during measurement sessions A and B to show the difference between two replications plotted against the average measurement for a) repeatability and b) reproducibility of forces applied during Grade IV

No significant linear relationships were demonstrated. The limits of agreement are given in Table 9-4. As these are an estimate of the true population values, it is appropriate to construct confidence intervals as were constructed for the mean difference d. The standard error of the limits of agreement is given by:

$$\text{SElimits} = \sqrt{\frac{3s^2}{n}} \qquad \text{equation 9.6}$$

From which, the confidence interval can be estimated:

$$95\% \text{ Confidence interval} = \pm t\, x \text{ SElimits} \qquad \text{equation 9.7}$$

Table 9-4: The limits of agreement for each grade of mobilisation for repeatability and reproducibility

Grade of mobilisation	A1 mean (N)	Repeatability A1/A2 Limits of agreement (N)	Reproducibility A1/B1 Limits of agreement (N)
End Feel	198	$d \pm 48$ (32 - 64)	$d \pm 64$ (43 - 85)
Grade I	35	$d \pm 22$ (15 - 29)	$d \pm 40$ (27 - 53)
Grade II	50	$d \pm 22$ (15 - 29)	$d \pm 47$ (32 - 62)
Grade III	137	$d \pm 56$ (38 - 75)	$d \pm 76$ (51 - 101)
Grade IV	158	$d \pm 51$ (34 - 68)	$d \pm 75$ (50 - 100)

It can be seen from Table 9-4 that on 95 % of occasions the force applied on the second repetition of a Grade III mobilisation was within ± 56 N of that applied during A1. When this is considered in light of the magnitude of the mean maximum force of 137 N, 56 N represents 40 per cent of the applied force. For Grades I, II, IV and End Feel this represented 63%, 44%, 32% and 24% of the mean force, respectively. The limits of agreement tended to be wider for reproducibility than for repeatability data.

This indicated that therapists became less reliable as the time between replications increased. For a Grade III mobilisation, on 95 per cent of occasions the force applied during B1 would be within ± 76 N of that applied during A1, representing ± 55% of the average force. For Grades I, II, IV and End Feel this represented 114%, 94%, 47% and 32 % of the mean force, respectively. In some cases, particularly for the lower grades of mobilisation, the error was greater than the mobilisation force applied during A1.

Therapists were relatively consistent in the frequency of oscillation between the first and second repetitions. Table 9-5 shows the mean frequency of oscillation of the mobilisation force calculated during each set of mobilisations.

Table 9-5: The mean frequency of oscillation recorded for each grade at each measurement session

Grade of Mobilisation	Frequency of Oscillation (Hz)			
	A1	A2	B1	B2
Grade I	1.1 (1.1-1.2)	1.2 (1.1-1.3)	1.2 (1.1-1.3)	
Grade II	1.1 (1.0-1.2)	1.1 (1.0-1.2)	1.1 (1.0-1.2)	
Grade III	1.1 (1.0-1.2)	1.2 (1.0-1.3)	1.2 (1.0-1.3)	1.3 (1.1-1.4)
Grade IV	1.5 (1.4-1.6)	1.5 (1.4-1.6)	1.6 (1.4-1.8)	

95 % confidence interval in parenthesis

An alternative presentation of the data involved estimating the percentage of therapists whose frequency of oscillation remained within 10% and 20% of the rate recorded at A1. Table 9-6 indicates that, when comparing A1 with A2, between 67 % and 87 % of therapists remained within 10 % of their original rate, and 93-100 % remained within 20%. When B1 was compared with A1, the recorded frequencies demonstrated greater variation, especially for Grades III and IV.

Table 9-6 : Repeatability and reproducibility of frequency of oscillation for each grade of mobilisation

Grade of Mobilisation	Repeatability		Reproducibility	
	A2 within 10% of frequency recorded at A1	A2 within 20% of frequency recorded at A1	B1 within 10% of frequency recorded at A1	B1 within 20% of frequency recorded at A1
Grade I	67 %	97 %	67 %	80 %
Grade II	57 %	93 %	63 %	80 %
Grade III	73 %	97 %	67 %	87 %
Grade IV	87 %	100 %	53 %	77 %

9.8.1 Sequence Effect

The effect of changing the sequence in which the mobilisations were performed was assessed during the second measurement session. A Grade III mobilisation was performed in isolation at the end of the session (B2). A comparison of B1 and B2 demonstrated significant bias of 34 N (Table 9-7). The mean difference between replications was far greater than that observed between repeated measurements of a Grade III mobilisation performed in the normal sequence, which differed by only 2-5 N (Table 9-3).

Table 9-7: Bias and limits of agreement between replications when a Grade III was performed out of sequence

Grade of Mobilisation	B1 mean (N)	B2 mean (N)	*d* (N)	Limits of agreement (N)
Grade III	132	166	34 ** (26 - 42)	± 44 (30 - 58)

95% confidence interval in parenthesis ** $p < 0.01$

The data indicate that on 95% of occasions, a Grade III performed out of sequence can be between 10 N smaller and 78 N greater (34 ± 44 N) than that applied when the mobilisation is performed after an End Feel, Grade I and II.

Analysis of the frequency of mobilisation demonstrated that 63% of therapists repeated a Grade III within 10% of the original rate, and 83% repeated it within 20% of the original rate. This suggested a slightly greater variation than shown by the assessment of reproducibility illustrated in Table 9-6.

9.9 GRADES OF MOBILISATION

Because bias was demonstrated in some of the lower grades of mobilisation, data from the three replications could not be combined. The data collected during A1 was used in all the subsequent calculations.

9.9.1 Normality

The data were analysed to examine the nature of their distribution using the Shapiro-Francia W' test of non-Normality. Figure 9-11 and Figure 9-12 illustrate two cumulative frequency distributions, the former demonstrating a positive skew and the latter compatible with a normal distribution. For frequency of oscillation during a Grade I mobilisation, illustrated in Figure 9-12, W' = 0.98 which is larger that the critical value of 0.93 signifying that the distribution is compatible with normality. The data associated with frequency of oscillation for the other grades was also normally distributed with W' values between 0.97 and 0.98. However, for the maximum force values exerted during Grade I mobilisation illustrated in Figure 9-11, W' = 0.76 which was smaller than the critical value and indicated a departure from normality. For End Feel, the data were normally distributed (W' = 0.93), but for Grades II, III and IV the data were skewed with W' values of 0.84, 0.90 and 0.92 respectively. Where the data were not normally distributed, non-parametric statistical analyses were used.

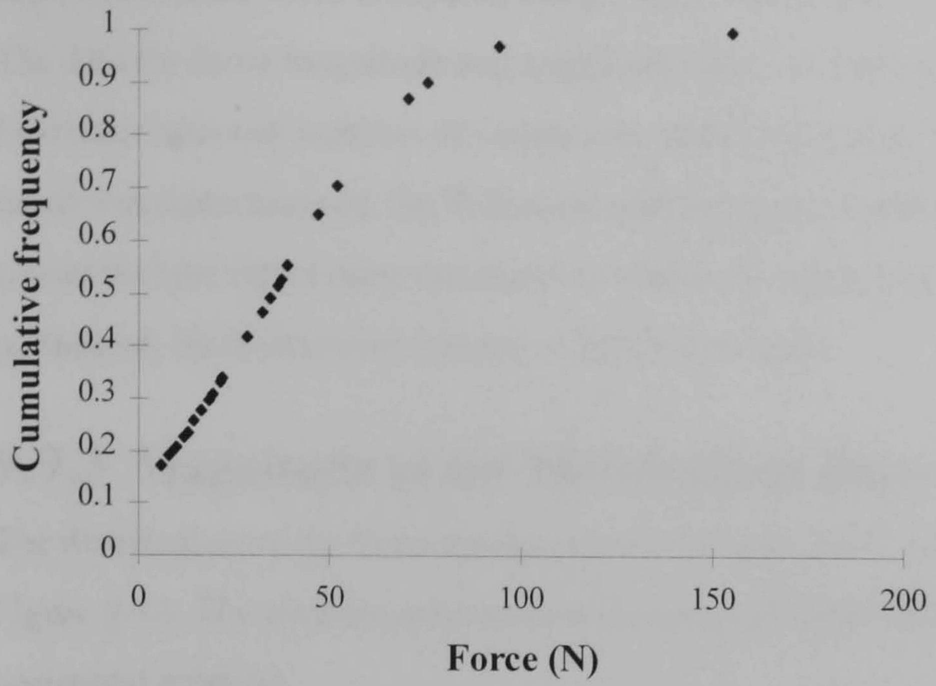


Figure 9-11: The cumulative frequency distribution of the maximum force exerted during a Grade I mobilisation (n=30, W' = 0.76)

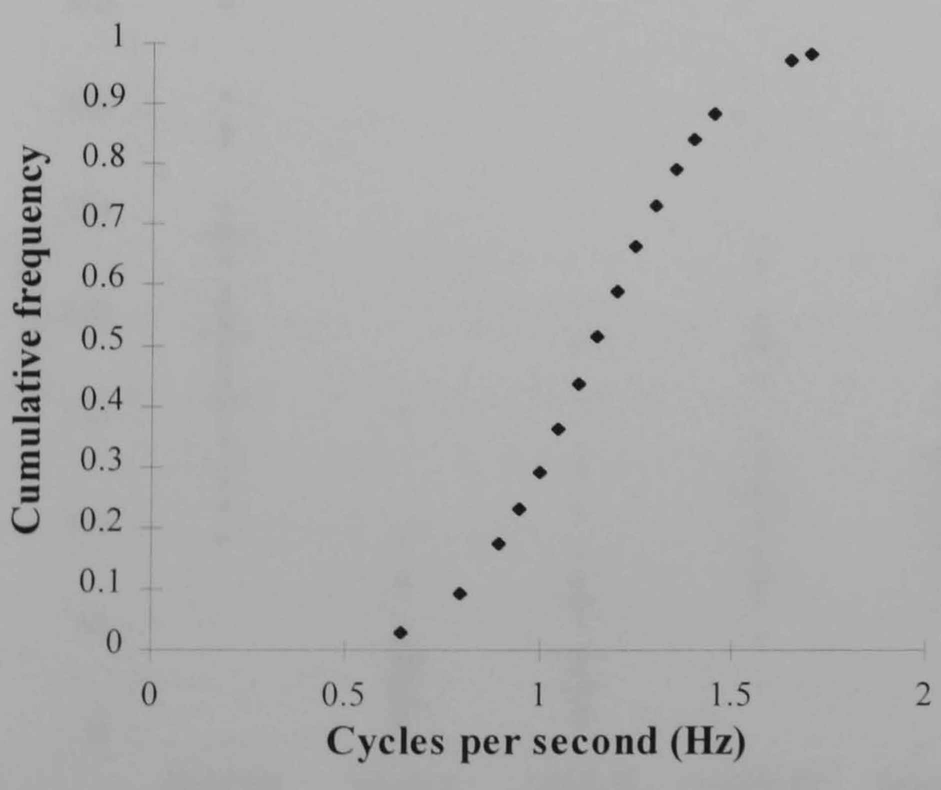


Figure 9-12: The cumulative frequency distribution of the frequency of oscillation for a Grade I mobilisation (n=30, W' = 0.98)

9.9.2 Comparison of group means

Analysis of variance (ANOVA) was used to compare the frequency of oscillation for each grade of mobilisation. Where the results of the F tests were significant, the individual means were compared using Tukey's post-hoc test for multiple comparisons. The data for force magnitude and amplitude were not normally distributed, so the Friedman two-way analysis of variance by ranks was used. Where an overall significant result was demonstrated, the Wilcoxon matched pairs, signed ranks test was used. To ensure that the type I error rate did not exceed 5% when multiple comparisons were performed, the Bonferroni correction factor was used.

9.9.3 Magnitude of the Mobilisation Force

The distribution of the force measurements for each grade of mobilisation is shown in Figure 9-13. The median and interquartile range of each distribution are illustrated by horizontal markers.

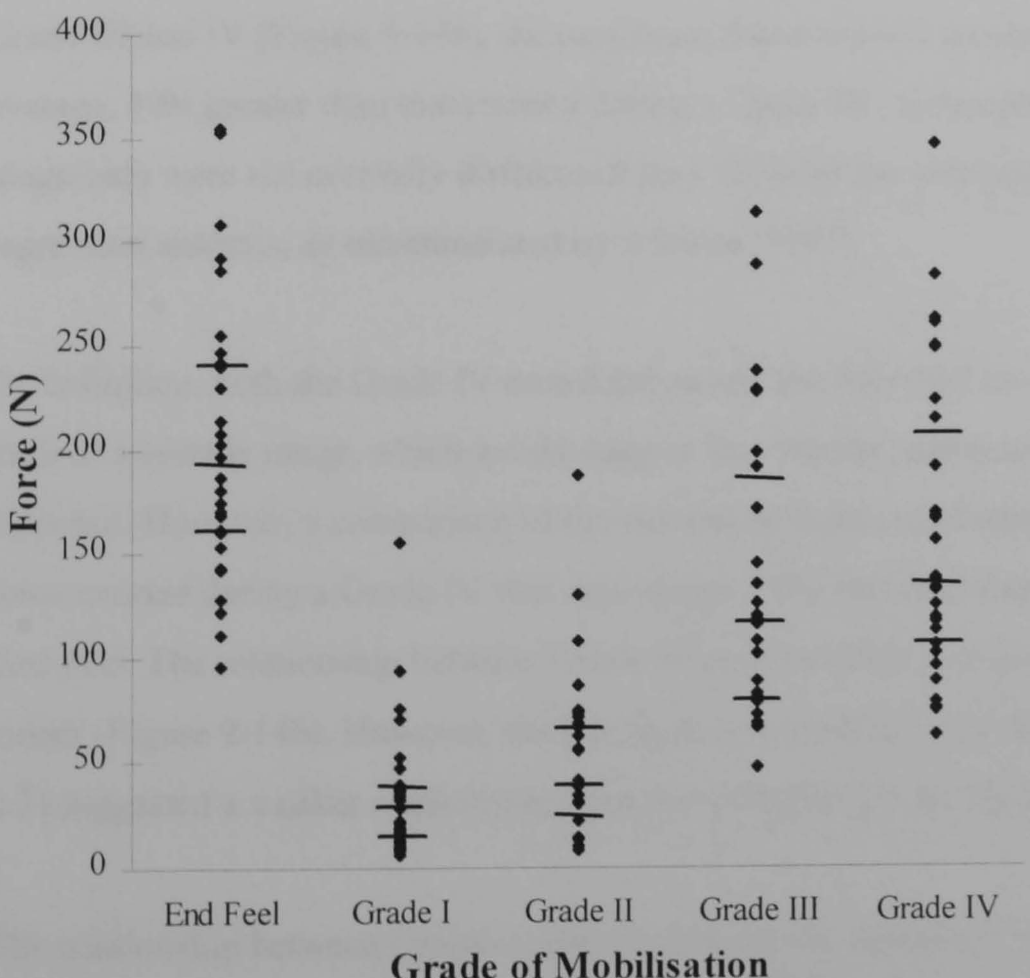


Figure 9-13: The distribution of maximum forces used for each grade of mobilisation by a group of 30 experienced therapists. Horizontal lines indicate median and interquartile values.

Friedman 2-way analysis of variance of the range of maximum forces used for each grade of mobilisation demonstrated an overall significant 'H' statistic ($p<0.001$). Post-hoc comparison of sequential pairs, i.e. Grade I with Grade II, Grade II with Grade III, Grade III with Grade IV and Grade IV with End Feel, demonstrated that all comparisons were highly significant ($p<0.001$).

Grades III and IV are both performed in resisted range. Some authorities recommend that the two higher grades move the target vertebra to the same point in intervertebral range, differing only in amplitude (*Magarey 1985*). It seems reasonable to suggest that one therapist would exert the same force during each procedure to reach the same point in joint range. However, although the results demonstrate a linear relationship between Grade III and IV (Figure 9-14a), the maximum force exerted during a Grade IV was on average, 30N greater than that exerted during a Grade III. Although the data for force magnitude were not normally distributed, they fulfilled the necessary assumptions for a regression analysis, as recommended by Altman (*1991*).

By definition, both the Grade IV mobilisation and the End Feel are performed at the limit of available range, which would suggest that similar maximum forces would be recorded. However, a comparison of the maximum forces used demonstrated that the force exerted during a Grade IV was on average, 25% less than that exerted during an End Feel. The relationship between Grade IV and End Feel was described by a linear model (Figure 9-14b). However, the relatively low coefficient of determination ($R^2 = 0.3$) suggested a weaker relationship than that between Grades III and IV.

The relationship between Grades I and II could also be described by a linear model (Figure 9-14c). The slope of the regression line suggested that the force exerted during Grade II was 20% greater than that exerted during a Grade I.

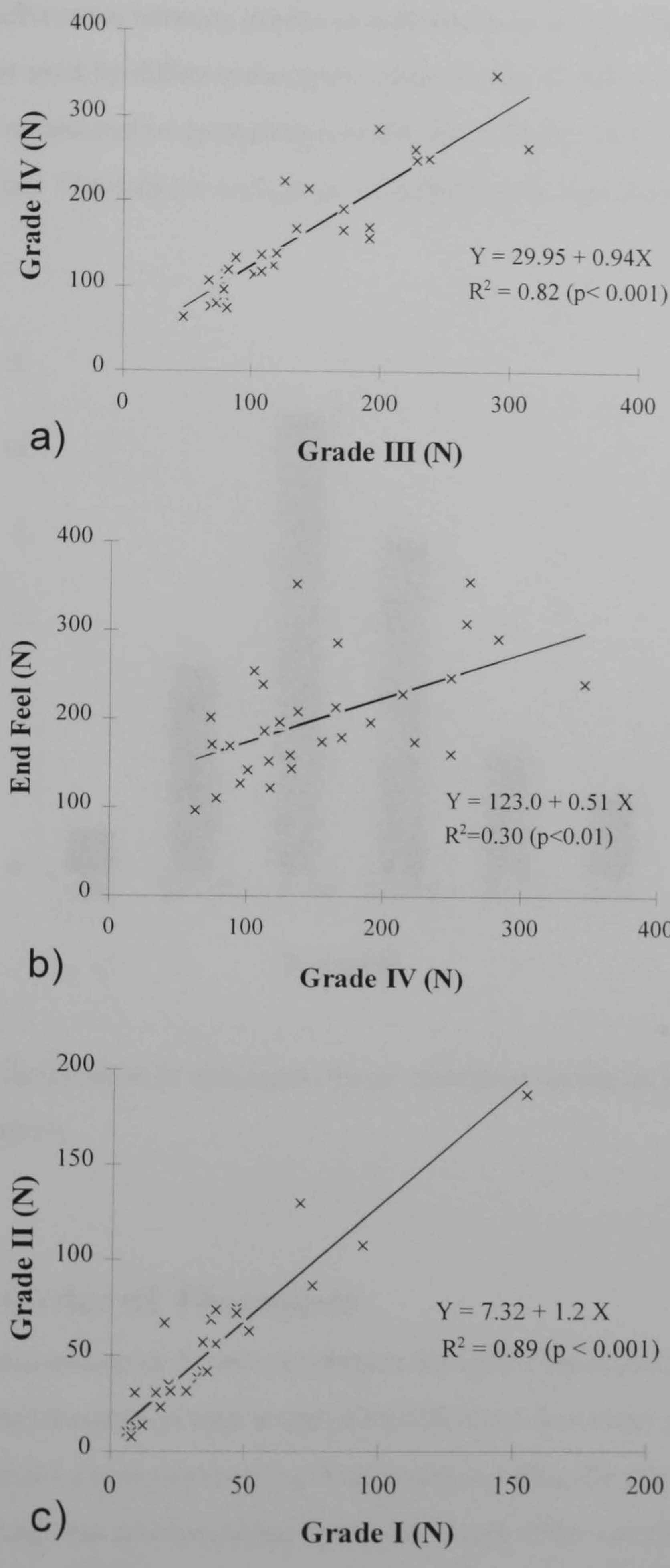


Figure 9-14: The relationship between the forces measured during a) Grades III and IV, b) Grade IV and End Feel and c) Grades I and II

Apart from the differences between grades of mobilisations, the results also show the variation in forces used by different therapists under similar conditions. The histogram in Figure 9-15 is an alternative presentation of the distribution of forces measured during the End Feel. The data for each grade of mobilisation demonstrated a similar positive skew.

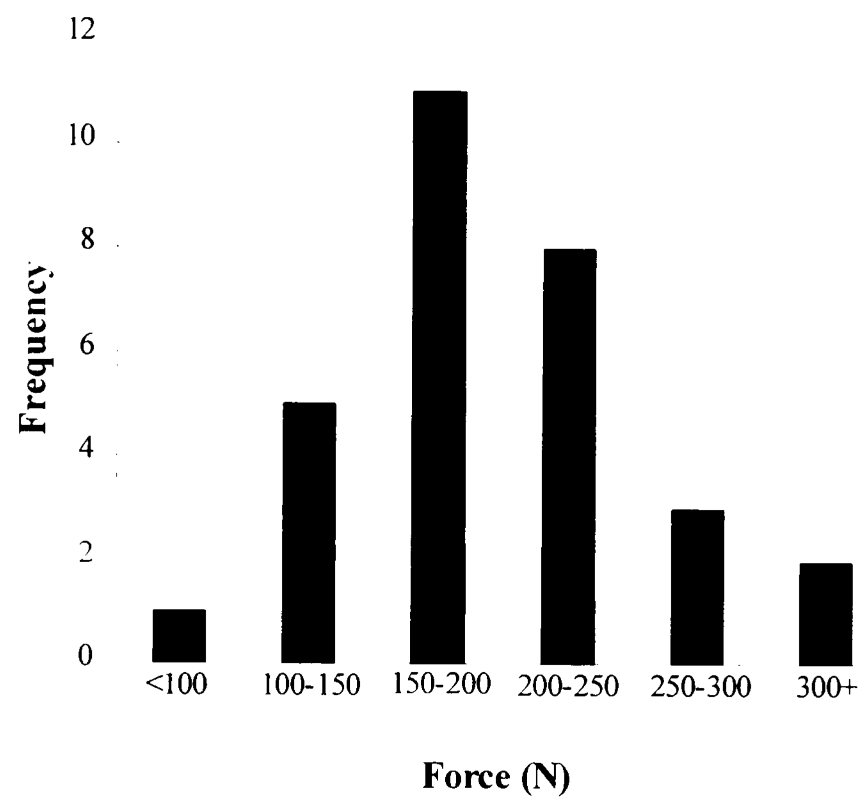


Figure 9-15: The distribution of maximum forces measured during an End Feel for all therapists

9.9.4 Rank Order of Therapists

An analysis was undertaken to determine whether therapists maintained the same rank order during the performance of each grade of mobilisation. For every grade, each therapist was assigned a rank representing their position within the group. Their relative rank within the group was assessed using Spearman's rank order correlation. The linear relationships shown in Figure 9-14 suggested that the therapists maintained a similar rank when performing related manoeuvres. However, those who ranked highest for the first group (Grades I and II) were not necessarily highest within the second group of techniques (Grades III / IV / End Feel). This is reflected by the relative magnitude of the coefficients given by the Spearman rank correlation shown in Table 9-8. The strongest association was found between each grade and the grade performed subsequently.

Table 9-8: Spearman rank order correlation matrix illustrating the relationship between the rank order of therapists during each grade of mobilisation

Grade of Mobilisation	ρ (rho)			
	Grade I	**Grade II**	**Grade III**	**Grade IV**
Grade II	0.93 **†			
Grade III	0.46 *	0.59 **†		
Grade IV	0.36 ns	0.50 **	0.92 **†	
End Feel	0.45 *	0.53 **	0.86 **†	0.87 **†

* $p < 0.05$ ** $p < 0.01$ **† $p < 0.002$ ns = not significant at 5% level

The characteristics of the therapists were also analysed to determine whether the build or experience of the therapist could be related to the maximum force applied. Table 9-9 summarises the results of the Spearman rank order correlation analysis to evaluate this relationship.

Table 9-9: Spearman rank order correlation matrix illustrating the relationship between the characteristics of the therapist and the forces used during spinal mobilisation

Characteristic	ρ (rho)		
	Weight	**Height**	**Experience**
Experience	0.04 ns	0.05 ns	
Force applied (Grade IV)	0.04 ns	0.10 ns	0.22 ns

ns not significant at 5% level

The Grade IV mobilisation is used for illustration in Table 9-9, but none of the coefficients associated with the remaining grades were significant at the 5 per cent level. The results indicate that the characteristics of the therapist account for only a small proportion of the variability in force.

9.9.5 Amplitude of the Mobilisation Force

The force amplitude is defined as the difference between the mean minimum and mean maximum force applied and represents the peak to peak range of the mobilisation cycle. A histogram of a typical distribution of force amplitude is shown in Figure 9-16. The histogram, based on data collected for a Grade I mobilisation, indicates a non-Gausian distribution. A similar finding of a skewed distribution, was also observed for the other three grades of mobilisation, implying that non-parametric statistics were appropriate for subsequent analyses.

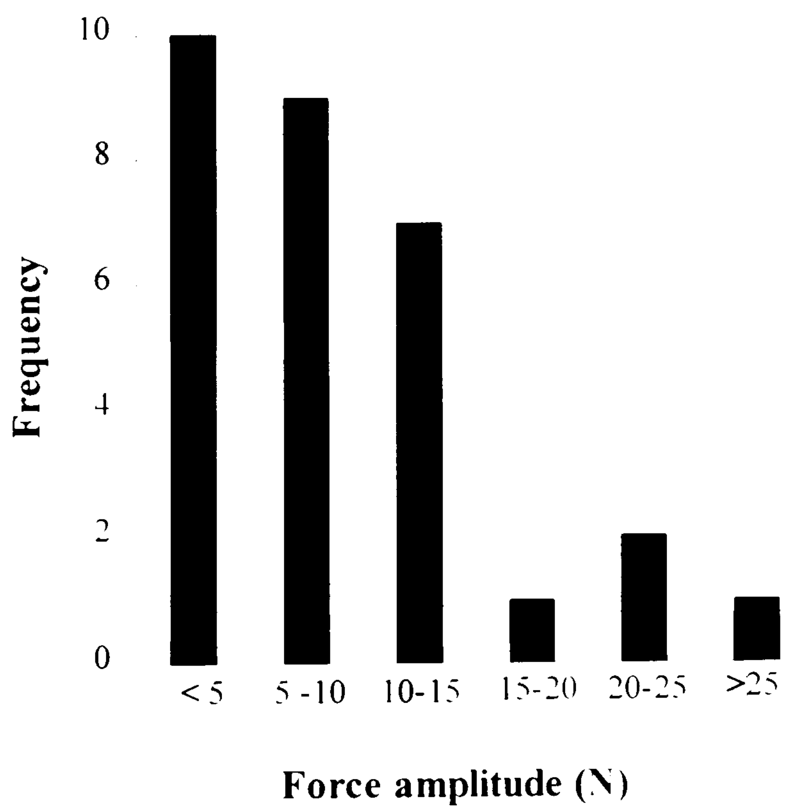


Figure 9-16: The distribution of force amplitude measured during a Grade I mobilisation for all therapists

The non-linear load-displacement response of the FSU (Figure 2.2) implies that the amplitude of the mobilisation force does not provide information on the amplitude of the resulting joint movement. Thus, a significantly greater force is necessary to displace the joint through 10% of its available range at the end of joint range than to move it a similar amount at the beginning of joint range. This is reflected in the relative magnitude and amplitude of the mobilisation force illustrated in Figure 9-17.

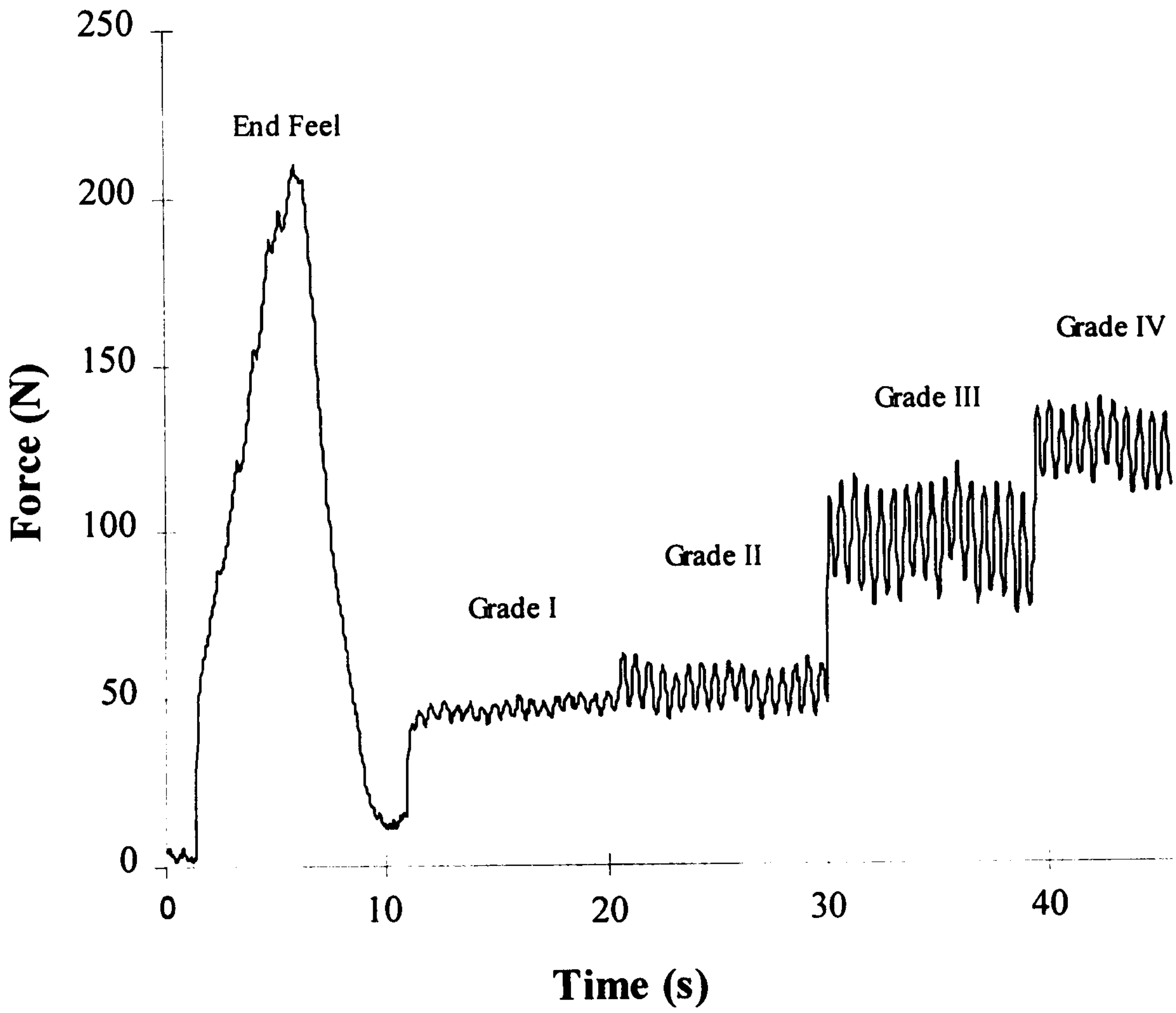


Figure 9-17: The relationship between grade of mobilisation and force amplitude for one therapist (Therapist 30)

A normalisation procedure was adopted where the force amplitude was expressed as a percentage of the maximum force applied by each therapist for each grade of mobilisation. Figure 9-18 illustrates the normalised force amplitude measured for each grade of mobilisation.

