An Ergonomic Appraisal of the Mechanical Loads Imposed on the Human Spinal Column During Impact Landing.

by

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Ph.D. Thesis.

Abstract

The aims of the thesis are to quantify the loads imposed on the spine by plyometric exercises entailing impact landings (bounding and drop jumping), training procedures used by athletes seeking to develop muscular power. Preventative methods prescribed to alleviate spinal loading and a warm-up routine were also assessed by recording small changes in stature (shrinkage). Spinal shrinkage for a drop jumping regimen was compared against biomechanical and electromyographical (EMG) data.

A five-link segment, dynamic, two-dimensional, biomechanical model estimated L5/S1 normal compression and shear forces during drop jump landing. Data from cine-film (100 Hz) and force platform were used to drive the model. Electromyography determined the activity of three trunk muscles - erector spinae, external obliques and rectus abdominis.

The fully automated, computer-linked stadiometer proved sensitive to spinal loading during bounding and drop jumping activities. Short duration (6 min) bounding and drop jumping regimens caused rapid stature loss, 1.85 and 1.74 mm respectively. Pre-exercise gravity inversion unloaded the spine and increased stature (2.7 mm). This was quickly lost during a subsequent bounding regimen (3.5 mm), exceeding the shrinkage following standing (1.7 mm, $p<0.05$). Post-exercise spinal unloading quickly restored stature lost during exercise. A 10 min warm-up routine caused shrinkage (0.89 mm), but this was not different from standing ($p>0.05$).

A 0.75 m drop jump regimen caused greater shrinkage than the 0.3 m regimen, 1.82 mm and 1.42 mm respectively. During the first land, the higher drop height resulted in a 29% increase in the cumulative spinal compressive load ($p<0.01$). Forefoot and heel impact produced large peak forces with short rise-times, a 78% greater mean peak occurring on landing from 0.75 m when compared with 0.3 m ($p<0.01$). A second, but more gradual rise in compressive force occurred as a result of trunk flexion, mean peak forces of 4927 and 5258 N being recorded for the 0.75 and 0.3 m drop lands, respectively ($p<0.05$). Erector spinae muscle activity predominated and normalized peak values were higher when 0.75 m drop lands were compared with 0.3 m drop lands ($p<0.01$).

During drop jumping, athletes must seek to cushion impact forces on landing and control trunk flexion. Increasing drop height increases spinal loading and training programmes should be structured to attenuate the resulting load elevation; alternatively, other training techniques should be sought. Effects from unloading the spine prior to exercise are short-lived and would be of greater benefit when interspersed throughout the training period.

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1. INTRODUCTION

. INTRODUCTION

ack pain is a problem of epidemic proportions which represents one f the major diseases of the musculoskeletal system. It is stimated that 50 to 80% of the population will at some time suffer rom this complaint (White and Gordon, 1982). The consequences of he disease to industry have been substantial, responsible for an stimated 12% of all the reported incapacities in 1989 (Tye and rown, 1990). It has been estimated that it is presently costing ndustry some 1,018 million pounds each year and the Health Service ome 156 million pounds per year (Tye and Brown, 1990). Statistics uggest that the disease is still on the increase, a 40% rise being eported in the number of man-days lost since 1983 (Tye and Brown, 990). Treatment of the disease is made increasingly more omplicated due to its multifactorial aetiology and the problems of redicting potential sufferers.

lthough, low back pain (LBP) has often been referred to as idiopathic', as in many instances its cause is unknown or ndeterminate, sufficient evidence exists to associate its aetiology ith a number of mechanical parameters. Andersson (1981), in a eview of epidemiological data collected over a period of 30 years, rew clear links between LBP symptoms causing work absenteeism and ix vocational factors. The common characteristic inherent to each as the mechanical load placed on the spine. Based on this riterion, many high risk occupations have been identified and rgonomic steps taken with the aim of prevention.

he extent to which sports participation may contribute towards back njuries is less well documented. The limited work does suggest hat LBP is a major complaint of sportsmen, is common to a number of ports and causes a large number of performers to withdraw from urther participation (Alexander, 1985). Brigham and Schafer (1988) eported that an estimated 5 to 10% of all athletic injuries are umbar spine related.

pidemiological surveys of the pathogenesis of sport-related njuries have often classified risk factors into two categories:- xtrinsic and intrinsic. Extrinsic factors are related to the type

of sporting activity, the manner in which the sport is performed and the environmental conditions. Intrinsic factors are associated with physiological and psychological characteristics of the individual. The majority of LBP experienced by athletes is felt to stem from the former and to be directly related to repetitive or exceedingly forceful trauma (Brigham and Schafer, 1988). The magnitude, direction and point of force application are the most important predeterminates of the type of injury sustained by the lumbar spine (Alexander, 1985). Landing on the feet at high velocity has been closely linked with lumbar spinal injuries (Williams, 1980).

In recent years, the development of new training techniques has been felt to contribute significantly to the advancement in athletic performance (Clutch et al., 1983). The possible damaging effects of prolonged or forceful exercise are often neglected. Therefore, it is important that any innovations in the development of new training protocols be fully assessed in terms of their potential to induce injury.

A relatively new training techniques, known as drop jumping or depth jumping, was first reported by Verhoshanski (1967). Drop jumping exercises, which involve subjects performing a vertical jump on landing from a raised platform, were performed by Russian athletes to improve their muscular power. These, along with other forms of exercises such as single response jumps (repeated jumps with pauses), bounding sequences (a number of consecutive jumps over flat ground) and box drills (a series of bounding and drop jumps), all seek to improve muscle function through a rapid muscle stretch-shortening cycle. Collectively, these activities have been termed 'Plyometric' exercises.

The implementation of such drills in the training of anaerobic muscular power has grown considerably. Studies have investigated the possible benefits to performance (Scoles, 1978; Blattner and Noble 1979; Clutch et al., 1983), for which there is conflicting evidence. Few have attempted to examine their potential damaging effects and none has sought to question the stresses placed on the

back. If such a question is to be addressed then it is necessary to provide estimates of the mechanical forces and loads the spine must contend with.

In occupational settings, ergonomists have provided the basic approach towards identifying potentially hazardous working conditions. An ergonomics approach seeks to study the worker in his environment with a view to improving safety, comfort and efficiency. In endeavouring to achieve these objectives, the ergonomist must first recognise the working capacity of the employee and the physical demands placed upon him. In terms of back related injuries, the latter has received the most attention, with emphasis placed upon measuring the loads on the spine *in-vivo* . Consequently, attention has focused around the development of effective and reliable measuring procedures.

Biomechanical modelling is at the forefront of these techniques, utilising the laws of physics and principles of engineering to describe motion and the forces acting on the human body. It provides an alternative technique to physical testing which could be destructive or potentially injurious to an individual (Miller, 1979).

Nachemson (1960) pioneered the use of intra-discal pressure measurements as a means of estimating *in-vivo* loads in the intervertebral discs. A pressure sensitive needle inserted into the disc recorded changes in pressure due to alterations in posture. Ethical and technical problems have limited the application of this technique, though Nachemson (1981) has shown with certainty that specific postures are more taxing for the spine than others. It remains inapplicable to assessment of spinal loading *in-vivo* under normal dynamic conditions.

Intra-abdominal pressure and measured muscle activity (electromyography) have been implemented as indirect means of estimating spinal loading. Whether either measure can be used as a reliable spinal load indicator is questionable. A solution was proposed by Fitzgerald (1972): the measurement of changes in height of the vertebral column. This phenomenon is associated with the

loss of disc height under static and dynamic loading (Kazarian, 1972; 1975). But the method of measurement now used was not developed until recently (Eklund, 1986; Corlett et al.; 1978). The technique is still in its infancy and requires considerable work to establish it as a reliable and effective procedure, not only to study loading activities, but also as a possible screening procedure for back pain sufferers.

2. AIMS AND OBJECTIVES

2. AIMS AND OBJECTIVES

This thesis adopts an ergonomic approach to estimate the loads placed *in-vivo* on the spine during plyometric activities. It will aim to:-

i) Develop and assess the reliability of the technique used to measure small changes in stature as an index of spinal loading

ii) Assess the spine's responses to certain plyometric activities and estimate the musculoskeletal forces generated about the lower back during landing from drop jumping.

Fulfilment of these aims will enhance an understanding of the potential for back injury during plyometric activities. As a result, constructive preventative guidelines or precautions which may be beneficial can be proposed.

These aims will be fulfilled by the following objectives :-

1. Further develop and update a stadiometer for measuring small changes in stature in order to improve the accuracy and reliability of the technique.

2. Apply this procedure to monitor spinal loading and unloading during bouts of physical activity incorporating plyometric routines.

3. Evaluate preventative procedures to minimise the likelihood of back injury by comparing rates of stature alteration.

4. Develop a biomechanical model to estimate the musculoskeletal forces about the lower back during landing from drop jumps.

5. In conjunction with point 4, undertake an electromyographical (EMG) analysis of trunk muscles and spinal shrinkage during drop jump landing in order to draw comparisons between techniques.

6. Based upon points 4 and 5, make recommendations to minimise the possibility of incurring back injury during the impact landing phase of drop jumping.

Hence, the project is concerned with validating the measurement of changes in stature as a reliable method to estimate mechanical loads on the spine. Attempts will be made to calculate the mechanical forces on the spine from which potentially hazardous activities can be identified. By doing so, the work will appraise ergonomically an aspect of plyometric training.

3. REVIEW OF LITERATURE

3 REVIEW OF LITERATURE

3.1 INTRODUCTION

The literature review will first attempt to identify the extent to which LBP afflicts members of the population. Reference will be made to the epidemiology of LBP and risk factors associated with the symptoms. This is not intended to be a comprehensive review; nevertheless, it will allow incidence rates to be compared with those resulting from sports participation. Following a review of sport-related back pain, a section will be devoted to identifying the possible mechanical causes and sites of back injury.

The project will attempt to determine the spinal loads imposed during plyometric activities, particularly during their impact landing phase. In order to undertake an ergonomic appraisal, it is important that a discussion is given to the principles underlying plyometrics and how they are implemented as a training stimulus. The review will conclude by investigating the effects of mechanical loads on the spine and examine the *in-vivo* measurement techniques used to study these loads.

3.2 LOW BACK PAIN

3.2.1 Incidence and Prevalence

Epidemiological documentation suggests that disorders of the low back are a problem of considerable magnitude afflicting a large proportion of the population at some time. It has been estimated that 50% to 80% of adults in Western countries will at some time experience an incidence of LBP, approximately half of those going on to encounter further episodes, and about a quarter of this total going on to consult their doctor (White & Gordon, 1982).

Questionnaires, insurance data, personnel interviews and attendances at general practitioners have all been techniques employed to establish the prevalence of LBP within the general population. Biering-Sørensen (1983) carried out an extensive study of 82% (489 males and 479 females) of all inhabitants aged 30, 40, 50 and 60 years old from a municipal suburb of Copenhagen, Denmark. Questionnaires and objective measurements were utilised to examine the prevalance of lower back disorders. Men showed a constant life time prevalence rate of between 68% and 70% across the age groups studied, whereas the value for women increased from 62% to 81% with increasing age. A follow-up questionnaire was administered a year later and completed by 99% of those initially studied. During the intervening period, 11% of the 30 year old sample population suffered LBP symptoms for the first time. This value decreased with age.

Valkenburg and Hanen (1982) also utilised questionnaire methods but in addition implemented standardised physical examinations. A study population of 6584 men and women (3091 males and 3493 females), aged 20 years and above, and from two districts (one urban and one rural district, Zoertermeer) showed LBP incidence rates of 51% and 57%, for males and females respectively. Twenty two per cent of males and 30% of females were found to have LBP at the time of examination.

Swedish National Health Insurance data, along with personal interviews, formed the basis upon which Svensson and Andersson (1982) conducted their study. A random sample of 940 men (40 to 47 years old) were contacted and personally interviewed, for which a LBP incidence and prevalence rate of 61% and 31%, respectively, was found. The study reported that 27% of participants who had been interviewed and not

reported ever having LBP were in fact found to have been absent due to the illness. This highlights one of the difficulties of trying to obtain reliable statistical data from such sources.

Data gathering from general practitioners has been an alternative method by which inferences have been gained as to the scale of back pain within a community. In Great Britain approximately 2 million adults per year visit their general practitioners with back pain and a total of about 3.4 million consultations are made per year because of back pain (Wood and Badley, 1980). Frymoyer et al. (1980) gathered information from the records of 3,920 patients (2,068 females, 1,852 males between 18 and 55 years old) entering a university-sponsored family practice in Vermont (U.S.A.) between 1975 and 1978. During this period, 11% of males and 9.5% of females visiting the practice had LBP symptoms. In 1958, Ward et al. (1958) reported on the number of LBP consultations at a general practice in the North Midlands (U.K.) with an estimated total population of 45,000. Low-back pain per 1,000 patients was 22.8 for males and 15.3 for females.

Whether attendances at general practitioners can provide a reliable estimate of back pain sufferers is questionable for these must be governed by the social and economic make up, and general ethos within the community in which the practice is based. Burton (1981) reported that one out of three patients with back pain and treated by a registered osteopath was not seen by a doctor.

It would seem that most of the reported epidemiological data is derived from occupationally related data. Some stem from national data which gives a discrete picture of prevalence in the population, including point prevalence (frequency of reported disorders at a specific period in time). However, none would appear to distinguish between occupationally related LBP and LBP from other causes.

The size of the problem, the demands placed upon the health services and the economic burden created by the LBP syndrome have led many researchers to identify possible risk factors.

3.2.2 Risk Factors

The epidemiology of LBP in industry has been well documented and many of the high risks associated with certain occupations are known. Industries identified as having a high work-handicap from back pain include coal and steel mining, dock work, nursing and driving (Andersson, 1981; Daniel et al., 1980; Blow and Jackson, 1971; Videman et al., 1984; Troup, 1978). Andersson (1981), in a review of the epidemiological aspects of LBP in industry collated over a period of 30 years, noted six vocational factors associated with low-back symptoms: physically heavy work, static work postures, frequent bending and twisting, lifting and forceful movements, repetitive work and vibrations. An increase in the load on the spine was suggested to be the common component linking each of these activities.

The extent to which heavy physical work is related to LBP symptoms is uncertain. Dales et al. (1986) stated that the differences in incidence rates between 'heavy' and 'light' occupations are only marginal, whereas, Troup and Edwards (1985) suggested that it is generally true that in most countries the heavier the work the more likely it is that back problems will prevail. Rowe (1969), in his ten year study of workers at a Kodak plant in New York, classified two groups of workers as heavy handlers and sedentary. He noted that 47% of the former group and 35% of the latter made visits to the medical department on the grounds of LBP complaints.

Svensson and Andersson (1983) identified a strong correlation between heavy physical work and the occurrence of LBP. This was not substantiated by Valkenburg and Haanen (1982) who found no trend with respect to prevalence of LBP and increasing job physical activity when jobs were classified as light, moderate and heavy. Unfortunately, they provide no description of the methods they used to distinguish between categories. As with many of these studies, information is often derived direct from the employee. This may produce misleading and often varied responses depending on the person's perception of the physical work. Task analysis provides a procedure by which to overcome some of these problems, quantifying spinal loading and facilitating comparisons between different working groups.

Kelsey (1975) studied the relationship between occupations and a specific LBP disease, acute prolapsed intervertebral disc. An increased risk of a prolapsed disc was reported for persons who had been employed in a sedentary occupation for several years. It was suggested (Kelsey and White, 1980) that this association may explain why some studies have failed to find differences in LBP incidence rates between differing occupations, as they have tended to consider the epidemiology of LBP in general rather than specific terms.

Incidences of disc protrusions amongst miners led Dales et al. (1986) to conclude that strength and physical fitness may have a protective role against back injury. They reported that a significantly lower proportion of miners attended a hospital with symptoms relating to disc protrusion than men from other occupations. It was found that this difference was more pronounced when consideration was given solely to those men admitted with the diagnosis, or those requiring discectomy. They proposed that the heavy manual work associated with mining may in some way protect the annulus fibrosus of the intervertebral disc from serious failure and posterior protrusion, or conversely, light work may cause a risk of disruption of the annulus.

Frymoyer et al. (1980), in their retrospective examination of patients' general practice records, assessed several occupational risk factors and identified lifting as the predominant factor responsible for 23% of the LBP symptoms. Truck driving, carrying, pulling, pushing, twisting and non-driving vibrational exposure were all found to significantly correlate with LBP. Svensson and Andersson (1983) also reported a significant correlation between occupations with high involvement in lifting and LBP. Also correlated with LBP were monotonous work, sitting, and work involving more walking and standing. In a study by Biering-Sørensen (1983), 37% and 24% of male and female LBP subjects, respectively, stated that heavy lifting was the causative factor responsible for their symptoms. Eighteen percent of men and 13% of women placed the cause on twisting movements.

More recent epidemiological findings (Riihimäki et al., 1989; Videman et al., 1990) would seem to support the view that heavy and light physical work carry similar risks of incurring back injury. Riihimäki et al. (1989) found that carpenters (dynamic physical work) had no excess risk

of low-back pain symptoms when compared with office employees (sedentary work). Videman et al. (1990) proposed a 'J-shaped' phenomenon where fewer pathological changes to the lumbar spine occur when physical loading falls between the extremes of heavy and sedentary work.

It is clear that many vocational factors may be regarded as high-risk in terms of their potential to cause discomfort to the lower back; however, the extent to which each contributes to the problem is uncertain. This may be because a majority of LBP cases share a gradual onset and that many occupations, either sedentary or heavy, encompass more than one of the identified vocational risk factors. A similar confusing picture emerges when one considers individual risk factors such as age, sex, anthropometric measures, strength and fitness.

From those studies which have quoted incidence and prevalence rates for LBP (section 3.2.1), it would appear that the men are as likely to contract the disease as women. Andersson (1981) stated that sex factors seem to be without importance when considering the likelihood of LBP symptoms. However, Krämer (1981) reported that the lumbar syndrome, described as symptoms and signs originating from degenerative changes in the lumbar discs, is more frequent in men than in women. The reason proposed was that men are involved in more activities of a stressful nature. He also believed an unknown 'species specific' factor may be at fault, such as a narrower lumbar canal in men.

In terms of age, Krämer (1981) stated that lumbar symptoms generally develop between 25 and 30 years old and reach a maximum at the age of 40 years in men and 50 years in women. Andersson (1981) reported that between the ages of 35 and 55 years the frequency of LBP symptoms reaches a maximum, whereas with increasing age, sickness absence and duration of symptoms increase. Biering-Sørensen's (1983) study of 30, 40, 50 and 60 year olds did show an increase in the lifetime prevalence of LBP with age for women (62% increasing to 81%), but for men the value remained almost constant (68% to 70%) for the age groups studied. Acute LBP is felt to be most common around the age of 25 years, but in 90% of cases it has often subsided within 2 months (White and Gordon, 1982). Witt et al. (1984), in a study of roentgenograms from the lumbar spine of 238 patients (above 20 years) with LBP, as compared with those from

66 patients without LBP, found in all subjects an increase in the incidence of spondylosis and disc degeneration with increasing age.

There exists no strong correlation between height, weight and body build, and LBP (Andersson, 1981). However, some studies have revealed a higher than average incidence of LBP in taller and more obese people. Heliövaara (1987) studied 332 men and women who had been discharged from hospital with a diagnosis of herniated lumbar intervertebral disc and 1205 matched controls. Men with a height of 180 cm or more and women with a height of 170 cm or more showed a relative risk of 2.3 (95% confidence limits, 1.4 - 3.9) and 3.7 (95% confidence limits, 1.6 - 8.6), respectively, when compared with those who were 10 cm shorter. Body mass index (weight/height) was shown to be an independent risk factor for the likelihood of a herniated lumbar disc in men but not women. For both sexes, skinfold thickness at the triceps site had no predictive significance.