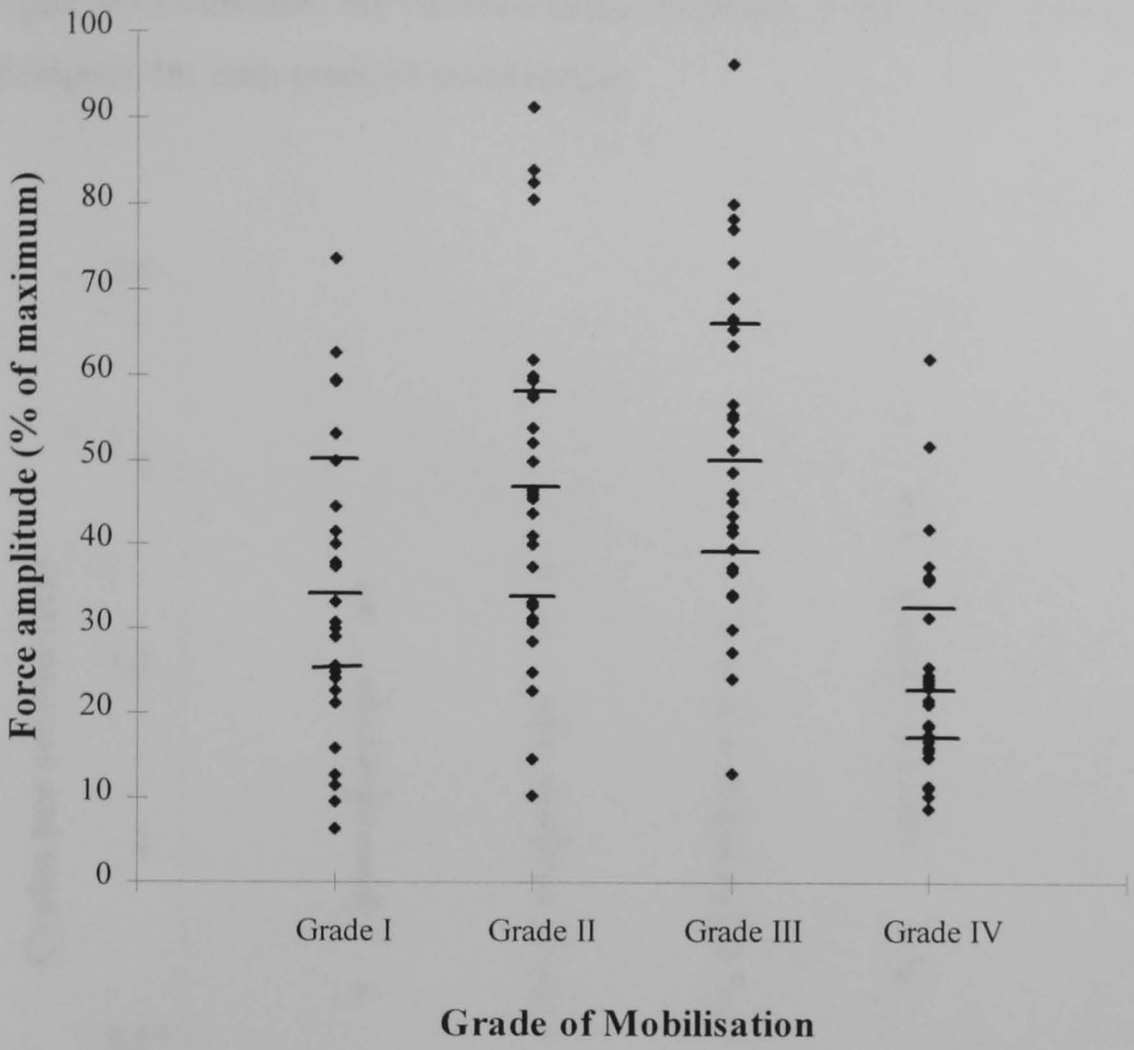


Figure 9-18: The normalised force amplitude used for each grade of mobilisation by a group of 30 experienced therapists. Horizontal lines indicate median and interquartile values.

The Friedman two-way analysis of variance by ranks, followed by appropriate post-hoc tests demonstrated that there were significant differences between the normalised force amplitude for each grade ($p<0.01$). This supports the Maitland philosophy discussed in Section 4.1.4. Maitland recommends that Grade I and Grade IV mobilisations are small amplitude movements where Grades II and III are generally large amplitude movements (*Magarey 1985*).

9.9.6 Frequency of Oscillation of the Mobilisation Force

Figure 9-19 illustrates the variation in the frequency of oscillation used by different therapists, for each grade of mobilisation.

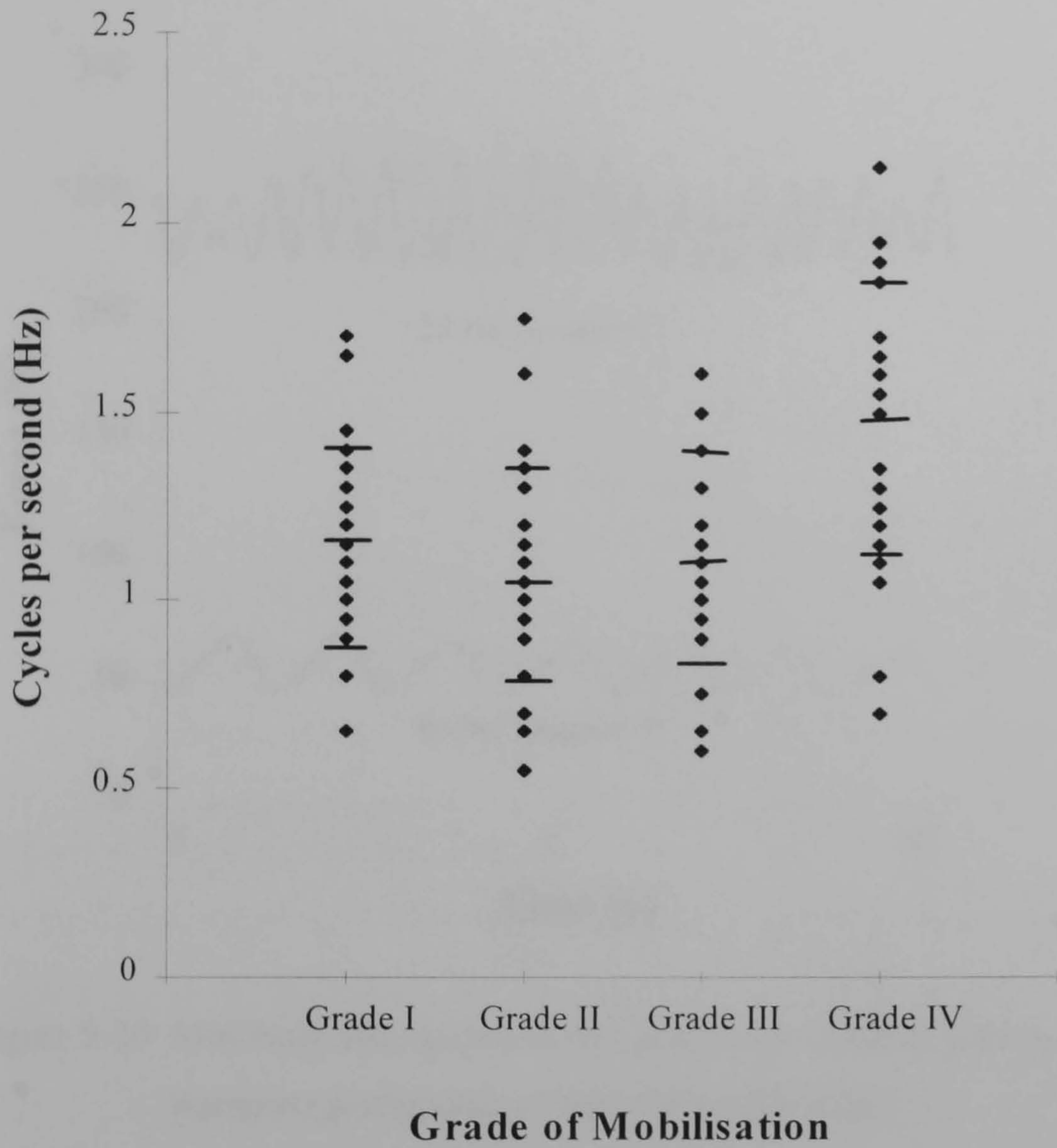


Figure 9-19: The frequency of oscillation used for each grade of mobilisation by a group of 30 experienced therapists. Horizontal lines indicate the mean ± 1 standard deviation

Although there were differences between the rates of oscillation used by different therapists, the mean rates for each grade appeared to be fairly similar. Analysis of variance, followed by Tukey's test for multiple comparisons demonstrated that there were significant differences between Grades I, II and III when compared to the rate of oscillation at Grade IV ($p < 0.001$). None of the other differences were found to be significant at the 5 per cent level. The waveform generated by different therapists differed considerably at every grade (Figure 9-20).

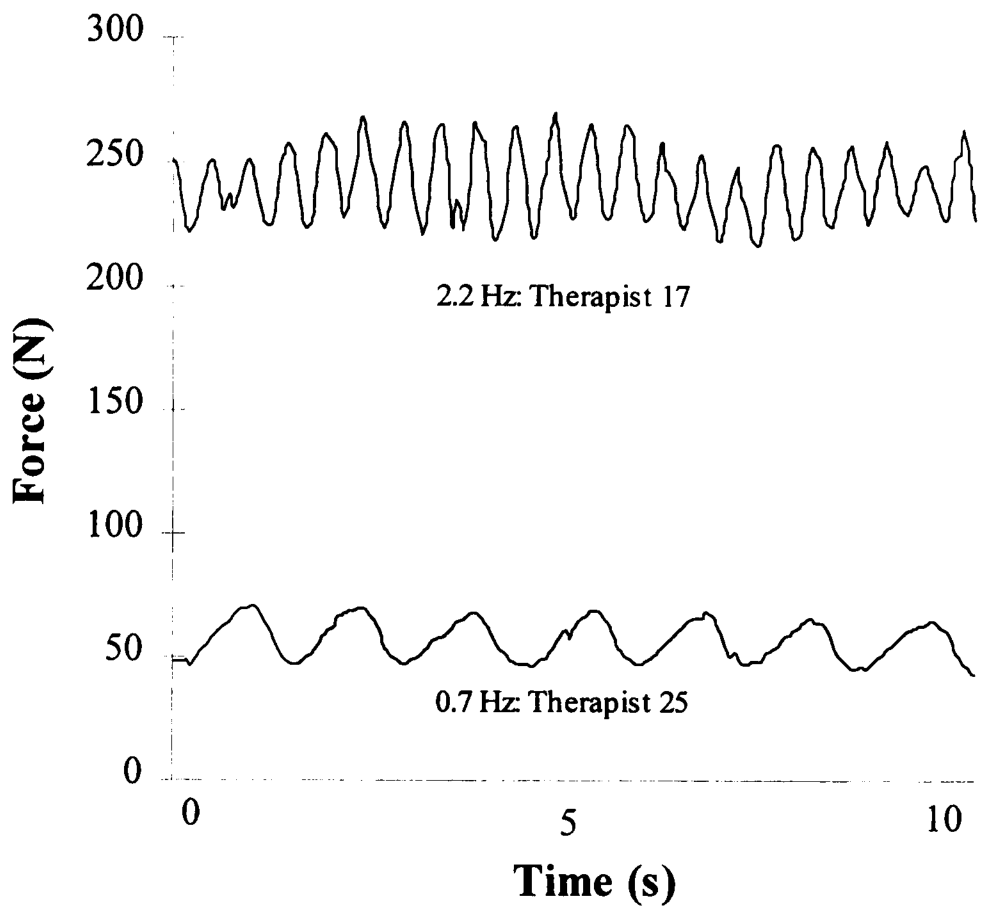


Figure 9-20: Minimum and maximum frequency of oscillation used by different therapists performing a Grade IV mobilisation

Figure 9-20 illustrates the wave form generated by one therapist performing the mobilisation technique at a high frequency of 2.2 Hz and with a relatively high force, and a second therapist who performed a Grade IV mobilisation at a relatively low applied force and low frequency. However, in general, there was no consistent relationship between the magnitude of the force applied by the therapist and the rate of oscillation as shown in Figure 9-21. Similar findings were apparent for all four grades of mobilisation.

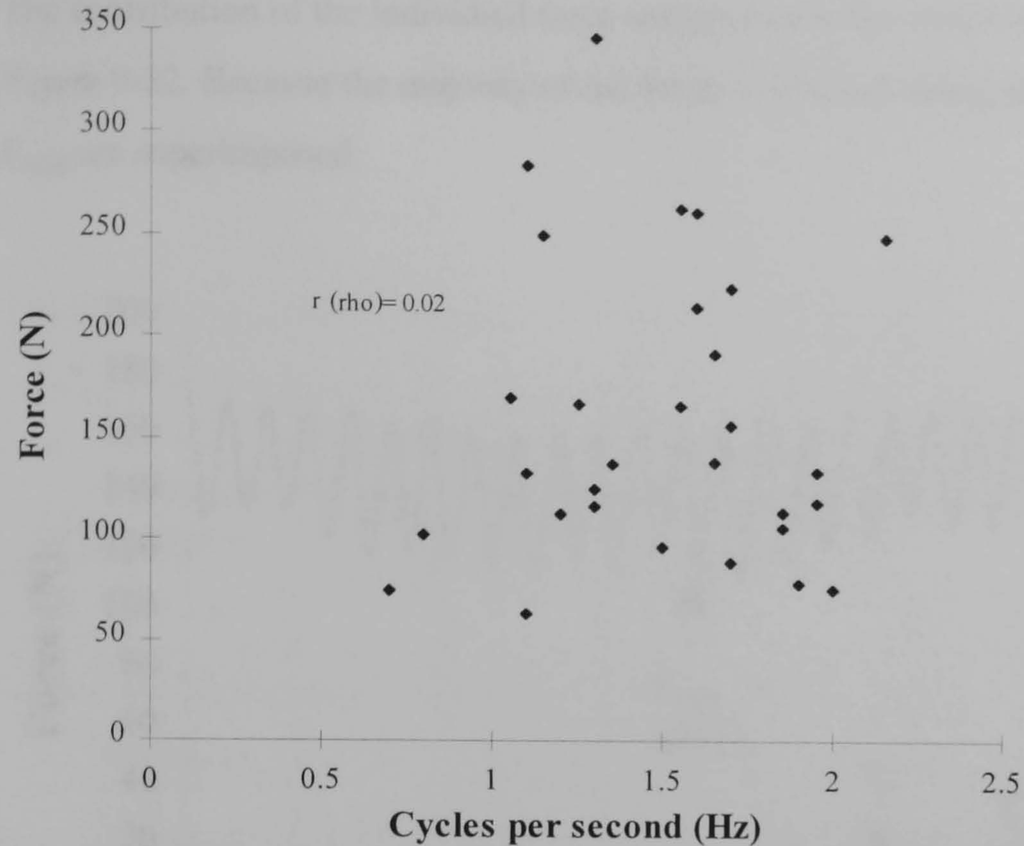


Figure 9-21: The relationship between force magnitude and frequency of oscillation used by different therapists during a Grade IV mobilisation. The Spearman rho for the relationship between frequency and force magnitude is shown.

9.9.7 Component Forces

The preceding sections have all discussed results involving the resultant force vector. A summary of the force values in the three principle axes is presented in Table 9-10.

Table 9-10: The range of forces directed along the three principle axes at each grade of mobilisation

Grade of Mobilisation	F_z (N)	F_x (N)	F_y (N)
End Feel	97 - 359	0 - 41	0 - 36
Grade I	4 - 152	0 - 12	0 - 18
Grade II	6 - 186	0 - 14	0 - 29
Grade III	46 - 314	0 - 32	0 - 29
Grade IV	64 - 348	0 - 60	0 - 32

The contribution of the individual force component to the resultant force is illustrated in Figure 9-22. Because the majority of the force is directed along the vertical axes, F_z and F_{total} are superimposed.

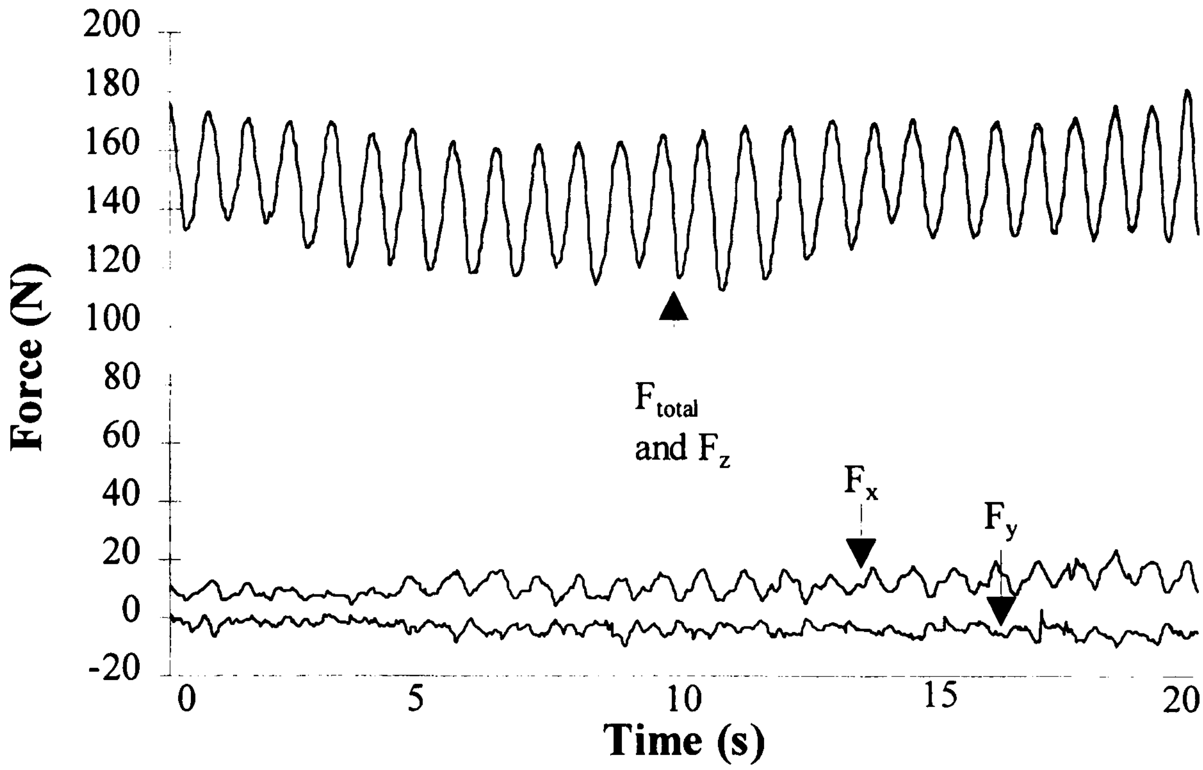


Figure 9-22: Relative magnitude of the forces directed along the three principal axes during a Grade IV mobilisation

These component forces are vector quantities. In order to estimate their relative contribution to the resultant force vector (F_{total}) the following equation must be employed:

$$F_{total} = \sqrt{F_z^2 + F_x^2 + F_y^2} \qquad \text{equation 7.4}$$

For example, the relative contribution of F_z to F_{total}

$$= \left(\frac{100}{F_z^2 + F_x^2 + F_y^2}\right) \times F_z^2 \qquad \text{equation 9.8}$$

A summary of the percentage contribution of the three components for each grade is given in Table 9-11.

Table 9-11: The relative contribution of F_z, F_x and F_y to F_{total}

Grade of Mobilisation	F_z (%)		F_x (%)		F_y (%)	
	Median	Q1-Q3	Median	Q1-Q3	Median	Q1-Q3
End Feel	99	98 - 99	1	0 - 1	1	0 - 1
Grade I	93	87 - 98	3	0 - 10	2	1 - 6
Grade II	96	94 - 99	2	0 - 5	1	0 - 4
Grade III	98	97 - 99	1	0 - 2	1	0 - 1
Grade IV	99	98 - 99	1	0 - 2	0	0 - 1

Q1-Q3: Interquartile range

It is clear that the vertical force (F_z) contributes predominantly to the total force. Both the F_x and F_y components are relatively small in comparison. This is perhaps not surprising when the choice of the mobilisation technique is considered. The postero-anterior technique was directed anteriorly and was performed on the central vertebra in the lumbar curve, where a force directed parallel to the vertebral body will lie close to the Z axis.

The direction of the resultant force with respect to the X and Y axes was also estimated for each of the 30 therapists. A summary of the results for each grade is given in Table 9-12.

Table 9-12: Direction of the resultant force vector with respect to the model and the therapist.

Grade of Mobilisation	Number of Therapists (%)					
	Positive F_x Caudate	Negative F_x Cephalic	$F_x < 1$ N	Negative F_y (towards therapist)	Positive F_y (away from therapist)	$F_y < 1$ N
End Feel	6 (20)	23 (77)	1 (3)	11 (37)	16 (53)	3 (10)
Grade I	3 (10)	19 (63)	8 (27)	6 (20)	19 (63)	5 (17)
Grade II	3 (10)	24 (80)	3 (10)	5 (17)	19 (63)	6 (20)
Grade III	6 (20)	21 (70)	3 (10)	6 (20)	21 (70)	3 (10)
Grade IV	6 (20)	20 (67)	4 (13)	8 (27)	21 (70)	1 (3)

The majority of therapists directed the mobilisation towards the head of the subject (cephalic) and across the subject, away from themselves. The trend is similar for each technique.

9.10 DISCUSSION

This study, using the instrumented mobilisation couch, has contributed considerable knowledge to the theoretical basis of spinal manipulative therapy. For the first time, reliable and valid data has been collected which has allowed a mobilisation waveform to be characterised. It has also allowed therapist reliability to be evaluated in an adequate sample of people.

In general the results support the Maitland philosophy that during Grades I and II less force is applied than during Grades III and IV (Figure 9-13). However, there was a significant variation in forces generated by the 30 different therapists and a correspondingly large area of overlap between the four grades. This suggests strongly that different therapists interpret feedback from the soft tissues in a diversity of ways. As discussed in Section 4.1.4, Grades I and II are performed within the range of joint movement, considered to be free of resistance. By contrast, Grades III and IV are performed within resisted range. The lack of consensus about the magnitude of force necessary to reach R1 and R2, shown by the maximum force exerted during a Grade II, and a Grade IV and End Feel respectively (Figure 9-13), suggests that interpretation of the characteristics of joint movement is highly subjective.

The therapist is often required to apply the same technique several times within one session, either as a means of treatment or to assess the effects of the treatment that has been administered. The techniques will also be repeated at subsequent appointments to evaluate the progress of the patient. The limits of agreement for data obtained for the individual therapist provide a conservative estimate of repeatability and reproducibility (Figures 9.6-9.10). Whilst 95% of replications will be within the calculated limits of agreement, many lie close to the line of unity indicating that some therapists repeat the grades of mobilisation with reasonable accuracy. The question still arises as to whether the differences described by these limits are clinically important. How disparate the two sets of measurements can be, and still be considered acceptable, remains a matter of judgement

There were large differences in the magnitude of the forces applied to test the end of joint movement. Due to the non-linear nature of the load displacement curve of the soft tissues (Figure 2.2), it seems reasonable to suggest that the force applied cannot be used to reflect the movement occurring at the FSU. The end of joint range may have been reached at a similar point by all therapists, whilst some therapists applied a greater force without any significant increase in joint movement.

The force applied during the isolated Grade III mobilisation was approximately 25% greater than when a Grade III was performed after the sequential procedures of an End Feel, Grade I and Grade II mobilisations (Table 9-7). This supports the arguments relating to the biomechanical properties of the FSU, discussed in Chapter 2, where a greater force would be required to reach the same point in range when the tissues had not been preconditioned by Grades I and II. Alternatively, it is possible that by sequentially performing the grades of mobilisation, the therapist increases the force applied by a significant amount at each grade, without respect to perceptual feedback.

The majority of therapists directed the mobilisation towards the head of the subject (cephalic). This may be the result of the mobilisation technique, where the pisiform bone of the right hand is used to apply the force, reinforced by the heel of the left hand. In this study, the therapist approached the couch from the left hand side of the subject. Therefore, the force applied through the right hand would tend to be directed towards the head of the model (Figure 9-23). Also, because the therapist stands to the side of the spine, it was hypothesised that the majority would exert some force away from themselves, this is substantiated by the results as shown in Table 9-12.

Figure 9-23: The position adopted by the therapists when mobilising the third lumbar vertebra

Chapter 10: Study Two - Mobilisation Forces in the Healthy Spine

The previous chapter described the characteristics of the mobilisation forces used by a range of experienced therapists on a single spinal model. Although many of the therapists were consistent in their force application over replications, there was considerable variation between the forces exerted by different therapists. Thus, it was important to extend this work to establish normal limits for mobilisation forces for defined populations. Indeed this approach was considered to be particularly important as Jull and Gibson (*1986*) had found significant variation in the mean forces applied by an experienced therapist on a range of subjects. This chapter describes a study of a group of 26 asymptomatic female subjects. This study also aimed to determine whether the position of the vertebra within the lumbar curve affected the application of these techniques. Therefore, in five subjects within the group, one technique was performed on the first lumbar vertebra (L1), in addition to the full set of techniques performed on L3.

10.1 OBJECTIVES

The objectives of Study Two were:

- To establish the characteristics of the mobilisation force used on the lumbar spine when no pathological changes were evident

- To determine the variation in the characteristics of the mobilisation force used on a healthy population

- To evaluate whether the position of the vertebra within the lumbar curve affected the application of mobilisation techniques

10.2 METHODS AND MATERIALS

10.2.1 Equipment Modification and Calibration

Following Study One, several terminals on the circuit boards within the amplifier required renewal. In addition, to protect the load cells from overload, two adjustable stands were constructed to enable the weight of the subject and the couch top to be supported whilst the subjects were positioning themselves.

A modified calibration protocol was undertaken to determine whether the characteristics of the couch had changed since the original calibration procedures had been carried out. As a result, small adjustments were made to the conversion factors (Table 10-1).

Table 10-1: Conversion factors used to transform load cell output to force measurement in three orthogonal directions

Axis	*X*	*Y*	*Z*
Previous conversion factor (mV/N)	5.46	5.79	8.10
New conversion factor (mV/N)	5.20	6.22	8.18

10.2.2 Measurement of Spinal mobility

The technique described as the modified Schrober method (*Macrea & Wright 1969*) was used to provide an estimate of spinal mobility. The therapist measured the range of sagittal movement of the lumbar spine in all subjects prior to performing the mobilisation protocol. With the subject standing and with their feet together, the therapist marked three lines on the skin overlying the spine. The first line was set at the level of the lumbosacral junction, between the posterior-superior iliac spines.

The second line was made 100mm above this and the third, 50mm below it, over the coccygeal region. The skin over the coccyx is considered to be reasonably well tethered and to move little during movement of the spine. Using radiographic techniques, Tillotson and Burton (1991) demonstrated that reference points marked on the skin overlying the twelfth thoracic and the second sacral vertebrae corresponded with the spinous processes in both flexed and extended postures. Although some skin movement over the upper level was unavoidable, the study by Macrea and Wright (*1969*) demonstrated that measurements made using this technique correlated well with x-ray measurement techniques.

The subject was asked to slide their hands down the front of their legs as far as possible, whilst keeping their knees straight. This was then repeated with the subject sliding their hands down the back of their legs. The distance between the upper and lower lines was measured during flexion and extension of the lumbar spine respectively. The measurements were performed twice on a group of normal subjects to provide an estimate of reliability of the measuring process. To ensure that the physiotherapist did not influence the second measurement by knowledge of the first, the measurements were made by marking a piece of cord held over the marks on the subject's skin. The marks were removed with alcohol wipes and reapplied between each set of measurements. The two pieces of marked cord generated for each subject were measured after the experimental session. The assessment of reliability requires a sample size of 30 subjects, as discussed in Section 9.3, of which 26 participated in the study of mobilisation procedures.

10.2.3 Experimental Design and Protocol for Study Two

Each subject was required to lie prone on the instrumented couch. The therapist applied one set of mobilisations to the third lumbar vertebra of each of the healthy subjects, according to the flow diagram in Figure 10-1. In this experiment, the period of data collection was increased from 10 seconds to 20 seconds for each grade of mobilisation. A period of familiarisation at each grade was incorporated before the recording started.