The size of the vertebral canal has been implicated in some cases of LBP resulting from neurogenic claudication and disc symptoms (Porter et al., 1978; 1980). A study, utilising ultrasound to measure the spine of 204 miners showed that the men who had the longest histories, taken the longest time off work, and had left the coalface or industry because of back pain, had significantly narrower canals than their co-workers (MacDonald et al., 1984). Over a three year period it was shown that 10% of 50 year old miners with the narrowest canals were responsible for 32.5% of sickness absence due to back pain.

Strength has been shown to be a risk factor for both spells and severity rates of back pain absence (Chaffin and Park, 1973). It has principally been assessed in terms of isometric strength relative to job demands. Chaffin and Park (1973) showed that workers have approximately a three times greater risk of back injury if their isometric strength capability is less than the maximum lifting strength required by the job. They defined a maximum lift strength ratio (LSR) of a job as the maximum load to be lifted divided by the predicted maximum lifting strength of a strong male worker (the strongest 2.5% of men) in the same lifting position. It was found that LSR values of 0.2 or more increased incidences of LBP. The difficulties of relating isometric strength

measures to specific job strength requirements make assessment of possible correlations difficult.

Further support for the beneficial effects of back strength were provided by Karvonen et al. (1980) and Biering-Sørensen (1984). Both found an inverse relationship between back muscle strength and LBP symptoms.

There is some evidence to suggest that physical fitness and muscular conditioning have a protective role against back injuries. Cady et al. (1979) evaluated the physical fitness of 1652 firefighters in California using measures of endurance capacity, isometric strength tests, spinal mobility, diastolic blood pressure during exercise at a heart rate of 160 beats/min, and heart rate recovery after exercise. The subject population was divided into three groups:- least fit, medium fit, and most fit - and the respective back injury rates were 7.1%, 3.2% and 0.8%. It was shown, however, that the most fit group suffered the most severe injuries, suggesting that this group may well have had a higher tolerance for minor injuries.

Work on cadaveric lumbar spines by Porter et al. (1989) would seem to support the view that physical activity could strengthen the spinal structures. Mechanical tests were conducted on 9 lumbar spines and the results compared with the men's occupational and recreational histories. For eight of the subjects aged 18 and over, the compressive strength of the spines tended to increase with an increase in estimated level of physical activity in life.

The potential benefits gained by leisure and sports participation was not identified by Videman et al. (1984, 1990). They found no relationship between these activities and reduced back pain symptoms.

Although a lack of strength and physical activity may be regarded as LBP risk factors, Kelsey (1975) found that participation in some sports (baseball, golf and bowling) may increase the risk of developing prolapsed lumbar discs. Thus there is a need to establish the prominence of back problems associated with sports participation, the sports most commonly linked with the injury, and the underlying causes

of these injuries, so that steps may be taken to reduce the likelihood of incurring injury.

3.2.3 Sport Injuries

The extent to which sport and exercise contribute towards the scale of back problems is far from well documented. The limited work suggests that low back problems are a common ailment amongst athletes and that certain activities are more likely to induce injury. A retrospective analysis of 3,920 patients (2,068 females, 1,852 males) attending a family practice between 1975 and 1978 showed that 4.1% of males and 3.7% of females with LBP attributed the cause to sport (Frymoyer et al., 1980). These values were similar to those attributing LBP to an automobile accident (4.1% and 5.3% for males and females, respectively).

Williams (1980) reported that each year in the United Kingdom there are some two million sport-related injuries of which 15% are spinal. An estimated 5 to 10% of those athletic injuries are believed to be purely lumbar spinal in origin (Brigham and Schafer, 1988).

Rowell and Rees-Jones (1988) directly compared the sports injuries at an Accident and Emergency Department (A & E) and a Sports Injury Clinic (SIC), in the same geographical area and over the same 12 month period. A total of 2,490 and 340 injuries were treated at the A & E department and the SIC, respectively. Spinal injuries accounted for 2.9% at the A & E department and 13.5% at the SIC, the difference being significant. This was suggested to stem from the nature of the majority of sport-related back injuries, which are chronic disc or muscular injuries and not acute in nature. Whether it is reasonable to make a direct comparison between these two sets of data is questionable given the time lapse between injury and appearance, and the severity of injury likely to be reported at each clinic.

Cannon and James (1984) undertook a retrospective study of the number of people attending a Middlesex athletic clinic due to back complaints. Over a period of four years, 197 patients, 6.8% of the total which attended, were seen with this complaint, of which 83% were male. The age range of these athletes was between 13 and 53 years, of which 65% were below the age of 30 years. The majority of patients (38%) were

diagnosed to have mechanical back pain which was defined as pain that was episodic and cyclic, radiating to the buttocks and upper thigh, and often associated with morning stiffness and pain. Acute incidences could be related to the mode of onset in 85 cases (43%), but the majority of cases gave a history of chronic or non-specific mode of onset. A total of 72% of the patients associated the pain with six popular sports:- soccer (18%), rugby (16%), track and field athletics (15%), squash (8%), hockey (8%) and cricket (7%). It is likely that these figures are more a reflection of popular sports within this community rather than of the risk inherent in the respective sports.

Smith (1977) reported that the Canadian medical staff at the 1976 Olympic Games found LBP to be a common problem amongst their highly trained athletes. It was also stated that the problem usually developed during training as a result of a neglect of certain anatomical areas. The abdominal region, particularly the abdominal muscles, was highlighted as one of these principal areas.

Billings et al. (1977) studied 87 men and 13 women who had been referred to a sports clinic with LBP complaints directly related to a sports activity. Fifty-two percent of the cases reported their injury took place while competing, while 37% reported the injury to be a direct result of training, weight training being the most frequent cause. At the time of injury the most common body action was rotation and combined rotation-flexion movements of the spine. Diagnosis following a full assessment suggested that 39% of the cases were due to intervertebral disc problems and 18% due to spondylolysis (defect or fracture of the pars interarticularis of the neural arch of a vertebra). The LBP had had a significant impact on the patients' lives: 65% had lost time from work/study (mean 11 days) and 39% eventually found that they had to partially or completely give up their sports activities.

Williams (1980) suggested that it is possible to associate injuries at different spinal levels with a number of sports. Gliding, parachuting and mountaineering, where landing on the feet at high velocity is a common feature of the sport, were implicated with injuries to the lumbar spine. Steinberg (1988), in an assessment of injuries to Dutch sport parachutists reported that 82% of all accidents among novices occurred during landing. Although, 11.6% of all the injuries were accountable to

the pelvis and spine, the authors did not state the exact nature of these injuries. Similar findings were obtained by Ellitsgaard (1987) who studied parachuting injuries from 110,000 sports jumps carried out in Denmark. Landing was responsible in 83.8% of the injury cases and spinal injuries accounted for 10.2% of all the injuries, 7.9% being identified as some form of fracture.

Alexander (1985), in a review of athletic lumbar spine injuries, reported that sports which commonly incur LBP include gymnastics, weightlifting and American football. The most common types of injuries were muscle strains, ligament sprains, lumbar vertebral fractures, disc injuries and neural arch fractures, with the latter being the most serious. McCarroll et al. (1986) followed 145 new recruited American college football players for a period of 5 years. A higher than average incidence of lumbar spondylolysis and spondylolisthesis was found, 15.2% as compared with approximately 6% in the general population.

A four times higher than average incidence of spondylolysis was reported by Jackson et al. (1976) in female gymnasts. Kotani et al. (1971) found that 30.7% of a weight-lifting group had developed spondylolysis after 4 to 5 years of weight-training activities. Twenty four out of the 26 weight-lifters (92%) had experienced an episode of LBP. In a study of 25 weight-lifters and 25 track and field athletes who used weight-training exercises as a major part of their training, 40% and 48% reported incidences of LBP, respectively (Aggrawal et al., 1979).

In addition to those sports already mentioned, back pain would appear to be common amongst runners (Zamzow and Feigel, 1982). Blair et al. (1987) used two clinical examinations to compare 2102 runners and 724 non-runners. A follow-up, mail questionnaire was completed after a period of approximately 4 years. Over this period back injury rates per 1000 person-years were 13.8 and 7.3, for runners and non-runners, respectively. Lysholm and Wiklander (1987) followed 60 runners (19 sprinters, 13 middle-distance and 28 long-distance runners) over a one year period. Two-thirds of the runners sustained at least one injury which varied depending on the type of runner. Middle-distance athletes reported the highest incidences of hip and back injury.

Active physical exercise has been linked with back problems in the young. Haapanen et al. (1985) reported on seven cases of anterior intervertebral disc herniation in children (mean age 12.75 years) active in sport. Kono et al. (1975) divided 1233 schoolboys into a sports group (891 boys) and non-sports group (342 boys) and a radiological survey revealed that spondylolysis was 3.4 times as frequent in the sports group.

Clearly, the epidemiological data concerning sport related LBP are limited and often subject to criticism. In many instances they fail to quantify the physical loading or seek to evaluate the mechanisms underlying injury. Activities within the training regimens, as opposed to the actual sport, may themselves present the risk factor. For example, weight training exercises are often undertaken by the majority of sports participants. Future epidemiological studies must seek to address these issues.

3.2.4 The Site and Mechanism of Pain

The aetiology of back pain is multifactorial and more often than not the causative factors are interactive (Troup, 1984). Troup (1981) suggested that injuries to the spine may be classified into three broad categories:

(i) injuries that affect non-bony tissue and the spine remains stable;

(ii) injuries that cause the spine to be unstable and put the spinal cord or nerve roots at risk;

(iii) injuries of the spinal column that cause gross neurological damage and imminent deformity.

Those belonging to the first group are the most common cause of back pain.

Although sport has been implicated as a risk factor in LBP, it must be realised that injury occurs as a result of a mechanism inherent to the activity, not as a result of the sport itself. Sport does not produce injuries, but many involve mechanisms which overload body structures in

some way (Williams, 1980) and it is this overloading that causes the injury. The mechanism in sport is usually a force or stress, the magnitude, direction and point of application of the force being the most important predeterminants of the type of injury sustained by the lumbar spine (Alexander, 1985).

Persistent LBP in athletes is felt to originate from a derangement of specific anatomical structures due to repetitive (micro) or exceedingly forceful (macro) trauma (Brigham and Schafer, 1988). As highlighted when discussing LBP risk factors (section 3.2.2), there is evidence to suggest that mechanical overload or fatigue is a likely cause of LBP. Thus, concern must lie with trying to identify the possible mechanical origins of pain. However, before one can appreciate the influence of mechanical factors in the onset of pain, it is essential to possess an appreciation of the structural components of the spine and their inherent mechanical properties.

The human spinal column provides a linkage system between the head, arms and legs, as well as a protective housing for the spinal cord and its many nerve roots. Functionally, it must act as a shock absorber while at the same time provide a combination of strength and flexibility, that affords maximal stability with minimal restriction on mobility. It consists of some 33 bony elements (Figure 1):- 7 cervical, 12 thoracic, and five lumbar vertebrae, plus the sacrum and coccyx (5 and 4 fused vertebrae, respectively). In the lateral or sagittal plane there are four distinctive curves - cervical, thoracic, lumbar and pelvic curves - which may well play a mechanical role in the absorption of forces (Finneson, 1973).

Except for the first two vertebrae in the cervical region the 24 presacral vertebrae are similar in architecture. Each vertebra consists of a central body or centrum and a posterior vertebral arch which encloses and protects the spinal cord. The spinal cord runs down the length of the spinal canal and is essentially a single body. On reaching the uppermost lumbar vertebra it separates into a bundle of nerve roots, the cauda equina. Figure 2 presents a diagrammatic view of the upper surface of a lumbar vertebra showing the respective location of the spinal canal, the nerve roots within the spinal dura, and the

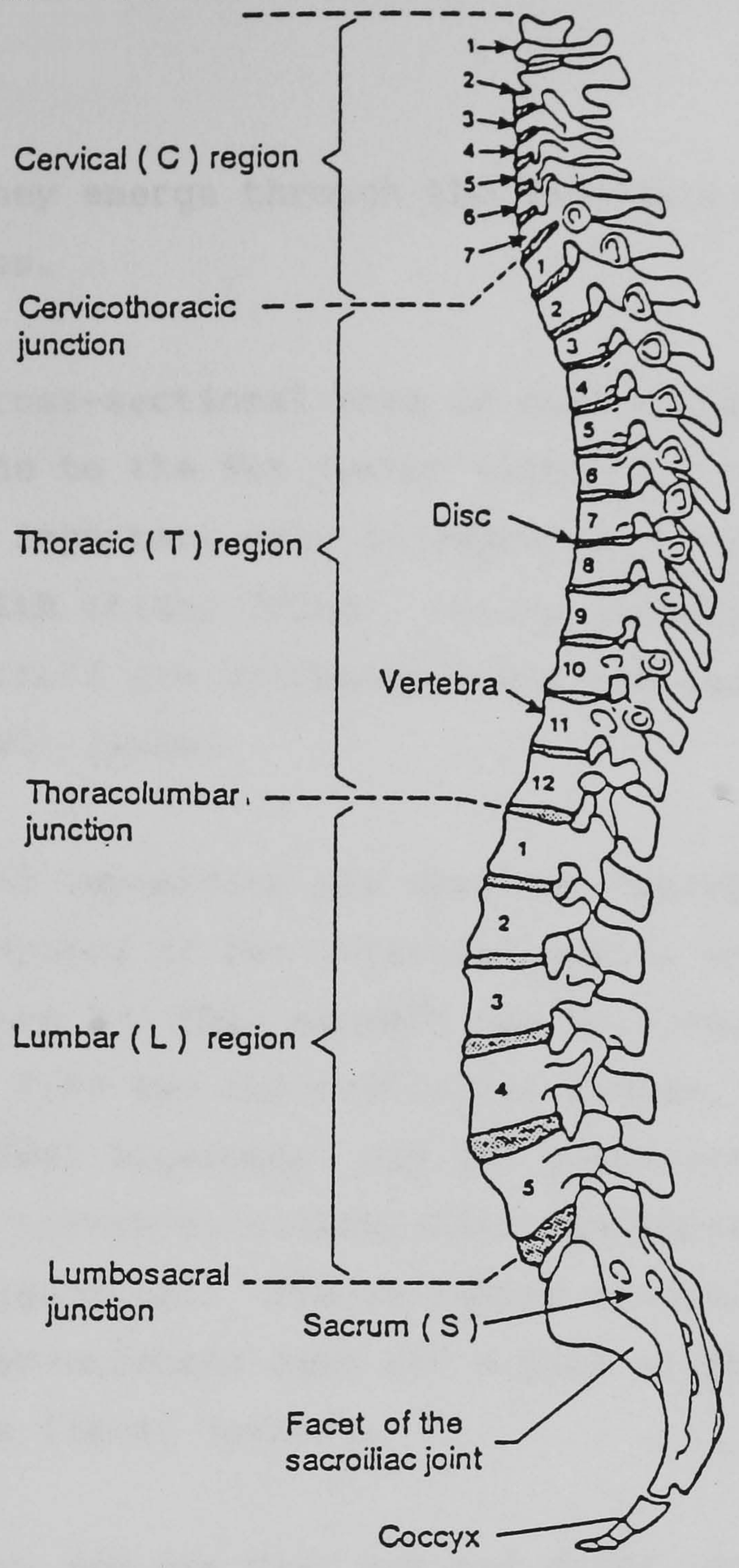


Figure 1. A lateral view showing the basic structures and curves of the adult spine.

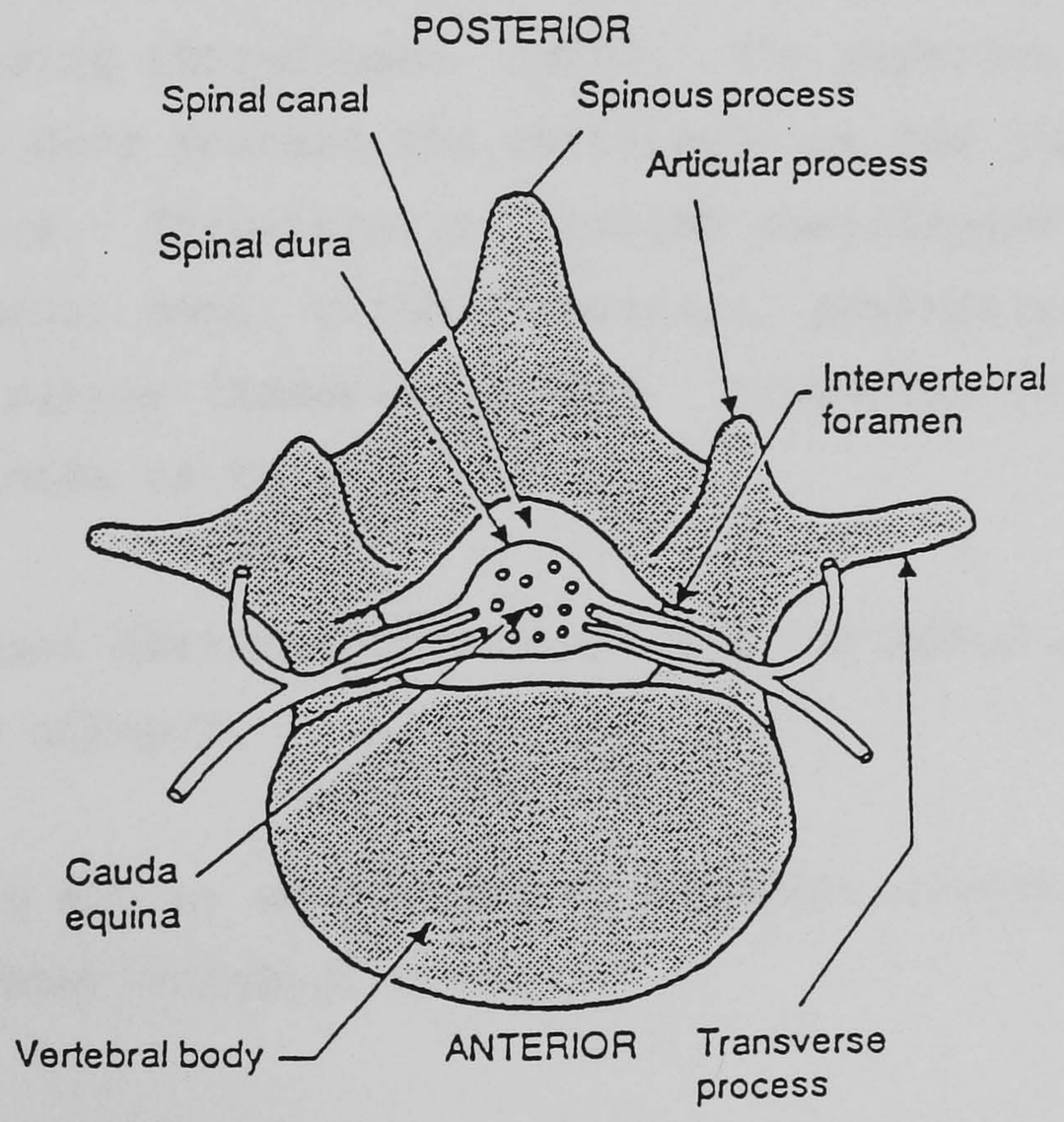


Figure 2. A diagrammatic view of the spinal nerves emerging from the spinal canal.

nerve roots as they emerge through the intervertebral foramina to their designated tissues.

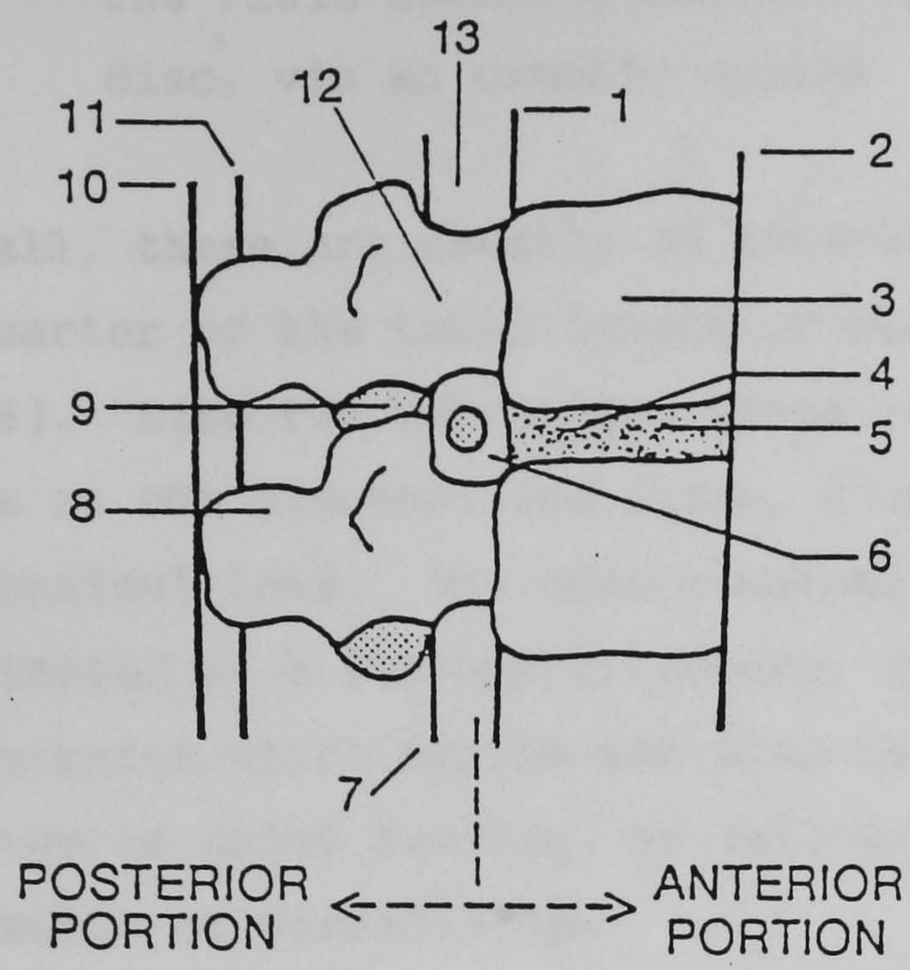
The height and cross-sectional area of each vertebra increases as one descends the spine to the 5th lumbar vertebra. This increase in size may well play an important role in supporting load, which also increases in conjunction with spinal depth. Forces resulting from the upper limbs, head and trunk are transmitted via the sacroiliac joints and pelvis to the lower limbs.

The motion segment represents the smallest functional unit of the spinal column and is composed of two vertebral bodies and their intervening soft tissue (Figure 3). This segment can be viewed in terms of its anterior portion (the two intervertebral bodies, the intervertebral disc and the longitudinal ligaments) and its posterior portion (vertebral arches, the intervertebral joints, the transverse and spinous processes, and associated ligaments). The vertebrae articulate via three joints:- the anterior intervertebral disc and a pair of posterior synovial apophyseal joints (facet joints).

The vertebral body and the disc are the major load bearing structures of the spine. Trabecular bone makes up the major part of the vertebral body and serves to resist any inward bending of the thin surrounding cortical bone. This structural arrangement makes the vertebrae extremely stable under axial loading (Brinckmann, 1985). The superior and inferior surfaces of the body possess the cartilaginous end plates made of hyaline cartilage. Structurally, healthy cartilaginous plates are made up of a homogenous, soft, pliable material, possessing a complex protein polysaccharide matrix (Kazarian, 1981). Armstrong (1958) listed three functional purposes of these plates;

(i) a structure designed to protect the vertebral spongiosa from pressure atrophy,

(ii) a body to act as an anatomical boundary confining the components of the intervertebral disc,



1. Posterior longitudinal ligament
2. Anterior longitudinal ligament
3. Vertebral body
4. Cartilaginous end-plate
5. Intervertebral disc
6. Intervertebral forearm with nerve root
7. Ligamentum flavum
8. Spinous process
9. Intervertebral joint and facet
10. Supraspinous ligament
11. Interspinous ligament
12. Transverse process
13. Vertebral canal

Figure 3. The motion segment and its component parts.

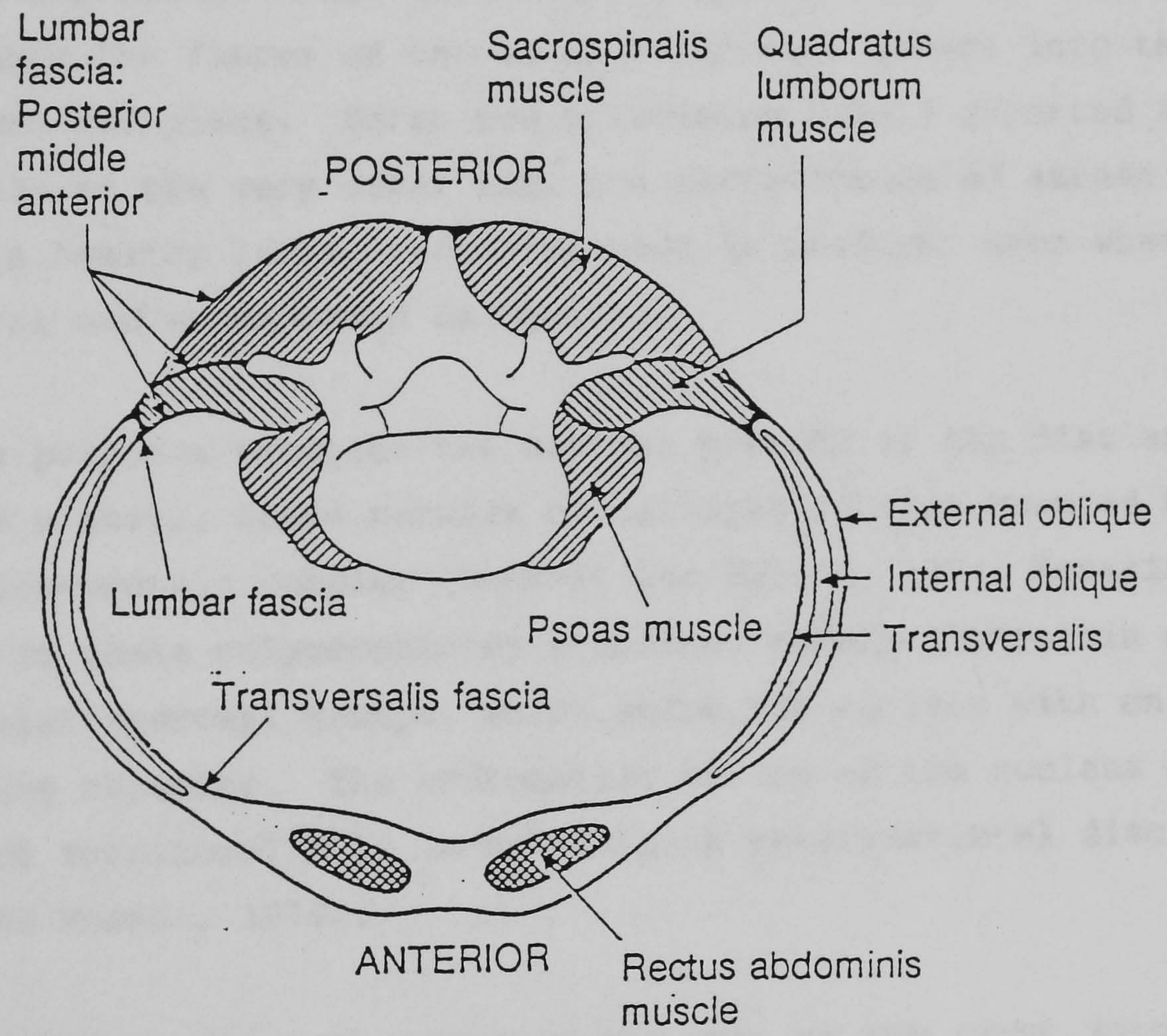


Figure 4. A cross-section through the lumbar spine showing the principle muscle groups and fascia.

(iii) a material which serves as a semi-permeable membrane facilitating the fluid exchange between vertebral body and intervertebral disc, via an osmotic system.

In all, there are usually 24 intervertebral discs which contribute about a quarter of the total length of the spinal column (Troup and Edwards, 1985). Like the vertebrae, these tend to increase in cross-sectional area as one descends the spine, accommodating the increases in mechanical load. The disc consists of two parts, a nucleus pulposus encircled by a fibrous structure, the annulus fibrosus. It is these structures which enable the disc to withstand and dissipate some of the energy of brief loading, as well as enabling the motion segment to have a degree of flexibility.

The fibres of the annulus fibrosus are oriented at an angle of about 60 degrees to the long axis of the spine, allowing the disc to limit both torsion and any radial bulge (Brinckmann, 1988). Near the rim of the vertebral body the fibres of the annulus fibrosus insert into the cartilaginous end plate. Horst and Brinckmann (1981) reported that with the exception of the very outer rim, the distribution of stress on the surface of a healthy lumbar vertebral body is uniform, even when loading is unilateral and accompanied by bending.

The nucleus pulposus occupies the central portion of the disc and is composed of a small, dense network of collagen fibrils covered by a polysaccaride-protein complex (Markolf and Morris, 1974; Kazarian, 1981). It is these polysaccharide-proteins, namely chonroitin sulphate with its polar hydroxyl groups, which endow the nucleus with an abundant water-binding capacity. The hydrophilic nature of the nucleus gives it an important functional role in maintaining intervertebral disc height (Markolf and Morris, 1974).

Often neglected in the literature is the role of the facet joints in load transmission. Lin et al. (1978) showed that the facet joints provide limited support during pure axial loading but provide a second load path in complex loading. The principal functional role of the facet joints is to limit the degree of directional freedom of the spinal motion segment. The plane of the joints varies in geometric orientation at each of the different spinal levels which directly influences spinal

mobility. They are true synovial joints enclosed by capsular ligaments.

Stability of the spine is felt to stem from an intrinsic and extrinsic mechanism. Intrinsic stability arises from the ligaments and discs, and extrinsic stability from the muscles around the spine. The ligaments' principal functions are to;

(i) Control the direction of motion,
(ii) Provide passive resistance to spinal motion,
(iii) Limit movement towards the end of its normal range,
(iv) Transfer tensile loads from one vertebra to another.

Major spinal stability is provided by the trunk muscles, of which the principal ones can be seen in Figure 4. There are numerous muscles which directly control the movement of the trunk by coordinated activity and are distinguished by their length and fibre orientation. The erector spinae muscles lie posterior to the vertebrae and constitute a group of muscles directly responsible for controlling extension, the rate of flexing under gravity, rotation and lateral flexion (Örtengren and Andersson, 1977). Lateral to the vertebrae are the quadratus lumborum, active in lateral flexion (Troup, 1979), and the psoas muscle, responsible for controlling hip/trunk flexion and lumbar posture (Nachemson, 1966). The abdominal muscles include the rectus abdominis, which resists extension and supports trunk flexion under gravity (Troup, 1979), and the oblique abdominal muscles, the external and internal obliques and transversalis, which are active during rotation and lateral flexion, and aid trunk flexion (Troup, 1979).

According to the literature, the most likely site of injury to the spine is at the junction between the last lumbar vertebra and the sacrum:- L5/S1 disc (Farfan, 1973; Armstrong, 1958). In mechanical terms two theories would seem to support this notion. Firstly, it represents the point at which the entire body mass above the pelvis acts. Secondly, in the normal anatomical position, a large lumbosacral angle is formed by the 5th lumbar vertebra and the head of the sacrum. As a result, compressive loading of the spine induces a higher shear force and consequently, places a larger strain on the annulus fibrosus of the disc and neural arch of the vertebrae (Alexander, 1985). Nachemson (1976)

highlighted six pieces of work which he proposed lend further support to the lower intervertebral disc being the most likely cause of pain.

As a mechanical structure, the vertebral column offers considerable rigidity and resistance to injury. The spine is able to withstand purely compressive, flexing or extensing forces easier than if these were combined in some way, or a torsional component existed (Roaf, 1960). Brinckmann (1988) has shown that the probability of a motion segment fracture is dependent on the size of the load and the number of load cycles, a rise in both these variables causing an increase in the probability factor. Perey (1957) reported an inverse relationship between the strength of the spine and the duration of the applied load. When a dynamic load was applied for a brief duration, Hodgson et al. (1963) found that the likelihood of injury was related to the peak acceleration and jerk (below 1000 $g.s^{-1}$).

In terms of the structures most susceptible to failure, the healthy disc would appear stronger than the vertebral body (Roaf, 1960). However, the age and degree of degeneration decrease the mechanical strength characteristics of the spinal units. Adams and Hutton (1982) found that a prolapsed lumbar disc can be a compressive injury of a joint flexed a few degrees beyond the normal limit. In addition, they found that slightly degenerated discs, from 40 and 50 year olds, were more likely to fail than were normal discs.

As to the origin of pain, degenerative joint disc disease is regarded by many as the most common cause of LBP (Krämer, 1981; Pope et al., 1980). The disc degeneration can have a direct or indirect influence on the cause of pain. As previously mentioned, spinal loads and the types of movements have been linked with increased risks of LBP and consequently, are felt to be strongly related to the aetiology of disc degeneration (Chaffin, 1974; Nachemson, 1976). White and Gordon (1982) constructed a schematic diagram to represent how they considered mechanical factors can lead to LBP symptoms (Figure 5). Three mechanisms were felt to have a direct or indirect influence:- structural/material changes, biochemical/nutritional changes, and inflammation. It was thought that at any one time, one or more of these mechanisms could be responsible for pain and they were often inter-related.

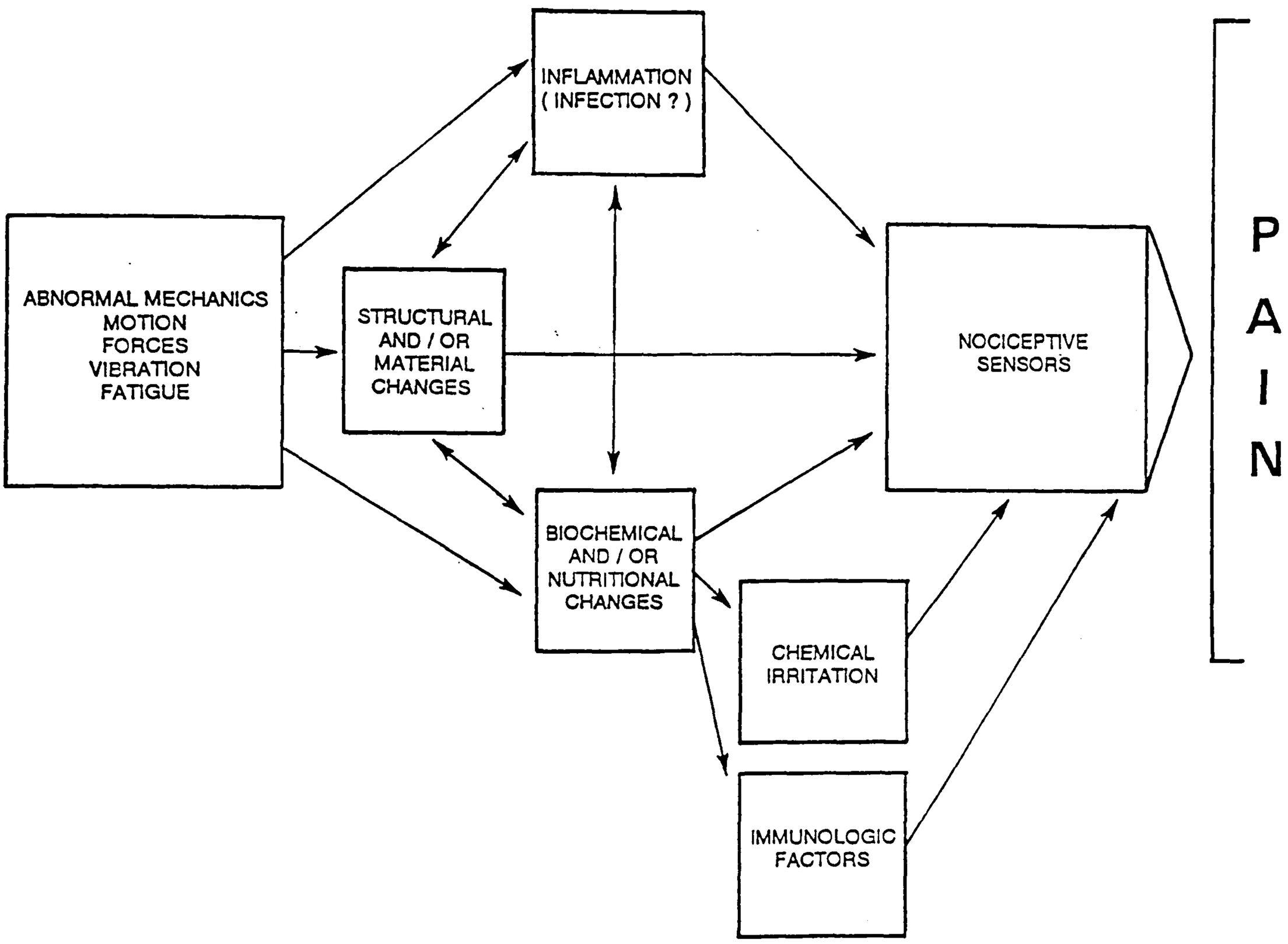


Figure 5. A schematic representation of the mechanisms involved in the mechanical onset of back pain.

(White and Gordon, 1982)

Structural or material changes are very much related to the close proximity of spinal structures and the large presence of nerves in the surrounding tissues. Nerve endings occur in the fibrous capsule of the facet joints, ligaments, periosteum, epidural sleeves of the neural foramina, the adventitia of paravertebral and epidural veins, and of vessels in the muscles and vertebral marrow (White and Gordon, 1982). Smyth et al. (1958) performed experiments which suggested that the peripheral layers of the annulus fibrosus of the disc may contain nerve endings.

Mechanical irritation of nerves can arise from a narrowing of the intervertebral foramen due to losses in disc height and encroachment of vertebral tissue. Disc height is principally governed by the water content of the nucleus pulposus which decreases with both age and spinal loading (Kazarian, 1981). Krämer (1981) reported that the foramen decreases by about a fifth during a day of constant loading. Also, extension and bending movements further reduce the size of the foramen.

Disc bulging, protrusion and prolapse are thought to be the most obvious cause of mechanical irritation. Under compressive loads up to 800 N and moments in flexion, extension, right lateral bending or torsions up to 11.8 Nm, Reuber et al. (1982) measured disc bulges of up to 2.7 mm, beyond the normal unloaded state. They also showed that degenerated discs bulge more than non-degenerated discs. Fluid injection into the disc or posterior element removal had little influence on disc bulge.

Intra-foraminal space can be further reduced if inflammation and swelling of the perineural tissues occur during disc protrusions or osteophytic reactions occur at the articular facets (White and Gordon, 1982). Krämer (1981) also identified the facet joints as a possible source of pain. He proposed that during disc height loss the joints move 'telescopically' over each other, if this was to be excessive then additional stress would be placed on the capsular ligaments.

A mechanical origin cannot be the only source for pain, as pain has been evident without any physically observable defects. Therefore, chemical factors have been proposed as possible sources of pain. Nachemson (1969) measured pH values in the intervertebral discs of patients with prolapsed discs and found them to be below normal, indicating an

increase in acidity. The theory put forward was that the proteolytic enzymes in the lysosomes of the connective tissue release amino acids from the protein polysaccharide complexes and the pH value is lowered. By loading and increasing the transport of fluid out of the disc, acid metabolites then cause inflammatory reactions in the adjacent nerve fibres.

Disc degeneration has been mentioned as a significant factor in the cause of LBP by affecting the functional properties of the spinal units. An alteration in disc nutrition is felt to be closely related to this degeneration which depends on a balance between the metabolic requirements of the disc's cells and the removal of waste products from it. Diffusion is the primary mechanism for this transport as the disc is avascular in nature.