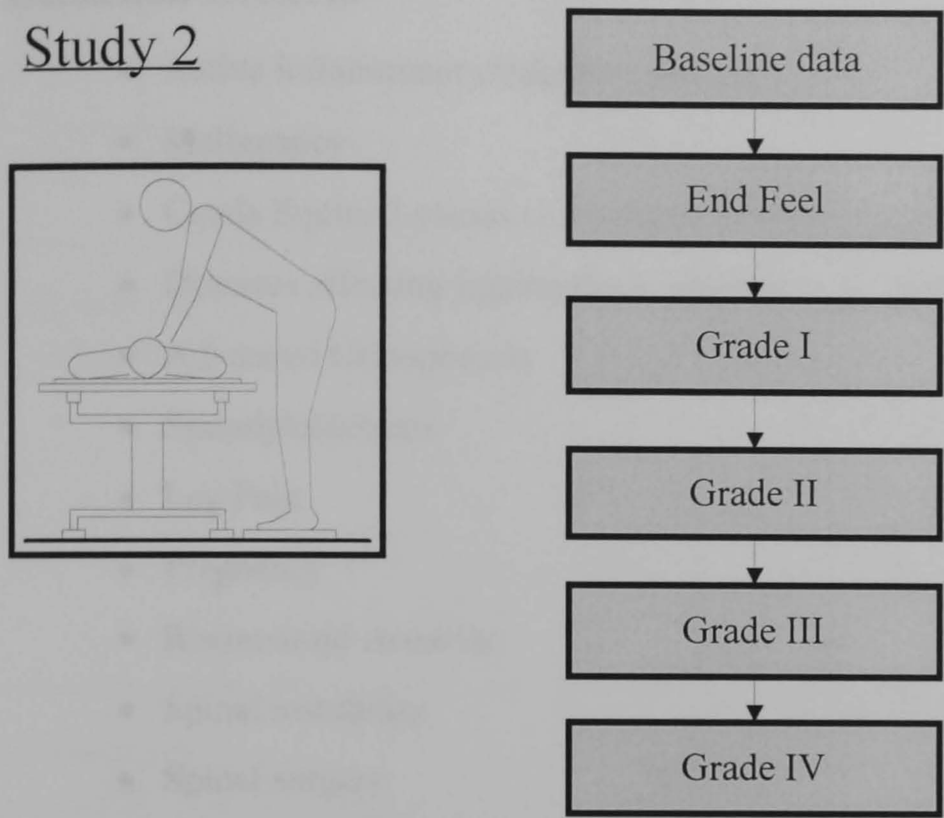


Figure 10-1: Flow diagram of the experimental protocol and sequence of data collection

A Grade III mobilisation was also performed on the first lumbar vertebra of a sample of 5 subjects. Eight new macros were written in Excel, similar to those described in Section 9.5.1 to prepare the data for analysis.

10.2.4 Therapist and Healthy Subjects

One therapist performed all the mobilisations in Study Two. The subjects who participated in the study were selected according to the following criteria:

Inclusion Criteria

- Aged 20-40 years in good general health
- No previous back problems requiring medical attention
- Members of staff within the physiotherapy department
- To limit variability all models were Caucasian females

Exclusion Criteria

- Active inflammatory/infective disease
- Malignancy
- Cauda Equina Lesions or evidence of spinal cord involvement
- Diseases affecting ligamentous integrity (e.g.: Ankylosing Spondylitis)
- Advanced Osteoporosis
- Spondylolisthesis
- Leg Pain
- Pregnancy
- Rheumatoid Arthritis
- Spinal instability
- Spinal surgery
- Bladder or bowel symptoms

10.2.5 Sample size

The data from Study One included a calculation of variance based on the results of one model. An initial estimate of sample size requirements for Study Two was made using this variance. However, calculations from this data alone would tend to underestimate the sample size required for a group study. Therefore, an appropriate sample size was recalculated once a subset of data from 15 models was available (Table 10-2).

Based on the formula $n \geq \frac{2k\sigma^2}{\Delta^2}$ Equation 10.1

Where n = sample size

k = 7.8 (constant where significance level α = 0.05 and power = $(1-\beta)$ % where β = 0.20)

σ = standard deviation of subset of models

Δ = magnitude of difference required between groups

Table 10-2: Data from a subset of normal subjects used for sample size calculations

Criterion	**Grade of Mobilisation**	**Mean**	**Standard Deviation** σ	Δ	**Estimated Sample Size** n
Maximum force (N)	End Feel	237	22	30	**8**
	Grade I	8	2	2	**16**
	Grade II	13	4	3	**25**
	Grade III	168	20	20	**16**
	Grade IV	184	24	20	**23**
Frequency (Hz)	Grade I	0.80	0.05	0.05	**25**
	Grade II	0.70	0.05	0.05	**19**
	Grade III	0.85	0.05	0.10	**9**
	Grade IV	1.45	0.15	0.15	**15**

Analysis of the data collected provided an approximation of the number of subjects required in each group to be studied. For an experimental design with a power of 80% at a 5 per cent level of significance, a sample of 26 subjects was required.

10.2.6 Data Analysis

The complete data set is included in Appendix 5. Data was checked for normality using the Shapiro-Wilks W' as described in Section 9.9.1. The data associated with the maximum resultant force and the frequency of oscillation were all normally distributed, except for the maximum force applied during Grade I and II mobilisations. This led to the selection of non-parametric or parametric statistical testing methods as appropriate.

10.3 RESULTS

10.3.1 Characteristics of the Therapist and Model Group

One female superintendent physiotherapist of average build (height 1.68 m, weight 59 kg), qualified for five years and with three years experience working in an outpatient department, performed all the techniques in Study Two. She was selected as being representative of the sample of therapists who had participated in Study One. The reliability results of this therapist (Therapist 13) calculated in Study One, are presented in Appendix 3.

The results of the reliability study of the Schrober method of measurement of the range of movement of the lumbar spine are given in Appendix 6. Figure 10-2 illustrates the results of the reliability study using the analysis technique of Bland and Altman (1986) described in Section 9.6. The limits of agreement are shown as dashed horizontal lines. The results suggest that on 95% of occasions, the physiotherapist could repeat the measurement of the range of sagittal mobility within 13 mm of the value obtained on the first measurement.

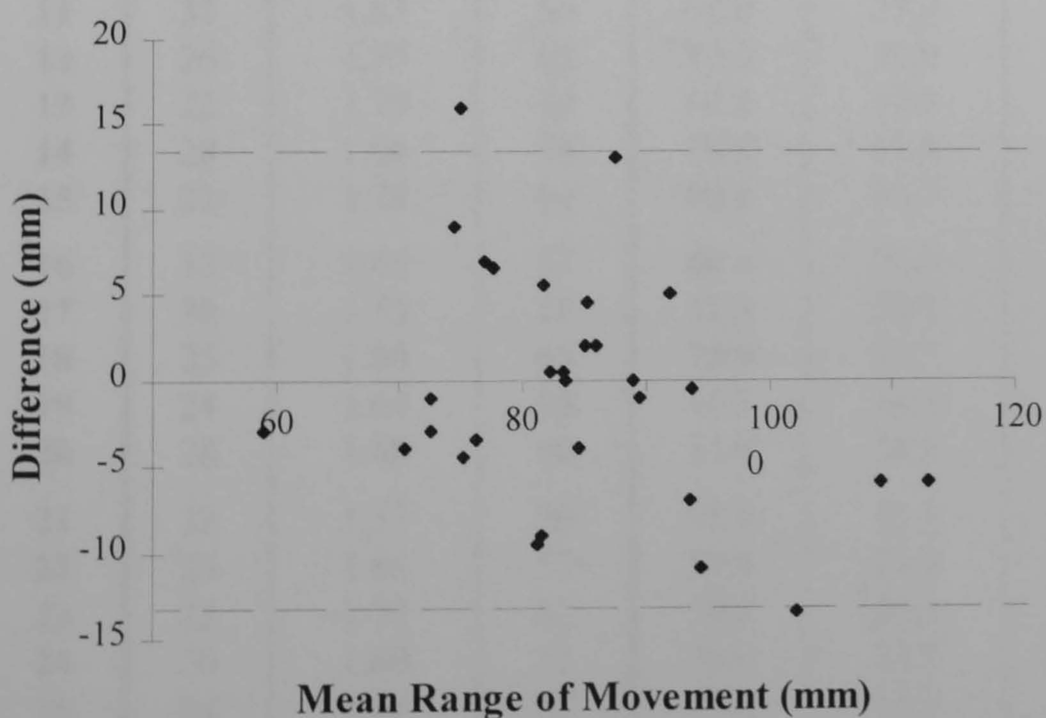


Figure 10-2: Assessment of repeatability for measuring the range of movement of the lumbar spine, based on the modified Schrobers method

The characteristics of the subjects who participated in the study are given in Table 10-3. The range of movement shown in this table corresponds to the sum of the ranges of flexion and extension measured by the modified Schrober method.

Table 10-3: Physical characteristics and range of movement of the lumbar spine of the subjects participating in the study

Subject	Age (yrs)	Height (m)	Weight (kg)	Lumbar Spine Mobility (mm)		
				Flexion	Extension	Range flex + ext
1	32	1.70	61	63.5	21.8	85.3
2	28	1.70	70	80.5	21.8	102.3
3	25	1.65	59	65.5	19.5	85.0
4	30	1.60	55	63.3	13.0	76.3
5	22	1.63	57	54.0	23.0	77.0
6	25	1.65	61	66.3	17.3	83.5
7	22	1.60	46	61.5	24.5	86.0
8	26	1.65	64	62.3	29.8	92.0
9	23	1.70	51	61.8	20.0	81.8
10	25	1.68	60	52.5	22.8	75.3
11	31	1.63	60	62.0	27.0	89.0
12	26	1.73	65	82.0	31.0	113.0
13	22	1.79	59	64.0	17.5	81.5
14	28	1.66	58	70.5	11.8	82.3
15	22	1.78	60	69.8	13.5	83.3
16	33	1.68	57	48.5	22.0	70.5
17	30	1.55	58	51.3	30.0	81.3
18	25	1.60	65	70.0	23.5	93.5
19	24	1.64	68	65.3	19.3	84.5
20	28	1.68	60	57.0	18.0	75.0
21	33	1.57	50	57.0	17.5	74.5
22	26	1.65	52	54.8	23.0	77.8
23	22	1.70	62	38.8	20.3	59.0
24	20	1.60	51	70.0	24.5	94.5
25	24	1.58	57	77.0	12.5	89.5
26	32	1.78	76	76.3	17.5	93.8
Mean	26.3	1.66	59.3	63.3	20.9	84.1
SD	3.9	0.06	6.5	10.1	5.2	10.6
Range	20 - 33	1.55 - 1.79	46 - 76	39 - 82	12 - 31	59 - 113

10.3.2 Magnitude of the Mobilisation Force

The range of maximum forces used during each grade of mobilisation are illustrated in Figure 10-3.

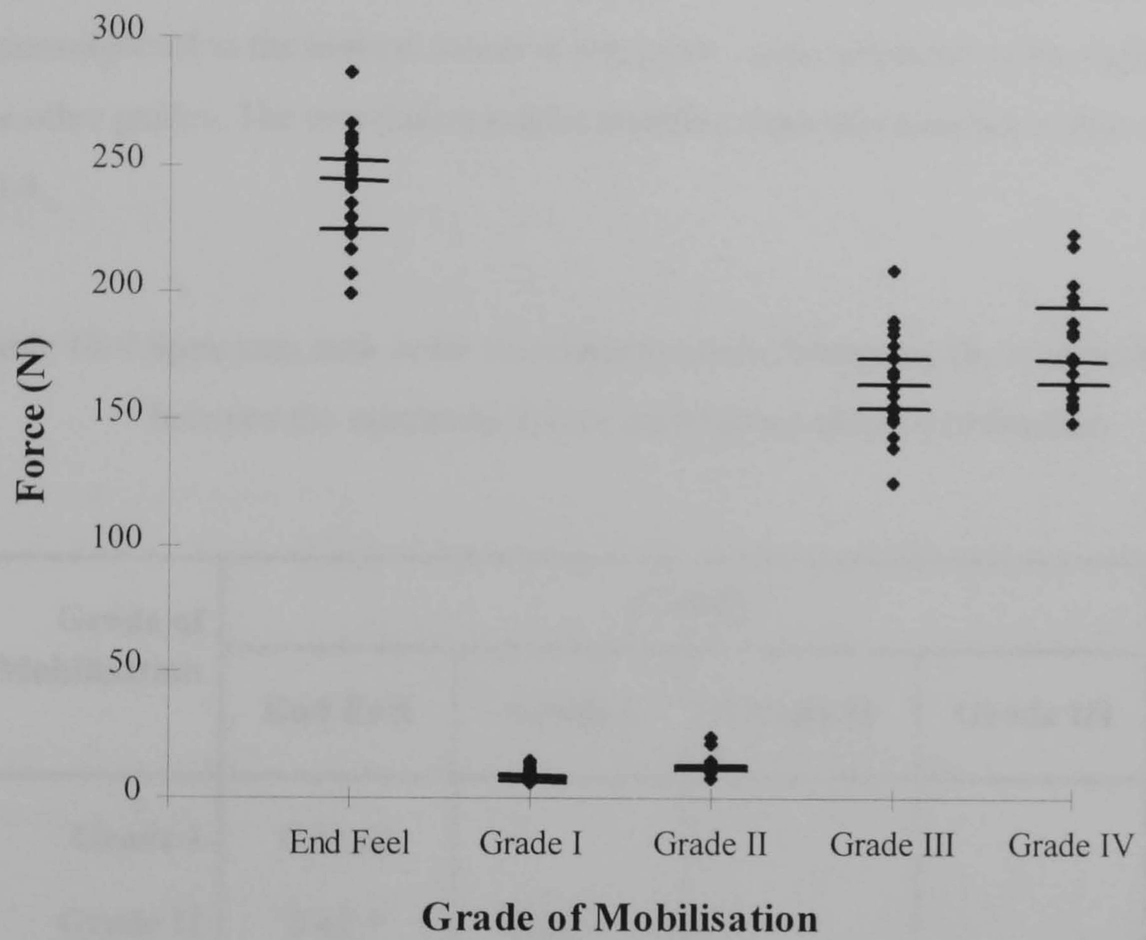


Figure 10-3: The distribution of maximum forces used for each grade of mobilisation performed on a group of 26 healthy subjects. Horizontal lines indicate median and interquartile values

A Friedman two way analysis of variance was used to examine the differences in the forces used for each grade of mobilisation. The analysis gave an overall significant 'H' statistic ($p < 0.001$). The Wilcoxon matched pairs, signed rank sum test, with the Bonferroni correction for multiple comparisons, showed that comparisons between sequential mobilisation procedures, as detailed in Section 9.9.3 were statistically significant ($p<0.01$).

Figure 10-3 illustrates that each of the higher grades of mobilisation incorporated an extreme data point. Inspection of the raw data confirmed that these were measured on the same model. The data were analysed to determine whether there was any relationship between the rank of forces applied for one grade of mobilisation with the equivalent rank for the other four procedures. This would determine whether spines that were subjected to the highest forces at one grade, were subjected to the highest forces at the other grades. The correlation matrix resulting from this analysis is shown in Table 10-4.

Table 10-4 Spearman rank order correlation matrix illustrating the relationship between the maximum forces used during spinal mobilisation

Grade of Mobilisation	ρ (rho)			
	End Feel	**Grade I**	**Grade II**	**Grade III**
Grade I	0.31 ns			
Grade II	0.42 *	0.55 **		
Grade III	0.39 *	-0.10 ns	0.17 ns	
Grade IV	0.58 **	-0.02 ns	0.23 ns	0.77 **

* = $p < 0.05$, ** = $p < 0.01$, ns = not significant at 5% level

The results illustrate that the relationships between the grades of mobilisation were significant at least at the 5 per cent level for Grades III, IV and End Feel. In addition, there was a significant relationship between Grade I and Grade II forces, and Grade II and End Feel. It should be noted however, that the rho values of 0.39 (0.42), although statistically significant at the 5 per cent level, indicate that only 15% (18%) of the variability of Grade III (Grade II) forces can be explained by the correlation with End Feel forces.

The association between the physical characteristics of the model, including their range of movement, and the maximum forces used during each grade of mobilisation was evaluated (Table 10-5).

Table 10-5 Spearman rank order correlation matrix illustrating the relationship between the physical characteristics of the model and the forces used during spinal mobilisation

Parameter	ρ (rho)			
	Age	**Height**	**Weight**	**Range of Movement**
Range of movement	-0.02 ns	0.07 ns	0.38 ns	
End Feel	0.13 ns	0.17 ns	0.32 ns	0.08 ns
Grade I	0.10 ns	-0.18 ns	-0.05 ns	-0.13 ns
Grade II	0.15 ns	-0.03 ns	0.22 ns	0.11 ns
Grade III	-0.14 ns	0.36 ns	0.18 ns	0.17 ns
Grade IV	-0.11 ns	0.21 ns	0.29 ns	0.36 ns

ns = not significant at 5 per cent level

There were no significant relationships between the characteristics of the model and the forces applied to their spines. This suggested that neither age, weight, height or spinal mobility of the subject influenced the magnitude of the force delivered during mobilisation.

10.3.3 Amplitude of the Mobilisation Force

The amplitude of the mobilisation force, expressed as a percentage of the maximum force applied at each grade is shown in Figure 10-4. A Friedman two-way analysis of variance gave an overall significant 'H' statistic ($p < 0.001$). The Wilcoxon matched pairs, signed ranks sum test, using the Bonferroni method to adjust the probability level for multiple comparisons, was used. All the comparisons were significant at the 1 per cent level, with the exception of the comparison between Grades II and III.

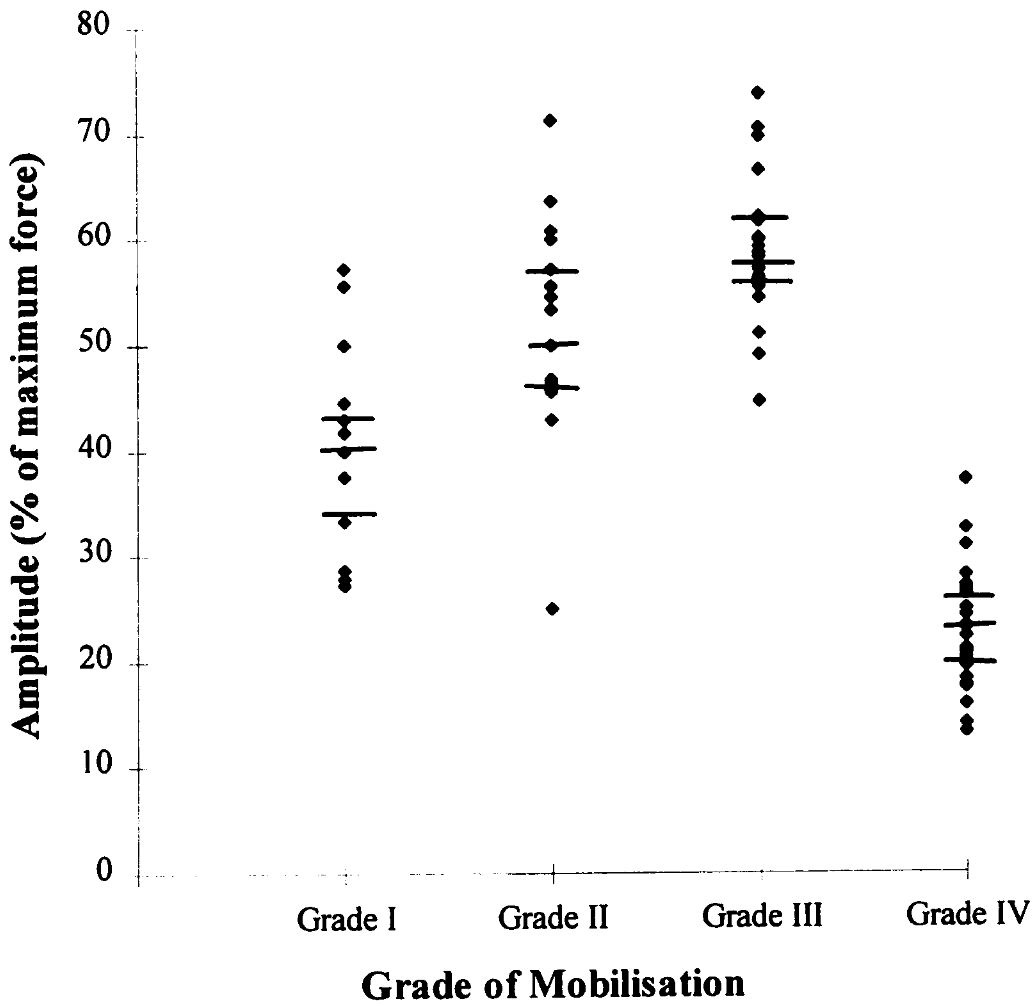


Figure 10-4: The normalised force amplitude used for each grade of mobilisation performed on a group of 26 healthy subjects. Horizontal lines indicate median and interquartile values

The results demonstrated that Grades II and III, regarded as large amplitude movements (*Magarey 1985*), were performed with a proportionally greater force amplitude than Grade I and IV.

10.3.4 Frequency of Oscillation of the Mobilisation Force

The frequency of oscillation used at each grade of mobilisation is illustrated in Figure 10-5. Clearly the frequency associated with the Grade IV mobilisation was higher than that associated with the other three grades. A two-way analysis of variance of the data gave a significant F statistic ($p < 0.001$). Tukeys post-hoc analysis to compare the differences between pairs of means demonstrated that five out of six comparisons were statistically significant at the 1 per cent level. The only exception was the comparison between Grades I and III where the values were clearly very similar.

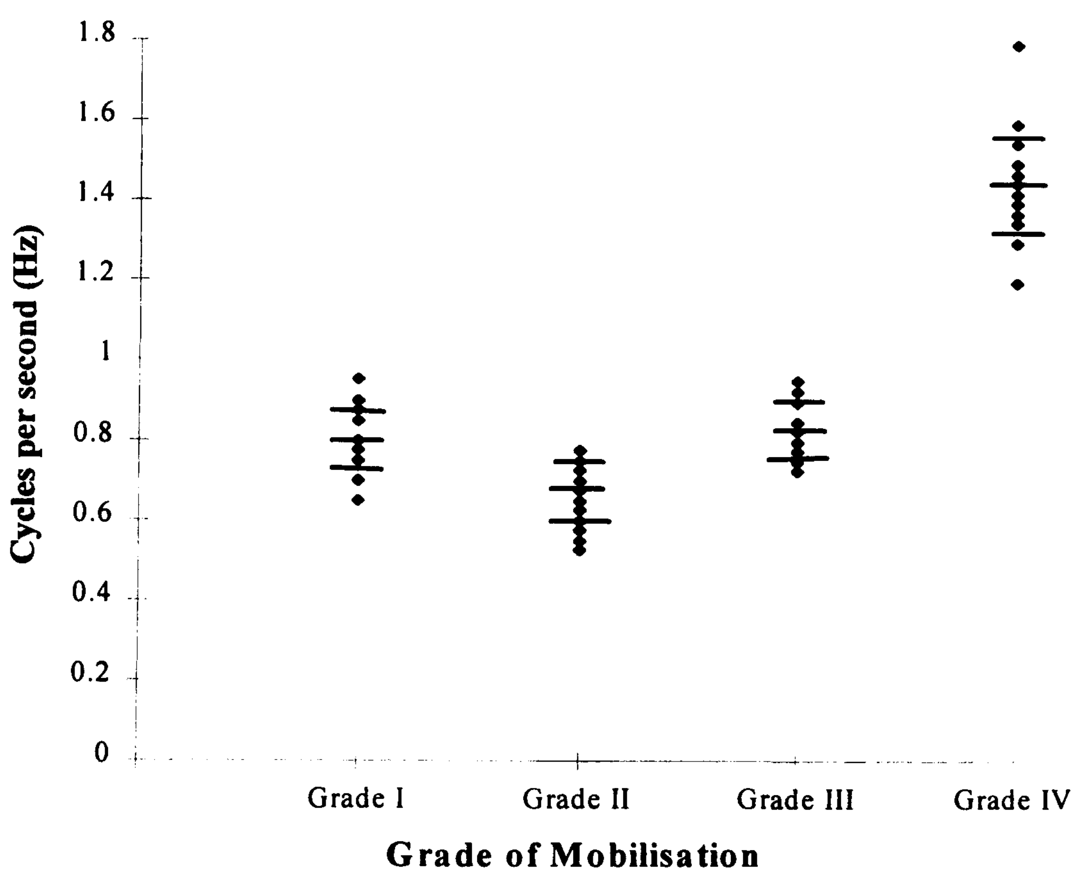


Figure 10-5: The frequency of oscillation used for each grade of mobilisation performed on a group of 26 healthy subjects. Horizontal lines indicate the mean ±1 standard deviation

10.3.5 Component Forces

When the range of forces directed along each of the three principal axes were assessed, the vertical force was found to contribute predominantly to the total force (Table 10-6).

Table 10-6: The range of forces directed along the three principal axes at each grade of mobilisation

Grade of Mobilisation	F_z (N)	F_x (N)	F_y (N)
End Feel	198 - 285	0 - 55	0 - 36
Grade I	5 - 14	0 - 5	0 - 10
Grade II	7 - 22	0 - 5	0 - 10
Grade III	134 - 205	1 - 57	2 - 26
Grade IV	132 - 225	0 - 59	2 - 30

The contribution of F_z, F_x and F_y to F_{total}, based on equation 9.8, are presented in Table 10-7. The results confirmed that the horizontal force components were relatively small. However, the median percentage contribution of F_y to F_{total} was found in some instances to be greater than 10%. This occurred as a result of the calculations involving small forces measured during Grades I and II. When F_{total} was less than 10N, the percentage of this force attributed to F_y would be within the error of the measuring system.

Table 10-7: The relative contribution of F_z, F_x and F_y to F_{total}

Grade of Mobilisation	F_z (%)		F_x (%)		F_y (%)	
	median	Q1-Q3	median	Q1-Q3	median	Q1-Q3
End Feel	99.7	98.9 - 99.8	0.0	0 - 0.1	0.2	0.1 - 0.5
Grade I	77.0	67.3 - 91.8	1.9	0 - 10.2	11.0	4.0 - 22.3
Grade II	85.0	78.5 - 94.6	0.9	0 - 7.7	11.1	4.5 - 16.0
Grade III	99.1	98.5 - 99.4	0.0	0 - 0.3	0.6	0.2 - 1.3
Grade IV	99.1	98.6 - 99.5	0.0	0 - 0.5	0.5	0.2 - 1.1

Q1 - Q3 = Interquartile range

10.3.6 The effect of vertebral position on the mobilisation force

To determine whether the characteristics of the force changed when the procedures were performed on a different lumbar vertebra, principally with respect to the magnitude of the component forces, the therapist applied a set of Grade III mobilisations to the first lumbar vertebra of five healthy subjects (Subjects 8, 17, 21, 24 and 25 in Table 10.3). An End Feel was performed first to allow the therapist to determine the amount of joint movement and the characteristics of the soft tissue resistance to palpation. The data collected during the Grade III mobilisation was analysed and the results are shown in Table 10-8, where it can be compared to the corresponding data from L3. The complete data set is included in Appendix 7.

Table 10-8: The characteristics of a Grade III mobilisation applied to the first and third lumbar vertebrae

Parameter	**First Lumbar Vertebra (n=5)**	**Third Lumbar Vertebra (n=26)**
Maximum force (N) Median (Q1 - Q3)	170 (170 - 190)	164 (155 - 174)
Frequency of oscillation (Hz) Mean (SD)	0.93 (0.06)	0.83 (0.07)
Force amplitude as % of maximum force Median (Q1 - Q3)	50 (47 - 55)	58 (56 - 62)
Contribution of component forces to resultant force (%) Median (Q1 - Q3)		
F_z	99.2 (99.2 - 99.3)	99.1 (98.5 - 99.4)
F_x	0 (0 - 0)	0 (0 - 0.3)
F_y	0.8 (0.7 - 0.8)	0.6 (0.2 - 1.3)

Because the procedure on L1 was only performed on five subjects, further data analysis was unwarranted. However, the similarities between the characteristics of the mobilisation force applied to L1 and L3, illustrated in Table 10-8, are apparent.

10.4 DISCUSSION

The second study using the instrumented mobilisation couch has contributed new information on the range of forces used for mobilisation procedures. The inclusion of a group of 26 healthy female volunteers has given an indication of the characteristics of the procedures used on spines unaffected by pathological changes.

It had been hypothesised that those subjects who exhibited stiffness of the FSU would record higher forces for every grade of mobilisation. The correlation coefficients shown in Table 10-4 support this hypothesis by demonstrating that there was a significant correlation between the magnitude of the forces used during each grade of mobilisation. However, as shown in Section 10.3.2, even though the results of the analysis were statistically significant, only a small proportion of the variability could be explained by the statistical model. This may be because the small variation in the forces used for each grade, resulting from the selection of a homogenous group of subjects, diminished the usefulness of the rank order correlation. The close grouping of the data, especially for Grades I and II suggested that even small differences in the force exerted between subjects would have resulted in a change in the rank order of the data.

It was anticipated that the physical characteristics of the model would influence the force used during mobilisation. However, the variation of 75 - 87 N between the highest and lowest values of the maximum forces exerted on different subjects during the higher grades of mobilisation (Figure 10-3) could not be explained by the age, weight, or height of the model, or attributed to the physiological range of sagittal movement in the lumbar spine. It was also argued that the range of lumbar spine movement was associated with the stiffness of the FSU, and that the range of movement would therefore be related to the forces used during mobilisation. Because no relationship was demonstrated, it is possible that the sagittal mobility measured by Schrobers method did not reflect the stiffness or mobility of the segmental level. The relationship between the postero-anterior accessory range of the vertebral joints described in Section 2.2, and the physiological sagittal range with which it is thought to be coupled (*Panjabi et al 1977*), may be more complex than generally described. This indicates that there are limitations to using physiological movements to monitor changes in the spine following treatment at a segmental level.

Because of the central position of L3 within the spinal curve, it had been hypothesised that F_z would be the main component force. However, when the same techniques were performed on L1, the magnitude of the component forces did not alter appreciably despite the different position of the vertebra within the lumbar curve. The results suggest that there may be few differences between the postero-anterior mobilisation techniques used on different vertebrae in the lumbar spine.

Chapter 11: Study Three - The Effect of Age Related Changes on Spinal Mobilisation Forces

The data in Chapter 10 defined the characteristics of a typical mobilisation force profile applied to a normal, asymptomatic spine. With age, the spine undergoes degenerative changes (*Butler et al 1990*) which may precipitate many of the clinical disorders discussed in Chapter 3. It is therefore important to determine whether these degenerative changes affect the performance of mobilisation techniques. This chapter involves a study of the range of forces applied to the spines of a group of 16 patients aged between 45 and 65 years.

11.1 OBJECTIVES

The objectives of Study Three were:

- To examine the effect of age related changes in the lumbar spine on the characteristics of the mobilisation force profile

- To determine the variation in the characteristics of the mobilisation force used on a clinical sample

11.2 METHODS AND MATERIALS

11.2.1 Experimental Design and Protocol for Study Three

The study followed the protocol described in Section 10.2.3. A single measurement of spinal flexion and extension using Schrobers method was performed on each subject in the patient group. The data were analysed according to the criteria described in Section 10.2.6.

11.2.2 Therapist and Patient Group

The therapist described in Section 10.3.1 performed all the mobilisations in Study Three. Subjects were recruited from patients attending the Physiotherapy Outpatient Department at the Middlesex Hospital. Ethical approval was granted by Camden & Islington Community Health Services NHS Trust Ethics Committee. The study conformed to the Declaration of Helsinki. All subjects were provided with information sheets and were required to give written consent prior to inclusion (Appendix 8).

Inclusion Criteria

- Aged 45-65 years
- Patients attending physiotherapy for treatment
- Asymptomatic L3 level which did not reproduce symptoms on palpation
- To limit variability all subjects were Caucasian females

Patients were excluded according to the criteria listed in Section 10.2.4. with the exception of those with leg pain. This excluded subjects who had conditions which may have had diverse effects of the spine, and affected the way in which the mobilisation techniques were performed. It also ensured that patients whose conditions contraindicated the use of spinal techniques were excluded from the study.

11.3 RESULTS

11.3.1 Characteristics of the Patient Group

The characteristics of the patients who participated in the study are summarised in Table 11-1. The full data set for Study Three is included in Appendix 9.

Table 11-1: Physical characteristics and range of movement of the lumbar spine of the patients participating in the study.

Patient	Age (yrs)	Height (m)	Weight (kg)	Lumbar spine mobility (mm)			Diagnosis
				Flexion	Extension	Range flex + ext	
1	54	1.60	57	27	23	50	Back Pain
2	57	1.57	51	59	17	76	Back Pain
3	53	1.60	82	61	16	77	Back Pain
4	59	1.60	73	92	2	94	Anterior knee pain
5	54	1.68	64	51	19	70	Anterior knee pain
6	49	1.70	64	66	29	95	Achilles Tendonitis
7	56	1.63	54	63	10	73	Lateral epicondylitis
8	51	1.63	83	70	19	89	Neck Pain
9	64	1.68	67	43	9	52	# Neck of Femur
10	59	1.65	67	66	14	80	Shoulder capsulitis
11	55	1.6	76	49	12	61	Shoulder capsulitis
12	62	1.66	70	31	15	46	Osteoarthritis, knee
13	55	1.65	95	70	26	96	Calcaneal Spur
14	61	1.68	83	68	16	84	Osteoarthritis, knee
15	47	1.65	54	54	10	64	Osteoarthritis, knee
16	59	1.60	57	69	16	85	Osteoarthritis, knee
Mean	55.9	1.60	68.6	58.7	15.8	74.5	
SD	4.7	0.04	12.7	16.1	6.7	16.2	
Range	47 - 64	1.57 - 1.70	51 - 95	27 - 92	2 - 29	46 - 96	

11.3.2 Magnitude of the Mobilisation Force

The range of maximum forces applied during each grade of mobilisation are illustrated in Figure 11-1. A Friedman two-way analysis of variance gave an overall significant 'H' statistic ($p<0.001$), indicating that there were differences between the grades of mobilisation. The Wilcoxon matched pairs signed rank sum test with the Bonferroni correction demonstrated significant differences between all grades of mobilisation, with one exception. Despite the apparent trend in the data which suggested that a higher force was applied during a Grade IV mobilisation than during a Grade III, the lack of significance suggests that a combination of the variation within the sample and the sample size, reduced the power of the experimental design. In addition, the use of non-parametric statistical tests, although more robust to anomalies in the distribution of the data, tend to be less sensitive than the parametric equivalent.

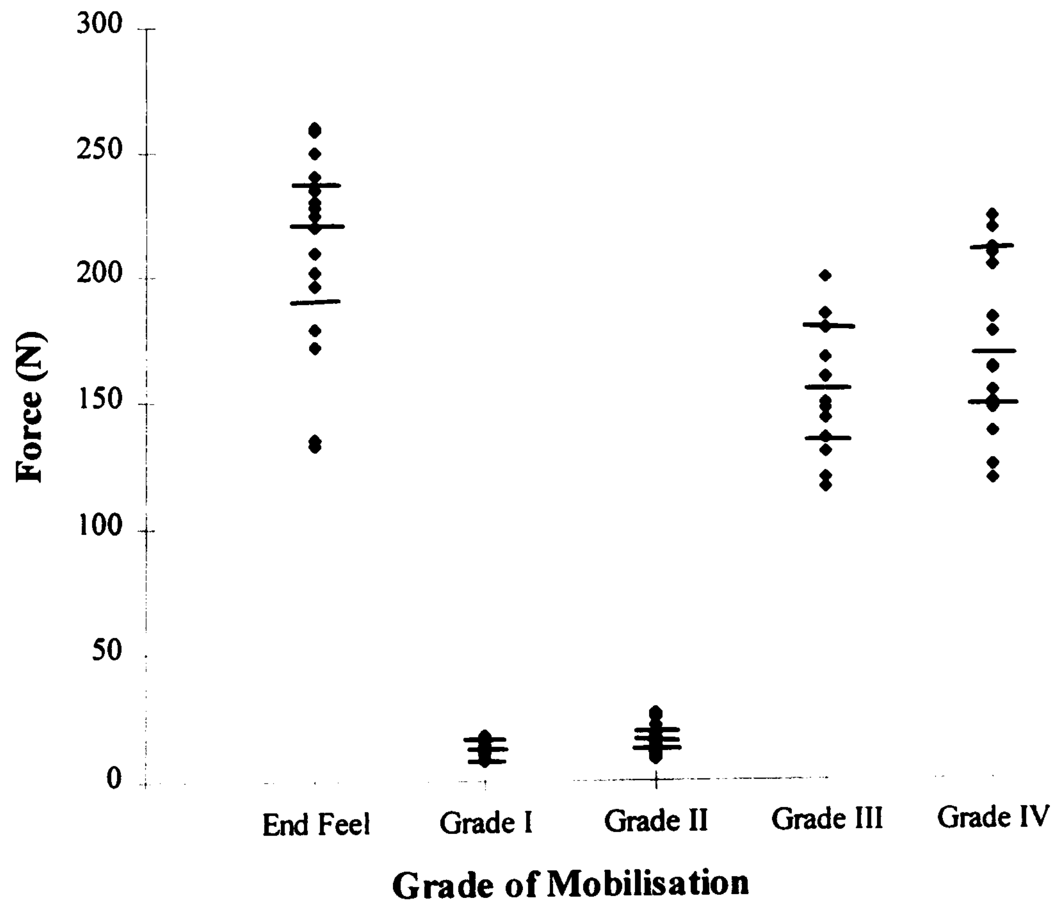


Figure 11-1: The distribution of maximum forces used for each grade of mobilisation performed on a group of 16 patients. Horizontal lines indicate median and interquartile values

The data of the patients occupying extreme positions in Figure 11-1 were examined to determine whether any measured characteristics were associated with these values. However, those patients who were subjected to the highest and lowest forces could not be identified by any physical characteristic, nor did their diagnosis appear to influence the magnitude of the force applied to their spine. The results of the subjects with back pain were spread evenly through the range of values (Appendix 9: Table A9-1).

The results of a Spearman rank order correlation demonstrated that there was a statistically significant relationship, at least at the 5 per cent level, between the forces used for each grade of mobilisation (Table 11-2). These were highly significant between Grade I and the other three grades and between Grade III and IV. This indicated that those patients who were subjected to the highest forces at one grade were also subjected to higher forces during the other four procedures. It was interesting to note that the coefficients associated with the End Feel procedure were generally lower than those within the four grades of mobilisation, and the highest coefficient was found between Grades III and IV.

Table 11-2: Spearman rank order correlation matrix illustrating the relationship between the maximum forces used during spinal mobilisation

Grade of Mobilisation	ρ (rho)			
	End Feel	**Grade I**	**Grade II**	**Grade III**
Grade I	0.58 *			
Grade II	0.50 *	0.83 **†		
Grade III	0.63 *	0.78 **†	0.59 *	
Grade IV	0.59 *	0.81 **†	0.68 **	0.97 **†

* $p < 0.05$, ** $p < 0.01$, **† $p < 0.002$

Figure 11-2a-e illustrate the linear relationships between the forces applied during these procedures. Figure 11-2a suggests that the force applied during a Grade IV procedure was approximately 20% greater than that applied during a Grade III.

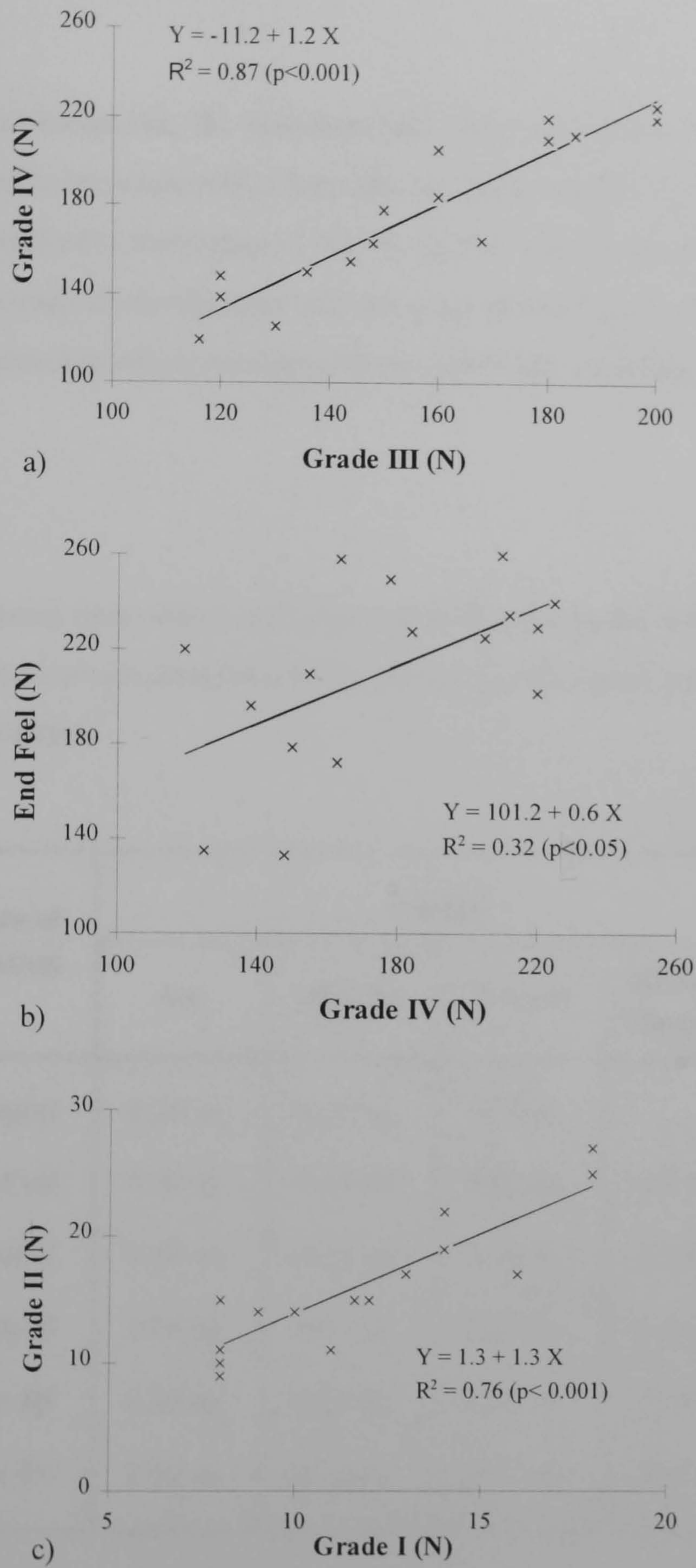


Figure 11-2: The relationship between the forces measured during a) Grades III and IV, b) Grade IV and End Feel and c) Grades I and II

In all comparisons, except one, the Spearman rank order correlation demonstrated that there were no significant relationships between the characteristics of the patient and the maximum force applied to their spine (Table 11-3). The relationship between the force exerted during a Grade III mobilisation and the range of movement of the lumbar spine was the only relationship which was found to be statistically significant at the 5 per cent level.

Table 11-3: Spearman rank order correlation matrix illustrating the relationship between the physical characteristics of the patient and the forces used during spinal mobilisation

Grade of Mobilisation	ρ (rho)			
	Age	Height	Weight	Range of Movement
Range of Movement	-0.15 ns	0.02 ns	0.37 ns	
End Feel	0.34 ns	-0.17 ns	0.01 ns	-0.45 ns
Grade I	0.14 ns	0.13 ns	-0.20 ns	-0.45 ns
Grade II	0.06 ns	0.03 ns	-0.07 ns	-0.26 ns
Grade III	0.05 ns	-0.09 ns	-0.29 ns	-0.54 *
Grade IV	0.04 ns	-0.14 ns	-0.21 ns	-0.41 ns

* $p < 0.05$, ns = not significant at 5 % level

11.3.3 Amplitude of the Mobilisation Force

The amplitude of the mobilisation force was expressed as a percentage of the maximum force applied as described in Section 9.9.5 (Figure 11-3).

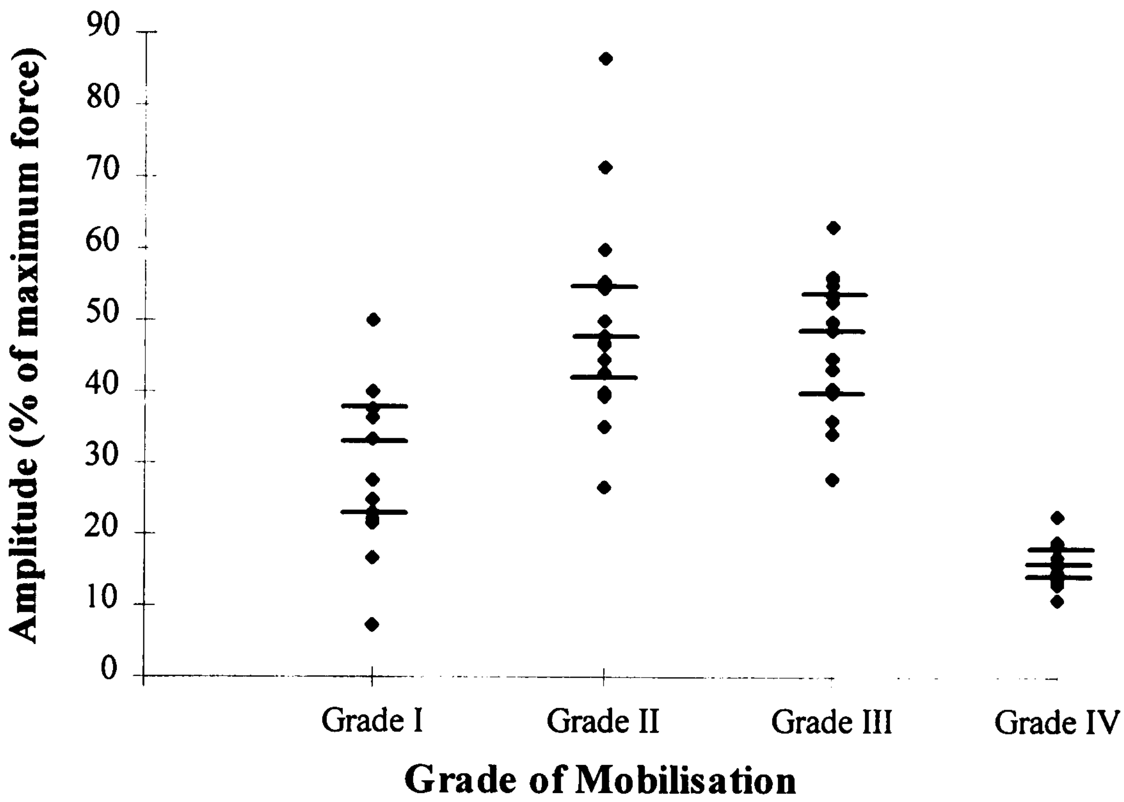


Figure 11-3: The normalised force amplitude used for each grade of mobilisation performed on a group of 16 patients. Horizontal lines indicate median and interquartile values

Figure 11-3 illustrates that there are differences between the grades of mobilisation with respect to the amplitude of oscillation and that Grade II and III mobilisations have a greater force amplitude than Grades I and IV. A Friedman two-way analysis of variance yielded a significant 'H' statistic ($p < 0.01$), demonstrating significant differences between the grades. Wilcoxon matched pairs signed rank sum test with the Bonferroni correction factor demonstrated that the comparisons between pairs of grades of mobilisation were all statistically significant at the 5 per cent level, with the exception of the comparison between Grades II and III , where the values were clearly similar.

11.3.4 Frequency of Oscillation of the Mobilisation Force

The oscillation frequency used at each grade of mobilisation is presented in Figure 11-4. It is apparent that the frequency associated with a Grade IV mobilisation is substantially higher than that associated with the other three grades.

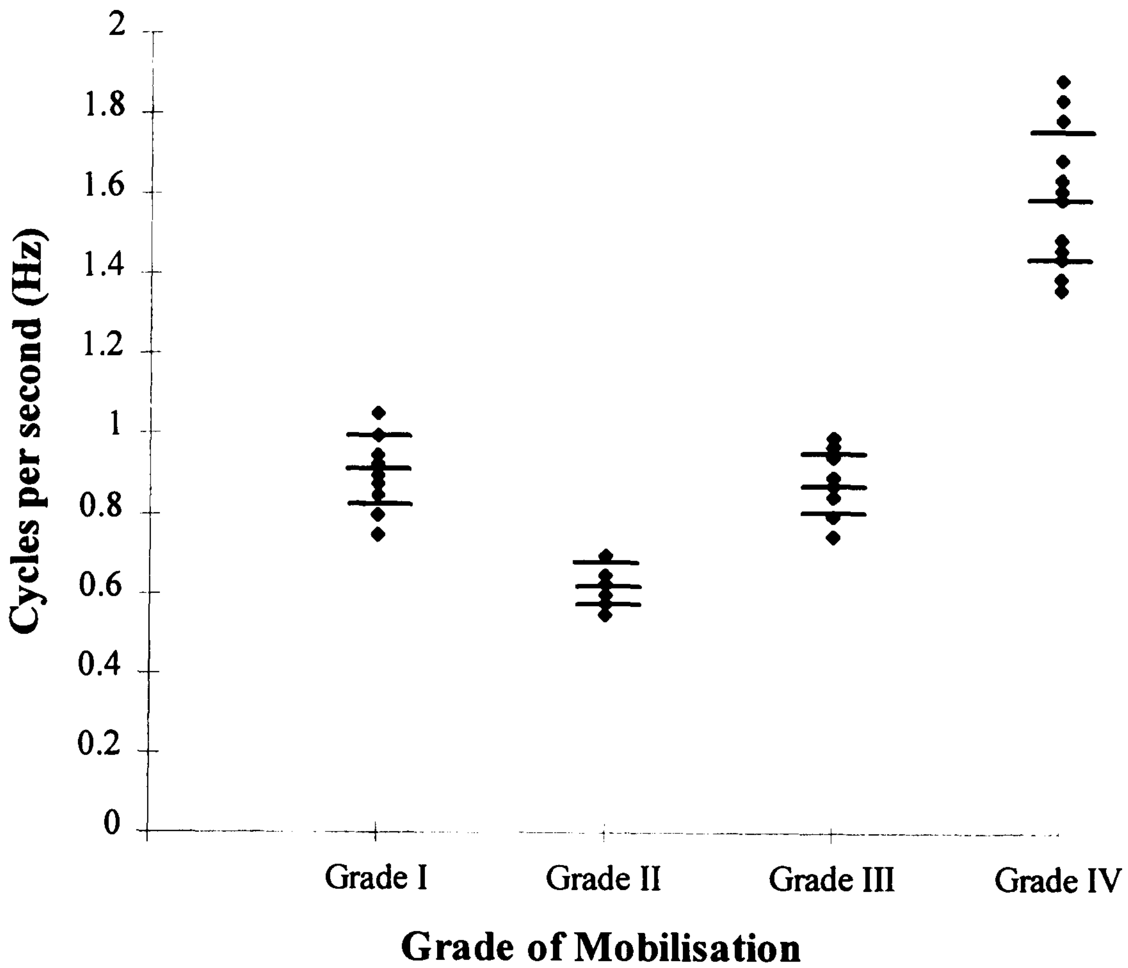


Figure 11-4: The frequency of oscillation used for each grade of mobilisation performed on a group of 16 patients. Horizontal lines indicate the mean ± 1 standard deviation

An analysis of variance yielded a significant F ratio ($p<0.001$) demonstrating differences between the frequencies used for the grades of mobilisation. Tukeys post-hoc test showed that all comparisons between the grades of mobilisation were significant ($p<0.01$) with the exception of the comparison between Grades I and III where the two distributions were similar.

11.3.5 Component Forces

Table 11-4 summarises the relative magnitude of the component forces along the vertical and horizontal axes, at each grade of mobilisation. For Grades III, IV and End Feel the results suggest that the magnitude of the force directed along the two horizontal axes was small in comparison to the magnitude of the force applied along the vertical axis.

Table 11-4: The range of forces directed along the three principle axes at each grade of mobilisation

Grade of Mobilisation	F_Z (N)	F_X (N)	F_Y (N)
End Feel	128 - 260	0 - 34	0 - 30
Grade I	4 -16	0 - 7	0 - 10
Grade II	5 - 26	0 - 7	2 - 11
Grade III	104 - 195	1 - 45	2 - 48
Grade IV	110 - 220	0 - 33	2 - 46

Table 11-4 suggests that for some patients, the therapist applied forces in the range of 45 to 48 N along the X and Y horizontal axes. However, when the raw data was examined, it became apparent that these were extreme values and were less remarkable when viewed in light of the magnitude of the vertical force component. For the majority of subjects, the vertical force component predominated. This is shown with greater clarity by the data in Table 11-5.

Table 11-5: The relative contribution of F_z, F_x and F_y to F_{total}

Grade of Mobilisation	F_z (%)		F_x (%)		F_y (%)	
	Median	Q1-Q3	Median	Q1-Q3	Median	Q1-Q3
End Feel	99.2	98.3 - 99.7	0.0	0.0 - 0.4	0.4	0.2 - 1.0
Grade I	83.3	71.2 - 91.8	3.3	1.3 - 7.4	6.4	2.1 - 20.9
Grade II	83.5	70.1 - 89.7	5.0	0.5 - 7.5	8.7	4.6 - 23.9
Grade III	99.0	95.5 - 99.4	0.2	0.1 - 0.7	0.5	0.2 - 3.3
Grade IV	98.5	97.3 - 99.4	0.1	0.0 - 0.6	0.7	0.2 - 2.6

Q1 - Q3 = Interquartile range

The median values for Grades I and II suggested that F_y contributed between 6 and 9 % of the total force. This finding was not consistent for all subjects. It seems reasonable to suggest that because the total forces measured during Grades I and II were relatively small, the contributions made by F_x and F_y were within the error of the measuring system calculated, in Section 8.2.6, to be 1.9 N for the horizontal axes.

11.4 DISCUSSION

The experimental design has been extended from the study involving healthy subjects to allow data from a patient group to be collected. The instrumented couch has provided a successful method of measuring mobilisation forces in a clinical sample. The measurement system has been shown to be adaptable to the clinical setting and appropriate for use with a patient group. The modifications made to the couch did not interfere with the procedures normally carried out during a treatment session and the system proved easy to use even with an older subject group. The data presented substantiates the results of the previous studies and provides new information on the characteristics of the techniques used on the spines of subjects with age related degenerative changes.

It had originally been intended to study patients with back pain, but whose third lumbar vertebrae was asymptomatic. It had been reasoned that the patients past experience of mobilisation techniques at other spinal levels would facilitate recruitment. However, it soon became apparent that the rate of inclusion of patients was inadequate for the time constraints of the study. Subjects were therefore recruited from all female patients who fulfilled the inclusion and exclusion criteria listed in Sections 11.2.2 and 10.2.4, respectively. All subjects were examined by a physiotherapist prior to inclusion. This helped to ensure that the third lumbar vertebra was asymptomatic, even when taken to the limit of joint range, as assessed by the End Feel procedure.

In this study, the relationship between the magnitude of the mobilisation force applied at each grade suggested that there were characteristics, apparent on palpation, which influenced the forces used during each procedure (Table 11-2). The data illustrated that the force applied was not influenced by the physical characteristics of the subject. Yet despite the fact that the magnitude of the mobilisation force did not appear to be greatly influenced by the range of movement of the lumbar spine, it is interesting to note that the correlation coefficients between these two variables were all greater than those associated with the patient characteristics with respect to the magnitude of the forces applied. The correlation coefficients were also all negative, indicating that the magnitude of the force applied varied inversely with the mobility of the spine of the subject (Table 11-3). This suggests that where the spine demonstrated greater mobility, less force was required to displace the joint to the appropriate point in range. In contrast, where the joint was less mobile, greater forces were applied by the therapist to achieve the desired effect.

Chapter 12: Discussion

The development of an instrumented couch to facilitate the measurement of forces applied during spinal mobilisation procedures has been described. This is an important advance in an area where the use of these techniques and their efficacy, remains controversial. Calibration has demonstrated that the system can be used to measure the characteristics of a mobilisation force, and the consistency of the results obtained from the individual studies, suggest that the findings can be interpreted with confidence. Measurement error was found to be acceptable and the sensitivity of the instrumented couch allowed measurement of the range of forces used for spinal mobilisation. It is important to note that the forces measured were those applied to the trunk and as will be discussed further in Section 12.7, the total force applied may not be transmitted solely through the FSU.

12.1 THERAPIST RELIABILITY

The results demonstrate substantial variation in the reliability shown by therapists in all the parameters of the mobilisation force that were assessed. In Chapter 9, limits of agreement were used to illustrate the boundaries of reliability for 95% of the therapists for each technique studied. Using this method of analysis, it will be possible in the future to establish whether the instrumented couch can be used to improve the ability of the therapist to replicate a defined force.

The data on therapist reliability generated in the previous studies and summarised in Table 5.1 cannot be used for comparison with the present data in any useful way, due to inappropriate statistical analysis. The absence of raw data reported also precludes simple comparisons. Jull and Gibson (*1986*) reported that experienced therapists could replicate mobilisation forces with reasonable accuracy, which is inconsistent with the results of this study. However, Matyas and Bach (*1985*) found large variation between the forces used by different therapists and between replications made by the same therapists.

Therapists tend to repeat the technique found to provoke the patients symptoms, at several points during a treatment session to determine the success of their intervention. Similarly, one technique or group of techniques can be used repeatedly within one session to treat the patient's condition. In this study, the assessment of repeatability gives an indication of the consistency with which the therapist replicates each technique during a treatment session. Whilst some therapists were able to reproduce the forces with reasonable accuracy, others showed wide variation (Figures 9.6-9.10).

Therapists use these techniques both for assessment and treatment over a course of care which may last several weeks. The assessment of reproducibility demonstrated that whilst some therapists were relatively consistent over a two week period, others showed considerable variation. In contrast, the results of Studies 2 and 3 where one therapist performed the techniques on two subject groups, suggest that it is possible for individual therapists to reproduce mobilisation forces with consistency (Figure 10.3 and Figure 10.5).

If the procedures used on two occasions are dissimilar, the response of the tissues will be inconsistent. The accuracy of the results of an assessment where these techniques have been used to assess joint characteristics must therefore be questionable. Similarly, the sensitivity of the therapist to detect the relatively small changes thought to result from treatment remains uncertain.

The effect of changing the sequence in which the mobilisations were performed was substantial (Table 9.7). The median force applied by the group of 30 therapists during a Grade III mobilisation, performed outside the normal grading sequence, was 34N greater than when the same procedure succeeded an End Feel and Grades I and II. It is possible that when the grades are performed sequentially, the therapist is influenced by the force applied during the preceding grades. An alternative suggestion is that the lower grades may have preconditioned the tissues prior to the Grade III mobilisation, resulting in less force being required to achieve the same effect than when a Grade III was performed in isolation. This indicates that mobilisations which progress through the grades will produce different effects in the tissues than when a single grade is performed in isolation.

To limit variability and standardise the conditions under which the therapists were operating, only one subject was used in the study of repeatability and reproducibility. Having been subjected to multiple mobilisation procedures it was conceivable that the condition of the spine of the subject changed over the period of the study and the characteristics of the soft tissues underwent inestimable changes. However, any systematic change which may have resulted from the repeated mobilisation procedures would have been revealed by the assessment of bias described in section 9.6.1. The results indicate that systematic bias was not a cause for concern in this study.

12.2 MAGNITUDE OF THE MOBILISATION FORCE

The differences between the magnitude of the force applied during each grade of mobilisation were evident in all the separate studies. This implies that most therapists maintain a similar relationship between the grades. In Chapter 4, Grades I and II were defined as being performed within the range of joint movement offering minimal resistance to the force applied by the therapist. In the studies involving both normal and patient groups, the relatively small forces measured suggest that Grades I and II are carried out within the neutral zone of the joint, or the toe region of the load displacement curve, supporting the suggestion of Lee and Evans (*1994*) discussed in Section 4.1.2. These grades are therefore unlikely to stretch the fibres of the surrounding soft tissues to any significant extent.

Similarly in Chapter 4, Grades III and IV were reported to be within the range of movement offering resistance to movement and may reach the end of the available joint range. In all three studies the relatively high forces recorded during Grades III and IV, suggested that the soft tissues were offering resistance to vertebral displacement. This implied that the range in which the joint was moving was within the linear region of the load displacement curve of the FSU (Figure 2.2). These techniques will therefore stretch the soft tissues which resist displacement of the vertebra.

Magarey (*1985*) suggests that Grades III and IV reach the same point in range, and differ only in amplitude. Similarly, because an End Feel procedure and a Grade IV mobilisation are performed at the limit of the available range, a comparable force magnitude was expected. To clarify these relationships, the protocol for Studies 2 and 3 specified that the Grade IV mobilisation should be performed at the end of the available range. However, the mean maximum forces measured in these studies increased significantly from Grade I through to Grade IV, with an additional increase in force recorded during the End Feel procedure. It is possible, however, that the Grade IV and End Feel procedures may have reached the same point in range and the additional force exerted during the End Feel may not have resulted in further movement.

A summary of the median forces generated by each mobilisation grade in the three main studies is presented in Table 12-1. The results of Therapist 13 are set apart from the main data in Study 1, to allow comparison with the results generated by the same therapist during the subsequent studies. Therapist 13 was shown to be representative of the range of therapists participating in Study 1 as discussed in Chapter 9.

Because the principal force component was directed along the vertical axis, the magnitude of the mobilisation force can be compared with the results of previous studies where only the vertical force component was measured.