Perey (1957) observed that discs subjected to high loads suffer micro-fractures of the cartilage endplates which may well influence the metabolic exchange. Holm and Nachemson (1982) investigated the effects of spinal fusion on metabolic movements within the discs of dogs. They observed that along with a decrease in the water content of the nucleus pulposus (approximately 15%) the fused discs became hypoactive and the discs above and below the fusion were hyperactive.

The motion segment is a highly complex functional unit whose mechanical characteristics can be altered by any slight change in its structural components. Brinckmann (1985) concluded that the maximum incidences of LBP will occur when there is coincidence of a certain level of pathological change, increased abnormal motion resulting from the pathological change, and considerable stress on the spine from environmental factors, such as a blow to the spine during landing from a fall. The inter-relationship is complex and support for this notion would seem to come from the fact that a majority of acute LBP is said to be unprovoked. Brinckmann et al. (1988) later went on to state that it is not the violence itself that decides the extent of the damage but the trauma combined with the condition of the disc at the time of injury. The more a disc is compressed the less additional trauma may be required to cause lesions. When LBP is the first unforeseen event the question is whether the back was already vulnerable or whether it was overloaded.

As outlined earlier, the physical mechanisms inherent to many sporting activities ensure that potentially stressful environmental factors are present. The review will now go on to consider one such activity, plyometrics, which is relatively new and advocated to those athletes seeking to enhance muscular power. Increasingly more and more athletes are participating in these exercises. However, the literature is clearly devoid of information pertaining to the forces generated during their execution and whether they can constitute a risk to the back.

3.3 PLYOMETRIC EXERCISES

Innovations and advancements in training regimen are felt to be a major contributor towards present athletic achievement (Clutch et al., 1983) and as such must be closely assessed to gauge, not only potential benefits, but also possible limitations. Just such a new training procedure was brought to the attention of athletes and coaches by Verhoshanski (1967) and has since been implemented in the majority of training programmes designed to enhance athletic muscular power.

Verhoshanski (1969) reported that depth jumps, often referred to as drop jumps, were a 'shock' method employed by Russian athletes to improve the reactive ability of the nerve-muscle apparatus. The drop jump required athletes to drop from a raised platform, to the ground, and upon contact perform an immediate jump for maximum height. From related exercises the term 'plyometrics' has emerged. Wilt (1976) provided a preliminary definition of the term as:

> "Exercises or training drills used in producing an overload of isometric-type muscle action which invokes the stretch reflex in muscles".
>
> Wilt (1976)

Whether isometric-type muscle action should be included in this definition is questionable as the training stimulus depends on a highly dynamic muscle stretch-shortening cycle to store and release the necessary elastic energy.

Besides drop jumping, other forms of plyometric exercises include single response jumps (repeated jumps with pauses), bounding sequences (a number of jumps undertaking over flat ground) and box drills (a series of bounding and drop jumps). In one form or another, these types of exercises have been advocated for athletes seeking to develop muscular power in events such as the shot, discus, javelin, hammer, long jump, high jump, sprinting, rowing, basketball, and volleyball (Verhoshanski, 1969; Johnson, 1976; Sitz, 1982; Wilt, 1976; White, 1982; Lohmann, 1983; Miller, 1981; Boase, 1983; Young and Marino, 1985). It has even been suggested that the young (10 to 15 year olds) should utilise such activities to develop the effective execution of take-offs in jumping

events (Mekhonoshin, 1984). Considering that these jumping and bounding activities require the human body to control and absorb considerably large ground reaction forces (Bobbert et al., 1987b), their potential for back injury must be high.

3.3.1 Plyometrics and Muscle Mechanics

Plyometric exercises utilise the sequential timing of muscle actions to generate greater forces. To appreciate this concept an understanding of muscle mechanisms is required.

Gasser and Hill (1924) suggested a three-category system of classification to describe muscle actions:- concentric, static and eccentric contractions. A concentric contraction refers to a muscle action which involves the development of tension in conjunction with a shortening of the muscle. Thus, the muscle force overcomes an opposing resistance. A static contraction occurs when a muscle develops tension with no apparent change in overall muscle length, ie, the force and resistance are equal. Eccentric contractions involve the development of tension within a muscle whilst there is an increase in muscle length. Eccentric muscle contractions do not cause movement since the movement occurs in a direction opposite to that in which tension is developed.

Muscle-force graphs calculated from experiments on isolated toad sartorius, frog gastrocnemius and the forearm flexor muscles in man have shown that muscles display a unique characteristic (Cavagna et al., 1968) with regard to the force and speed of contraction (Figure 6). The graph shows that for negative velocities of contractions (muscle stretching) the maximum force created is much greater than that during positive contraction velocities (muscle shortening). The maximum isometric force represents the point on the curve which intersects the vertical line, representing zero velocity of muscle shortening.

It is possible to assume that if an active muscle is initially allowed to stretch prior to contraction then the force developed in the subsequent shortening will be increased. This assumption has been proved experimentally by the work of such authors as Cavagna et al. (1968) and in practice, for example, during trunk rotation (Grieve and Arnott, 1970). It is this fundamental concept that underlies the

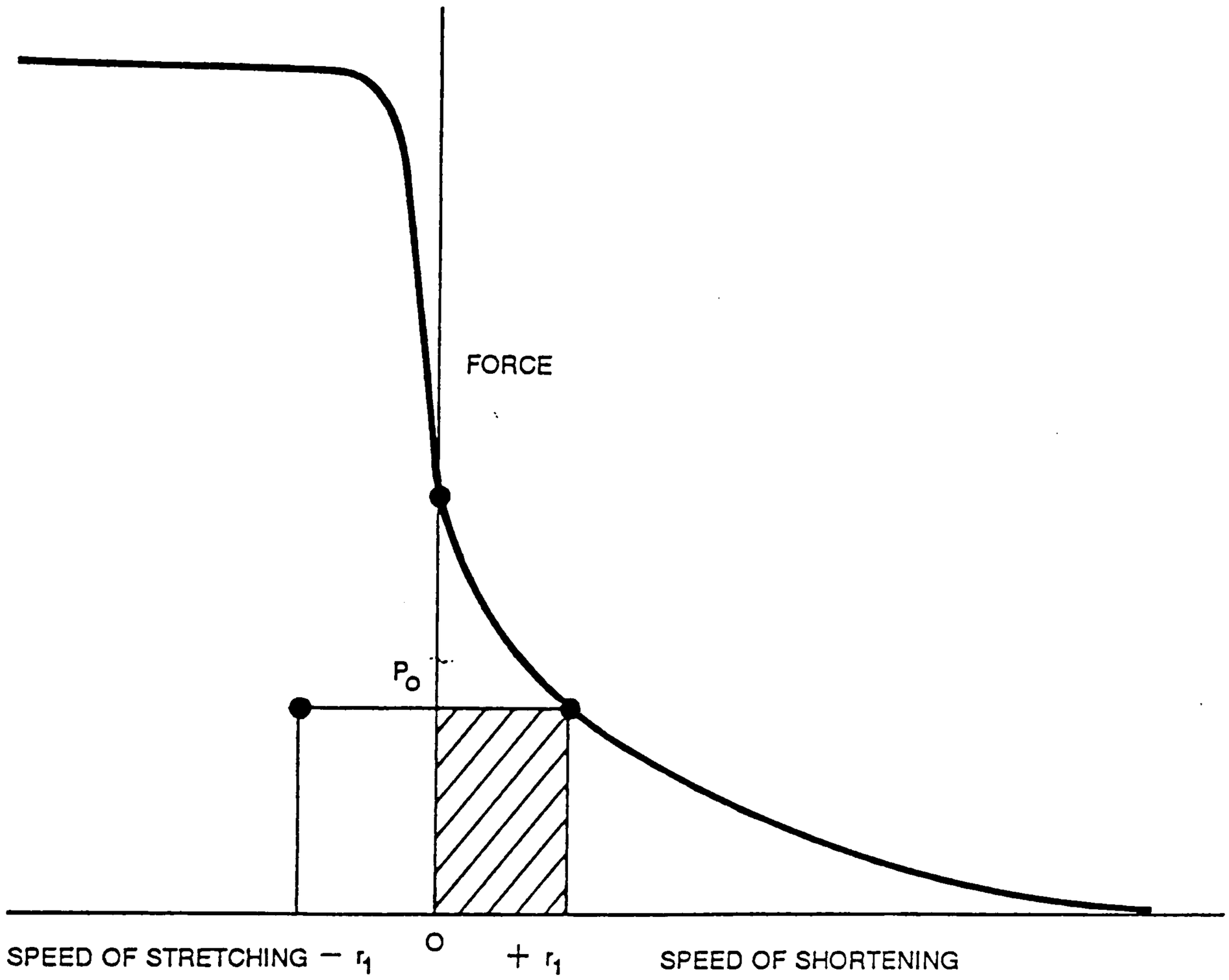


Figure 6. The force-velocity relationship during eccentric (stretching) and concentric (shortening) muscle actions.

principles behind plyometric training which seeks to induce a stretch-shortening muscle action. Thus, such exercises endeavour to create an eccentric muscle contraction prior to a concentric contraction.

Explanation for the increased muscular force is believed to lie partly with the storage of elastic energy in the series elastic component (SEC) of the muscle when it is pre-stretched (Cavagna et al., 1968). The SEC is thought to be present not only in tendons but also in the cross-bridges between the actin and myosin filaments of muscle fibres (Bosco and Komi, 1979). A critical factor in the muscle's ability to utilise stored elastic energy is the shortness of latency within the stretch-shortening cycle (Bosco et al., 1981). Bosco et al. (1981) hypothesised that if the stretched muscle position is held too long it leads to a detachment of actin and myosin cross-bridges and will, therefore, reduce subsequent energy output during muscle shortening. Thus, the time between the muscle's eccentric and concentric contraction must be minimal.

As the velocity of muscle contraction plays an important role in force production, then the structural and compositional differences within skeletal muscle must have an influence on the ability to store and re-utilise strain energy. Such compositional differences within muscle fibres have been identified and loosely classified into fast-twitch (white) and slow-twitch (red) fibre groups. The fast-twitch fibres work largely anaerobically, whereas slow-twitch fibres are capable of prolonged oxidative processes (refer to Reilly, 1981c, for a more detailed description of the characteristic differences which separate these two fibre groups).

Viitasalo and Bosco (1982) found that subjects with a higher proportion of fast-twitch muscle fibres jumped significantly higher when performing a vertical jump from a static squat position (concentric muscle contraction) than a slow-twitch fibre group. With counter-movement jumps and drop jumps (eccentric-concentric muscle contractions) there was no significant difference between muscle fibre groups. Jumping performance from a static squat position would seem to support the theory that subjects with predominantly more fast-twitch fibres can recruit motor units more explosively than predominantly slow-twitch fibre subjects. The fact there was no difference between the fibre type

groups in performance of the plyometric exercises, suggested to the authors that the slow-twitch fibre type group could utilise stored elastic energy more efficiently.

Plyometric activities are proposed to have a two-fold effect during training:

(i) improve the ability of muscles to utilise stored elastic energy.

(ii) enhance the stretch reflex neuro-muscular system.

(Mann, 1981; Clutch et al., 1983).

To appreciate how they may aid the latter, consideration must be given to the function and action of muscle spindles.

Muscles possess mechanoreceptors known as muscle spindles which are capable of detecting changes in muscle length. The spindles are composed of two parallel fibres which move relative to each other when a muscle is stretched and the resulting movement propogates an action potential, via the afferent nerve supply, to the spinal cord. Some of the nerve impulses are transmitted monosynaptically (ie, directly back to the muscle and not via the brain) causing excitation and hence, a concentric contraction (Åstrand and Rodahl, 1986). It is thought that plyometric activities facilitate this reflex, establishing a better neural pathway and allow a greater number of afferent impulses to be transmitted to the motor (alpha) neurons of the stretched muscle (Clutch et al., 1983).

3.3.2 Drop Jumping and Bounding: A Training Stimulus

The effectiveness of plyometric training in improving jumping performance has been demonstrated in several studies (Scoles, 1978; Blattner and Noble, 1979; Steben and Steben, 1981; Clutch et al., 1983). Scoles (1978) investigated the influence of an 8 week drop jump training programme on vertical and standing broad jump performances. Subjects were divided into 3 groups:- a drop jumping group, a flexibility group and a control group. The drop jumps were performed from a height of 0.75 m and after the training period were found to increase vertical and standing broad jump performances by 0.02 and 0.08 m, respectively. The control and flexibility groups showed no significant improvements.

Blattner and Noble (1979) compared drop jumping exercises from a height of 0.86 m with exercises performed on an isokinetic leg press machine (Model 16BX 'Leaper' leg press machine manufactured by Mini-Gym, Inc, Independence, No.). This entailed three sets of 10 repetitions with subjects encouraged to exert a maximum effort. At the start of each repetition the subjects' knees were bent at an angle of 90° and movement speed was the maximum allowable by the machine (0.45 $m.s^{-1}$). Over the 8 week training period the male drop jumping subjects were progressively given additional load in the form of a weighted vest. For the first two weeks of training no resistance was used. During weeks 3 and 4, a 45.8 kg mass was added and then increased by 23.1 kg every two weeks, thereafter. A 0.05 m mean increase (10% improvement) was noted in a vertical jump test which was comparable to that experienced by the isokinetically trained subjects.

Steben and Steben (1981) investigated whether specific plyometric drills could facilitate improvements in field event performances of young subjects as compared with 'normal' physical education class warmups. The population consisted of 160 students enrolling in physical education classes at a California school. These were divided into four groups; a group performing drop jumps from 0.25 m, a group who used box drills specifically designed to improve triple jump performance, a group who trained with a series of flexibility-agility exercises utilising hopping, bounding and high knee lift movements in a linear direction, and a control group not involved in specific training exercises but did take part in physical activity. All subjects undertook training for 10 min per day, five days per week, for seven weeks. The results showed that those training with the plyometric drills did improve their high, long and triple jump performances.

Clutch et al. (1983) reported on two experiments which investigated:

(i) the benefits of certain drop routines when combined with weights;

(ii) whether athletes engaged in training for their sports could benefit from drop jumping exercises.

In the first experiment, 12 male students, enrolling on a weight training class, trained with weights and participated in three jumping programmes; a maximum vertical jump, and 0.3 m, 0.75 m and 1.10 m drop jumps. Training was carried out twice weekly and for a total of four weeks. The three jumping programmes all resulted in an increase in one repetition of maximum squat strength, isometric knee extension strength and vertical jump, but there was no significant difference between routines. The authors concluded that drop jumps, when combined with weights, are no more effective than a programme of regular maximum jumps. Whether a noticable difference in performance can be expected within the training period conducted is a question which needs to be addressed.

The second experiment employed 16 subjects from a weight training class and 16 members of a volleyball team. All subjects trained twice a week, for 16 weeks, and were randomly assigned to train either with weights and drop jumps, or just with weights. It was found that there was a significant increase in vertical jump performance for all groups, with the exception of those weight lifters who did no jumping. The active volleyball players, who were involved in both weight training and jumping, made improvements in performance that was equivalent to those experienced by the volleyball players who were only weight training. Thus, it was felt that drop jumps are useful for athletes who do no other jumping but have little benefit for athletes who use jumping exercises as a principle component of their normal practice.

To invoke the necessary rapid eccentric-concentric muscle contraction, drop jumping uses the body mass, falling under gravity, to pre-stretch the lower limb muscles. The range of heights employed has varied between 0.12 to 1.1 m (Fahmi, 1986) and there would appear to be controversy as to the necessary optimum height to induce a training effect. Verhoshanski (1969) stated that drop jumps should be performed from a height of between 0.75 and 1.1 m. A height of 0.75 m was felt to facilitate the maximum speed for which the muscles could switch from yielding work to overcoming work. At 1.1 m it was suggested that maximum dynamic strength is developed.

Asmussen and Bonde-Petersen (1974) found that maximum vertical rebound jumps were obtained when the drop height was 0.4 m. In contrast, Fahmi

(1986) studied both power and endurance athletes and found on two occasions that a height of 0.12 m produced the maximum rebound jump.

Although there are contrasting views as to the optimum drop height, many agree that maximum knee bend during the initial drop land should not be excessive (White, 1982; Boase, 1983; Young and Marino, 1985). Boase (1983) stated that excessive knee flexion on landing will result in a slower eccentric muscle action leading to muscle spindles being stimulated asynchronously and consequently, reduce explosive power output. This may not be the only explanation, excessive knee flexion is likely to reduce the biomechanical efficiency of the drop jump. During deep knee bends, considerably more work will be required to raise the body's centre of gravity.

3.3.3 Adverse Effects from Eccentric Based Muscular Actions

Exercise, particularly unaccustomed strenuous activity involving high mechanical forces, can result in muscle fibre damage and pain. Plyometric activities induce such forces and incur problems such as morphological changes, delayed-onset muscle soreness, performance decrements, and elevation of muscle protein levels in the blood (Ebbeling and Clarkson, 1989).

The severity of the muscle actions in plyometric training was realised by Verhoshanski (1969) who commented that drop jumping should be discontinued by athletes ten to fourteen days before competition. The damage associated with such exercises is not permanent and tissues are repaired (Friden et al., 1983). It has also been noted that a muscle adaptation takes place with repeated exercises such that subsequent exercise bouts provoke fewer symptoms (Schwane et al., 1987).

Generally, there are two types of pain associated with severe muscular exercise:- a temporary soreness, characterised by moderate pain and felt during the final stages of fatiguing exercises, and a delayed-onset muscle soreness, unrelated to fatigue and described as a dull, aching pain, combined with tenderness and stiffness. The latter is of primary concern to this study.

The pathophysiological mechanism underlying delayed muscle soreness is unclear but several theories have been put forward. A view supported by

many is that damage occurs to the connective elements of the muscle and tendons (Abraham, 1977; Friden, 1983). The damage is felt to originate from an over-stretching of the muscle's connective tissue among the fibres and myofibrils, leading to a subsequent inflammation and pressure build up around nerve endings, which would explain why there is a close relationship between eccentric exercise and delayed onset muscle soreness (Abraham, 1977; Ebbeling and Clarkson, 1989). Friden et al. (1983) have reported extensive damage to the z-band of the myofibrils in the form of marked streaming, broadening and, in places, total disruption following intense eccentric muscular contractions.

Indirect measures of muscle damage have focused on soluble intramuscular enzymes in the blood, particularly creatine kinase (CK), which is believed to stem from damage to the membranes of muscle fibres. Tiidus and Ianuzzo (1983) identified a positive relationship between serum CK activity and muscle soreness 24 hours post-exercise, following a dynamic leg exercise involving concentric and eccentric muscle contractions. Increased intensity and duration of exercise produced increased enzyme levels and muscular soreness, with intensity having the more pronounced effect. The highest enzyme activity occurred 8 to 24 hours post-exercise, whereas peak muscular soreness sensation occurred 48 hours post-exercise.

This inconsistency has been reported elsewhere (Newham et al., 1983) and others have found that the delay can vary in length depending on the type of exercise (Ebbeling and Clarkson, 1989). Three to six hours after downhill running or isometric exercises, CK activity levels have been shown to rise significantly, reaching a peak 18 to 24 hours post-exercise. However, local muscular eccentric exercise may not cause a significant increase in CK activity until 48 hours post-exercise and may not peak in some subjects until 7 days after exercise.

The inconsistency of measured CK activity and perceived pain may well be influenced by the fact that CK concentrations in blood represent both the efflux and clearance of the enzyme. Thus, peak values may be misleading. Generally, muscle soreness develops during the first 24 to 48 hours following eccentric exercise and peaks between 24 and 72 hours. Within 5 to 7 days post-exercise, induced pain subsides and disappears (Abraham, 1977; Newham et al., 1983; Ebbeling and Clarkson, 1989).