Table 12-1: The median maximum forces exerted for each grade of mobilisation, measured during the research programme

Study Number	**1**		**2**		**3**
Therapist	Therapists 1-30	Therapist 13	Therapist 13		Therapist 13
Model Group	1 normal subject		26 normal subjects	5 normal subjects	16 Patients
Vertebral level	L3		L3	L1	L3
Grade of Mobilisation	**Median Force (N)**				
End Feel	191	239	243		223
Grade I	30	34	9		11
Grade II	40	41	13		15
Grade III	117 167*	102	164	170	155
Grade IV	135	113	175		171

* Grade III performed outside normal sequence

During a Grade IV mobilisation, forces of between 28 N (*Grieve & Shirley 1987*) and 329 N (*Matyas & Bach 1985*) have been reported (Table 6.1). In Study 1, forces of between 39 N and 353 N were recorded for the same procedure. However, although the results of previous studies are not refuted by the results of this study, the criticisms relating to the experimental methods discussed in Chapter 6 are still relevant.

The wide range of forces measured in Study 1 demonstrate the variation that can exist within a sample of experienced therapists. Clearly, the results generated by the small groups of therapists employed in previous studies, can not be assumed to be representative. This highlights the importance of using a large sample of therapists in any study of reliability.

The magnitude of the median mobilisation force applied to L1 during a Grade III procedure was 6N higher than when the same procedure was performed on L3. This difference represented only 4% of the median force applied. The results suggest that the position of the vertebra within the lumbar curve has little influence on the magnitude of the force applied (Table 10.8).

12.2.1 The Effects of Therapist and Model Characteristics

In the study of therapist reliability, it had been hypothesised that the physical characteristics of the therapist would have had some bearing on the forces that they applied. However, the results suggest that the variation in force could not be attributed to the weight or height of the therapist, or to their experience in the use of these techniques (Table 9.9).

In the study involving normal subjects, it had been anticipated that the nature of the resistance of the spinal tissues to movement, as determined by the therapist during palpation, would influence the characteristics of the mobilisation force. Thus, those subjects who were subjected to the highest forces during one grade, would also be subjected to the higher forces at subsequent grades. This relationship would lead to significant correlation coefficients between the five techniques. Indeed, five of the ten pair-wise comparisons were significant at the 5% level (Table 10.4), but the coefficients of determination for the statistically significant pairs were only 15 - 59 %. Generally the results suggested that the characteristics of joint movement were not interpreted by the therapists in a consistent way for each grade of mobilisation. Similarly the relationship between the physical characteristics of the subject and the force applied to their spine was examined to determine whether the weight, height or movement of the lumbar spine could account for the variation in the magnitude of the mobilisation force. However, no such relationships were demonstrated (Table 10.5).

The results of the patient group did demonstrate significant relationships between the magnitude of the force applied at one grade with those applied during the other four procedures (Figure 11.2). This indicated that those patients who were subjected to the highest forces during one grade of mobilisation were also subjected to the highest forces during the other four procedures. This implied that where pathological changes had affected the nature of joint movement, the quality and quantity of that movement influenced the characteristics of the mobilisation force in a similar way at each grade. This also indicated that where pathological changes were present, different therapists interpreted the nature of tissue resistance to movement with greater consistency. The differences between the results generated by healthy subjects and those with age related degenerative changes, emphasised the importance of substantiating the results of normal subjects with data from a clinical sample.

Again, no physical characteristic of the patient, including their age, height or weight, could account for the variation in the magnitude of the mobilisation force. Although not statistically significant, the range of spinal mobility tended to be inversely related to the magnitude of the force applied. This suggested that patients who had reduced movement in their lumbar spine, despite being asymptomatic at the L3 level, required greater force to achieve the required displacement of the mobilised vertebra. This finding is consistent with those of other studies where the relationship between the range of spinal movement and the forces required to alter the alignment of the spine has been examined (*Chase et al 1989*). It is possible that, had the patients experienced pain or clinical symptoms as a result of spinal degeneration, a stronger relationship between physical characteristics and the forces applied to their spines might have been demonstrated.

12.3 FREQUENCY OF OSCILLATION OF THE MOBILISATION FORCE

The mean frequencies recorded in the main studies are summarised in Table 12-2. The results demonstrate some consistent trends between the three studies. Grade I and III mobilisation procedures were performed at a similar frequency, while Grade II was performed generally at a slightly lower frequency and Grade IV at a higher frequency This relationship is clearly illustrated in Figures 9.19, 10.5 and 11.4. In the latter two figures, it can be seen that the Grade IV procedure is performed at almost twice the frequency of the other three grades.

Table 12-2: The mean frequency of oscillation for each grade of mobilisation, measured during the research programme

Study Number	**1**		**2**		**3**
Therapist	Therapists 1-30	Therapist 13	Therapist 13		Therapist 13
Model Group	1 normal subject		26 normal subjects	5 normal subjects	16 Patients
Vertebral level	L3		L3	L1	L3
Grade of Mobilisation	**Mean Frequency (Hz)**				
Grade I	1.1	1.1	0.8		0.9
Grade II	1.1	0.7	0.7		0.6
Grade III	1.1 1.3*	0.8	0.8	0.9	0.9
Grade IV	1.5	1.2	1.5		1.6

* Grade III performed outside normal sequence

In Chapter 4, the frequency of mobilisation advocated by various authorities was discussed. Maitland (*1986*) advised that the frequency should vary between 0.5 to 3Hz. Blake (*1984*) recommended a frequency of 2 Hz and Jull and Gibson (*1986*) reported that one experienced therapist studied performed every technique at a rate of 1.8 Hz. Whilst the range measured in this study was considerable, only one therapist approached the highest value reported by Maitland, with a recorded frequency of 2.9 Hz for a Grade IV procedure. Indeed, the mean frequencies recorded for individual therapists were generally markedly lower than those reported by Blake (*1984*) and Jull and Gibson (*1986*).

Because of the viscoelastic nature of the tissues, the rate at which spinal mobilisations are performed will affect the response of the tissues. The results of Lee and Evans (*1992*) who delivered a force to the spine in vivo, at two distinct frequencies of 0.03 and 2 Hz, suggest that lower frequencies achieve greater vertebral displacement.

Gal et al (*1994*) also suggest that manoeuvres which are carried out rapidly invoke a reflex muscular contraction and thereby increase the stiffness of the soft tissues. The results of all three studies demonstrated that the Grade IV procedure was performed at a higher frequency and with greater force than Grade I, II and III procedures. Because a Grade III and Grade IV procedure are thought to reach the same point in intervertebral range, it is possible that the increased frequency at which Grade IV was performed necessitated the higher force, reported in Table 12-1, to overcome the increased stiffness of the soft tissues. It also suggests that those therapists who perform these techniques at a lower frequency, may achieve greater intervertebral displacement for a specified force.

12.4 AMPLITUDE OF THE MOBILISATION FORCE

The subjective definitions used to describe the relative characteristics of each grade of mobilisation were given in Table 4.1. When the amplitude of each grade was considered, it was recommended that Grades I and IV should be small amplitude movements and Grades II and III, large amplitude movements (*Maitland 1986*). It is not appropriate to make a direct comparison between the force amplitude measured during each grade because of the nature of the load displacement curve of the FSU. Displacement resulting from a given amplitude at one point in the range will not result in the same displacement at another point in the range. Therefore, it cannot be assumed that a large force amplitude will result in a large amplitude movement.

This concept is illustrated in Figure 12-1 where the hypothetical position of each grade can be seen in relation to the load displacement curve of the soft tissues.

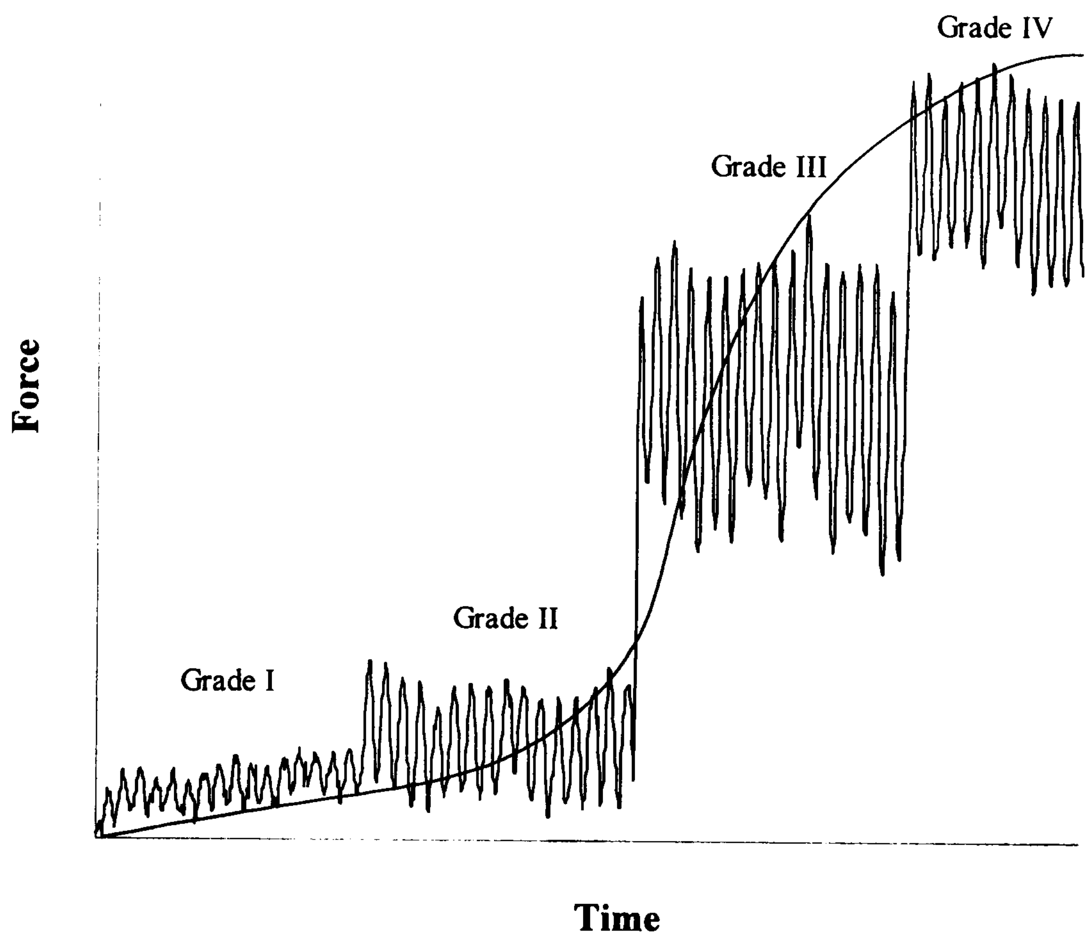


Figure 12-1: A schematic representation of a typical set of mobilisation forces superimposed on a load-displacement curve

To allow a more equitable comparison between the Grades, the normalised force amplitude was calculated as discussed in Section 9.9.5. A comparison of the results of the three studies is presented in Table 12-3. The results demonstrate that in all the studies, Grades I and IV had a smaller normalised force amplitude than Grades II and III. This suggests that the amplitude of the vertebral joint displacement is likely to be greater for Grade II and III procedures than for Grade I and IV procedures.

Table 12-3: The median normalised force amplitude of each grade of mobilisation measured during the research programme

Study Number	1		2		3
Therapist	Therapists 1-30	Therapist 13	Therapist 13		Therapist 13
Model Group	1 normal subject		26 normal subjects	5 normal subjects	16 Patients
Vertebral level	L3		L3	L1	L3
Grade of Mobilisation	**Median Force Amplitude (% of maximum)**				
Grade I	33	26	40		33
Grade II	46	62	50		48
Grade III	49 46 *	57	58	54	49
Grade IV	22	22	23		16

* Grade III performed outside the normal grading sequence

12.5 COMPONENT FORCES

The results of the three studies demonstrated that the principle force component was directed along the vertical axis. The magnitude of the horizontal components were comparatively small, contributing to only 0 to 11 % of the total force applied. This finding is probably a result of both the central position of the third lumbar vertebra within the lumbar lordosis and the choice of a posteroanterior technique.

The profile of the posteroanterior mobilisation force was measured on two lumbar vertebrae, L1 and L3 to determine whether their relative positions within the lumbar lordosis affected the force characteristics. The angle between L1 and L3 has been measured from x-rays taken with the subjects in the standing position, and was estimated to be approximately 20 degrees (*Bogduk & Twomey 1991*), although this may be less pronounced when the subject is lying prone. An anteroposterior mobilisation force is thought to result in anterior translation of one vertebra with respect to adjacent vertebrae To achieve comparable results at L1 with those reported at L3, the force would need to be angled by up to 20 degrees from the vertical axis, towards the head of the subject and would have been reflected by a greater force component along the X axis. However, no increase in the horizontal component was noted (Table 10.8), implying that the therapists made no allowance for the change in angle by modification of their technique.

12.6 THE EFFECT OF AGE RELATED CHANGES ON MOBILISATION FORCES

As discussed in Chapter 3, approximately 97 % of discs show degenerative changes in people aged over 40 years, often accompanied by osteoarthritic changes in the zygapophyseal joints (*Butler et al 1990, Eisenstein & Parry 1987*). The patients in Study 3, aged between 47 to 64 years, would therefore be expected to exhibit some signs of disc degeneration.

The data generated by the healthy group and the patient group were compared to determine the effect of age, and the degenerative changes occurring in the spine, on the profile of the mobilisation force. The Mann Whitney test, a non-parametric test for comparing data for two independent groups, was selected as it allows for inequality in sample sizes and was used in the following analyses.

12.6.1 Magnitude of the Mobilisation Force

The median maximum forces applied to the two groups during each grade are shown in Figure 12-2. The data demonstrates that the maximum forces applied to both groups by the same therapist were similar for Grades II, III and IV. However, the forces applied during the End Feel procedure were 8% lower in the patient group than in the normal group ($p<0.05$) and 18% higher in the patient group during a Grade I ($p<0.05$).

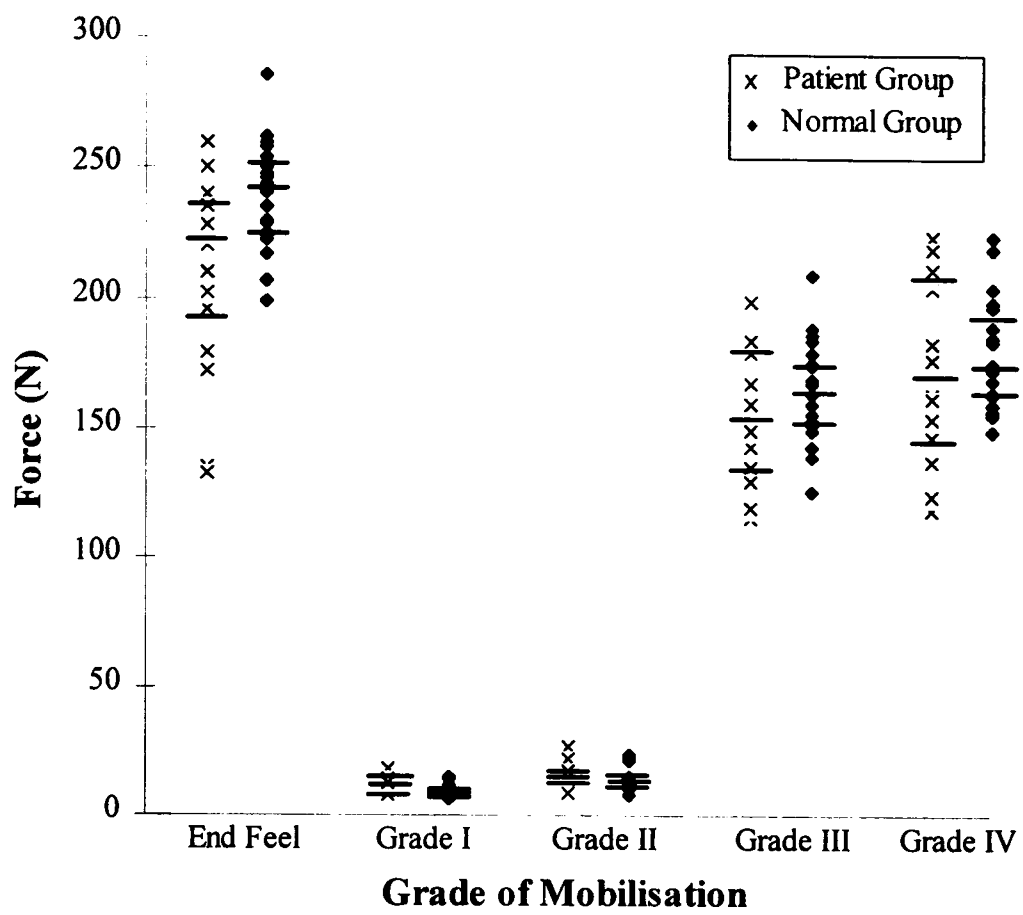


Figure 12-2: The distribution of maximum forces used for each grade of mobilisation performed on the patient and normal groups. Horizontal lines indicate median and interquartile values

The similarities between Grades II, III, IV applied to the two groups suggest that the therapist may have been guided by the magnitude of the force applied, and not by the amount of joint movement occurring. However, this was not substantiated by the difference in the forces exerted during the End Feel procedure, where a smaller force was recorded in the patient group. As discussed in Section 3.3.2, the degenerated disc deforms more rapidly than the normal disc when subjected to the same force. This suggests that in an aged spine, a smaller force would displace the joint to the end of range. In contrast, the greater force applied to the patient group during the Grade I procedure may have resulted from the nature of the soft tissues overlying the spine, or from an increased initial resistance to movement due to age related changes in the synovial fluid within the zygapophyseal joint, equivalent to an increased static coefficient of friction, or in the nucleus of the disc. As this increase is not common to the other grades and involves relatively small forces, it is unlikely to be of great clinical significance.

12.6.2 Amplitude and Frequency of Oscillation of the Mobilisation Force

The normalised amplitude of the mobilisation force was also compared for the two groups as illustrated in Figure 12-3. It was clear that the derived parameter was greater in the normal group for each of the mobilisation grades.

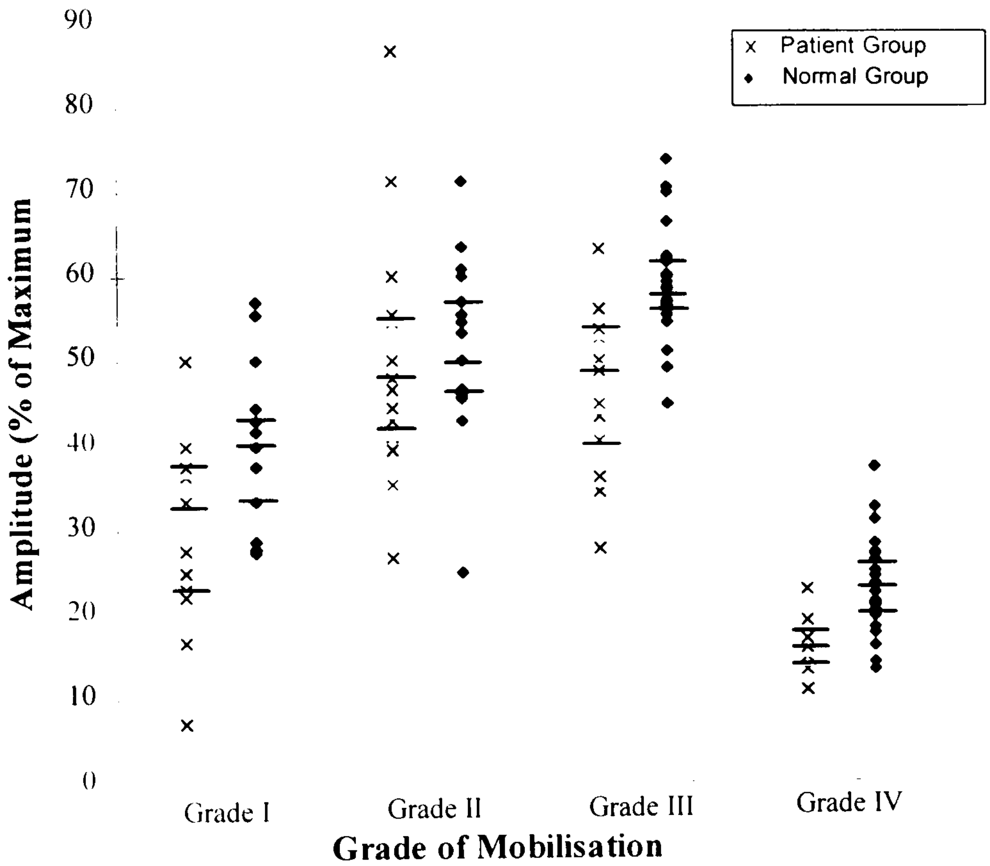


Figure 12-3: The normalised force amplitude used for each grade of mobilisation performed on the patient and normal groups. Horizontal lines indicate median and interquartile values

The differences between the groups of subjects were statistically significant for Grade I ($p < 0.01$) and Grades III and IV ($p < 0.001$), although they were not statistically significant at the 5 per cent level for Grade II mobilisation.

The frequency of oscillation for both patient and normal groups are shown in Figure 12-4. The frequencies recorded for the patient group were higher than for the normal group during Grade I ($p<0.001$), Grade III ($p<0.05$) and Grade IV ($p<0.01$). In contrast, the frequency of oscillation during a Grade II procedure was lower in the patient group than in the normal group ($p<0.05$).

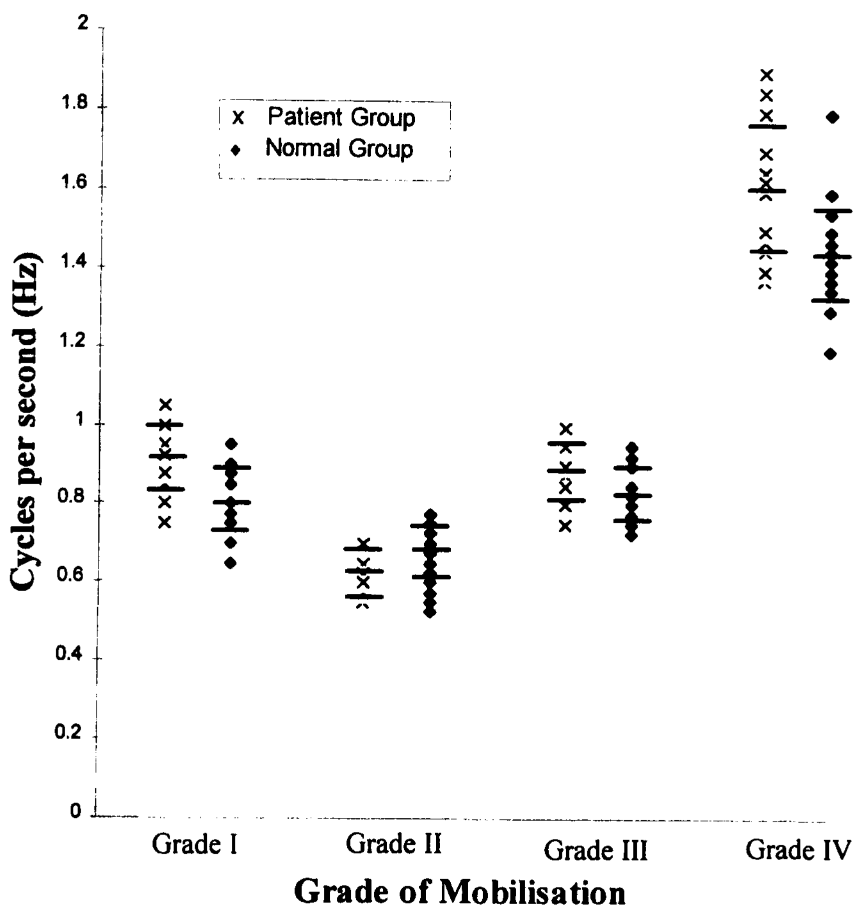


Figure 12-4: The frequency of oscillation used for each grade of mobilisation performed on a patient and a normal group. Horizontal lines indicate the mean ± 1 standard deviation

Age related changes in the spinal tissues would suggest a reduction in the compliance of the FSU. As a result, a lower frequency of oscillation in the patient group may have been expected. In fact, a higher frequency was recorded for the patient group in all but the Grade II mobilisation. It is possible that the reduced amplitude of oscillation for three of the four grades in the patient group may have resulted in a higher frequency of oscillation, despite an increase in stiffness of the FSU associated with age. This suggestion is substantiated by the results of the Grade II mobilisation, where neither the amplitude or frequency of the mobilisation force were significantly different between the groups.

12.7 ANATOMICAL CONSIDERATIONS

To predict the likely effects of spinal mobilisations and the resulting movement of the vertebral components, it is necessary to draw on the results of previous research (e.g. *Lee & Evans 1992, Schneider 1993*). Because many studies have isolated individual structures for examination and used cadaveric material, where the biomechanical properties of the tissues will be different to those in vivo, any extrapolation to intact human tissue should be undertaken with caution. However, movement of the vertebrae resulting from a postero-anterior mobilisation is likely to be:

- Extension of the lumbar spine, causing a gliding movement in both the superior and the inferior zygapophyseal joints

- Anterior translation of the mobilised vertebra with respect to adjacent vertebrae, resulting in distraction of the superior zygapophyseal joint surfaces and impaction of the inferior zygapophyseal joint surfaces (Table 2-1)

- Posterior sagittal rotation of the vertebral body. Figure 12-5 illustrates the relationship between the spinous process and the vertebral body, where a force exerted on the former is likely to result in rotation of the vertebral body

Forces of between 5 to 383 N were measured in this study, yet the effects that these forces and the resulting movement have on the surrounding ligamentous structures are uncertain. Anterior translation is thought to stretch the annular fibres of the disc, with the greatest effect on the lateral fibres (*Bogduk & Twomey 1991*). Many of the spinal ligaments are not orientated in a direction to resist anterior translation directly, but when this movement is combined with extension of the lumbar spine, several structures may be affected. A combination of anterior translation and extension, which takes the joint beyond the neutral zone of 1-2°, will result in a moderate strain on the anterior

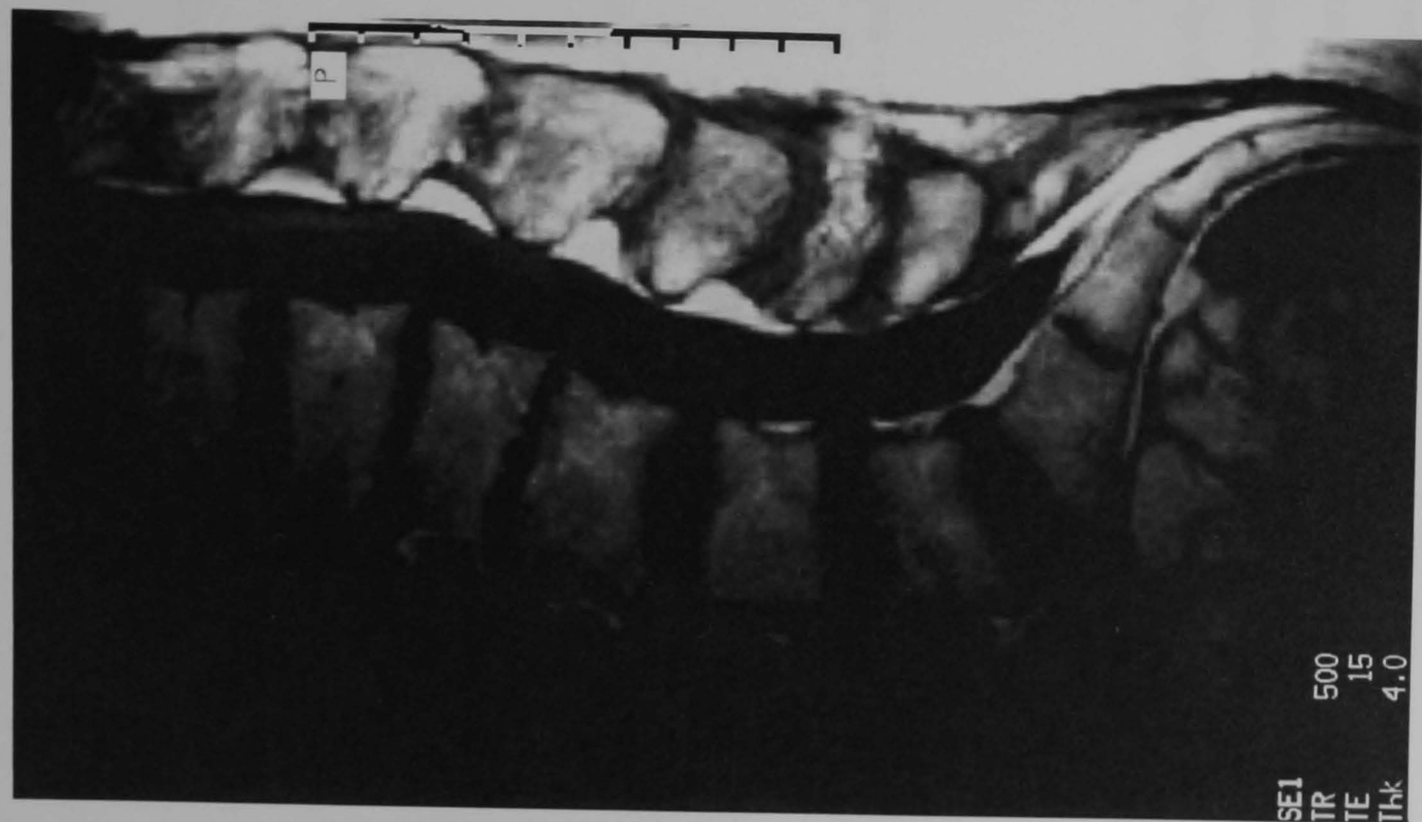


Figure 12-5: Magnetic Resonance Image of the lumbar spine illustrating the relative positions of the spinous process and vertebral body

longitudinal ligament and the anterior portion of the disc (*Panjabi et al 1982*). Similarly, because both the anterior and posterior ligaments and the ligamentum flavum are under some degree of pretension when the spine is in a neutral position (*White & Panjabi 1990*), these ligaments will tend to be influenced by any movement of the FSU. White and Panjabi (*1990*) also reported that with the exception of the Ligamentum Flavum, the spinal ligaments deformed by 1.0 to 2.5mm when subjected to a force of 150 N. Although many therapists in this study exerted forces in excess of 150 N during the higher grades of mobilisation, no single ligament bears load in isolation and the force would be distributed between several structures, which makes extrapolation difficult.

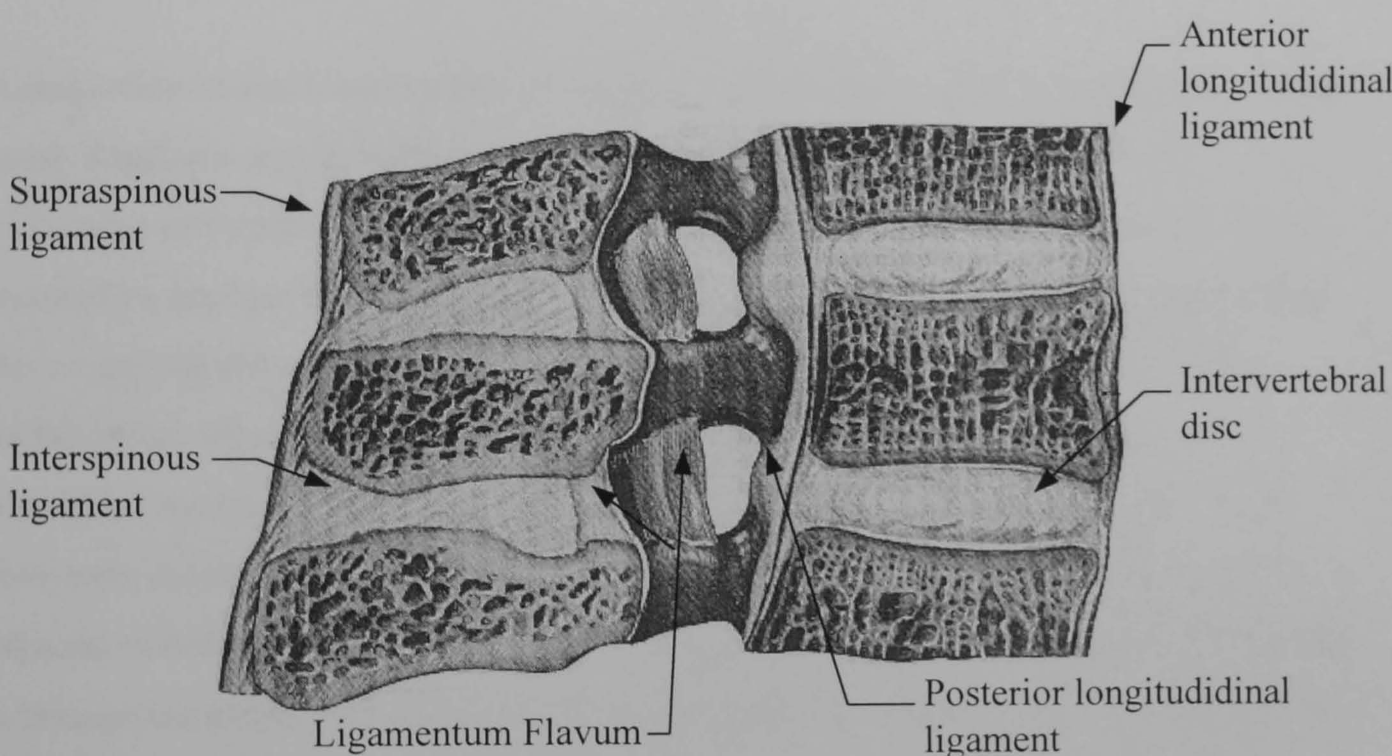


Figure 12-6: The main ligaments of the spinal column

Stokes (*1988*) demonstrated that when a horizontal force of 200 N was applied to the middle vertebra of three in a mounted jig, the resulting anterior translation was only 0.5 mm. This study also reported a maximum of 1mm of intervertebral translation under both anterior and posterior shear force. White and Panjabi (*1990*) also evaluated shear stiffness of the FSU in the horizontal plane. Anteroposterior and lateral stiffness were found to be approximately 260N/mm, indicating that relatively large forces were required to produce abnormal displacement.

The measurement of intervertebral displacement in vivo, without the use of imaging techniques, has had limited success. Lee and Evans (*1992*) measured the displacement of three vertebrae when the middle vertebra was subjected to an anteroposterior force of 150N. The displacement of the mobilised vertebra and adjacent vertebra was reported to be between 9 to 13mm, measured by linear displacement transducers resting on the overlying skin. However, although this was recorded as relative intervertebral displacement, it represents the displacement of each segment from its starting position. Therefore, because the majority of this displacement resulted from compression of the abdominal contents and the overlying skin, the study provides little information on the relative displacement between adjacent vertebrae.

A proportion of any force applied to the spine will be expended in extending the lumbar spine. Similarly, a proportion will be transmitted up and down the spine and a proportion will go towards compressing the abdominal contents. Therefore, the forces required to displace the FSU by 1mm in vivo are likely to be considerably greater than those reported above. It seems reasonable to suggest that the forces applied during mobilisation, which ranged from 5 to 383 N, would result in less than 1 mm of translatory movement, although the resulting degree of lumbar spine extension may have been considerable. The anterior translation of 1mm of one vertebra in respect to adjacent vertebrae is unlikely to compromise the neural integrity of the spinal cord. This is because the mean internal diameter of the spinal canal at the level of L3 is reported to be 17.5 mm and the spinal cord and Cauda Equina only occupy half of this space (*Panjabi at al 1992, Butler 1991*).

Using a pressure mat, Conway et al (*1993*) recorded mean forces of 364 N (SD 106N) during manipulation, at the point of cavitation of the zygapophyseal joints. Cavitation was detected by skin mounted accelerometers. Several the therapists in the study of therapist reliability applied forces which were within this range (Figure 9.13) however, no audible cracks were heard. This indicates that the high-velocity thrust used in the study by Conway et al (*1993*) may have very different effects to the same forces applied in an oscillatory, low velocity procedure.

12.8 THE EFFECT OF MOBILISATION FORCES ON THE PATHOLOGICAL CONDITION

The movement resulting from mobilisation procedures may have many effects on the components of the FSU, influencing the disease processes in diverse ways. Because of the extensibility of normal ligamentous structures. their elastic properties suggest that they are unlikely to be affected significantly. However. where scar tissue is present within the soft tissues, mobilisation procedures may help to restore normal structure and function.

During the chronic inflammatory process, movement, in conjunction with an increase in blood flow and lymphatic drainage within the tissues, is likely to encourage the removal of the by-products of inflammation and improve tissue rehabilitation. Similarly, the deposition of collagen which occurs as part of the healing process can restrict joint movement as it consolidates and contracts. Mobilisation during the healing process subjects the newly forming tissue to normal tensions, orientating the collagen fibres in line with the normal tissue fibres, helping to promote a stronger bond (*Videman 1987*).

The effect of mobilisations on the articular cartilage of the facet joint is unclear. These techniques are performed for a relatively short period, and are therefore unlikely to invoke the regenerating effects of continuous passive motion described by Salter (*1981*). However, by influencing fluid movement within the joint, improving the flow of nutrients and tissue metabolism, the general cartilage environment will be improved. In the disc, mobilisation procedures are likely to cause fluctuations in the pressure within the nucleus (*see: White & Panjabi 1990*), thereby improving diffusion and nutrition in the surrounding annulus.

It seems unlikely that mobilisations would have any gross mechanical effects on joint derangements like nerve root entrapment or meniscoid fragment displacement. However, it is possible that mobilisation techniques may have indirect effects on these conditions by altering the condition of the local environment.

Pain may be affected through several mechanisms, consistent with the theories discussed in Section 4.1.6.1. The rhythmical movement may modulate pain by stimulating receptors within the mobilised structures, resulting in a sedating effect on painful stimuli. Similarly, the mobilisation procedures may facilitate the relaxation of muscle spasm through sensory modulation. Pain will also be indirectly affected by the reduction of adhesions and inflammatory materials, and by the stretching of contracted tissues.

As discussed in section 2.4, the soft tissues demonstrate hysteresis, which is likely to occur during mobilisation procedures, producing heat within the soft tissues (*McGill & Brown 1992*). It is possible that this has a therapeutic value as a rise in temperature is known to increase the rate of tissue metabolism and the flow of fluids within the tissues (*Collins 1996*).

12.9 IMPLICATIONS FOR CLINICAL PRACTICE

The variation in forces applied by different experienced therapists at each grade of mobilisation was substantial (Figure 9.13), exceeding any differences between individual subjects and between groups of subjects with different spinal characteristics. This implies an individual interpretation of the feedback from the tissues as the joint is displaced. This undermines much of the theoretical basis of the Maitland Concept, which assumes that all therapists are applying similar forces and interpreting the characteristics of spinal movement in a similar manner. The results suggest that when the same technique is used by different therapists in the clinical environment, neither the forces used, the interpretation of the perceptual feedback, nor the results, will be comparable.

This has broader implications when the number of professional groups who use the techniques classified under the generic term of "manipulation", including osteopaths, chiropractors and physicians, are considered. The wisdom of combining the techniques of different therapists or professions in one study group, or comparing or contrasting the effects of each technique before it can be adequately defined, should be questioned. Before further progress can be made, there needs to be a new approach to the definition, teaching and standardisation of these techniques. The present study has now made it possible, with the use of an instrumented mobilisation couch, to develop a feedback system for therapists, to improve standardisation.

Whilst empirical evidence implies that mobilisation techniques are effective in treating back pain, the variation in the characteristics of the mobilisation forces suggests that if all therapists are to be considered effective, they may be using the techniques to elicit a range of therapeutic effects. For example, one therapist might apply a Grade III mobilisation slowly with a large force amplitude which would stretch the tissues, whilst another could perform it rapidly within a small part of the joint range, altering the circulatory environment. Alternatively, if only forces of a certain magnitude were to be considered effective, it must be accepted that some therapists would have little influence on the spinal components, producing mainly a placebo effect through contact with the patient. The third possibility is that small forces were sufficient to relieve the patients symptoms and that those therapists applying high forces under the same conditions, would not be producing any advantage and might indeed be causing unnecessary trauma.

The variation between therapists, illustrated in this study demonstrates the lack of standardisation in the use of these techniques. Therefore, the validity of clinical trials (e.g. *Meade et al 1990, Koes et al 1992*) which report the efficacy of these forms of treatment should be questioned, not least because the nature of the treatment administered could not be defined. The results of any trial conducted without measuring the profile of the mobilisation forces could not be interpreted with confidence and would clearly be of limited value.

Before a definition of an optimum mobilisation force profile can be offered, evidence of the effect of the posteroanterior mobilisation force on the spinal components is needed. Until further research establishes the relationship between mobilisation force and effect, it is not possible to resolve this dilemma.

12.10 LIMITATIONS AND FUTURE WORK

As reported in Table 8.16, the measurement system was sensitive to minimum forces of 0.8 N along the vertical axis and 1.9 N along the horizontal axes of the couch. This sensitivity conformed to the specification described in Section 7.1 and was thought to be appropriate for recording the minimum peak mobilisation forces, reported to be between 6.9 to 24.5 N (*Grieve & Shirley 1987, Matyas & Bach 1985*). In this study the smallest recorded force along the vertical axis was 1.6 N, which was still within the capacity of the measuring system. However, during the Grade I procedure, approximately 20-26 % of therapists applied forces of less than 1.9 N along the X and Y axes, which were smaller than the estimated sensitivity. It is therefore possible that the measurement of the horizontal force components in some of the Grade I procedures may have been influenced by measurement error.

This study has provided substantial evidence about the profile of the mobilisation force used during five techniques performed on the lumbar spine. However, whilst the movement of the spinal components under this force can be estimated, the data provides no conclusive evidence. Therefore, to extend this work, it would be necessary to measure the displacement of the vertebrae during mobilisation procedures. This could be achieved in two ways. Firstly by using a cadaveric model to measure the movement occurring directly. Secondly, by recording images of the lumbar vertebrae during mobilisation in vivo, using fluoroscopy.

Using a cadaveric specimen, the lumbar spine could be removed leaving the intervertebral discs, ligaments and overlying soft tissues intact. Forces corresponding to typical mobilisation forces could be applied to the spinous process of L3. Rigid wires could be attached to the articular facets, the spinous processes and other anatomical landmarks, projecting vertically above the spine. Angular and linear displacement of the wires could be monitored using an opto-electronic system like the 3SPACE Isotrack system (*Pearcy & Hindle 1989*). Three dimensional analysis of the displacement of the vertebra in space and in relation to adjacent vertebrae could be calculated from the relative movement of the wires. Load-displacement curves of angular and linear vertebral movement could be generated to enable an assessment of the effects of mobilisation techniques.

The stress generated within the connective tissues when the vertebrae are subjected to mobilisation forces could be measured. Transducers have been used to measure stresses within the ligaments and facet joint capsules of the spine with minimal tissue disruption. In a cadaveric model, Behrsin (*1988*) used modified strain gauges to measure microscopic deformation of the soft tissues in response to stress. Force transducers based on this principle could be used in three structures:

- Anterior Longitudinal Ligament
- Supraspinous Ligament
- Facet Joint Capsule (Dorsal and Superior fibres)

This would provide information on the stresses generated within one anterior, one posterior and one central spinal component, accessible without disruption of the FSU (*Behrsin 1988*). During these procedures, x-rays could also be taken which would enable the movement between the joint surfaces and the horizontal displacement of the vertebrae to be calculated.

Although correct storage of cadaveric material and carefully controlled laboratory conditions ensure similar biomechanical characteristics to the living model, it is recognised that neural and muscular tissue respond differently in vitro. Therefore, the results of this study will be substantiated by a study involving a clinical sample.

To enable mobilisation forces to be applied to the spines of a group of patients without exposing the therapist to ionising radiation, a device would be required to be developed and validated which could enable average mobilisation forces to be applied to the patient. The system could consist of a force applicator and a plunger joined by a tensioned cable. Forces applied to the plunger by the therapist would be transmitted along the cable and reproduced by the applicator. The applicator, mounted in a frame which could be fixed to the instrumented couch or the fluoroscopy table, could locate on the spinous process of the appropriate vertebra of the prone subject. Although the therapist may receive some feedback from the subject's tissues, the actual forces delivered by the applicator would be monitored by an in-line load cell. The load cell signal would be displayed on a monitor to provide visual feedback for the therapist. A reference force waveform generated from the data collected in Studies 2 and 3, would be superimposed on the load cell signal. By following the reference waveform, the therapist using the plunger would be able to replicate the previously measured pattern of "hands-on" application. This device would be validated and the accuracy with which the therapist could replicate a standardised mobilisation force would have to be assessed

In a second phase of this work, Video and static fluoroscopy could be used to record images of the lumbar spine whilst measured forces were applied to the spines of a range of patients with different spinal problems. Computerised digital movement analysis techniques, based on the procedures described by Brinckmann et al (*1994*) could allow measurement of small changes in vertebral alignment.

Having established the forces necessary to produce defined effects in the spinal components, therapists could then be taught, with the aid of the instrumented couch, to reproduce the forces required to produce these effects. If, for example, the aim of the technique was to stretch the capsular ligament, then it would be necessary to calculate the force magnitude and direction which would produce the requisite movement of the zygapophyseal joint surfaces. Once an acceptable degree of standardisation between therapists had been achieved, it would be possible to undertake valid studies on the efficacy of these procedures.

12.11 CONCLUSION

The instrumented couch has been shown to provide a valuable method of measuring the characteristics of the mobilisation forces used by therapists when treating the lumbar spine. This is the first system reported where the *modus operandi* of the therapist has not been affected by the measuring system. Its use in the clinical situation demonstrates that it is an important advancement in the measurement of force characteristics during clinical procedures, with considerable potential in both research and teaching applications. A data analysis protocol has also been developed and described which will allow direct comparison of the reported data with that collected in future studies.

The results demonstrated a wide variation in the forces used by different therapists performing the same technique, causing differing physical and physiological effects in the spine. The theoretical basis of these techniques suggests that therapists are guided by perceptual feedback from the soft tissues. However, the variation in forces used by a group of experienced therapists, when mobilising the same spine, suggests that this feedback is open to subjective interpretation and is largely unreliable. The implications of this will become apparent as the effects of loading the FSU in vivo are better understood.

The results of the comparison between the patient and normal subject groups demonstrate that although the characteristics of the forces applied differ in many respects, these differences are not as great as would have been expected, given the alterations in the constitution of the nucleus and the reduction in compliance of the soft tissues associated with the ageing process. The analysis suggests that whilst age and concomitant spinal degeneration effect the force amplitude and frequency of oscillation, they have little effect on the magnitude of the mobilisation force.

The dearth of basic information on the most fundamental and frequently used techniques has impeded the development of a scientific basis for the professions who use manipulative techniques, and has hindered the evaluation of the efficacy of treatment practices. In the light of changes within the Health Service and the emphasis on research based practice, improvements in treatment practices have considerable potential for making cost-effective improvements in service provision. The success of this research programme may lead to advancements in the methods of training available to many physiotherapists and other professionals. The instrumented couch promises to be a versatile tool, with many applications in the field of manual therapy. Further experimental work using an instrumented mobilisation couch will make it possible to develop a rationale and framework to facilitate the evaluation of procedures in common use.

Appendix 1: Load Cell Performance Data

Revere Transducers Europe BV

Ramshoorn 7, POBox 6909, The Netherlands

High Accuracy Load Cell - Type SHBxM

Date:	05-01-94	**Output resistance:**	349.29 Ω
Model number:	SHBxM-0.05-CC	**Isolation resistance:**	500000 MΩ, measured with 50 V DC
Part number:	604705-13		
Serial number:	168767-94	**Hysteresis:**	0.010 % of full scale
Classification:	CC	**Non-linearity:**	-0.009 % of full scale
Capacity:	50 kg	**Zero balance:**	0.057 mV/V
Input resistance:	435.16 Ω	**Cable length:**	3.00 m

Load Cell connection

The SHBXM load cell is provided with a shielded 4 Conductor cable:

Attention never cut the load cell cable

Wiring for positive polarity:

Excitation (+)=	**Green**	**Excitation (-) =**	**Black**
Output (+) =	**White**	**Output (-) =**	**Red**
Shield =	**Orange/tranp (floating)**		

Dimension	A	B	C	D	E	F	G	H	J	K	L	M	N	P
SHBxM 0-200 kg	120	38.0	22.9	31.7	21.2	42.0	10.0	82.0	18.0	20.0	10.0	20.0	8.2	8.2
Tolerance	±0.8	±0.8	±0.2	±0.3	±0.2	±0.3	±0.1	±0.1	±0.2	±0.2	±0.2	±0.2	±0.1	±0.2

Mounting	**Fixation bolts**	**Torque for the 2 fixation bolts**
SHBxM 20-200 kg	2 * M8	23 Nm

Appendix 2: Load Cell Mounting Blocks

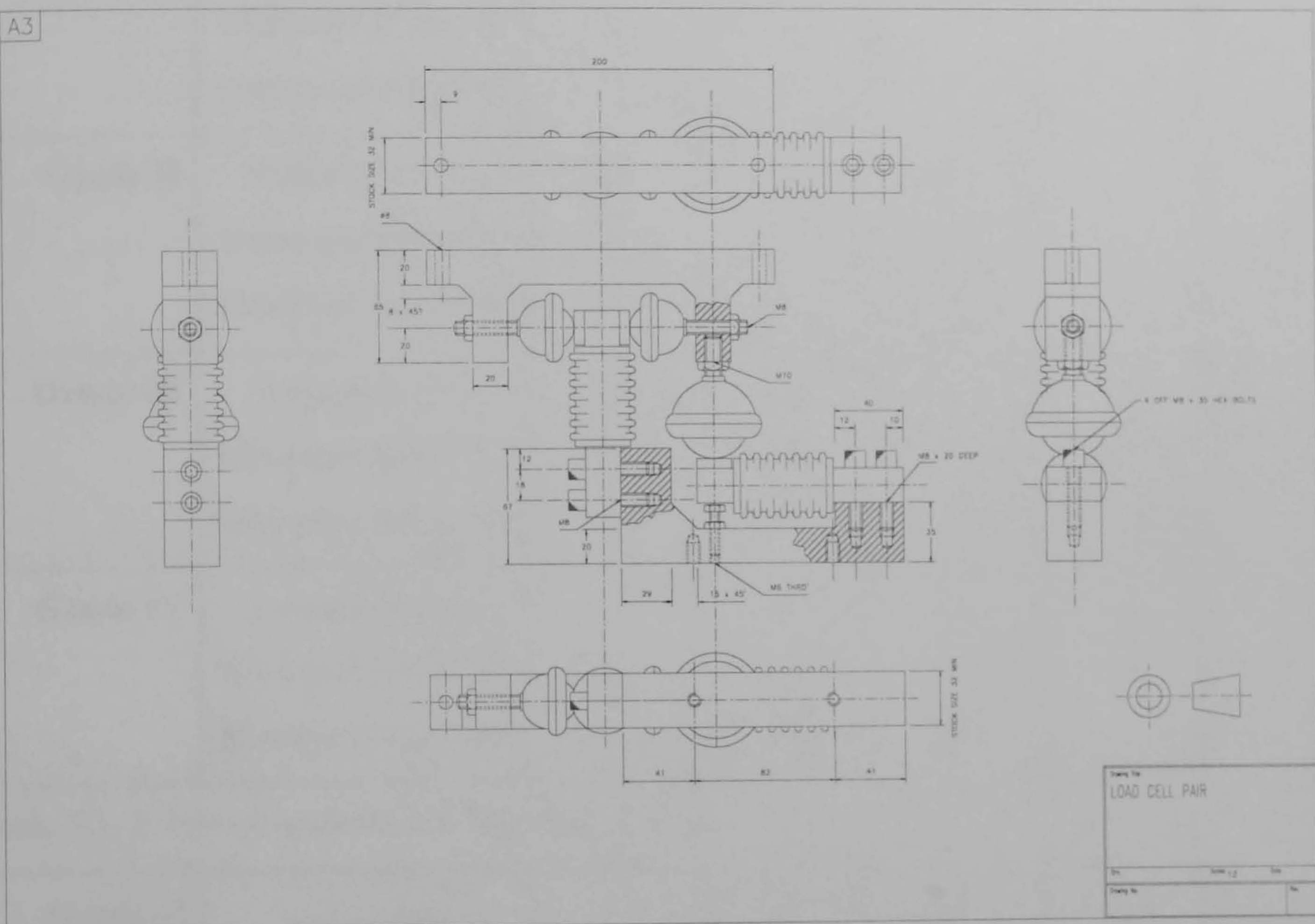


Figure A2- 1: Engineering Drawings of Load Cell Mounting Block

Appendix 3: Study One: Therapist 13

Table A3- 1:: The resultant forces recorded for Therapist 13 (F_{total})

Grade of Mobilisation	Parameter	A1	A2	B1	B2
End feel	Maximum force (N)	239	211	225	
Grade I	Frequency (Hz)	1.1	1.2	1.0	
	Minimum force (N)	26	25	11	
	Maximum force (N)	34	34	22	
Grade II	Frequency (Hz)	0.7	0.8	0.8	
	Minimum force (N)	16	20	6	
	Maximum force (N)	41	46	26	
Grade III	Frequency (Hz)	0.8	0.8	0.8	1.1
	Minimum force (N)	44	50	47	117
	Maximum force (N)	102	112	114	177
Grade IV	Frequency (Hz)	1.2	1.3	1.4	
	Minimum force (N)	88	93	101	
	Maximum force (N)	113	120	126	

Table A3- 2: Forces recorded for Therapist 13 (F_z)

Grade of Mobilisation	Parameter	A1	A2	B1	B2
End feel	Maximum force (N)	238	211	225	
Grade I	Minimum force (N)	23	22	12	
	Maximum force (N)	32	31	18.6	
Grade II	Minimum force (N)	14	18	5	
	Maximum force (N)	38	44	26	
Grade III	Minimum force (N)	41	47	44	118
	Maximum force (N)	104	114	108	177
Grade IV	Minimum force (N)	88	92	103	
	Maximum force (N)	109	119	123	

Table A3- 3: Forces recorded for Therapist 13 (F_x)

Grade of Mobilisation	Parameter	A1	A2	B1	B2
End feel	Maximum force (N)	-16	-10	10	
Grade I	Minimum force (N)	-6	-5	-2	
	Maximum force (N)	-10	-7	-6	
Grade II	Minimum force (N)	-5	-4	-2	
	Maximum force (N)	-9	-7	-6	
Grade III	Minimum force (N)	-7	-8	-5	0
	Maximum force (N)	-13	-13	-12	-6
Grade IV	Minimum force (N)	-6	-7	-7	
	Maximum force (N)	-9	-11	-11	

Table A3- 4: Forces recorded for Therapist 13 (F_y)

Grade of Mobilisation	Parameter	A1	A2	B1	B2
End feel	Maximum force (N)	17	12	13	
Grade I	Minimum force (N)	1	5	0	
	Maximum force (N)	5	10	4	
Grade II	Minimum force (N)	2	5	0	
	Maximum force (N)	7	10	4	
Grade III	Minimum force (N)	7	7	2	-2
	Maximum force (N)	14	14	7	5
Grade IV	Minimum force (N)	5	5	4	
	Maximum force (N)	10	10	7	

Appendix 4

Study One: Repeatability and Reproducibility

Table A4- 1: Mean minimum and maximum forces recorded for all therapists (F_{total})

Therapist	End Feel (N)			Grade I Minimum (N)			Grade I Maximum (N)			Grade II Minimum (N)			Grade II Maximum (N)		
	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1
1	170.7	196.2	204.0	25.5	33.4	28.4	32.4	39.2	34.8	25.5	33.4	27.5	38.3	46.1	41.2
2	110.9	130.5	133.4	4.9	3.9	16.7	9.8	7.8	21.6	5.9	4.9	18.6	12.8	13.7	28.4
3	142.2	147.2	157.9	4.9	5.9	3.9	18.6	22.1	20.6	3.9	3.9	6.9	22.6	33.4	37.3
4	143.2	192.3	167.8	8.8	11.8	14.2	21.6	27.5	31.4	2.9	4.9	5.9	34.3	44.1	45.1
5	168.7	164.8	200.1	9.8	19.6	12.8	14.7	23.5	16.7	9.8	17.7	11.8	13.7	23.5	17.7
6	127.5	147.2	188.4	11.8	19.6	22.1	19.6	29.4	29.4	17.7	22.6	23.5	31.4	42.2	43.2
7	196.2	211.9	202.1	31.4	33.4	18.6	37.3	41.2	23.5	41.2	43.2	35.3	68.7	76.5	74.6
8	355.1	376.7	278.6	13.7	15.7	14.2	22.1	23.5	19.1	12.8	18.6	11.8	31.4	39.2	31.9
9	292.3	315.9	304.1	14.7	10.8	5.9	39.2	43.2	13.7	8.8	7.8	4.9	55.9	49.1	26.5
10	196.2	211.9	268.8	13.7	9.8	33.4	19.6	17.7	39.2	28.4	24.5	45.6	66.7	54.9	69.2
11	159.9	166.8	204.0	7.8	6.9	5.9	16.7	13.7	10.8	5.9	5.9	4.9	30.4	25.5	13.7
12	186.4	167.8	249.2	26.5	49.1	47.1	34.3	54.9	56.9	27.5	54.4	45.1	39.7	70.6	66.2
13	239.4	210.9	224.6	25.5	24.5	10.8	34.3	34.3	21.6	15.7	19.6	5.9	41.2	46.1	26.5
14	308.4	282.5	248.2	70.6	85.2	4.9	75.3	91.2	9.2	77.5	97.1	6.4	86.3	110.4	16.7
15	153.0	166.8	185.4	35.3	53.0	49.1	47.1	67.7	62.8	40.2	56.9	56.9	74.6	91.2	94.2
16	121.6	107.9	141.3	8.8	8.8	4.9	12.8	12.6	9.8	7.8	9.0	2.9	13.3	12.6	8.8
17	241.3	298.2	257.0	135.4	175.6	152	155.0	198.2	174.6	159.9	167.8	144	187.4	200.1	173.6
18	215.8	382.6	352.2	41.2	53.0	48.1	70.6	94.2	84.4	70.6	82.4	66.7	129.5	160.9	155.0
19	180.5	184.4	185.4	2.9	3.9	2.0	5.9	6.9	4.9	3.9	4.9	2.4	9.8	10.8	7.7
20	174.6	188.4	192.3	4.9	5.9	2.9	7.8	7.8	5.5	5.9	6.9	3.9	9.8	9.8	8.6
21	229.6	257.0	251.1	4.9	7.8	34.3	8.8	12.8	49.1	12.8	14.7	53.0	30.4	37.3	92.2
22	286.5	237.4	251.1	7.8	7.8	15.7	11.8	13.7	31.4	6.9	9.8	26.5	13.7	16.7	53.0
23	213.9	202.1	182.5	20.6	23.5	20.6	34.3	39.2	34.8	30.4	35.3	41.2	56.9	63.8	75.5
24	202.1	223.7	184.4	33.4	58.9	23.5	47.1	73.6	34.3	48.1	64.7	29.9	71.6	96.1	58.9
25	174.6	162.8	145.2	14.7	17.7	13.2	36.3	41.2	33.4	16.7	13.7	11.8	41.2	54.0	47.1
26	255.1	217.8	237.4	27.5	43.2	16.7	39.2	53.0	25.5	35.3	49.1	21.6	73.6	93.2	50.0
27	160.9	149.1	160.9	3.9	3.9	2.9	7.8	7.8	6.9	4.9	4.9	2.9	7.8	9.8	5.9
28	247.2	270.8	211.9	83.4	61.8	37.3	94.2	78.5	48.1	83.4	60.8	54.0	107.9	92.2	88.3
29	96.1	92.2	94.2	21.6	26.5	7.8	28.4	35.3	12.8	23.5	30.4	8.8	31.4	44.1	17.7
30	208.0	194.2	202.1	47.1	37.3	46.6	52.0	46.1	54.0	43.2	56.9	63.8	62.8	83.4	85.3
Median	191.3	195.2	202.1	14.7	19.6	16.2	30.4	34.8	27.5	17.2	21.1	20.1	39.0	46.1	44.2
Min	96.1	92.2	94.2	2.9	3.9	2.0	5.9	6.9	4.9	2.9	3.9	2.4	7.8	9.8	5.9
Max	355.1	382.6	352.2	135.4	175.6	152	155.0	198.2	174.6	159.9	167.8	144	187.4	200.1	173.6

Table A4- 2: Mean minimum and maximum forces recorded for all therapists (F_{total} continued....)