The effects of exercise-induced muscle damage have been shown to influence certain motor performance parameters, particularly muscle strength. Directly following eccentric exercise there is an immediate decrease in maximum force which gradually recovers (Friden et al., 1983; Newham et al., 1987). Maximum strength may not be achieved for a week or longer. The decrease in strength could well be secondary to the muscle soreness, or as is more widely thought, stem from the morphological changes that have taken place (Komi and Viitasalo, 1976; Newham et al., 1987; Ebbeling and Clarkson, 1989). There appears to be no relationship between the time interval to develop muscle soreness and loss of strength (Friden et al., 1983).

Physical training of skeletal muscle over a period of time reduces morphological changes, delayed muscle soreness and CK activity in the blood following an acute bout of eccentric exercise (Ebbeling and Clarkson, 1989). The changes that take place are not fully understood, but alterations in energy metabolism and structural adaptations have been proposed. Schwane et al. (1987) studied the effects of progressive short-term training on both delayed muscle soreness and serum CK levels. Downhill running (predominately eccentric muscle contractions) and uphill running (predominately concentric muscle contractions) were used as the training exercise for a total of either 5 or 10 days. Delayed onset soreness was reduced, but not totally, by the fifth day of training and more noticeably by the tenth day of training, when running downhill. Uphill running had little effect on preventing muscle soreness and neither training had any significant effect on CK levels. It was concluded that delayed onset muscle soreness may be difficult to avoid totally but prevention is more likely if training encompasses a substantial eccentric component.

3.4 SPINAL LOADING

3.4.1 The Spinal Motion Segment's Response to Load

The impacts of landing during bounding and drop jumping are known to induce high joint reaction forces at the ankle, knee and hip joints (Smith, 1975; Bobbert et al., 1986; Bobbert et al., 1987a, 1987b), but little is known about the forces the spine must withstand. Berme et al. (1984) did estimate the compressive forces acting on the third lumbar disc during a simple drop landing from 0.33 m and found peak values ranged between 1360 and 4750 N. This spread in peak forces may reflect the diverse techniques people employ when seeking to absorb impacts of this nature. Unfortunately, the authors provided no data for individual kinematic differences during landing.

Several studies have examined the *in-vitro* response of the intervertebral discs to varying loads (Virgin, 1951; Hirsch, 1955; Perey, 1957; Kazarian, 1975; Cyron and Hutton, 1981, Koeller et al., 1984, 1986) and shown that the discs respond in a viscoelastic manner. Thus, when subjected to a compressive load for short periods of time the intervertebral disc loses and regains height immediately, with the application and subsequent removal of the load. Extending the loading period causes, in addition to the elastic response, a narrowing and stiffening of the disc (Markolf, 1972). The rate of reduction in disc height, often termed 'creep', is non-linear (Kazarian, 1975) with the greatest loss occurring in the early stages of loading. The rate of reduction decreases with respect to time until the disc and load are in equilibrium. Hysteresis and axial deformability both decrease during creep (Kazarian, 1984; Lin et al., 1978).

The mechanism or mechanisms underlying creep are not fully known, but both fluid loss and structural alterations have been proposed as possible explanations. The fluid mechanism theory proposes that the intra-discal space, cartilaginous plates, annulus fibrosus, paravertebral tissue and spongiosa of the adjacent vertebrae make-up a osmotic system (Krämer, 1985). Fluid is extruded from the disc when the combination of imbibition pressure of the protein-polysaccharide complex of the nucleus pulposus and the osmotic pressure difference across the cartilage endplates is exceeded by the compressive load. When the load is removed the pressure gradient is reversed, fluid is drawn back, and disc height regained.

The fluid shifts are proportional to the pressure gradient and the critical load dictating a fluid flow reversal is proposed to be between 70-80 kp (Krämer, 1985). Thus, as the disc is avascular, a regular loading and unloading above and below this value will promote the exchange of nutritional and waste products to and from the disc, and consequently, maintain its health.

Experiments by Koeller et al. (1984) showed creep and recovery can occur without fluid shifts. They identified extension and contraction of the annular fibres as the cause of disc height alterations. Brinckmann (1988) suggested that the time constant for fluid exchange and the elastic response of the disc may well be different, the former being in the order of hours, the latter in terms of minutes.

Koeller et al. (1984) also reported that creep behaviour is similar during static or dynamic loading. This is contrary to Kazarian (1972) who found increased creep response to dynamic loading when compared with static loading. The larger the peak to peak amplitude during the dynamic loading the greater was the creep response.

Hirsch (1955) examined motion segments, complete with skeletal and ligamentous parts, and subjected them to dynamic and static loads. Static loads were found to produce a reaction in the course of minutes, as compared with fractions of a second for dynamic loads. To simulate impulse loading, 1 kg weights were dropped onto pre-loaded segments (loads between 10 and 130 kg) from heights of 0.1 and 0.5 m. This additional force was found to create a vibration within the disc which lasted a few seconds. The greater the pre-load the less was the amplitude of the shock wave induced. These results prompted Hirsch (1955) to conclude that a loaded spine is less able to adapt to a sudden extra strain than if it were previously unloaded.

Perey (1957) adapted a similar approach to study the consequences of impulse loads on localised end-plate fractures. Dropping 15 kg weights from a height of 0.5 m on to pre-loaded cadaveric spines was found to cause peak compressive forces between 1050 and 1350 kp, over a period of 6 ms. Twenty incidences (26%) of end-plate fractures were observed in the specimens studied. Static compression tests were also performed in

which spinal specimens were compressed to breaking point. Motion segments, comprising of two and three vertebrae and their intervening disc, were noted to have visible end-plate fractures in 32 and 42% of cases, respectively. In addition, a greater resistance was found for those segments which were younger and when the duration of the compressive load was kept to a minimum. In cases of excessively high impulse loading there is evidence to suggest that the latter relationship may change and become dependent on the rate of change of acceleration or jerk. Hodgson et al. (1963) subjected seated cadavers to vertical accelerations and observed that at jerks of less than 1000 $g.s^{-1}$ the vertebral deformation was related to the magnitude of the acceleration.

The literature would suggest that motion segments vary considerably, particularly in terms of the forces they can withstand. Failure loads as low as 3 kN to as high as 12 kN have been reported by experimentors (Brinckmann et al., 1988). Perey (1957) found that failure of cadaveric specimens, aged forty years old and under, occurred when an average force of 780 kp was applied, the range being between 510 to 1100 kp. Specimens over the age of sixty failed when exposed to an average force of 425 kp, the range extending between 290 and 530 kp. Similarly, Lin et al. (1978) reported failure loads ranging between 1560 and 5800 N for 19 fresh lumbar intervertebral joints, although, questions may be raised as to the reliability of his findings given that his tests followed a series of loading experiments.

Brinckmann et al. (1987) undertook a preliminary study on 23 motion segments and found a strong correlation between the ultimate compressive strength and the product of bone density and endplate area. As Brinckmann et al. (1987) stated, if such a relationship can be substantiated then there is the potential for making *in-vivo* estimates of compressive strength by utilising such techniques as computer tomography and magnetic resonance imaging.

Age, sex, disc level and degree of disc degeneration have all been studied with regard to their influence on the mechanical properties of the motion segment. Nachemson et al. (1979) subjected 42 fresh cadaver lumbar motion segments to shear forces of up to 205 N, and bending and torsion moments up to 205 Nm, on top of a 400 N static preload. Age and

disc level did not significantly affect the mechanical behaviour of the motion segment. Female segments were found to be more flexible than males in response to bending moments and compressive loads. Discs which were grossly degenerated deformed more in compression and less in flexion-extension.

Under static axial loading the rate of disc height loss for degenerated disc has been shown to be greater than that for non-degenerated discs (Kazarian, 1975). Koeller et al. (1986) identified three distinct age ranges in which disc mechanical properties were seen to alter. They subjected 178 disc specimens (96 lumbar, 81 thoracic), aged between 5 to 84 years, to axial dynamic compression loads of 950±540 N. From the first to the middle of the third decade, the lumbar disc creep response was seen to decrease, whereas from the middle of the third decade to the beginning of the sixth decade biomechanical behaviour remained unaltered. After the sixth decade there was an increase in creep response. For all ages and lumbar discs, axial deformability was independent of age. The degenerative state of the lumbar disc was found to be linearly related to an increase in age, whereas the mean water content of the nucleus decreased significantly with age. Paradoxically, their findings implied that discs behave most efficiently within the age ranges that coincide with the period of life that has the highest reported incidences of back pain.

Several *in-vitro* studies have attempted to mimic injuries to the disc in order to investigate the effects on the mechanical functioning of the spinal motion segment. The work by Adams and Hutton (1982) suggested that a single high compressive load applied to a hyperflexed disc can result in a sudden disc prolapse. However, it is possible to criticise their work given that disc failure often occurred beyond the normal physiological range of the motion segment. A gradual disc prolapse was also shown to occur when a slightly flexed motion segment was subjected to a period of cyclic compressive loading.

Markolf and Morris (1974) examined the responses of 24 vertebral specimens to compressive loading following the injection of saline solution into the discs, puncturing of the discs through the annular wall, removal of the disc's nucleus alone, and removal of the disc nucleus along with central end-plates and supporting bone. The injected

saline caused axial distension (0.5 to 2.5 mm) with no increase in disc bulge and no significant alteration in compressive behaviour. Following injury to the annulus, mechanical behavioural characteristics were quickly restored. An initial load cycle appeared to cause an extrusion of nuclear material through the lesion which led to its subsequent sealing. Injection of a solution into the sealed disc revealed a water-tight system had been formed. The removal of the nucleus and annular wall did not appear to influence the disc's compressive characteristics and it was concluded that the annulus acts as the major structure towards providing the disc with its mechanical properties.

These findings were not supported by Panjabi et al. (1984) who observed a marked alteration in the mechanical functioning of the disc following annular injury and nucleus removal. In addition, no self-sealing mechanisms were seen to take place. These contrasting findings were felt to result from the method by which Panjabi et al. (1984) sought to gauge functional performance, namely three-dimensional flexibility and creep, as opposed to pure axial deformation.

Brinckmann (1986) attempted to replicate an internal division of the annulus, leaving only a 1 mm thick peripheral layer, to simulate the effects of a traumatic disc failure or degenerative process. Disc bulge at the site of injury was measured under axial loading, within mechanical load ranges, and after axial compressive fractures and intra-discal injections. Application of a 1000 N load caused a bulge of less than 0.5 mm and following fracture, there was no increase of disc bulge beyond 1 mm. In all instances there was never any extrusion of disc material. These findings suggested that a disruption of the annulus was not sufficient to produce a clinically relevant disc herniation.

The extent to which observations from *in-vitro* studies can be extrapolated to predict failures in the *in-vivo* situation is open to debate. Individual differences in bone material storage, test conditions, removal of important musculature and bone structure, can all significantly influence the biomechanical properties of the motion segment. Brinckmann et al. (1988) stated that a major problem with the majority of *in-vitro* motion segment tests are that they are not performed with adjacent discs present and forces are applied direct to the vertebrae. They felt that this could severely influence the

mechanism of failure and stress distribution.

Brinckmann et al. (1988) have performed *in-vitro* experiments which they claimed more closely represent *in-vivo* situations and can be used to predict the probability of a fatigue fracture in living subjects. A number of cyclic loading tests were carried out on lumbar vertebrae from which a probability table for fatigue fracture was established. The table accounted for loads cycles of 10, 100, 500, 1000 and 5000 N and different relative loads between 20 and 70%. For example, a lumbar vertebra with compressive strength of 6 kN when subjected to a cyclic load of 2 kN (equivalent to a moderate lifting exercise) was estimated to have an 18% chance of a fatigue failure after 500 load cycles. Following 5000 repetitions the probability was found to rise to 27%. It was stressed that when evaluating such findings the load cycles must have accumulated in a short time interval. Longer time intervals allow natural bone repair mechanisms to take place and can therefore, significantly enhance the number of load cycles that can be tolerated. However, they felt this was not a major limitation considering that a majority of activities, such as marathon running, accumulate 5000 load cycles in a relatively short time span. A more influential limitation could well stem from the fact that a majority of activities involve loading cycles which are often intermittent and have a continually changing force amplitude.

In-vitro studies have provided some important clues as to how spinal structures react to mechanical forces and the possible mechanisms of spinal injury. However, the major criticism is that such studies cannot be related to *in-vivo* situations. For this reason, researchers have sought to find methods by which to estimate *in-vivo* spinal load responses.

3.4.2 Measures Adopted to Study *In-Vivo* Spinal Loads

As the literature would suggest, injuries to the intervertebral disc are a major cause of LBP symptoms. Taking this into consideration, in conjunction with the epidemiological data identifying spinal loading as a predominant risk factor associated with LBP, the *in-vivo* techniques employed to investigate spinal disorders have focussed on the measurement of lumbar disc loads. Direct and indirect techniques have been implemented with varying degrees of success. These have sought to

measure intervertebral disc pressure, intra-abdominal pressure, spinal shrinkage, trunk muscle activity (electromyography), and compressive disc forces estimated from biomechanical models.

Intra-discal pressure measurements were pioneered by Nachemson (1959) to directly measure pressure within the actual disc of living subjects. A specially constructed hollow needle was connected to an electromanometer and was sensitive to pressure changes. Insertion of the needle into the nucleus of the intervertebral disc allowed the measurement of disc pressure, which due to the hydrostatic conditions within the disc, facilitated estimates of spinal loads. *In-vivo* investigations have centred on the influence of various tasks and working positions on disc loads while standing, sitting, and lifting (Nachemson 1966, 1976; Andersson et al., 1976, 1977, 1978). From these studies it has been possible to state that one particular working posture is more taxing for the spine than another. For example, disc pressure when standing was shown to be about 40% higher than when lying supine and increased a further 40% when sitting (Nachemson and Elfström, 1970). Appropriate backrests and arm supports were found to reduce the pressure values when sitting.

The instrument has been used on cadaveric specimens and it was shown that normal discs exert pressure of 1.3 to 1.6 times the vertical load per unit area. In addition, spinal compression was the major determinant of disc pressure and there was a linear increase in pressure for external loads up to 2000 N (Nachemson, 1960).

Because of the technique's inherent ethical limitations, it is restricted to a laboratory based environment. Consequently, only a limited range of activities (particularly dynamic) can be studied, and only selective subjects (ie, those without degenerated discs) can be experimented on. Thus, the method has received limited application throughout the literature.

For a number of years intra-abdominal pressure (IAP) has been reported to play an important role in the functional mechanics of the spine (Davis, 1956). It is proposed that under certain circumstances pressure rises inside the abdomen serve to reduce compression loads on the spine. Certain activities cause contraction of the diaphragm and muscles of the

abdominal wall, principally the transversus abdominis and the internal and external obliques. This serves to contract the abdominal contents into a semi-rigid mass, thereby making the abdominal cavity a semi-rigid cylinder (Morris et al., 1961). Consequently, an upward force acts upon the diaphragm and a downward force on the pelvic floor, indirectly causing a pressure on the diaphragm. The resulting extensor torque will ultimately reduce stress on the extensor tissue and alleviate the compressive load on the spinal components.

The theoretical justification of IAP as a relieving mechanism, in conjunction with observed increases in IAP during lifting activities (Bartelink, 1957; Davis, 1981), led many researchers to propose that spinal stress could be measured indirectly from IAP changes. The devices employed to record these pressure changes have ranged from indwelling catheters to implanted balloons connected to pressure tambours (Davis, 1985). However, the technique which has been found to be most simple, accurate and place only a limited restriction on an individual or his activity is the radio pressure pill. This is swallowed by the subject and once inside the stomach is sensitive to pressure changes.

The strong correlations between intradiscal pressure and applied load *in-vitro* and calculated load *in-vivo* (Nachemson and Elfstrom, 1970), along with the finding that IAP correlated well with intradiscal pressure (Andersson et al., 1977; Davis, 1981), supported the use of IAP as an indicator of spinal stress. However, more recent studies have questionned the validity of such an assumption. Schultz et al. (1982a) found that simultaneously measured IAP and intradiscal pressure were poorly correlated (r=0.36) during seated tasks. Furthermore, the recorded IAP values and estimated compression values from a biomechanical model possessed a low correlation (r=0.24).

Similarly, Troup et al. (1983) compared calculated lumbar compression values and IAP during different lifting and lowering techniques and found no consistency. They reported a large variation in peak IAP values with standard deviations being up to 86% of the mean. In general, the back lifts, in which the trunk underwent greater flexion, produced the lowest IAP values when compared with erect trunk postures. Thus, IAP appears to be influenced by trunk posture. This finding would

also seem to support the proposal that IAP is at a greater mechanical advantage when the spine is flexed as the moment arm it possesses is larger than normal (Davis and Troup, 1965). Mairiaux and Malchaire (1988) have added further support to the argument by stating that IAP cannot be used as an index of biomechanical stress unless the individual IAP to lumbar moment relationship is considered.

The associated limitations with both intradiscal and intra-abdominal pressure measurements have limited their application as a spinal stress indicator. Researchers have, therefore, tended to focus their attention on electromyography (EMG) and biomechanical modelling, and more recently, measurements of alterations in stature as a means to determine loads placed upon the spine. The next section of this review will be devoted to these techniques.

3.4.3 Spinal Shrinkage as a Measure of Spinal Load

The component tissues that contribute towards the length of the human body include three primary structures:- the length of the bones, the cartilages between the bones and the soft tissues covering the bones. They are all elastic to a varying degree. Soft tissue covering bone, such as that covering the scalp and soles of the feet, responds promptly to external forces and reaches a stable level of compression in a short period of time. Under constant body weight, the heel pad thickness equilibrates within a 2 min period (Foreman and Linge, 1989). Thus, to a large extent it is possible to control the response of soft tissue to loading. By contrast, joint cartilage, particularly the intervertebral joint cartilage, responds in a visco-elastic manner which is dependent upon the load and period of application, as discussed in section 3.4.1.

As 24 intervertebral discs make up the spinal column, accounting for some 25% of its total length, and as few other joint cartilages contribute towards the length of the human body, it is justifiable to assume that any alterations in bodily stature are a direct result of spinal column alterations, and by inference spinal loading. It is also likely that the rate of stature loss (shrinkage) is influenced by the type of static or dynamic loading during the course of the day and that stature recovery will occur whenever the person adopts a posture that unloads the spine.

According to Fitzgerald (1972), prerequisites for measuring the height of an erect body as a reflection of alterations in disc height include:

a) a constant individual posture during and between subsequent measures;

b) a measuring apparatus sensitive enough to record minute changes in stature to within 1 mm.

It is only recently that an accurate repeatable method of measuring changes in stature has been developed (Eklund and Corlett, 1984). Although DePuky (1935) presented literature describing changes in the standing height of individuals and its relation to intervertebral disc loading, he presented no explanation of any experimental protocol. Methods outlined by Stricklands and Shearin (1972), Goode and Theodore (1983) and Pope and Klingenstierna (1986) possess sensitivity values and reproducibility errors outside acceptable limits. The reliability claimed by Fitzgerald (1972) could not be reproduced and improvements were necessary to achieve the repeatability that was needed (Eklund and Corlett, 1984; Tyrrell et al., 1985).

A device for measuring stature (stadiometer) which was capable of maintaining a standard reproducible posture and recording small changes in stature was reported by Eklund and Corlett (1984). Repeated observations (1110) on 15 subjects were found to exhibit a standard deviation (SD) of 0.63 mm. Tyrrell et al. (1985), using a similar technique on 8 young male adults, reported repeatability with a SD of 0.28 mm and maximum range within 1 mm, when 10 consecutive height measurements were conducted. Similarly, Leatt et al. (1986) found that 10 consecutive measures had SD values of 0.36 mm and 0.33 mm for two groups of 9 and 7 subjects, respectively.