Therapist	Grade III Minimum (N)				Grade III Maximum (N)				Grade IV Minimum (N)			Grade IV Maximum (N)		
	A1	A2	B1	B2	A1	A2	B1	B2	A1	A2	B1	A1	A2	B1
1	45.1	53.0	46.1	38.3	68.7	72.6	64.7	55.9	63.8	66.7	61.3	75.5	78.5	71.1
2	37.3	56.9	58.9	68.7	72.6	100.1	113.8	149.1	63.8	61.8	93.2	78.5	74.6	113.8
3	15.7	17.7	17.7	29.4	68.7	78.5	80.4	109.9	49.1	66.7	106.9	102.0	123.6	166.8
4	3.9	3.9	11.8	13.7	115.8	141.3	151.1	180.5	111.8	110.9	130.5	133.4	145.2	157.0
5	56.9	72.6	80.4	113.8	78.5	82.4	98.1	133.4	80.4	102.0	132.4	88.3	109.9	142.2
6	35.3	35.3	47.1	74.6	78.5	81.4	105.9	137.3	65.7	78.5	97.1	96.1	107.9	107.9
7	68.7	66.7	78.5	90.3	117.7	117.7	145.2	190.3	95.2	92.2	97.1	124.6	131.5	136.4
8	107.9	103.0	63.8	93.2	313.9	279.6	250.2	284.5	202.1	189.3	184.4	262.9	266.8	252.1
9	39.2	17.7	9.8	27.5	198.7	139.3	92.2	184.4	107.9	68.7	72.6	284.5	204.0	190.3
10	83.4	64.7	93.2	127.5	171.7	137.3	163.8	209.9	142.2	113.8	143.2	191.3	164.8	188.4
11	23.5	17.7	47.1	76.5	88.3	77.5	102.0	141.3	84.4	51.0	118.7	132.4	113.8	149.1
12	55.9	94.2	146.7	139.8	80.0	121.6	160.9	171.7	101.5	138.3	146.7	113.3	151.6	160.9
13	44.1	50.0	47.1	117.7	102.0	111.8	113.8	176.6	88.3	93.2	101.0	112.8	120.2	125.6
14	172.7	192.3	66.7	98.1	227.6	247.2	149.1	153.0	231.5	237.4	125.6	260.9	276.6	157.0
15	60.8	83.4	88.3	123.6	107.9	132.4	127.5	162.8	96.1	98.1	105.0	116.7	125.6	133.4
16	45.1	45.1	27.0	41.2	82.4	88.3	65.3	86.3	105.0	98.1	74.1	118.7	114.8	91.2
17	199.1	180.5	169.7	228.6	228.6	214.8	217.8	302.1	220.7	202.1	218.8	249.2	227.6	264.9
18	166.8	142.2	127.5	171.7	289.4	274.7	284.5	323.7	220.7	206.0	219.7	347.3	318.8	353.2
19	39.2	54.9	43.2	54.9	107.9	143.2	105.9	135.4	94.2	92.2	90.3	137.3	137.3	133.4
20	58.9	47.1	35.3	31.4	191.3	158.9	139.3	139.3	141.3	110.9	84.4	169.7	143.2	122.6
21	55.9	60.8	88.3	117.7	125.6	166.8	158.9	206.0	139.3	147.2	190.3	223.7	255.1	227.6
22	74.6	58.9	70.6	78.5	145.2	137.3	145.2	176.6	162.8	139.3	139.3	214.8	192.3	192.3
23	62.8	74.6	70.6	90.3	135.4	133.4	141.3	190.3	125.6	119.7	139.3	166.8	172.7	184.4
24	92.2	90.3	54.9	93.2	171.7	162.8	131.5	158.9	134.4	121.6	79.5	165.8	160.9	118.7
25	17.7	19.6	22.6	19.6	82.4	81.4	88.3	86.3	43.2	36.3	42.2	74.6	78.5	68.7
26	63.8	99.1	72.6	83.4	191.3	171.7	145.2	191.3	122.6	123.6	113.8	156.0	152.1	145.2
27	40.2	29.4	25.5	55.9	66.7	51.0	46.1	86.3	86.3	73.6	66.7	105.9	91.2	80.4
28	157.0	93.2	115.8	132.4	237.4	152.1	200.1	220.7	212.9	126.5	180.5	250.2	174.6	223.7
29	29.4	38.3	17.7	21.6	47.1	56.9	25.5	39.2	40.2	49.1	24.5	62.8	74.6	39.2
30	75.5	93.2	105.0	149.1	119.7	142.2	146.2	193.3	108.9	122.6	160.9	138.3	147.2	181.5
Median	56.4	59.8	61.3	86.8	116.7	135.4	135.4	167.3	106.4	106.4	110.4	135.4	144.2	147.2
Min	3.9	3.9	9.8	13.7	47.1	51.0	25.5	39.2	40.2	36.3	24.5	62.8	74.6	39.2
Max	199.1	192.3	169.7	228.6	313.9	279.6	284.5	323.7	231.5	237.4	219.7	347.3	318.8	353.2

Table A4- 3: Mean minimum and maximum forces recorded for all therapists (F_z)

Therapist	End Feel (N)			Grade I Minimum (N)			Grade I Maximum (N)			Grade II Minimum (N)			Grade II Maximum (N)		
	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1
1	168.7	196.2	202.1	24.5	31.9	27.5	31.4	38.3	34.3	23.5	32.9	27.5	37.3	44.1	41.2
2	104.0	125.6	133.4	4.7	2.7	16.1	6.9	6.3	20.8	5.7	4.5	18.6	11.0	12.0	28.4
3	138.3	143.2	157.0	3.9	3.9	2.9	11.8	20.6	18.6	2.9	2.9	6.9	21.6	32.4	36.3
4	139.3	192.3	165.8	5.9	10.8	13.2	19.6	24.5	28.4	0.0	0.0	4.9	32.4	41.2	43.2
5	166.8	164.8	196.2	9.8	19.1	11.8	11.8	24.5	15.7	8.6	17.7	11.8	11.8	22.6	15.7
6	125.6	147.2	187.4	10.8	18.6	20.6	17.7	28.4	28.4	16.7	22.1	23.5	30.9	39.2	41.2
7	198.2	211.9	200.1	30.9	31.4	17.2	37.3	39.2	26.5	40.2	44.1	35.3	69.7	77.5	74.6
8	359.0	380.6	279.6	10.8	15.7	12.8	19.6	20.6	17.7	12.8	16.2	10.8	29.9	37.3	31.9
9	294.3	315.9	306.1	12.8	10.8	1.0	39.2	42.2	10.8	5.9	2.0	0.0	56.9	47.1	25.5
10	194.2	208.0	268.8	11.8	10.3	32.9	17.7	16.2	38.3	29.4	24.5	44.1	64.7	53.0	67.7
11	158.9	165.8	203.1	6.9	5.9	7.1	15.2	11.8	10.2	4.9	0.0	4.3	29.4	23.5	12.8
12	181.5	161.9	249.2	25.5	47.6	47.1	33.8	55.9	56.9	22.1	49.1	45.1	33.8	64.7	64.7
13	238.2	211.0	224.6	22.9	22.0	11.8	32.0	31.2	18.6	13.9	17.6	4.9	38.3	43.8	25.5
14	304.1	281.5	247.2	64.7	80.4	2.4	71.6	85.8	6.7	71.1	81.4	5.1	90.7	105.5	14.9
15	153.0	166.8	184.4	35.8	50.5	50.0	46.6	66.7	61.8	40.2	56.9	58.9	74.6	91.2	93.2
16	118.2	101.0	140.3	3.9	3.9	2.0	6.9	7.8	5.4	2.9	2.9	0.0	8.8	8.8	6.9
17	241.3	298.2	257.0	134.4	170.7	152	152.1	198.2	170.7	159.9	168.7	147	186.4	201.1	174.6
18	215.8	382.6	351.2	40.2	54.0	47.1	68.7	90.3	81.4	67.7	78.5	66.7	122.6	158.9	155.0
19	178.5	184.4	183.4	0.0	1.0	0.0	3.9	3.9	2.0	1.0	1.5	0.0	7.8	9.3	6.1
20	171.7	184.4	192.3	2.2	1.6	2.0	4.7	3.5	3.1	2.7	2.0	2.4	6.3	4.3	7.5
21	224.6	255.1	251.1	3.9	4.9	9.8	7.8	11.3	19.6	11.8	13.7	35.3	28.4	35.3	86.3
22	286.5	238.4	252.1	5.9	6.9	17.7	9.3	11.3	32.4	6.4	7.8	27.5	12.8	14.7	52.0
23	212.9	200.1	182.5	19.1	23.1	19.6	33.4	36.3	33.4	29.4	33.4	40.2	56.9	62.8	74.6
24	201.1	219.7	183.4	33.4	54.8	24.0	48.1	71.1	33.4	45.1	64.7	29.4	69.7	94.2	56.9
25	169.7	162.8	141.3	13.7	15.7	11.8	34.8	44.1	33.4	15.7	12.8	11.8	42.2	53.0	45.1
26	255.1	217.8	236.4	28.9	43.2	16.2	38.3	52.5	24.0	35.3	49.1	21.6	72.6	91.2	49.1
27	158.9	147.2	158.9	2.4	2.0	2.0	4.7	3.9	3.9	2.0	2.9	2.0	5.9	7.8	3.9
28	243.3	267.8	209.0	82.9	60.8	36.3	92.7	76.0	47.1	82.4	60.3	53.0	105.9	90.7	85.3
29	96.5	92.2	95.2	21.6	27.0	6.9	27.5	34.8	12.3	23.2	30.9	8.8	31.4	43.7	16.7
30	206.0	192.3	225.6	42.7	37.3	45.1	47.6	44.1	54.0	42.2	54.0	63.8	58.9	80.4	83.4
Median	187.9	194.2	201.1	13.2	18.9	14.7	29.4	33.0	25.3	16.2	19.9	20.1	35.6	44.0	42.2
Min	96.5	92.2	95.2	0.0	1.0	0.0	3.9	3.5	2.0	0.0	0.0	0.0	5.9	4.3	3.9
Max	359.0	382.6	351.2	134.4	170.7	152	152.1	198.2	170.7	159.9	168.7	147	186.4	201.1	174.6

Table A4- 4: Mean minimum and maximum forces recorded for all therapists (F_z continued...)

Therapist	Grade III Minimum (N)				Grade III Maximum (N)				Grade IV Minimum (N)			Grade IV Maximum (N)		
	A1	A2	B1	B2	A1	A2	B1	B2	A1	A2	B1	A1	A2	B1
1	44.1	51.5	46.6	34.8	66.7	72.6	63.8	53.5	62.8	64.7	60.8	74.1	79.5	70.6
2	38.7	54.9	58.9	69.7	71.6	98.1	113.8	153.0	62.8	61.8	94.2	77.5	74.1	112.8
3	13.7	19.6	15.7	27.5	66.7	80.4	77.5	113.8	49.1	66.7	101.0	102.0	125.6	163.8
4	3.9	4.9	7.8	11.8	109.9	141.3	147.2	174.6	110.9	110.9	126.5	130.5	145.2	155.0
5	56.9	71.6	78.5	112.8	77.5	81.9	94.2	132.4	80.0	101.0	129.5	87.8	107.4	136.8
6	32.4	34.3	47.1	70.6	78.5	81.4	107.9	139.3	66.7	76.5	86.3	95.2	106.9	107.9
7	67.7	62.8	76.5	94.2	119.7	113.8	146.2	186.4	95.2	89.3	96.1	127.5	132.4	134.4
8	107.9	98.1	58.9	93.2	313.9	279.6	245.3	279.6	202.1	192.3	184.4	260.9	270.8	247.2
9	6.9	19.6	7.8	25.5	201.1	141.3	90.3	184.4	117.7	72.6	78.5	284.5	200.1	194.2
10	82.4	62.8	90.3	127.5	166.8	131.5	160.9	208.0	138.3	113.8	140.3	185.4	157.9	193.3
11	21.6	14.7	43.2	76.5	88.3	76.5	102.0	138.3	88.3	51.0	118.7	133.4	113.8	148.1
12	50.5	93.2	96.1	139.8	71.1	121.2	125.6	169.2	94.7	134.4	145.2	107.4	146.2	158.9
13	40.6	47.4	44.1	117.7	98.7	113.8	107.9	176.6	87.7	91.9	103.0	109.4	118.5	122.6
14	168.7	225.6	68.7	94.2	191.3	248.2	147.2	151.1	230.5	260.0	123.6	237.4	274.7	151.1
15	59.8	79.5	86.3	120.7	108.9	130.5	127.5	163.8	84.4	97.1	104.0	116.7	124.6	132.4
16	40.2	43.2	23.5	41.2	78.5	88.3	62.8	86.3	97.1	90.7	72.1	111.8	108.9	88.8
17	200.1	180.5	170.7	225.6	228.6	216.8	216.8	301.2	220.7	203.1	217.8	248.2	228.6	262.9
18	166.8	142.2	127.5	166.8	294.3	274.7	284.5	318.8	220.7	206.0	225.6	348.3	315.9	348.3
19	37.3	51.0	45.1	53.0	105.9	141.3	105.0	133.4	93.2	86.3	91.2	137.3	135.4	129.5
20	58.9	53.0	35.3	33.4	186.4	160.9	141.3	141.3	141.3	109.9	85.3	168.7	143.2	122.6
21	51.0	58.9	78.5	107.9	121.6	155.0	155.0	206.0	135.4	147.2	164.8	217.8	245.3	230.5
22	72.6	62.8	68.7	76.5	145.2	133.4	136.4	176.6	162.8	143.2	135.4	213.9	190.3	194.2
23	68.7	74.6	68.7	86.3	131.5	127.5	141.3	182.5	123.6	116.7	138.3	162.8	167.8	182.5
24	94.2	88.3	54.9	92.2	170.7	160.9	127.5	157.0	134.4	119.7	80.4	166.8	158.9	117.7
25	17.7	17.7	22.6	19.6	77.5	78.5	88.3	86.3	43.2	36.3	42.2	72.6	76.5	67.7
26	63.8	96.1	72.6	82.4	191.3	172.7	145.2	192.3	122.6	125.6	113.8	157.0	152.1	147.2
27	38.3	27.5	24.5	54.0	65.7	52.0	43.2	83.4	83.4	72.1	65.2	104.0	87.3	83.4
28	157.0	92.2	117.7	132.4	237.4	147.2	196.2	220.7	210.9	122.6	179.5	248.2	171.7	224.6
29	28.4	37.8	15.7	20.6	46.1	55.4	25.5	37.3	40.2	48.6	23.5	63.8	73.6	39.2
30	74.6	93.2	103.0	145.2	111.8	138.3	143.2	187.4	105.9	121.6	155.0	133.4	141.3	176.6
Median	54.0	60.8	58.9	84.4	110.9	131.0	127.5	166.5	101.5	105.5	108.9	133.4	142.2	147.6
Min	3.9	4.9	7.8	11.8	46.1	52.0	25.5	37.3	40.2	36.3	23.5	63.8	73.6	39.2
Max	200.1	225.6	170.7	225.6	313.9	279.6	284.5	318.8	230.5	260.0	225.6	348.3	315.9	348.3

Table A4- 5: Mean minimum and maximum forces recorded for all therapists (F_y)

Therapist	End Feel (N)			Grade I Minimum (N)			Grade I Maximum (N)			Grade II Minimum (N)			Grade II Maximum (N)		
	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1
1	0.0	0.0	3.9	0.0	5.9	0.0	5.9	10.8	0.0	1.0	7.1	0.0	5.9	10.6	2.5
2	-11.8	-11.8	-12.8	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
3	3.9	-2.0	8.8	0.0	0.0	0.0	0.0	0.0	0.0	1.0	1.0	1.6	-2.9	-2.9	-3.1
4	0.0	-5.9	12.6	1.0	0.0	6.7	2.9	0.0	7.8	0.0	0.0	0.0	0.0	0.0	5.9
5	-25.5	-11.4	-27.5	0.0	0.0	0.0	0.0	-2.9	-3.9	0.0	0.0	0.0	0.0	0.0	0.0
6	14.7	5.9	16.7	0.0	-2.0	2.9	0.0	-4.9	6.9	0.0	0.0	1.0	0.0	0.0	5.9
7	10.8	8.8	2.9	0.0	1.0	0.0	2.9	3.9	2.9	1.0	2.9	0.0	3.9	5.9	2.0
8	-32.4	-25.5	13.7	5.1	1.0	0.0	7.8	3.9	2.9	0.0	1.6	0.0	0.0	3.9	0.0
9	-24.5	-16.7	-13.7	0.0	0.4	0.0	3.9	4.3	3.1	0.0	1.0	0.0	3.9	3.9	2.0
10	-24.5	-27.5	-27.5	-4.3	0.0	-3.9	-6.7	-2.9	-5.9	-5.9	-3.5	-3.9	-11.8	-7.8	-7.8
11	-15.7	-15.7	-11.8	-2.0	-1.0	0.0	-3.9	-2.9	0.0	-1.0	-1.0	0.0	-2.9	-2.9	-2.0
12	-7.8	-7.1	-20.6	-2.0	0.0	0.0	-4.9	0.0	0.0	-0.8	0.0	-4.9	-3.5	0.0	-6.9
13	17.3	12.1	-12.8	0.9	5.2	0.0	5.2	10.4	3.9	1.7	5.2	0.0	6.9	10.4	3.9
14	-9.8	-9.8	-18.6	5.9	9.8	0.0	9.8	13.7	2.9	5.4	6.4	0.0	8.8	9.8	2.5
15	7.8	7.8	11.8	0.0	0.0	0.0	4.4	4.7	3.9	0.0	1.2	0.0	7.8	5.5	3.9
16	15.7	13.3	17.7	0.0	0.0	0.0	0.0	0.0	0.0	-2.0	0.0	-2.0	2.0	0.0	2.0
17	-7.8	-11.8	-19.6	-1.0	0.0	0.0	-2.9	0.0	-4.9	0.0	0.0	0.0	0.0	0.0	-4.9
18	15.7	17.7	2.9	-7.8	-8.8	-4.9	-14.7	-14.7	-11.8	-18.6	-14.2	-10.8	-29.4	-25.0	-21.6
19	13.7	12.8	25.5	0.0	0.0	0.0	2.0	2.9	2.9	0.0	1.0	0.0	2.0	2.9	3.9
20	23.5	25.5	9.8	2.7	3.5	0.0	4.7	5.5	2.9	3.5	3.9	2.0	6.3	6.5	3.9
21	15.7	14.7	-14.7	1.0	2.0	2.0	2.9	4.9	-2.0	0.0	2.9	0.0	2.9	6.9	19.6
22	-5.9	-3.9	0.0	2.9	2.9	2.0	4.9	4.9	3.9	2.9	2.9	2.0	4.9	5.9	3.9
23	2.9	2.0	-7.8	1.0	1.0	-1.0	2.9	3.5	-3.9	1.0	2.0	2.0	2.9	5.9	-2.9
24	4.9	25.5	16.7	1.0	4.9	2.9	3.9	8.8	6.3	0.0	4.9	0.0	2.0	12.8	9.8
25	4.9	7.8	11.8	0.0	0.0	0.0	3.9	2.9	2.9	0.0	0.0	0.0	3.9	3.9	5.9
26	-4.9	6.9	12.8	0.0	0.0	0.0	-2.0	2.9	2.9	-2.9	-2.0	0.0	2.9	6.9	2.9
27	13.7	17.7	15.7	0.0	0.0	0.0	2.9	2.9	2.5	0.8	1.0	0.0	2.6	2.9	2.9
28	36.3	32.4	29.4	13.7	13.2	6.9	17.7	17.7	11.4	12.8	9.8	8.8	21.6	17.1	19.6
29	0.0	-5.9	0.0	0.0	-1.0	0.0	2.9	2.0	2.9	0.0	-1.0	0.0	2.9	2.0	2.9
30	23.5	21.6	26.5	4.9	3.4	5.9	7.8	6.4	9.8	5.9	5.4	9.8	12.8	12.3	17.7
Median*	12.8	11.8	13.2	0.9	1.0	0.0	3.9	3.7	2.9	0.9	1.4	0.0	2.9	4.7	3.9
Min*	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Max*	36.3	32.4	29.4	13.7	13.2	6.9	17.7	17.7	11.8	18.6	14.2	10.8	29.4	25.0	21.6

* Absolute values

Table A4- 6: Mean minimum and maximum forces recorded for all therapists (F_y continued....)

Therapist	Grade III Minimum (N)				Grade III Maximum (N)				Grade IV Minimum (N)			Grade IV Maximum (N)		
	A1	A2	B1	B2	A1	A2	B1	B2	A1	A2	B1	A1	A2	B1
1	3.9	6.3	1.0	1.6	7.8	9.4	4.9	5.1	5.9	8.8	2.4	11.8	12.8	4.7
2	0.0	0.0	-2.9	-4.9	-2.0	-2.0	11.8	-9.8	-1.0	0.0	-4.9	-3.9	-2.9	-9.8
3	-1.0	2.4	5.9	5.9	-4.9	-1.6	-2.0	-2.7	0.0	1.0	10.8	-3.9	5.9	25.0
4	0.0	1.0	0.0	0.0	7.8	-5.9	13.2	11.8	2.4	-3.9	7.8	7.5	-7.8	13.2
5	-3.5	-9.8	-8.8	-13.7	-6.7	-14.7	-13.7	-8.8	-7.8	-13.7	-26.7	-11.8	-17.7	-29.8
6	1.0	-1.0	0.0	-2.0	4.9	-4.9	4.9	4.9	1.0	-1.0	0.0	5.9	-4.9	0.0
7	3.9	2.0	1.0	3.9	6.9	4.9	7.8	17.7	4.9	4.9	4.9	7.8	8.8	9.8
8	-4.9	-4.9	0.0	0.0	4.9	4.9	25.5	34.3	2.0	6.9	6.9	5.9	11.8	13.2
9	0.0	0.0	-5.9	-2.0	0.0	3.9	5.9	7.8	-1.0	2.0	-3.9	-4.9	7.8	2.9
10	-10.8	-9.8	-13.7	-15.7	-26.5	-26.5	-25.5	-28.4	-21.6	-20.6	-23.5	-32.4	-28.4	-33.4
11	-6.9	-4.9	0.0	-1.0	-12.8	-7.8	-2.0	-3.9	-9.8	-3.9	-3.9	-13.7	-6.9	-7.8
12	0.0	0.0	-13.7	-11.8	0.0	0.0	-17.7	-15.7	0.0	-2.0	-22.6	0.0	-3.9	-26.5
13	6.9	6.9	2.0	-2.0	13.8	13.8	7.4	4.9	5.2	5.2	3.9	10.4	10.4	6.9
14	-2.0	1.0	-2.0	-1.0	4.9	4.9	-7.8	-5.9	6.9	0.0	-3.9	13.7	7.8	-7.7
15	5.5	2.0	1.6	3.9	10.8	8.8	6.3	9.8	6.9	6.3	2.9	12.8	10.6	9.8
16	-1.0	2.0	0.0	0.0	5.9	12.8	4.9	10.8	14.7	9.8	3.9	19.6	16.7	9.8
17	-1.0	0.0	0.0	0.0	-4.9	0.0	-4.9	-4.9	-3.9	0.0	0.0	-6.9	0.0	-4.9
18	-2.9	-5.9	-6.9	-7.8	4.9	-13.7	-12.8	20.6	-3.9	-9.8	-14.7	-13.7	-14.7	9.8
19	1.0	0.0	7.4	4.9	12.8	14.7	19.6	20.6	5.1	2.9	10.8	10.2	9.8	21.6
20	5.9	3.9	0.0	2.9	19.6	20.6	6.9	-4.9	3.9	9.8	1.0	8.8	14.7	4.9
21	2.9	4.9	0.0	9.8	13.7	28.9	0.0	29.4	6.4	7.8	9.8	28.9	37.3	29.4
22	1.0	1.0	2.9	0.0	5.4	8.8	5.9	-5.9	2.9	2.0	2.9	6.4	8.8	5.9
23	1.0	3.9	-2.0	4.9	10.8	12.3	14.7	30.4	7.8	6.9	7.4	13.7	14.7	17.2
24	0.0	2.9	0.0	2.0	2.9	17.7	15.7	18.1	1.0	8.8	3.9	4.9	16.2	14.7
25	0.0	0.0	1.0	0.0	7.8	5.9	8.8	2.9	2.0	0.0	2.9	6.9	6.9	5.9
26	-4.9	-2.0	0.0	-4.9	8.8	6.9	12.3	15.7	-2.0	-2.0	2.0	4.9	3.5	9.8
27	2.0	2.5	2.0	3.4	5.9	7.8	6.9	12.8	5.9	8.8	4.9	11.8	12.8	10.8
28	13.7	12.3	11.8	13.7	29.4	25.5	30.9	35.3	17.2	16.7	17.7	26.5	29.9	29.4
29	1.0	-2.0	0.0	1.0	3.9	2.0	2.9	4.9	1.0	0.0	0.0	3.9	2.0	2.9
30	11.8	9.8	15.7	25.5	25.5	25.5	35.3	39.2	16.7	16.7	32.4	27.5	25.0	40.7
Median*	2.0	2.2	1.8	3.2	6.8	8.3	7.8	10.3	4.4	5.0	4.4	9.5	10.1	9.8
Min*	0.0	0.0	0.0	0.0	0.0	0.0	0.0	2.9	0.0	0.0	0.0	0.0	0.0	0.0
Max*	13.7	12.3	15.7	25.5	29.4	28.9	35.3	39.2	21.6	20.6	32.4	32.4	37.3	40.7

* Absolute values

Table A4- 7: Mean minimum and maximum forces recorded for all therapists (F_x)

Therapist	End Feel (N)			Grade I Minimum (N)			Grade I Maximum (N)			Grade II Minimum (N)			Grade II Maximum (N)		
	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1	A1	A2	B1
1	-9.0	-19.6	-18.6	-2.0	-3.5	-2.9	-4.9	-7.5	-5.9	-2.4	-3.9	-3.3	-5.1	-7.5	-6.3
2	-4.7	-6.9	-3.9	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	-2.0	0.0
3	-11.8	-8.8	-19.6	0.0	1.0	0.0	-3.9	-2.0	-3.5	0.0	0.0	0.0	-4.9	-4.9	-5.9
4	-25.5	-23.5	-23.5	-3.1	-2.4	-3.5	-7.1	-7.1	-7.1	-1.0	-2.4	-2.6	-7.8	-8.6	-8.0
5	0.0	-7.8	-18.6	0.0	0.0	-1.0	3.9	3.9	-3.9	0.0	1.0	0.0	2.9	3.9	-2.9
6	-15.7	-7.1	-5.9	-1.0	-2.0	-2.9	-4.9	-4.9	-6.9	-2.0	-2.9	-2.9	-6.9	-5.9	-6.9
7	-16.7	-14.7	-16.7	-1.2	-2.7	-2.9	-4.3	-5.1	-5.9	-3.1	-2.9	-2.9	-5.5	-5.9	-6.9
8	25.5	19.6	-23.5	-1.0	0.0	-3.9	-3.9	0.0	-6.3	0.0	-2.0	-3.5	0.0	-3.9	-6.7
9	33.4	26.5	-7.8	0.0	-2.4	-2.9	0.0	-5.1	-5.9	0.0	-1.6	-2.9	-4.3	-5.5	-6.9
10	5.9	17.7	9.0	2.0	0.0	0.0	3.9	2.9	0.0	2.0	0.0	0.0	4.9	2.9	2.9
11	-9.8	-5.9	-16.7	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
12	12.6	19.6	-11.8	0.0	2.4	-2.0	2.9	4.3	-5.9	2.0	5.5	-2.9	4.3	7.5	-5.9
13	-16.5	-10.1	9.8	-6.4	-4.6	-2.0	-10.1	-7.3	-5.9	-4.6	-3.7	-2.0	-9.2	-7.3	-5.9
14	-21.6	-28.4	-19.6	-2.4	-5.7	-2.5	-5.9	-17.7	-4.9	-3.9	-3.9	-2.9	-7.8	-8.8	-6.7
15	-6.9	-7.8	-7.8	0.0	0.0	-3.9	0.0	0.0	-6.9	-1.6	0.0	-2.0	-4.3	0.0	-5.9
16	-15.7	-13.7	-19.6	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	-2.9	0.0	0.0
17	-14.7	3.9	-13.7	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	-4.9	-4.9	-4.9
18	-21.6	-24.5	-26.5	-1.0	1.0	-2.0	-3.9	-2.9	-4.9	0.0	-2.0	-4.9	-2.9	-4.9	-6.9
19	-9.8	-9.8	-22.6	-1.0	-1.0	0.0	-3.9	-3.9	-3.9	-1.0	-2.0	0.0	-3.9	-3.9	-3.9
20	-16.7	-25.5	-13.7	-1.0	-1.6	0.0	-3.9	-3.5	-2.9	-1.0	-2.0	0.0	-3.9	-3.9	-2.9
21	-41.2	-33.4	19.6	-1.0	-2.9	0.0	-3.9	-5.9	4.9	-2.9	-5.9	0.0	-5.9	-9.8	4.9
22	2.0	-6.9	3.9	-1.0	-2.9	-2.0	-3.9	-4.9	-4.9	-1.0	-2.9	-2.9	-2.9	-4.9	-5.9
23	-21.6	-30.4	-20.6	-4.9	-5.9	-3.9	-8.8	-9.8	-6.9	-6.9	-7.8	-7.4	-10.8	-12.8	-11.8
24	-23.5	-31.4	-15.7	-2.9	-7.8	-3.3	-5.9	-11.8	-5.7	-2.9	-7.8	-3.9	-6.9	-14.7	-7.8
25	-19.6	-13.7	-13.7	-3.9	-3.9	-2.9	-9.8	-9.8	-7.8	-4.9	-4.9	-2.4	-9.0	-8.8	-9.0
26	3.9	-13.7	-18.6	0.0	-2.2	-1.5	0.0	-4.7	-3.9	1.0	0.0	-2.0	-2.9	-8.8	-5.9
27	-14.7	-19.6	-14.7	-1.0	-1.0	-1.0	-3.9	-3.9	-2.9	0.0	0.0	-1.0	-3.9	-3.9	-2.9
28	-5.9	-2.9	-1.0	0.0	0.0	0.0	0.0	-2.9	2.9	0.0	0.0	2.0	-2.9	0.0	3.9
29	-9.8	-9.8	-4.9	-1.0	-1.0	0.0	-3.9	-4.9	-2.9	-2.0	-2.9	-1.0	-4.9	-5.9	-2.9
30	-29.4	-26.5	-20.6	-10.2	-7.8	-3.9	-12.2	-9.8	-6.9	-9.8	-8.8	-6.9	-13.7	-13.7	-9.8
Median*	15.2	14.2	16.2	1.0	1.8	2.0	3.9	4.5	4.9	1.0	2.0	2.0	4.6	5.2	5.9
Min*	0.0	2.9	1.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Max*	41.2	33.4	26.5	10.2	7.8	3.9	12.2	17.7	7.8	9.8	8.8	7.4	13.7	14.7	11.8

* Absolute values

Table A4- 8: Mean minimum and maximum forces recorded for all therapists (F_x continued...)

Therapist	Grade III Minimum (N)				Grade III Maximum (N)				Grade IV Minimum (N)			Grade IV Maximum (N)		
	A1	A2	B1	B2	A1	A2	B1	B2	A1	A2	B1	A1	A2	B1
1	-5.9	-3.9	-5.1	-10.8	-9.8	-7.8	-8.2	-13.7	-5.9	-3.1	-5.5	-10.6	-6.9	-9.0
2	0.0	-1.0	0.0	0.0	0.0	-3.9	4.9	0.0	0.0	-1.0	0.0	0.0	-3.9	0.0
3	-5.9	0.0	-3.3	0.0	-2.0	-5.9	-11.8	-8.2	-1.0	2.0	-12.8	-6.9	-2.0	-22.6
4	0.0	0.0	-2.9	-1.0	-17.7	-15.7	-23.5	-21.6	-7.8	-7.5	-14.1	-14.7	-13.7	-19.2
5	0.0	0.0	-5.9	-8.8	0.0	0.0	-8.8	-12.8	0.0	0.0	-8.6	0.0	-2.9	-12.2
6	-2.9	-2.0	0.0	1.0	-7.8	-5.9	-2.9	8.8	-3.5	0.0	2.9	-7.8	0.0	6.9
7	-4.9	-5.1	-3.9	-7.8	-7.8	-7.8	-6.9	-14.7	-5.9	-5.9	-6.3	-8.6	-9.8	-9.8
8	-3.9	0.0	-2.0	-6.9	5.4	17.7	-31.9	-21.1	0.0	-3.9	-12.3	2.9	-11.8	-18.6
9	-2.0	-5.9	-2.0	0.0	21.6	3.5	-4.9	7.8	4.9	1.0	0.0	22.6	16.7	7.8
10	8.3	6.4	9.8	10.3	22.6	17.2	20.1	22.6	23.5	21.6	19.6	37.3	35.3	30.4
11	0.0	-1.0	-2.7	-1.0	0.0	-4.9	-7.1	-3.9	0.0	0.0	-5.1	0.0	-3.9	-9.8
12	5.1	7.5	0.0	2.0	7.8	10.6	0.0	4.9	7.1	11.0	0.0	11.4	13.3	3.9
13	-7.3	-8.2	-4.9	0.0	-12.8	-12.8	-11.8	5.9	-5.5	-7.3	-6.9	-9.2	-11.0	-10.8
14	-13.7	-12.8	-5.9	-10.8	-26.5	-21.6	-16.7	-19.6	-18.1	-17.7	-9.3	-25.0	-23.5	-14.7
15	-2.7	1.0	-4.3	-6.7	-6.7	4.7	-9.8	-11.8	-3.9	0.0	-5.9	-6.9	2.0	-9.8
16	-4.9	-6.9	-6.9	-7.8	-10.8	-13.7	-15.7	-14.7	-13.7	-10.8	-7.8	-17.7	-15.7	-14.7
17	-1.0	-3.9	0.0	-4.9	-6.9	-9.8	-4.9	4.9	0.0	0.0	0.0	0.0	-4.9	9.8
18	-5.4	-10.8	-13.2	-13.7	-15.7	-22.6	-27.0	-35.8	-9.8	-18.6	-12.3	-17.7	-29.4	-28.4
19	-3.5	-4.9	-6.9	-4.5	-11.0	-13.7	-16.7	-9.6	-4.9	-2.9	-9.3	-6.9	-5.9	-15.7
20	-5.9	-6.9	-2.0	0.0	-13.7	-15.7	-11.4	-8.3	-7.8	-6.9	-2.0	-10.8	-10.8	-5.9
21	-11.3	-13.7	0.0	-2.0	-32.4	-42.2	-9.8	-11.8	-26.5	-22.6	-8.8	-59.8	-60.8	-17.7
22	2.9	-1.0	2.5	2.0	8.8	-6.4	5.9	11.8	7.8	2.0	11.8	12.8	5.9	16.7
23	-10.8	-12.8	-13.7	-15.7	-22.6	-24.0	-24.5	-31.4	-16.7	-19.6	-23.5	-24.5	-30.9	-35.3
24	0.0	-7.8	-2.5	-7.8	-10.8	-19.6	-19.6	-21.6	-2.9	-8.8	-7.4	-5.9	-16.7	-16.7
25	-4.9	-4.9	-2.9	-2.0	-12.8	-10.6	-12.8	-9.8	-6.7	-4.9	-4.9	-9.4	-9.8	-8.8
26	2.0	-1.0	-3.9	0.0	-9.8	-11.8	-13.7	-15.7	-3.9	0.0	-3.9	-7.8	-5.9	-7.8
27	-5.9	-5.9	-6.4	-8.8	-9.8	-9.8	-9.8	-14.7	-10.8	-8.8	-10.8	-14.7	-14.7	-13.7
28	0.0	0.0	2.9	2.0	3.4	3.4	4.9	4.9	0.0	1.0	2.0	2.9	3.9	4.9
29	-2.9	-3.9	-2.0	-3.9	-5.9	-7.8	-3.9	-6.9	-3.9	-4.9	-2.9	-5.9	-7.8	-5.9
30	-14.2	-10.3	-8.8	-18.1	-25.0	-20.6	-15.7	-25.0	-17.2	-11.8	-14.7	-25.5	-19.6	-18.6
Median*	4.4	4.9	3.1	4.2	9.8	10.6	10.6	11.8	5.7	4.9	7.1	9.3	10.3	11.5
Min*	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
Max*	14.2	13.7	13.7	18.1	32.4	42.2	31.9	35.8	26.5	22.6	23.5	59.8	60.8	35.3

* Absolute values

Table A4- 9: Frequency of oscillation of mobilisation force recorded for all therapists

Therapist	Grade I (Hz)			Grade II (Hz)			Grade III (Hz)				Grade IV (Hz)		
	A1	A2	B1	A1	A2	B1	A1	A2	B1	B2	A1	A2	B1
1	1.35	1.50	1.45	1.30	1.50	1.30	1.30	1.45	1.30	1.65	2.00	2.00	2.10
2	1.40	1.40	1.30	1.75	1.70	1.55	1.60	1.45	1.50	1.60	1.90	2.05	2.00
3	0.65	0.70	0.80	0.65	0.65	0.65	0.60	0.60	0.60	0.65	0.80	0.80	1.05
4	0.95	1.10	0.95	0.90	1.00	0.90	0.95	0.95	0.85	0.90	1.95	1.90	1.70
5	1.30	1.20	1.40	1.20	1.10	1.45	1.60	1.80	1.70	1.85	1.70	1.80	1.80
6	1.35	1.50	1.30	1.35	1.40	1.30	1.40	1.40	1.45	1.50	1.50	1.40	1.60
7	1.10	1.00	1.10	1.00	1.05	1.05	1.10	1.15	1.20	1.35	1.30	1.30	1.40
8	1.30	1.15	1.30	1.15	1.30	1.20	1.50	1.50	1.45	1.50	1.55	1.65	1.55
9	1.15	1.15	1.25	1.05	1.15	1.15	1.00	1.10	1.05	1.15	1.10	1.25	1.25
10	1.05	1.10	1.25	1.00	1.10	1.15	1.20	1.25	1.30	1.40	1.65	1.45	1.65
11	0.80	0.90	0.85	0.80	0.90	0.80	0.75	0.80	0.75	0.75	1.10	1.00	0.75
12	1.15	1.30	1.15	0.90	1.10	0.90	0.90	1.10	0.90	0.90	1.85	1.75	1.50
13	1.10	1.20	1.00	0.65	0.75	0.80	0.75	0.80	0.80	1.10	1.20	1.30	1.45
14	1.20	1.30	1.10	1.00	1.10	0.80	1.15	1.15	0.90	1.15	1.60	1.65	1.55
15	1.30	1.30	1.30	1.10	1.10	1.10	1.05	1.05	1.10	1.30	1.30	1.55	1.45
16	1.10	1.10	1.00	1.00	1.00	1.05	0.90	0.90	0.90	1.00	1.95	1.80	2.50
17	1.65	1.70	2.25	1.40	1.60	1.80	1.40	1.25	1.40	1.80	2.15	2.15	2.90
18	1.20	1.20	1.30	1.05	1.10	1.00	1.20	1.20	1.30	1.50	1.30	1.40	2.00
19	0.90	1.00	1.10	0.95	1.05	0.90	1.10	1.25	1.05	1.10	1.35	1.50	1.40
20	1.00	1.00	0.85	0.70	0.80	0.65	0.65	0.70	0.55	0.60	1.05	1.00	1.05
21	0.80	0.70	0.70	0.55	0.55	0.65	0.95	0.95	1.05	1.35	1.70	1.80	1.55
22	1.45	1.45	1.75	1.30	1.40	1.70	1.30	1.40	1.66	1.50	1.60	1.65	1.75
23	1.25	1.25	1.20	1.20	1.20	1.20	1.15	1.20	1.20	1.20	1.25	1.20	1.20
24	0.90	1.40	1.40	1.00	1.45	1.50	1.15	1.35	1.50	1.70	1.55	1.60	1.80
25	0.65	0.70	0.60	0.65	0.60	0.50	0.60	0.50	0.50	0.50	0.70	0.75	0.95
26	1.10	1.25	1.15	1.10	1.30	1.20	1.10	1.30	1.30	1.35	1.70	1.80	1.65
27	1.35	1.35	1.35	1.30	1.40	1.40	1.20	1.25	1.55	1.30	1.85	1.95	2.20
28	0.90	0.95	0.95	1.00	1.00	0.95	1.10	1.00	1.00	1.05	1.15	1.05	1.05
29	1.10	1.10	1.35	1.10	1.00	1.05	1.00	1.00	1.25	1.15	1.10	1.00	1.20
30	1.70	1.65	2.00	1.60	1.60	1.85	1.60	1.60	1.85	1.85	1.65	1.65	1.85
Mean	1.14	1.19	1.22	1.06	1.13	1.12	1.11	1.15	1.16	1.26	1.49	1.51	1.60
SD	0.26	0.26	0.35	0.28	0.30	0.35	0.28	0.30	0.35	0.36	0.37	0.38	0.46
Min	0.65	0.70	0.60	0.55	0.55	0.50	0.60	0.50	0.50	0.50	0.70	0.75	0.75
Max	1.70	1.70	2.25	1.75	1.70	1.85	1.60	1.80	1.85	1.85	2.15	2.15	2.90

Appendix 5: Study Two: The Healthy Subjects

Table A5- 1: Mean minimum and maximum forces recorded for healthy subjects (F_{total})

	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
Subject	**Maximum**	**Min**	**Max**	**Min**	**Max**	**Min**	**Max**	**Min**	**Max**
1	286	5	9	8	15	80	210	180	225
2	248	6	10	5	11	70	160	126	160
3	246	4	7	5	10	95	187	195	225
4	229	3	7	5	11	75	175	165	205
5	207	5	8	4	10	64	150	126	158
6	223	5	7	4	8	50	126	94	150
7	230	4	8	4	11	70	164	140	185
8	241	5.2	7.2	6	13.5	40	154	120	174
9	217	5	8	6	11	64	144	115	158
10	252	10	14	6	11	48	164	118	175
11	199	3	6	4.5	11.5	64	170	124	164
12	244	6	10	12	22	88	180	168	200
13	252	5	8	11.5	21.5	80	176	150	200
14	224	6	9	9	12	68	168	140	190
15	258	4	6	6	12	105	190	170	198
16	235	6	10	7	15	80	164	140	170
17	246	8	11	6	15	66	152	126	164
18	262	5	9	6	12	70	170	150	190
19	225	6	10	7	13	68	156	145	176
20	254	10	15	13	24	70	164	136	186
21	242	7	12	7	15	59	156	130	164
22	217	4	9	6	14	68	164	118	156
23	224	5	8	4	14	42	140	128	165
24	260	6	10	8	14	48	144	134	175
25	250	6	10	6	14	75	185	158	220
26	266	5	8	8	14	64	160	143	175
Median	243	5	9	6	13	68	164	138	175
Min	199	3	6	4	8	40	126	94	150
Max	286	10	15	13	24	105	210	195	225

Table A5- 2: Mean minimum and maximum forces recorded for healthy subjects (F,)

Subject	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	285	4	7	7	14	80	205	180	220
2	250	4	8	4	10	70	156	120	160
3	244	2	6	3	8	100	185	195	225
4	231	2	5	4	8	65	165	165	205
5	205	5	6	4	10	60	145	126	158
6	222	2	6	2	7	52	136	96	132
7	229	2	5	3	9	70	160	138	180
8	240	4	6	5	12	40	156	120	170
9	215	4	6	4	9	56	134	108	148
10	245	6	8	2	10	46	156	112	162
11	198	3	5	4	10	64	168	119	163
12	239	6	9	12	18	88	176	165	200
13	248	4	6	10	18	84	176	150	200
14	222	5	8	8	16	68	170	140	190
15	258	3	6	6	11	110	185	165	200
16	234	5	9	6	14	80	162	140	170
17	243	8	11	6	14	64	152	126	160
18	262	6	10	4	12	72	170	145	185
19	225	5	9	6	12	74	154	145	176
20	253	10	14	13	22	70	162	136	183
21	242	6	11	6	15	58	156	130	164
22	215	5	7	6	13	66	160	114	152
23	221	2	6	2	14	42	140	130	166
24	261	5	8	6	12	50	144	132	176
25	244	4	8	6	14	75	180	156	210
26	265	5	7	6	13	64	162	143	176
Median	241	4	7	6	12	67	161	137	176
Min	198	2	5	2	7	40	134	96	132
Max	285	10	14	13	22	110	205	195	225

Table A5- 3: Mean minimum and maximum forces recorded for healthy subjects (F_y)

Subject	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	10	1	4	3	6	-2	2	-2	2
2	12	2	5	0	3	9	23	13	20
3	10	0	2	0	4	-2	2	0	3
4	14	0	3	0	4	-3	-8	0	3
5	5	0	3	0	2	2	7	2	7
6	8	1	3	1	3	0	4	0	-4
7	10	0	2	0	4	-2	-6	-4	-10
8	-7	0	1	0	3	0	-4	0	-6
9	8	0	4	1	5	-4	-11	-3	-9
10	-36	6	10	1	4	-6	-23	-13	-26
11	6	0	0	0	4	0	-5	-3	-6
12	7	0	3	2	6	6	14	10	14
13	26	0	3	4	8	12	26	14	25
14	32	0	-4	-2	2	8.5	20	14	20
15	18	0	0	0	0	4	14	12	15
16	30	-1	-3	0	-2	8	18	12	18
17	7	0	0	0	-2	6	14	7	12
18	14	0	-2	0	-4	10	24	16	22
19	8	-2	2	-2	-4	2	12	6	8
20	11	3	5	3	10	11	23	20	30
21	7	0	0	-2	2	1	14	12	15
22	28	0	2	1	5	4	20	12	20
23	22	0	-4	0	3	4	14	10	14
24	14	0	-3	-3	-6	3	15	10	14
25	0	0	0	0	3	2	12	12	14
26	8	0	4	2	8	6	12	6	12
Median*	10	0	3	1	4	4	14	10	14
Min*	0	0	0	0	0	0	2	0	2
Max*	36	6	10	4	10	12	26	20	30

* absolute values

Table A5- 4: Mean minimum and maximum forces recorded for healthy subjects (F_x)

Subject	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	-16	0	-2	-1	-3	0	11	9	11
2	-2	0	0	0	0	-2	-4	-2	-4
3	-6	0	-2	-1	-3	2	6	2	5.5
4	-5	0	-2	0	-4	6	16	3	6
5	-24	-1	-3	-1	-3	-11	-17	-12	-18
6	5	-2	-5	-1	-4	0	6	5	10
7	-8	-1	-4	-1	-4	1	12	8	15
8	11	-2	-5	-2	-5	0	14	6	18
9	22	0	-3	-2	-3	17	48	33	51
10	55	0	2	-1	-3	17	57	35	59
11	9	-1	-3	-1	-3	3.5	10	7.5	15
12	0	-1	2	1	-2	0	-2	-0.8	-2
13	-3	-2	2	0	-2	0	-3	-1	-3
14	-3	0	1	0	0	0	-2	-1	-3
15	-2	0	1	0	1	0	-1	-1	-2
16	-2	0	0	0	0	0	-1	0	-2
17	-1	0	0	-1	1	0	-2	0	-2
18	-1	-1	1	0	1	0	-3	0	-2
19	0	0	1	0	1	0	-2	0	-2
20	0	0	0	0	0	0	-3	-1	-3
21	-1	0	0	0	0	0	-2	-1	-2
22	-3	0	0	0	0	0	-2	0	-2
23	-3	0	0	0	0	0	-2	0	-2
24	0	0	1	0	2	0	-2	-1	-2
25	0	0	0	0	0	0	-2	0	-2
26	0	0	0	0	0	0	-1	0	0
Median*	3	0	1	0	2	0	3	1	3
Min*	0	0	0	0	0	0	1	0	0
Max*	55	2	5	2	5	17	57	35	59

* absolute values

Table A5- 5: Frequency of oscillation of mobilisation force recorded for healthy subjects

Subject	Grade I (Hz)	Grade II (Hz)	Grade III (Hz)	Grade IV (Hz)
1	0.8	0.6	0.8	1.8
2	0.8	0.7	0.8	1.4
3	0.8	0.7	0.8	1.6
4	0.7	0.7	0.7	1.3
5	0.8	0.8	0.9	1.5
6	0.9	0.8	0.8	1.2
7	0.8	0.8	1.0	1.5
8	0.7	0.8	0.8	1.4
9	0.8	0.8	0.8	1.4
10	0.9	0.7	0.9	1.4
11	0.8	0.7	0.8	1.3
12	0.9	0.6	0.8	1.5
13	0.8	0.6	0.9	1.5
14	0.9	0.7	0.9	1.5
15	0.8	0.7	1.0	1.6
16	0.8	0.6	0.9	1.6
17	0.8	0.6	0.9	1.5
18	0.8	0.7	0.9	1.5
19	0.9	0.7	0.8	1.5
20	0.9	0.8	0.9	1.4
21	0.9	0.7	0.8	1.6
22	0.8	0.6	0.9	1.5
23	0.8	0.6	0.8	1.4
24	0.8	0.7	0.8	1.4
25	0.8	0.7	0.8	1.5
26	1.0	0.5	0.8	1.4
Mean	0.8	0.7	0.8	1.5
SD	0.1	0.1	0.1	0.1
Min	0.7	0.5	0.7	1.2
Max	1.0	0.8	1.0	1.8

Appendix 6

Reliability of Shrobers method of measurement

Table A6- 1: Repeated measurement of range of movement of lumbar spine

Subject	Measurement 1 (mm)			Measurement 2 (mm)		
	Flexion	Extension	Range	Flexion	Extension	Range
1	66	22	88	62	22	83
2	76	20	99	86	24	109
3	68	18	86	63	21	84
4	66	9	75	61	17	78
5	50	31	81	58	16	74
6	66	18	84	67	17	84
7	64	23	87	59	26	85
8	68	27	95	57	33	90
9	59	26	85	65	14	79
10	52	21	73	53	25	78
11	63	26	89	61	28	89
12	82	28	110	82	34	116
13	57	20	77	71	15	86
14	71	12	83	70	12	82
15	71	13	84	69	14	83
16	50	19	69	47	26	73
17	53	24	77	50	36	86
18	69	21	90	71	26	97
19	66	17	83	65	22	87
20	61	22	83	53	14	67
21	59	20	79	55	15	70
22	53	28	81	57	18	75
23	36	22	58	42	19	61
24	68	21	89	72	28	100
25	75	14	89	79	11	90
26	78	16	94	75	19	94
27	44	28	72	43	30	73
28	81	25	106	81	31	112
29	91	3	94	83	-2	81
30	46	25	71	54	20	74
Mean	64	21	84	64	21	84
SD	12	6	11	12	8	13
Min	36	3	58	42	0	60
Max	91	31	110	86	36	116

Appendix 7: Grade III Mobilisation performed on First Lumbar Vertebra

Table A7- 1: Characteristics of a Grade III mobilisation performed on L1 of five healthy subjects

	F_{total}		
Subject	Minimum (N)	Maximum (N)	Frequency (Hz)
16	90	170	1.0
19	108	166	0.9
21	95	190	1.0
23	90	200	0.9
24	76	170	0.9
Median	90	170	0.9
Min	76	166	0.9
Max	108	200	1.0

	F_z	
Subject	Minimum (N)	Maximum (N)
16	90	166
19	104	168
21	100	190
23	90	200
24	78	170
Median	90	170
Min	78	166
Max	104	200

	F_y	
Subject	Minimum (N)	Maximum (N)
16	6	15
19	10	15
21	-9	-20
23	0	11
24	-8	-14
Median*	8	15
Min*	0	11
Max*	10	20

	F_x	
Subject	Minimum (N)	Maximum (N)
16	0	0
19	0	2
21	0	2
23	0	0
24	0	3
Median*	0	2
Min*	0	0
Max*	0	3

* Absolute values

Appendix 8: Information Sheets and Consent Forms for Patients and Healthy Subjects

Information for Healthy Volunteers

Physiotherapy Management of Low Back Pain: The Effects of Spinal Mobilisation Techniques

A study is currently underway, undertaken by members of the Physiotherapy Service, to look at the effects of techniques used to treat back pain. The purpose of this trial is to measure the range of forces used by therapists to mobilise the spine in both normal subjects and patients.

If you are happy to participate, you will be required to lie prone on a couch which contains measuring equipment. The Physiotherapist will then perform a short series of mobilisations on your spine, which will not last longer than one minute. It is not anticipated that you will experience any discomfort as a result of these techniques, which will be felt as a firm pressure over your spine.

You do not have to take part in this study if you do not want to. If you decide not to take part you may withdraw at any time without having to give a reason. All proposals for research using human subjects are reviewed by an ethics committee before they can proceed. This proposal was reviewed by Camden and Islington Community Health Services Trust Ethics Committee.

Thank you for your help

Information for Patients

Physiotherapy Management of Low Back Pain: The Effects of Spinal Mobilisation Techniques

The staff in the Physiotherapy Department are conducting a research project to look at the effects of techniques used to help people with back pain. These techniques are used to treat people with many different types of spinal problems and although we know that they help back pain sufferers, it has not been possible to show how they affect the spine.

The procedures in which we are interested, involve the therapist applying a rhythmic pressure to one vertebra in the low back. This technique may have been used on your spine as part of your treatment programme, or during the assessment of your back at your first visit..

You will be asked to lie on your front on a bed which allows us to measure the pressure applied to your spine by the Physiotherapist. The Physiotherapist will then perform a short series of mobilisations on your back. This will be felt as a firm pressure and should involve no discomfort and take less than a minute to complete. The techniques applied during the measuring session do not form part of your current treatment programme

If you are happy to be included in the study, the procedure will be linked with one of your normal appointments. It will be carried out in the Middlesex Hospital Physiotherapy Department.

You do not have to take part in this study if you do not want to. If you decide not to take part you may withdraw at any time without having to give a reason. Your decision whether to take part or not will not affect your care and management in any way.

Thank you for your help

All proposals for research using human subjects are reviewed by an ethics committee before they can proceed. This proposal was reviewed by Camden & Islington Community Health Services NHS Trust Ethics Committee

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Contraindications for healthy volunteers

Do you have:

1.	Any problem with your back sufficient for you to seek medical (or other) attention	YES/NO
2.	Do you have leg pain	YES/NO
3.	Are you pregnant	YES/NO
4.	Do you have: Malignancy Osteoporosis Rheumatoid Arthritis Ankylosing Spondylitis Spondylolisthesis Spinal instability	YES/NO
5.	Do you have any other condition that could be affected by mobilisations ? YES/NO	
6.	Have you had spinal surgery ? YES/NO	
7.	Do you have any bladder of bowel problems consistent with spinal problems ?	YES/NO
8.	Are you in good general health ?	YES/NO
9.	Are you unwell (cold, flu etc.)	YES/NO

Signed..Date...

Name in Block Letters..

Age....................   Height......................   Weight................................

Confidential

Physiotherapy Management of Back Pain: The effects of spinal mobilisation techniques

Michele Harms MSc MCSP

Superintendent Research Physiotherapist

Camden & Islington Community Health Services NHS Trust

To be completed by the patient/volunteer

1. I have read the information about this study YES/NO

2. I have had an opportunity to ask questions and discuss this study YES/NO

3. I have received satisfactory answers to all my questions YES/NO

4. I have received sufficient information about this study YES/NO

6. I understand that I may refuse to be included in this study without giving a reason for withdrawing and without affecting my future medical care YES/NO

7. I agree to take part in this study YES/NO

Signed..Date..

Name in Block Letters...

Age.................... Height...................... Weight................................

Appendix 9: Study Three: The Patient Group

Table A9- 1: Mean minimum and maximum forces recorded for patient group (F_{total})

	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
Patient	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1*	258	7	11	5	11	78	168	136	164
2*	172	5	8	5	11	70	148	142	163
3*	220	4	8	2	15	52	116	92	119
4	179	6	10	4	14	50	136	122	150
5	135	4	8	4	10	60	130	102	125
6	133	8	12	11	15	68	120	120	148
7	228	8.4	11.6	9	15	105	160	164	184
8	196	5	8	4	9	60	120	116	138
9	202	11	14	11	22	110	200	190	220
10	250	10	13	9	17	66	150	152	178
11	240	12	16	11	17	120	200	195	225
12	225	7	9	8	14	70	160	175	205
13	260	12	18	13	25	115	180	170	210
14	210	5	8	8	15	104	144	132	155
15	230	15	18	15	27	92	180	185	220
16	235	13	14	11.5	19	110	185	176	212
Median	223	8	11	9	15	74	155	147	171
Min	133	4	8	2	9	50	116	92	119
Max	260	15	18	15	27	120	200	195	225

* Patients with back pain

Table A9- 2: Mean minimum and maximum forces recorded for patient group (F_z)

	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
Patient	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	256	4	6	3	8	76	164	136	164
2	172	4	7	4	13	70	148	142	162
3	216	4	8	0	16	48	104	86	110
4	179	6	9	1	10	52	134	122	150
5	132	4	7	4	10	60	124	100	124
6	128	2	4	0	5	64	117	117	138
7	228	7	9	7	14	108	160	168	187
8	194	2	4	0	8	60	119	110	135
9	198	10	13	10	21	110	195	186	214
10	248	9	11	8	14	68	152	153	178
11	240	10	14	9	16	120	195	195	220
12	220	7	9	6	13	72	160	175	200
13	260	12	15	10	21	110	180	165	205
14	218	5	7	6	14	104	140	130	155
15	228	14	16	14	26	90	178	185	215
16	235	12	14	10	17	110	185	173	206
Median	219	7	9	6	14	74	156	148	171
Min	128	2	4	0	5	48	104	86	110
Max	260	14	16	14	26	120	195	195	224

Table A9- 3: Mean minimum and maximum forces recorded for patient group (F_y)

Patient	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	-12	-4	-6	-4	-8	-4	2	0	4
2	5	0	-4	-2	-4	-4	2	2	5
3	18	0	4	1	6	12	48	31	46
4	-8	0	-2	4	10	0	10	0	8
5	30	0	0	0	2	4	26	18	26
6	20	7	10	8	10	17	37	19	21
7	18	0	2	2	4	4	7	16	18
8	10	2	6	3	6	10	30	20	31
9	10	0	-2	0	3	0	5	-2	2
10	0	0	0	0	2	-2	2	0	3
11	26	0	2	1	4	6	12	14	16
12	27	0	1	2	3	5	7	8	10
13	25	4	6	6	11	22	32	28	40
14	12	0	2	3	5	6	10	6	10
15	14	2	4	4	6	14	25	16	20
16	17	3	4	4	6	15	19	14	22
Median*	16	0	3	3	5	6	11	14	17
Min*	0	0	0	0	2	0	2	0	2
Max*	30	7	10	8	11	22	48	31	46

* absolute values

Table A9- 4: Mean minimum and maximum forces recorded for patient group (F_x)

Patient	End Feel (N)	Grade I (N)		Grade II (N)		Grade III (N)		Grade IV (N)	
	Maximum	Min	Max	Min	Max	Min	Max	Min	Max
1	0	0	1	0	1	-1	1	0	0
2	0	0	1	0	1	0	1	0	0
3	-2	-1	1	-1	1	-1	-6	-4	-6
4	0	0	0	0	0	-2	1	-1	1
5	-3	0	0	0	0	0	-3	-1	-3
6	-2	0	-2	0	-2	0	-3	-2	-3
7	-2	-4	-6	-4	-6	-3	-6	-3	-6
8	0	0	0	0	0	10	30	20	31
9	34	-2	-4	-3	-5	22	45	28	33
10	15	-4	-5	-4	-6	4	13	4	6
11	-12	0	-3	-3	-5	-8	-12	-25	-28
12	-30	-1	-2	-2	-4	-11	-13	-13	-15
13	3	-5	-7	-5	-7	0	-6	2	-2
14	3	0	-2	-2	-4	0	5	4	8
15	-17	-1	-3	-4	-6	-15	-27	-16	-20
16	2	-3	-5	-4	-5	-6	-8	-4	-6
Median*	3	0	2	2	4	3	6	4	6
Min*	0	0	0	0	0	0	1	0	0
Max*	34	5	7	5	7	22	45	28	33

* absolute values

Table A9- 5: Frequency of oscillation of mobilisation force recorded for patient group

Patient	Grade I (Hz)	Grade II (Hz)	Grade III (Hz)	Grade IV (Hz)
1	0.9	0.6	1.0	1.7
2	0.8	0.7	0.8	1.7
3	0.9	0.6	0.9	1.5
4	0.9	0.7	0.9	1.6
5	0.9	0.6	0.8	1.5
6	1.1	0.6	0.8	1.5
7	1.0	0.7	0.9	1.9
8	0.8	0.6	0.9	1.7
9	0.9	0.6	1.0	1.8
10	0.9	0.7	0.9	1.9
11	0.9	0.6	1.0	1.5
12	1.0	0.7	0.9	1.7
13	1.0	0.6	1.0	1.4
14	1.0	0.7	1.0	1.4
15	1.1	0.6	0.9	1.6
16	0.9	0.6	0.9	1.5
Mean	0.9	0.6	0.9	1.6
SD	0.09	0.05	0.07	0.16
Min	0.8	0.6	0.8	1.4
Max	1.1	0.7	1.0	1.9

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