To achieve these levels of accuracy it was found that two important methodological requirements had to be fulfilled. Firstly, subjects had to participate in a training period to familiarise themselves with equipment and procedures. Secondly, subjects had to be in a normal, relaxed standing position which could be easily maintained and reproduced. To allow subjects to relax, it was found that the apparatus needed to be inclined backwards from the vertical by a few degrees.

Several factors were found to require constant monitoring and control if spinal posture was to be precisely reproduced. These included:

(i) head and limb positions;

(ii) height and horizontal projections of the supports for the spine;

(iii) angle of the head in the sagittal plane;

(iv) position and curvature of spinal curves;

(v) weight distribution between heels and forefeet;

(vi) the phase in the repiratory cycle.

The stadiometer developed by Eklund and Corlett (1984) consisted of a central pillar set at right angles to a base plate and inclined backwards by 10 degrees from the vertical. This angle of inclination has varied in other studies from 5 (Reilly et al., 1984) to 15 degrees (Corlett and Eklund, 1986). On the base plate, weighing scales and a heel rest were positioned so that weight distribution in a relaxed position could be standardised. The central pillar supported a back-location frame consisting of adjustable rods, a few cm square, which contacted the prominent points and curves of the spine (ie, sacrum, lumbar, thoracic and cervical regions of the spine, and the back of the head). Head angle in the sagittal plane was controlled by the subjects themselves using spectacles marked with lines on the side of the frames. Repositioning the same preferred horizontal line in a mirror in front of the subjects ensured the same head angle between measures. The measuring head, a wooden disc 15 cm in a diameter, rested on top of the subject's head. A vertical rod attached to the disc and running on linear bearings activated a transducer (accurate to within 0.1 mm) which, via a UV recorder, measured the vertical displacement of the head disc.

In recent years advancements have been made to the basic design but the principal structure and procedures have remained the same. Wilby et al. (1987) fixed an infra-red emitter to the spectacles which was matched to an infra-red receiver mounted on an adjustable box directly in front of

the subject. This sought to improve the accuracy of head alignment in the sagittal plane. Also mounted on the box were LED's associated with microswitches attached to the ends of the adjustable rods, weighing-scales and spectacle alignment system. Thus, subjects were now able to obtain a visual feedback of the required posture. The measuring head was replaced by a 15 cm plastic disc connected to a Mercer dial gauge with a precision of 0.01 mm. Two strain gauges were mounted back-to-back on a spring riding which rested on top of the head disc. Vertical displacements were monitored by a chart recorder which could be calibrated using the dial gauge.

Further adaptations have been made to the head measuring and alignment system with Corlett and Eklund (1986) replacing the solid head disc with five cylindrical pins (1 mm in diameter and 17 mm in length) in order to penetrate the hair and rest directly on the scalp. To improve head location, a head support was introduced which consisted of two wooden plates (12 cm by 10 cm) fixed vertically in a 'V' with 90 degrees between each. Bonney (1988) has recently implemented a nose pointer to replace the spectacles and align the head.

No protocol has yet established the necessary time period required to train a subject in using the apparatus. Eklund and Corlett (1984) suggested about 20 min, whereas Leatt et al. (1986) and Wilby et al. (1987) used a criterion of 10 successive measures with a SD of less than 0.5 mm.

The development of the stadiometer as a device for measuring changes in stature has progressed to a level where sensitivity is high and repeatability error is low. The information being gathered on stature alterations is now providing valuable insight into intervertebral disc responses to loading.

3.4.3.1 Circadian and Diurnal Changes in Stature

A diurnal variation in stature was first reported by DePuky (1935) whose studies on over 1200 Bucharest residents, aged between 5 and 90 years, found a loss in stature during the course of a day. From the time of rising in the morning to retiring to bed at night males and females were reported to lose an average of 17.1 mm and 14.2 mm respectively (1.1 and 0.9% of overall height, respectively). DePuky (1935) attributed the

stature loss to intervertebral disc loading from upper body weight brought about by the person adopting an upright posture. An unloading of the spine was thought to take place at night when the person adopted a supine posture allowing the influx of fluid into the discs and consequent regain in disc height.

DePuky (1935) also identified age as being a variable influencing height losses, the youngest age group showing the largest daily oscillations, with the value steadily decreasing with increasing age. This was felt to stem from a decrease in overall disc height with age, possibly reflecting disc degeneration.

Reilly et al. (1984) confirmed the findings of DePuky (1935) and showed a circadian variation in adult male body height of 1.1% (19.3 mm) of overall stature. Approximately 50% of the total stature loss occurred within the first hour after rising and as long as the spinal loading remained constant, the rate of shrinkage decreased and became relatively asymptotic by nightfall. After retiring to bed there was a rapid recovery in stature, 70% being regained within the first four hours. Several other studies (Fitzgerald, 1972; Stricklands and Shearin, 1972; Leatt et al., 1986; Wilby et al., 1987) have all observed circadian variations in stature. For males, values range between 0.83% (14.4 mm) to 1.1% (19.3 mm) of overall height (refer to Table 1). Wilby et al. (1987) and DePuky (1935) provided the only values relating to female subjects, 0.92% (15.4 mm) and 0.88% (14.2 mm) of stature, respectively.

Clearly, the circadian and diurnal changes in stature reported by these authors depends on constant loading from 'normal' daily activities. Physical loading and unloading of the spine from various forms of exercise or activity at different periods during the day have been shown to influence the rate of stature loss.

3.4.3.2 Static Loading and Shrinkage

Fitzgerald (1972) was one of the earliest investigators to implement shrinkage as a method to study postural loads. He monitored the effects of an externally applied shoulder weight on body height changes of young adult males. Results showed that a shoulder load in excess of that normally experienced caused an increase in the rate of stature loss. The amount of shrinkage was also shown to be related to the magnitude of

load; 9.98 kg caused an average height loss of 10.94 mm compared to 6.33 mm for a 2.27 kg, for an identical time period. Tyrrell et al. (1985) observed a similar effect when 8 young males were required to hold 10, 20, 30 and 40 kg barbells across the shoulders and carry 2.5 and 10 kg rucksacks for 20 min. The 40 kg barbell caused an 11.2 mm loss in stature as against 5.14 mm for a 10 kg barbell. A similar loss in stature for the 10 kg barbell and 10 kg rucksack (5.45 mm) suggested that load was the important variable governing height loss. Corlett and Eklund (1986) found a linear relationship between shoulder load (0 to 25 kg) and stature changes. The regression equation governing this relationship was:

$$y = -0.093x - 1.68$$

where y = stature change (mm)
x = shoulder load (kg)

with a standard deviation of 0.17 mm.

3.4.3.3 Dynamic Loading and Shrinkage

Tyrrell et al. (1985) studied the influence of dynamic loading exercises using the 'dead lift' weight lifting technique with 10 and 40 kg loads. The exercise was repeated 12 times per minute for 20 min. Again, load had an effect on shrinkage, 7.59 mm greater shrinkage stemming from the 40 kg load. On comparing static loading against repetitive lifting, it was shown that repetitive lifting led to greater shrinkage and the difference was more pronounced the greater the load. Vincent et al. (1987) examined a repetitive lifting task involving a two second lift, three second hold, one second lower and four second rest, for a total of 30 min. Load was measured in terms of a lumbar torque and ranged between 0 and 57.74 Nm, the two extremes of load resulting in 3.85 and 7.26 mm shrinkage, respectively.

Stålhammar et al. (1989) monitored changes in stature during 30 min of fixed paced (4 lifts per min) and self-paced repetitive lifting. Ten subjects (5 females, 5 males) lifted a box with handles from a 10 cm shelf to knuckle height, the weight of the box being determined by the 'rating of acceptable load'. Neither sex nor lifting rate was found to influence the rate of stature loss.

3.4.3.4 Stature Changes During Recovery and Unloading Postures

Following the repetitive lifting tasks, Stålhammer et al. (1989) instructed subjects to rest for 30 min in the Fowler position, a standard orthopaedic posture designed to unload the spine, while stature measurements were taken 5, 15 and 30 min into this rest period. At the end of 30 min, following the self-paced lifting exercise, there was a regain in stature of 81% (males) and 90% (females) of that which had been lost during exercise. Following the fixed paced lifting exercise the total regain was 76% and 105% for males and females, respectively. Significantly, on both occasions the greatest regains in height occurred within the first 5 min of the recovery period. Thus, it was suggested that frequent rest periods should be encouraged during repetitive lifting tasks.

Tyrrell et al. (1985) also monitored recovery directly following their experiments involving static and dynamic loading. For a 10 min standing rest period, following 20 min of various static shoulder loads, the regains in stature increased with respect to the amount of shrinkage that had occurred during exercise. Thus, a 40 kg barbell not only caused the greatest shrinkage but the regain in height directly following this activity was also greater. A comparison of standing recovery against recovery in the Fowler position showed that the latter is more effective in promoting disc height recovery (Tyrrell et al., 1985).

Another procedure designed to promote a rapid disc height recovery is that of gravity inversion. This involves a person being inverted in such a manner that the forces of gravity are utilised to extend the spine. Such forms of postures, along with various forms of traction, have been advocated for athletes and non-athletes alike, in an effort to prevent and relieve back pain symptoms (Temple, 1984). By encouraging an expansion of the discs it is felt that any nerve root entrapment will be lessened and the shock absorbing and functional ability of the motion segments will be improved.

Leatt et al. (1985), using changes in stature as a measure of spinal loading, compared the Fowler position with 50, 70 and 90 degrees (with respect to the horizontal) gravity inversion. As expected, gravity

inversion invoked the greater regains in stature when compared with the Fowler position. When the angles of inversion were compared it was found that 50 degrees inversion caused a greater recovery in stature than either 70 or 90 degrees inversion. Considering that an angle of 90 degrees should induce the largest force (gravity) component, there would appear to be some other factor influencing the unloading process. Ratings of perceived discomfort provided a possible explanation in that subjects found 50 degrees inversion to be more relaxing and might therefore, have created less muscular tension in the trunk to oppose unloading.

Bridger et al. (1990) conducted a study which served to compare conventional traction at 1/3 body weight using pelvic and thoracic harnessess against lying supine with the knees flexed (90 degrees) and feet supported. Measurements of changes in stature were used not only to compare procedures but also to investigate the effects of duration by conducting measurements after 5, 15 and 25 min of traction. Traction was found always to cause increases in stature, 8.94 mm and 3.33 mm for 25 min lumbar traction and supine lying, respectively. Stature increases under traction were shown to be time-dependent, the greatest changes occurring within the first 15 min.

3.4.3.5 Shrinkage Resulting From Different Forms of Physical Exercise

Other forms of exercise which have been found to effect spinal loading include weight-training and running. Leatt et al. (1986) measured stature prior to and following two weight-training circuits of 9 exercises lasting an average of 24.6 min. The exercises caused a loss in stature of 5.4 mm which was less than the value reported by Tyrrell et al. (1985) for the repetitive lifting of a 10 kg barbell. Although the exercises were chosen to load the spine directly or involve back muscles, the very dynamic nature of some of the exercises may well have promoted unloading. This, as well as the rest breaks between exercises, could provide a possible explanation for the reduced shrinkage.

Wilby et al. (1987) also studied circuit-weight training employing 10 females to repeat sequences of 8 exercises lasting 20 min. The study investigated the influence of diurnal changes in stature on the magnitude of exercise-induced shrinkage by conducting exercise periods immediately after rising and at 22:00 h. A significantly greater

($p<0.001$) mean loss in stature occurred in the morning (5.4 mm) than in the evening (4.33 mm). This supported their initial hypothesis that a loss of disc height from a given load would be reduced when body height is in the trough of circadian variation. Consequently, they concluded that heavy exercise conducted towards the later part of a day offers an increased threat of spinal injury. A 20 min recovery in the Fowler position, directly following both the morning and evening exercise periods caused increases in stature, the former being the greater (4.5 mm and 3.4 mm respectively) in an absolute sense but similar with respect to previous losses (83% and 79% respectively). Interestingly, they found that measures of isometric back strength were inversely related to the magnitude of shrinkage ($r=0.60$, $p<0.05$). It was suggested that the musculature of the back may in some way be better able to compensate for disc height losses in the evening than in the morning. This would correspond to a circadian rhythm in muscular strength whose peak is close to that of the body temperature curve (Reilly, 1990).

Leatt et al. (1986) conducted experiments to investigate the influence of running on spinal loading and whether running experience affects the rate of shrinkage. They studied two groups of runners, experienced and inexperienced, running on a treadmill at a speed of 12.2 km/h for a distance of 6 km. The exercise caused no significant difference in the amount of shrinkage, although the novices did tend to lose slightly more height, 3.25 mm as opposed to 2.35 mm. Making comparisons of this nature are difficult as running speed was fixed and may well have indirectly altered running technique. The experienced runners ran for a further 19 km at 14.6 km/h and lost an additional 7.79 mm, indicating that the duration of the run was a dominant factor in loading the spine.

3.4.3.6 Stature Change Following Seated Work Tasks

As an ergonomic tool for the assessment of work-related stress Eklund and Corlett (1984) implemented spinal shrinkage to assess seat design. A comparison of 1.5 h sitting on a stool without back support, on a office chair with support and in an easy chair inclined at 110 degrees, showed that chair design has a significant effect on spinal loading. The stool caused 4.6 mm shrinkage, whereas the other two chairs had only a negligible effect on stature, suggesting that the normal diurnal

shrinkage had been halted. These results were found to be in agreement with the disc pressure measurements reported in the literature (Nachemson, 1981).

Further studies of seated work (Eklund and Corlett, 1987) not only examined the influence of chair design on spinal loading but also the effects of work postures and task requirements. Whilst seated in chairs with a 38 cm and 18 cm high back support, 8 subjects were required to perform a pushing task (25 N force) using both hands for a period of 45 min. Significantly less shrinkage resulted from sitting in the high back rest than when sitting in the chair with lumbar support, 0.66 and 1.37 mm respectively ($p<0.05$). When a task which involved the operator rotating the head 90% to one side was studied, the low back rest (30 cm) was found to induced less shrinkage than the high back rest (47 mm), although the difference was not significant (0.88 mm and 1.44 mm, respectively). The authors suggested that in such a situation the low back rest offered less of an impedance to upper body rotation and therefore, imposed less stress on the spine.

Within recent years the design of chairs has undergone some dramatic changes in attempts to relieve many work-related back pain symptoms. The introduction of the sit-stand chair and knee support chair are typical examples. These chairs reflect the ideology that the prolonged static flexed lumbar curve during seated work is a major underlying factor leading to injury (McKenzie, 1982).

Results of a field study of three office chairs by Ericson and Goldie (1989) suggested that the knee support chair did in fact induce higher spinal loads than the more conventional chairs. Spinal shrinkage values were 3.1 mm for the knee support chair compared to 1.3 mm for a conventional office chair. Higher muscular loading and increased discomfort ratings were the suggested reasons for the increased loading. In a study of an assembly task with restricted knee room (Eklund and Corlett, 1987) a sit-stand chair was found to be more favourable than the normal conventional chair as it induced significantly less shrinkage (0.93 mm and 2.41 mm, respectively).

3.4.3.7 Stature Change Following Seated Vibration

Bonney (1988) has also implemented shrinkage measurements to assess both seat design and vibration during simulated driving. For a 30 min period there was found to be no significant alterations in stature at vibrational frequencies of 0, 4 and 8 Hz for either car or lorry driving positions. However, a back rest position of 110 degrees allowed an increase in stature (0.81 mm) as opposed to a decrease in stature for a seated posture with a 90 degrees back rest, the difference being significant. To study the influence of sinusoidal vibration further, Bonney (1988) extended the exposure time and introduced an additional vibrational frequency of 6 Hz, whilst back rest position was maintained at 90 degrees. The static posture caused a decrease in stature (1.1 mm), whereas both 6 and 8 Hz appeared to prevent any stature loss and 4 Hz significantly increased stature (1.85 mm). The apparent unloading effect from vibration was felt to stem from forces created by internal organs. At 4 Hz the spine and internal organs are close to their natural frequency and therefore, it was suggested that the reaction of the internal organs acting on the spine creates an upward force which exceeds the natural loading.

In contrast to the findings of Bonney (1988), Sullivan and McGill (1990) found increased shrinkage (9 mm) following 30 min seat vibration at a frequency of 5 Hz (peak-to-peak amplitude of 3 mm and peak acceleration less than 2 $m.s^{-2}$) in comparison to a control group (1 mm). Interestingly, they showed vibrational exposure to have an influence on the normal diurnal shrinkage response; significantly reduced shrinkage (3.6 mm) was recorded at the end of the day when subjects had been exposed to vibration than when no vibration exposure had taken place (10.6 mm). The authors hypothesised that the vibration may have induced an inflammatory response within the spine encouraging the uptake of fluid by the disc. Criticism can be directed at the technique which Sullivan and McGill (1990) employed to measure changes in spinal length, as no control of spinal curvature was undertaken during measurements.

3.4.3.8 Nursing Activities and Shrinkage

Foreman and Troup (1987) used shrinkage to study the workload imposed during nursing. Twelve nurses from different hospital wards were measured during the course of an early shift, late shift and day off. In addition, a permanent time-based record of the frequency and duration

of each posture and activity was recorded during the working periods. Less stature was found to be lost during the late shift (9.8 mm, 11:30-20.30 h) than the early shift (10.2 mm, 07.45-16.30 h), although the difference was not significant. During the 12.5 h period of their day off the total height loss was less than that which had occurred during the early shift (8.75 h), in 8 out of 10 subjects, and in 7 out of 10 subjects for the late shift (9 h). When analysing the postures for each shift it was found that the duration for which the spine was off-loaded was inversely related to losses in stature, particularly for the early shift ($p<0.01$). Shrinkage was also found to be directly related to the duration of lean/stoop and lifting during the early shift.

3.4.4.9 Shrinkage and Other Variables

Relationships with changes of stature have also been found when comparing ratings of discomfort for static shoulder loading (Troup et al., 1985), and perceived exertion following running, circuit-weight training and repetitive lifting (Troup et al., 1985; Wilby et al., 1987; Vincent et al., 1987). Application of the procedure as a diagnostic tool for screening possible back pain patients has been studied by Hindle et al. (1987). They found that diurnal variations in stature for a group of patients diagnosed with ankylosing spondylitis was significantly less than that for a group of normal patients ($p<0.01$). This was attributed to osseous changes within the intervertebral discs and spinal ligaments reducing the normal functioning of the motion segment. Reduced diurnal changes in patients with only a short case history of the disease suggested that this method could be used as an early screening procedure.

Table 1 provides a summary of the experimental findings which have been discussed in this section and have focussed on measurements of stature as a method of studying spinal loading. Comparisons must be undertaken with care as studies were often performed at differing times of day.

3.4.4 Electromyography as a Predictor of Spinal Load

3.4.4.1 Electrical Muscle Activity and its Detection

Electromyography (EMG) utilises electrodes, intramuscular or superficial, to detect the level of electrical activity of a contracting muscle and is, therefore, a study of muscle function (Basmajian and DeLuca, 1985). The electrical activity stems from the depolarisation

able 1. Summary of studies which have measured changes in stature as an index of spinal loading

INVESTIGATORS	SUBJECTS	ACTIVITY	STATURE CHANGE (mm) (+ Denotes increase) (- Denotes decrease)		
DePuky 1935	1216 (5-90 yr)	Circadian variation	17.1 14.2	(1.16%)* (0.88%)	M+ F
Fitzgerald 1972	52 (20-68 yr)	Circadian variation	15.6		M
Strickland et al. 1972	100 (3-14 yr)	Circadian variation in the young	15.4		M & F
Reilly et al. 1984	8 (19-21 yr)	Circadian variation	19.3	(1.1%)	M
Leatt et al. 1986	11 ($\bar{x}$ = 20.3 yr)	Circadian variation	14.4	(0.83%)	M
Wilby et al. 1987	10 (20-30 yr) ($\bar{x}$ = 22.2 yr)	Circadian variation	15.4	(0.92%)	F
Hindle et al. 1987	40 Normals = 18 AS = 22	Diurnal variation (50 min after rising) for normals Ankylosing spondylitis (AS) group	 -11.44 -5.23	 (0.68%) (0.34%)	 M & F M & F
Foreman & Troup 1987	11 (19-24 yr)	Diurnal variation of nurses Day off (12 h 30 min) Early shift (8 h 45 min) Late shift (9 h)	 -8.14 -10.2 -9.8		
Sullivan & McGill 1990	5 (23-25 yr)	Diurnal variation (within 3 h after rising) Control Day Day with 30 min seated vibration exposure(5 H_Z)	 -10.6 - 3.6		 M M
Fitzgerald 1972	32 (20-23 yr)	20 min static shoulder load 2.27 kg 9.98 kg	 -6.33 -10.94		 M M
Tyrrell et al. 1985	8 (19-21 yr)	20 min static shoulder load 2.5 kg rucksack 10 kg rucksack 10 kg barbell 20 kg barbell 30 kg barbell 40 kg barbell	 -3.87 -5.45 -5.14 -7.11 -9.42 -11.22		 M M M M M M
Tyrrell et al. 1985	8 (19-21 yr)	10 min standing following static shoulder load 2.5 kg rucksack 10 kg rucksack 10 kg barbell 20 kg barbell 30 kg barbell 40 kg barbell	 +2.94 +4.23 +3.9 +5.24 +7.37 +8.9		 -M -M -M -M -M -M

Percentage values denote change with respect to overall height
M - Males F - Females

Table 1 (Continued).

INVESTIGATORS	SUBJECTS	ACTIVITY	STATURE CHANGE (mm) (+ Denotes increase) (- Denotes decrease)	
Corlett & Eklund 1986	8 (18-34 yr) ($\bar{x}$ = 30 yr)	45 min static shoulder load		
		0 kg	-1.8	M & F
		25 kg	-4.1	M & F
Tyrrell et al. 1985	8 (19-21 yr)	20 min repetitive lifting (crouch to knuckle height -12 lifts/min)		
		10 kg barbell	-6.90	M
		40 kg barbell	-14.49	M
		10 min standing following repetitive lifting		
		10 kg barbell	+4.04	M
		40 kg barbell	+8.37	M
		20 min static shoulder load and 20 min repetitive lifting (12 lifts/min)		
		(static) 10 kg barbell	-5.35	M
		(repetitive) 10 kg barbell	-6.85	M
		10 min recovery in Fowler position following static & repetitive loads		
		(static) 10 kg barbell	+6.82	M
		(repetitive) 10 kg barbell	+7.62	M
Vincent et al. 1987	8 (22-43 yr)	30 min repetitive lifting (300 mm below shoulder height to shoulder height 6 lifts/min)		
Fitzgerald 1972	32 (20-23 yr) ($\bar{x}$ = 30.3 yr)	57.74 Nm lumbar torque	-7.26	M
		43.3 Nm lumbar torque	-5.59	M
		28.87 Nm lumbar torque	-4.76	M
		0 Nm	-3.85	M
Stålhammar et al. 1989	10	30 min repetitive lifting rating of acceptable load, 10 cm shelf to knuckle height)		
	(M 35-43 yr) ($\bar{x}$ = 38 yr)	Self paced	-5.8	M
			-5.1	F
	(F 29-49 yr) ($\bar{x}$ = 37 yr)	Fixed paced (4 lifts/min)	-6.8	M
			-5.8	F
		30 min rest following repetitive lifting in Fowler position		
		Self paced	+4.7	M
			+4.4	F
		Fixed paced	+5.2	M
			+6.1	F

Table 1 (Continued).

INVESTIGATORS	SUBJECTS	ACTIVITY	STATURE CHANGE (mm) (+ Denotes increase) (- Denotes decrease)	
Leatt et al. 1986	10 ($\bar{x}$ = 19.8 yr)	Circuit-weight training (9 exercises repeated twice, mean time 24.6 min)	-5.4	M
		20 min standing recovery following circuit-weight training	-0.7	M
Wilby et al. 1987	10 (20-30 yr) ($\bar{x}$ = 22.2 yr)	Circuit-weight training (20 min, 8 exercises)		
		at 7:30 h	-5.4	F
		at 22:00 h	-4.3	F
Leatt et al. 1986	9 Novices ($\bar{x}$ = 19.9 yr) 7 experienced ($\bar{x}$ = 31 yr)	Running (Inexperienced & experienced athletes)		
		6 km(12.2 km/h) Novices	-3.25	M
		6 km(12.2 km/h) Experienced	-2.35	M
		6 km(12.2 km/h) + 19 km (14.7 km/h) Experienced	-10.15	M
		20 min standing recovery following running exercises		
		6 km Novices	+0.75	M
		25 km Experienced	+1.45	M
Leatt et al. 1985	10 (19-25 yr)	30 min posture unloading		
		Gravity inversion 50°	+5.57	M
		70°	+4.39	M
		90°	+4.57	M
		Fowler position	+3.58	M
		20 min standing following unloading postures		
		Gravity inversion 50°	-4.97	M
		70°	-3.4	M
		90°	-4.21	M
		Fowler position	-3.71	M
Bridger et al. 1990	10 (20-31 yr) ($\bar{x}$ = 20.4 yr)	25 min		
		Traction - 1/3 body weight	+8.94	M & F
		Control - Crook lying	+3.33	M & F

Table 1 (Continued).

INVESTIGATORS	SUBJECTS	ACTIVITY	STATURE CHANGE (mm) (+ Denotes increases) (- Denotes decreased)	
Eklund & Corlett 1984	3	1½ h sitting		
		Stool without back support	-4.6	M & F
		Office chair with back support	-0.1	M & F
		Easy chair inclined at 110°	+0.4	M & F
	15 (22-39 yr)	1 h sitting		
		14 kg shoulder load	-3.1	M & F
		0 kg shoulder load	-1.4	M & F
	2	30 min sitting pushing (25N) with alternate hand		
		with back rest	+1.1	M & F
		without back rest	-1	M & F
		45 min sitting pushing(25N) with alternate hand		
		38 cm high back support	-0.66	
		18 cm high lumber support	-1.37	
Eklund & Corlett 1987	8 (24-42 yr) ($\bar{x}$ = 32 yr)	45 min sitting with restricted knee room (assembly task)		
		Sit stand chair	-0.93	
		Conventional chair	-2.41	
	8 (24-42 yr) ($\bar{x}$ = 34 yr)	45 min sitting with eyes 90° to left		
		High back rest	-1.44	
		Low back rest	-0.88	
	8 (18-61 yr) ($\bar{x}$ = 34 yr)	45 min sitting (Grinding work)		
		High back rest	-2.54	
		Low back rest	-2.81	
		45 min sitting (Punch press work)		
		Conventional seat, low support	+0.19	
		Sit stand seat	+0.48	

Table 1 (Continued).

INVESTIGATORS	SUBJECTS	ACTIVITY	STATURE CHANGE (mm) (+ Denotes increases) (- Denotes decreased)	
Bonney 1988	10	30 min driving simulation		
		8 Hz vibration	+0.14	M
		0 Hz vibration	+0.13	M
		4 Hz vibration	-0.55	M
		Car driving position	-0.44	M
		Lorry driving position	+0.13	M
		110° back rest	+0.81	M
		90° back rest	-1.27	M
	8	1 h driving simulation with 90° back rest		
		0 Hz vibration	-1.1	M
		4 Hz vibration	+1.85	M
		6 Hz vibration	-0.05	M
		8 Hz vibration	-0.1	M
Sullivan & McGill 1990	5 (23-25 yr)	30 min seated vibration (5 Hz, peak-to-peak 3 mm, peak acceleration <2ms^{-2})		
		Control (no vibration)	-1	M
		Vibration	-9	M
Ericson & Goldie 1989	8 (27-40 yr) ($\bar{x}$ = 30.5 yr)	3 h VDU work using 3 different chairs		
		Conventional chairs	-1.3	M & F
		Ullman chair	-1.8	M & F
		Balam chair	-3.1	M & F

and repolarisation of the surface membrane of the muscle fibre (sarcolemma). When a nerve fibre is activated acetylcholine is released at the nerve ending (myoneural junction) which causes a depolarisation of the sarcolemma. A depolarising wave is propagated in both directions along the fibre (DeLuca, 1979), accompanied by a flow of ions. It is these ions which create an electrical field in the vicinity of the muscle which is detectable by a suitably positioned electrode. As muscle contraction involves the stimulation of several muscle fibres at once, the EMG signal is essentially an accumulation of action potentials.

Intramuscular electrodes are preferable to surface electrodes as they are more selective of the electrical signal; however, the associated technical and ethical problems often limit their application. Surface electrodes can be made of any conducting material but most often silver or silver chloride discs are used, as they are cheap, readily available, easy to use and present little discomfort to the subject.

The detection of an EMG signal, particularly from the skin surface, is dependent upon a number of influential factors which makes the reliability and repeatability of a detected signal difficult. These factors include:

(i) conducting properties of the human body between the muscle and skin;

(ii) electrical properties of the skin;

(iii) properties of the detecting electrode;

(iv) placement of the electrode over the muscle mass.

All of these influence the quality of the signal. It is these inconsistencies which form the major criticisms for its use as a reliable predictor of muscle force.

The voltage commonly detected by surface electrodes tends to range between 100 μV and 5 mV (Winter, 1979) which necessitates the use of a specially designed biological amplifier to record the signal. These

design requirements are of importance if a 'clean' signal, free of noise or artefacts and linearly amplified, is to be recorded. Noise can originate from man-made sources, such as power lines (hum) and machinery, and biological sources like ECG signals. Artefacts refer to false signals generated by skin, electrode and cable movements.

The biological amplifier used must take account of:

(i) amplification gain;
(ii) input impedance;
(iii) frequency response;
(iv) common mode rejection.

The amplification gain is dependent upon the signal-to-noise ratio through the system (Mcleod, 1973) and in order to maximise this relationship the maximum amplification that can be tolerated by the rest of the myoelectric recording system should be used (Cabri, 1989). The input impedance (resistance) of the amplifier must be high, at least 10 times greater than the impedance at the source electrode in order to prevent attenuation of the EMG signal (Cabri, 1989). Unfortunately, electrode impedance can not always be determined and so a much higher amplifier impedance than normal is used. However, the greater the impedance the greater is the chance of detecting radiated energy from the main electrical power lines. This is not a problem if a differential amplifier, incorporating a common mode rejection is incorporated into the system. In such instances, a signal is extracted from two active electrodes, and by subtraction the common hum is removed leaving only the difference in the EMG signal between the two electrodes.

All frequencies within the EMG signal must be amplified equally and in order to do so the amplifier must possess an appropriate bandwidth. The EMG spectrum has been reported to span from 5 to 2000 Hz, with most of the signal concentrated between 20 and 200 Hz. A significant component does extend up to 1000 Hz (Winter, 1979). Thus, the amplifier must be able to detect and linearly amplify all frequencies that are likely to be present within the EMG signal. Both ECG signals (100 Hz) and power line hum (50 Hz) possess major frequencies inside the EMG bandwidth and little can be done to eradicate such noise. However, noise originating

from movement artefacts have a major frequency component between 0 and 10 Hz and can therefore be easily filtered from the signal.

3.4.4.2 EMG Signal Characteristics

The magnitude of a nerve muscle stimulation will ultimately determine the force of a muscle contraction (Bigland-Ritchie, 1981). The EMG is a direct measure of the level of muscle excitation and so it would seem logical to assume that EMG can provide a measure of muscle force. However, the precise relationship between the mechanical force a muscle exerts and the muscle's electrical output is uncertain. Findings have conflicted, with reports of both a linear (Ringelberg, 1985; Aoki et al., 1986) and non-linear (Komi and Viitasalo, 1976; Thorstensson et al., 1976; Hakkinen and Komi, 1981; Peres and Maton, 1987) relationship. The only certainty is that it is monotonic, an increase in myoelectric activity corresponding to an increase in force but of unknown magnitude. It is not the intention of this chapter to discuss in detail those studies which have examined this relationship, as this has been comprehensively covered elsewhere (Cabri, 1989). However, it is felt that the possible reasons for these conflicting findings should be considered. This will serve to aid an understanding of why certain approaches have been adopted to quantify EMG signals, thereby enabling inter-subject comparisons to be made.

The EMG signal is characterised by a time dependent, cyclic repetition of the same standard waveform about a baseline, with both positive and negative values (Basmajian and DeLuca, 1985). Oscillations arise from the summation of the asynchronously firing motor units and movement of the depolarising wave past the active electrodes, which produces frequencies and amplitudes that are continuously changing (Basmajian and DeLuca, 1985). Signal characteristics and evaluation are further complicated by the signal's changing nature with respect to anatomical and physiological properties of the muscle, the control scheme of the central nervous system, as well as the nature of the observing and detecting instrument (Cabri, 1989).

As previously mentioned, signal detection is governed by a number of influential factors (ie., electrical properties of the skin, electrode placement, etc) which all affect the pattern of the signal. In addition, EMG signal comparisons can be complicated by intrinsic muscle

variables, such as muscle fibre composition and length, and the state of fatigue. Thus, the raw signal is only suitable for a visual qualitative inspection in order to identify whether or not a muscle is active. Its use has therefore been restricted mainly to the validation of biomechanical models (Morris et al., 1961). In an attempt to improve the predictive power of the EMG signal and enable comparisons to be made between subjects, various methods have emerged to quantify the muscle activity.

3.4.4.3 EMG Signal Processing

Linear full-wave rectification forms the basis for most quantification procedures and generates the absolute value of the EMG signal, usually with positive polarity (Winter et al., 1980). Full-wave rectification, whereby all negative voltages are reversed, is preferred to half-wave rectification, in which negative values are eliminated, as it contains all the energy of the signal (Basmajian and DeLuca, 1985). By itself, the full-wave rectified signal has limited use but it does serve as an input to many other processing procedures.

An alternative approach to rectification is the root mean square (RMS). Using this technique, the square of the wave is first calculated, thereby making it all positive, and then the square root is calculated.

A semi-quantitative assessment of a rectified signal can involve a visual inspection of amplitude fluctuations which gives a good indication of changing muscle contraction levels (Winter, 1979). A processing procedure which provides a more quantitative evaluation of the rectified signal is linear enveloping. It has been best described as a moving average which closely follows the trend of the muscle contraction encapsulating it in an 'envelope'. With the linear envelope, input is linearly related, in time, to the output, by analogue or digital filtering. Selection of the appropriate low pass filter is an important criterion of the procedure. Winter (1979) stated that a filter with a cut-off at about 6 Hz and of the second-order type should be implemented if the envelope is to represent muscle tension. The reason for this is that the frequency of human movement does not normally exceed 6 Hz, so any signals which represent muscle tension should not have a frequency above 6 Hz.

A linear envelope is not a true mathematical integration of the signal as the raw signal is smoothed to establish the moving average. Pure mathematical integrators can be, and have been used as a procedure to quantify the rectified signal (Winter, 1979).

An integrator seeks to measure the 'area-under-the-curve' and will, therefore always increase whenever EMG activity is present. Several types of integration are possible, the simplest form initiates an integration at specific periods of time, usually at the start of a muscle activity and continues throughout the muscle action. Other forms reset the integrated signal to zero at specific intervals, either time or voltage related. With a constant time reset a series of peaks is produced which represents the trend of the EMG amplitude with time. Each peak is essentially an average EMG over the previous time interval. If this method of analysis is to be used then the time reset is a critical factor. Setting a large time interval will not provide a true picture of muscle activity when rapid fluctuations occur in the EMG signal and too small a reset will produce a signal that is highly succeptible to noise.

The variability of the EMG signal and the need to make subject/trial comparisons have led to the development of a number of normalisation techniques (Yang and Winter, 1984). Signal normalisation relates the recorded muscle activity to a specific signal characteristic, usually the EMG of maximum effort or highest EMG value. In most cases a subject is asked to perform a maximum voluntary isometric contraction of the muscle, or muscle groups under investigation and the amplitude of the resulting raw or rectified signal is used as the reference (100%) for all other activities. It has been shown that maximum isometric contraction can often result in lower amplitudes than certain dynamic contractions. This led Yang and Winter (1984) to compare four normalisation procedures, two utilising an isometric calibration and two derived from the muscle activity trace itself. It was the latter two, normalisation with respect to the peak within-subject ensemble average and the mean of the within-subject ensemble average, that drastically reduced inter-subject variability by between 12 to 73%.

3.4.4.4 EMG and Spinal Loading

It is only relatively recently that authors have began to extend the role of EMG from a technique to validate spinal muscle model assumptions to a method which can establish the degree of compressive spinal loading. Andersson et al. (1980) examined both the estimated erector spinae tension force and calculated EMG activity during asymmetrical static seated work. Ten male subjects held loads of 0, 20 and 40 N in ten different arm positions above a table with the head in a fixed, but unsupported position. The myoelectric signals from 12 bipolar recessed surface electrodes on both sides of the back were passed, via multi-channel RMS detectors with low-pass filters (0.8 Hz), to the analog-to-digital converter of a computer. For each 15 s of recorded activity the mean signal amplitude (μV) was calculated and compared with the mathematically derived estimate of erector muscle force. A highly significant correlation was found between the left and right lumbar erector spinae EMG activity and force predictions, r = 0.984 and 0.988 respectively.

In two reported studies by Schultz et al. (1982a, 1982b), in which they implemented an identical EMG processing procedure to that of Andersson et al. (1980), EMG data were used to validate a biomechanical model of lumbar disc compression forces. Schultz et al. (1982a) studied 25 isometric tasks, 10 standing and 15 sitting, in which loads (between 0 and 20 kg) were applied horizontally and vertically to 4 subjects. Twelve bipolar recessed surface electrodes were placed 3 cm lateral to the midline at various levels down the back (T8, L1, L3 and L5) and over the external obliques (3 cm anterior and above iliac spine) and rectus abdominal muscles (2 cm lateral to the midline on both sides at the level of the umbilicus). For the erector spinae, a regression analysis of the mean predicted lumbar erector spinae tension against mean measured myoelectric signal amplitude showed significant correlation coefficients of 0.94 and 0.91 for the left and right sides, respectively. The predicted left and right rectus abdominis force had a respective correlation coefficient of 0.56 and 0.55 with mean measured EMG amplitudes. Correlations for the left and right internal and external oblique muscles were 0.57 and 0.20, respectively.

In a similar study (Schultz et al., 1982b), 10 subjects performed isometric weight-holding and force-resisting work tasks while standing,

with and without trunk twists, with a lateral trunk bend, and in postures involving combinations of bending and twisting (a total of 20 tasks with loads varied between 0 and 40 kg). When the linear programming model assumed a minimum muscle contraction intensity the linear correlation ranged from 0.67 to 0.88 over the six muscle force-activity correlations. For a minimum compression assumption, these correlations ranged between 0.34 and 0.92. It was noted that the abdominal oblique muscles showed the poorest force-activity correlations when the minimum compression objective function was used to predict the muscle tensions.

Seroussi and Pope (1987) also studied the relationship between trunk muscle EMG and lifting moments in the sagittal and frontal planes. Using a simplified biomechanical model, they hypothesised that the sum of the left and right erector spinae processed EMG depends on the sagittal plane moment, and the difference between the left and right erector spinae processed EMG signal depends on the frontal plane moment. To test these hypotheses ten subjects stood in apparatus which restricted gross body movements and held a 7.5 kg mass in 28 different frontal and sagittal positions. Four pairs of surface electrodes were attached to the left and right erector spinae and external obliques. At a sampling frequency of 200 Hz, data were collected from each muscle and postural position for 1.5 s. Software was used to perform a digital RMS to DC conversion of the 300 raw data points. A normalising procedure was adopted to prevent preferential weighting of one subject's data when averaging across all subjects and to eliminate the effects of arm and trunk weight. This took the following form:

$$E-E(min)/(E(max)-E(min))$$

where: E = RMS EMG
E(max) = RMS EMG maximum (For all postures)
E(min) = RMS EMG minimum

The results did support the proposed hypotheses, with the average sum of the right and left erector spinae EMG signal significantly correlating ($r=0.96$) with the sagittal plane moment, but not with the frontal plane moment. Similarly, the frontal plane moment strongly correlated ($r=0.95$) with the difference between the right and left erector spinae

EMG signal, but not the sagittal plane moment. The external oblique muscles were found to be unrelated to either the sagittal or frontal plane moments.

Freivalds et al. (1984) applied EMG to validate a dynamic biomechanical model of lifting. Monopolar surface electrodes were used to record erector spinae activity at the L3/L4 level for 6 male subjects lifting acceptable loads. A differential amplifier was used to collect the raw signal which was then rectified, smoothed and digitised in 40 ms intervals. A correlation of the estimated compression forces and the EMG over time for all trials yielded an average correlation coefficient which was significant ($r=0.42$, $p<0.05$). The authors concluded that myoelectrical activity can be used as an indirect indication of muscle contraction force and low-back compression forces during dynamic lifting.

A study which sought to make direct comparisons between measured intradiscal pressure and EMG data was performed by Örtengren et al. (1981). Eighteen different static postures involving symmetrical and asymmetrical loading of the trunk in upright, flexed, lateral flexed and rotated trunk postures, along with back and leg lifting exercises, were studied. An EMG data collection and processing routine identical to that mentioned earlier (Andersson et al., 1980) was used. For the symmetrical and asymmetrical postures it was found that mean disc pressure values correlated well with the mean myoelectric signal amplitudes, the coefficients of determination being 0.95 and 0.97, respectively. However, it was found that these linear relationships could not be applied to the heavy lifting tasks which caused the myoelectrical signal amplitudes to vary, depending upon the back posture and handle heights. It was suggested that during the heavy lifting tasks, structural changes occur within the spine which change the mechanical conditions and therefore, the muscle activity.

3.4.5 Biomechanical Modelling of Spinal Loads

3.4.5.1 Biomechanical Modelling as an Investigative Tool

Biomechanical models provide a mathematical, non-invasive, indirect means of gaining an understanding of musculoskeletal functioning which would otherwise be difficult, if not impossible to measure. The direct comparison of a model's behavioural characteristics with those of the

actual system enable a further insight into how the system's components function. More importantly, they serve as a means to predict potentially hazardous loading conditions for certain musculoskeletal components (Chaffin and Andersson, 1984). If one accepts that mechanical disruption of normal tissue function is a major cause of LBP, then biomechanical modelling would seem a logical method by which to investigate the problem.

Modelling provides an ideal investigative tool from the point of view of versatility. Various parameters can be manipulated to facilitate an understanding of the complex nature of musculoskeletal loading, without risking the health of the individual. In return, information is gained as to the parameter's functional role within the movement cycle.

Subjecting cadaveric motion segments to a variety of different types of loading configuration is beginning to improve our understanding of tissue failure, particularly in the case of gross force failure (Brinckmann et al., 1987). However, it is only through biomechanical modelling, inclusive of relevant musculature, that we can relate cadaver failure data to the likelihood of injury in real task situations.

The National Institute for Occupational Safety and Health (NIOSH) was one of the first organisations to provide guidelines for maximum load lifting (NIOSH, 1981). Based principally on the crush values of cartilage endplate of cadaver spines, it was stated that L5/S1 compression values of 3425 N or above were potentially hazardous for some workers. It was recommended that values should not exceed 6361 N, as these were felt to be hazardous to most workers. Thus, biomechanical models have been used, in conjunction with *in-vitro* measures to provide an alternative tool to overcome the limitations of direct, invasive measurements of lumbar spinal loads.

3.4.5.2 Approaches Adopted in Biomechanical Modelling

Functionally, the spine has to respond to quite diverse circumstances. Certain situations require it to be highly flexible, while others require it to be rigid and support considerable stress. Consequently, it exhibits large displacements, rotations and constantly changing inertial and stiffness properties. To accommodate these changing conditions numerous analytical models have been developed to simulate

and estimate forces on the human spinal column. These models can be classified into two broad categories - deformable element representation and rigid link segment.

Models comprising of deformable elements generally fall into lumped parameter, discrete parameter, finite element or continuum categories. Within the lumped parameter classification, subunits are modelled as springs, masses and dampers, depending on their dynamic response characteristics. Burns and Kaleps (1980) adopted such an approach to analysis the load-deflection behaviour of the intervertebral discs under axial compression.

Liu et al. (1971) used a discrete parameter model to investigate spinal responses to impact during aircrew ejection. This simulated the mechanical properties of the vertebrae and discs with alternate rigid and deformable bodies, respectively.

The finite element approach sub-divides a body into smaller homogeneous components so that the system's geometry can be modelled more precisely. King and Yang (1986) constructed a finite element model of the lumbar motion segment using 8-noded hexahedral (brick) elements, thick/thin shell elements (4-noded), and elastic bar elements (2-noded) - in total 207 individual segments. These were then assigned specific geometric and mechanical properties. Under different loading conditions it was possible to assess facet joint loading, disc bulge, nucleus pressure and annulus stress.

Cramer et al. (1976) used a continuum model to investigate the mechanical responses of the spine to acceleration. In this approach, the system's components are portrayed in a continuous form, a curved beam being the form adopted by Cramer et al. (1976).

A relatively new model of the spine, the arch model, which would seem to fall into the deformable element category has been proposed by Aspden (1987, 1988). He considered that the flexibility afforded to the spine means that in most postures it will conform to a curve. Therefore, he felt that the spinal load bearing mechanisms are better described by an arch, a structure that relies on its curvature for stability; abutments of the arch being the sacrum at its base and some combination of body

weight, muscular and ligamentous forces at the neck. The model applies the plasticity theorem and considers that the thrust line must be within the cross-section of the arch, along its entire length if it is to be stable under a particular load. Further work is needed to assess the validity of such a model.

Deformable element models have yielded important information in terms of spinal mechanics. However, rigid-link segment models take a much wider view of the human body and enable greater inferences to be made about postures and external forces. Thus, in terms of an ergonomics investigative tool, they have received a much wider application. Future discussion of modelling techniques will therefore, be restricted to such an approach.

3.4.5.3 Rigid-Link Segment Modelling

Rigid-link segment modelling represents the human body as a spatial linkage system, assuming body components (limbs or body structures) to be rigid segments rotating about hinge joints. Mathematically, using a Newtonian analysis, the human body can then be treated as a 'rigid body dynamic linkage' problem. Kinematics of the linkages, and any external forces acting on the system are combined to generate the forces and moments acting about the joint hinges. In the human body these are equivalent to joint reaction and muscle moments acting about a joint. Assumptions inherent to a link model approach are:

(i) the mass of each segment is fixed and located at its centre of gravity;

(ii) during movement the location of the link's centre of gravity remains fixed;

(iii) hinge or pin joints are used to represent segment articulations;

(iv) a segment's mass moment of inertia remains constant about either end of the segment or about its centre of mass.

(Winter, 1979)

In general, the link segment modelling approaches which have been implemented to investigate lumbar muscular and compressive forces (Schultz et al., 1982b; Andersson et al., 1980; Leskinen, 1985; 1983a, 1983b, 1985; McGill and Norman, 1985; Cappozzo, 1983, 1984, 1985; Cappozzo et al., 1985) have undertaken two basic steps:-

(i) determine the net reaction forces and muscle moments across a section of the lower trunk;

(ii) determine the internal force distributions that supply this net reaction.

3.4.5.4 Lumbar Net Reaction Forces and Muscle Moments

To calculate joint reaction and muscle moments it is necessary to obtain accurate posture and kinematic data. This information is then combined with anthropometric data (segmental masses, centre of mass location, and mass moment of inertia) and any external forces. The techniques for the assessment of human kinematics and their associated limitations have been extensively reviewed elsewhere (Plagenhoef, 1971; Miller and Nelson, 1973; Atha, 1984; Harrison, 1989). From such reports it would still appear that high speed cinematography offers the most effective and affordable means to capture and record human movements.

Estimating segmental masses, segmental centre of gravity and segmental moments of inertia have proved problematic. The literature identifies three approaches that have been implemented to study these parameters:

(i) cadaver studies;
(ii) experiments on living subjects;
(iii) mathematical modelling.

Dissection studies on cadavers are attractive in that they provide a basis from which to make predictions of living subject's segmental parameters. Because of the difficulties encountered when trying to perform such studies, the data are not always a true reflection of the average adult population, with respect to age, height and weight. Therefore, the scope of any inferences will be limited.

Dempster's (1955) data are extensively referred to throughout the literature and have been implemented by several modellers of spinal mechanics (Chaffin, 1969; Leskinen et al., 1983a, b, 1985; Freivalds et al., 1984; Occhipinti et al., 1985). Although, previous cadaveric studies had been conducted, Dempster (1955) was the first to perfect and standardise techniques. Eight cadavers were dissected into 13 segments (trunk and head representing one segment) from which information was gained as to anthropometric measurements, segmental masses, centre of mass, density, and moments of inertia. Segmental masses were expressed as a percentage of total body mass, centres of gravity as a percentage of link lengths, and moments of inertia as constants. Widule (1976) identified the possible shortcomings of expressing segmental moments of inertia in such a manner (ie., only representative of subjects within the height and weight of Dempster's (1955) population) and consequently, developed separate equations with height and weight as independent variables.

Clauser et al. (1969) utilised Dempster's (1955) techniques but sought to improve the predictive power of cadaver data by establishing stepwise regression equations from anthropometric measurements. Thirteen male cadavers were studied from which anthropometric measurements, sufficient to describe the length, circumference, and breadth or depth of each segment, were taken. After ascertaining the total body volume and centre of mass, each was dissected into 14 segments for which weight, volume and mass centre were measured. The multi-step regression equations were developed by identifying the body dimension which had the maximum power to predict a given segmental parameter. In all, three or less dimensions were used in each equation, these being the most powerful predicting parameters. The implementation of more than three variables decreased the power of each equation, due to the small sample size.

Unlike mass, centre of mass or anthropometric measurements, there is no simple single measurement to describe moment of inertia. Therefore, deriving individual body segment moments of inertia has proved difficult. Chandler et al. (1975) provided for the first time data which related segmental principal moments of inertia values for body segments with standard anthropometric measurements. Six cadavers were dissected into 14 segments, and the moment of inertia of each segment

about six axes was derived using a pendulum technique. From these six values, the direction of the principal axes and corresponding principal moments of inertia were calculated. Thus, Chandler et al. (1975) were able to provide equations requiring length, width and perimeter measurements, which could then be used to estimate the segmental moment of inertia values for living subjects.

Several experimentation techniques have evolved which gather measures directly from the living subjects; however, they are difficult to administer and provide only limited data on a few segmental parameters. Two techniques which have been used with some success include stereophotogrammetry and radiation.

McConville et al. (1980) combined stereophotometric and anthropometric techniques to measure 31 subjects. Twenty four body parts were assessed in terms of their volume, centre of volume and principal moments of inertia about three principal axes. They measured 75 anthropometric variables and derived 10 additional dimensions. Using the most highly correlated dimensions, multiple regression equations were established to determine segmental volumes and principal moments of inertia.

Zatsiorsky and Seluyanov (1983) provided data on 16 segmental parameters (including three trunk sections) for 100 males (mean age = 23.8 yr) using a radioisotope technique. The absorbtion of gamma-radiation as it passes through body tissue provided the method by which body segment parameters were calculated. Knowing the intensity of radiation before and after it passes through each body segment, the authors were able to calculate underlying tissue mass. In addition, each segment's centre of mass, moment of inertia and radii of gyration were estimated. One hundred and fifty multiple regression equations were established with body mass and body height as the independent variables. The results from this study have interested many researchers on the basis that they provided a large quantity of data for a young, living population.

The advancements in computer technology and the need for personalised segmental parameters have encouraged some researchers (Hanavan, 1964; Jensen, 1976; Hatze, 1980) to develop mathematical models of the human body. Such models make simplified assumptions as to the structural compositions of body segments. Hanavan's (1964) computer model

represented body segments as simple geometric solids, with the head as an ellipsoid of revolution, the upper and lower torso as right elliptical cylinders, the hands as solid spheres, and all other segments as frusta of right circular cones. Each segment was constructed from 25 anthropometric dimensions and Barter's (1957) regression equations were employed to approximate segment masses.

More advanced computer models, requiring considerably more anthropometric data to construct each representative segment, have been developed by Jensen (1976) and Hatze (1980). The major criticism raised at these and other similar approaches is that by employing simple geometric solids they cannot cater for the highly compositionally diverse and changing body structures known to exist within the human body.

3.4.5.5 Trunk Muscle Modelling

Conceptually, determining the possible internal force distribution is much more difficult than the net reaction forces and muscle movements. The reason is that many internal body structures can exert forces that are unknown and cannot easily by measured. Therefore, biomechanical modelling must make simplified assumptions in order that estimates of gross force distributions can be made.

The lower trunk presents a major problem in that many muscles, muscle groups and ligaments play a functional role in maintaining trunk stability. From a mechanical perspective, there are at most six equations of equilibrium (three force and moment equations) to be satisfied. However, there are often more than six muscle forces to be calculated which makes the system statistically indeterminate. For example, Schultz et al. (1982b) identified ten muscles that play an important role in bending and twisting tasks, which means there is a total of 13 unknowns, when the three components of joint reaction force are included. The indeterminacy can be further increased if additional anatomical constraints are taken into account, such as ligamentous forces, the tangential path of some muscles, translation of the disc's centre of rotation, and the effects of muscle length and velocity of length changes. Consequently, researchers have adopted several different approaches to model the internal trunk forces (refer to Table 2).

The majority of models, particularly those associated with symmetrical sagittal plane lifting (Morris et al., 1961; Leskinen et al., 1983a, 1983b, 1985; McGill and Norman, 1985) have modelled truncal muscular forces as one major force component and assumed antagonistic muscle actions are negligible. Of the studies employing a single muscle action, it has been assumed that the activity is controlled by the main extensor muscle group of the back, the erector spinae. In such instances, as long as the muscle moment arm is known then there is only one indeterminate and the problem is solvable.

Cappozzo (1983, 1984) sought to model torso muscles with four single-equivalent muscles. Each represented all the muscles crossing one of four sectors into which the trunk section was divided by the X and Y pelvis axes. The line of action of each representative muscle was made to act at a point which was equivalent to the centroids of the torso muscle cross-sections within each sector. In order to solve the model, two assumptions were necessary. The first stated that muscles could only transmit tensile forces, while the second ignored the presence of any antagonistic muscle actions. As Cappozzo and Gazzani (1982) quite rightly pointed out, this modelling of muscular forces was highly simplified and open to inaccuracies. However, they felt that at the time it was a major advancement in spinal biomechanics which would yield useful information.

Andersson et al. (1980) used three single-equivalent force components to model internal trunk muscles while performing isometric weight holding exercises. Subjects were seated at a table and held their right hand at specified points above the table. Weights held in the hand were varied between 0, 40 and 80 N. The modelled muscle forces were equivalent to those exerted by the left and right erector spinae muscle groups, and an equivalent of the left abdominal oblique muscle. Each muscle force vector was located at the centre of each muscle mass. Again, Andersson et al. (1980) found it necessary to make the assumption that antagonistic muscle activity was minimal in order to solve the necessary equations.

Schultz et al. (1982b) sought to increase the accuracy of internal muscle modelling by including ten muscle force components (five

bilateral pairs of single forces representing the rectus abdominis, the internal and external oblique abdominals, the erector spinae, and latissimus dorsi muscles) into his model. Mathematical optimisation (linear programming) was the technique used to solve the numerous equations. This required systematically implementing all unknowns into the available equations until some criterion of acceptance was achieved. Minimum muscle contraction intensity and spine joint compression force were the criteria adopted by Shultz et al. (1982b).

Bean et al. (1988) used the muscle model of Schultz et al. (1982b) but improved the sensitivity and stability of solutions, and reduced computational requirements by adopting a procedure which formulated and solved two linear programs sequentially (double linear programming). Both modelling approaches employed by Schultz et al. (1982b) and Bean et al. (1988) offerred attractive means by which to tackle the problem of determining internal force distribution during complex tasks.

By far the most extensive anatomically detailed model of the truncal musculo-ligamentous-skeletal system was provided by McGill and Norman (1986). Induced lumbar (L4-L5) reactive moments during sagittal plane lifting were partitioned into moments provided by the disc in bending, ligamentous strain, and active muscle contraction. The internal trunk model incorporated 14 ligamentous and 36 muscular force components. The ligamentous force and restorative disc moment were calculated using exponential regression equations and summed to establish their total moment contribution. The remaining restorative moment was then partitioned into the various muscle force components using EMG as a guide. Muscle force calculations took account of instantaneous muscle length, velocity, and type of contraction (concentric/eccentric).

For the lifting activities studied, it was observed that muscular components generated approximately 99% of the total restorative moment, with the other 1% being provided by the bending of the disc. The ligament system was found to provide no restorative moment as subjects elected to maintain a 'flat backed' posture throughout the lifts (lumbar flexion rarely exceeded 34 degrees). Thus, ligaments were not placed in a state of tension.

Although, criticisms can be raised at the methods adopted to calculate various moment contributions, namely EMG to estimate muscle force, the model was felt to make significant inroads into modelling spinal functional anatomy (McGill and Norman, 1986). The dominance of muscle activity and the high forces calculated were thought to be consistent with the observed muscle strains associated with load lifting.

3.4.5.6 Spinal Compression Force Modelling

In this section, it is not the intention to review each individual modelling approach that has assessed lumbar compression loading during various physical activities. This would prove to be a time-consuming exercise of limited value. A comprehensive summary of such rigid-link segment models is provided in Table 2. The intention of the section is to outline the major problems encountered when trying to adopt such approaches.

The early models of lifting predicted spinal compression forces exceeding tissue tolerance data; however, these lifts appeared to have no detrimental effect on the performer. This suggested to many that the models incorporated inaccuracies and excluded mechanisms which reduce loads on the trunk extensor tissues. Intra-abdominal pressure, as mentioned in section 3.4.2., was originally felt by many to be the missing component and was included in many of the later models (Bartelink, 1957; Morris et al., 1961). More recently some authors (McGill and Norman, 1986) have began to criticise the significance of IAP in reducing spinal loads by arguing that to increase IAP there needs to be an accompanying increase in abdominal wall muscle activity, which would consequently add to the compression load. McGill and Norman (1986) revealed that during a lifting exercise the extensor forces and moment created by IAP did not offset the increased compression and flexor moment generated by the abdominal muscle activity.

Gracovetsky et al. (1981) proposed that muscular activation of, and the hydraulic action upon, the lumbodorsal fascia provide the missing sources of spinal support. However, this too seems unlikely following the detailed anatomical studies by Bogduk and MacIntosh (1984).

This seemingly perplexing situation has led researchers to re-examine the moment arm of the main extensor tissue, the erector spinae, as this

le 2. Summary of estimated lumbar compression forces during various activities using biomechanical models.

/ESTIGATORS	ACTIVITY	MAXIMUM COMPRESSION FORCE RANGE(N)	MAXIMUM SHEAR FORCE RANGE (N)	MODEL DESCRIPTION
ɔpozzo 83 = 5	Running 2.62 - 3.44 ms^{-1}	L3/L4 2.7 - 6.3 x body weight	-	Dynamic Muscles: 2 - extensors 2 - flexors
ɔpozzo 84 = 5	Level walking 0.99 - 1.30 ms^{-1}	L3/L4 357.3 - 863.9	-	"
ɔpozzo et al. 85 = 4	Half squat Loads 0.8 - 1.6 x body weight	L3/L4 4218 - 6697	-	"
:me et al. 84 = 4	Landing from 0.33m	L3/S1 1360 - 4750	-	Dynamic Muscles: 2 - extensors 2 - flexors
:k & Chaffin 74 erage man	Lifting - 5 cm from ankle to 25 cm of floor	L5/S1	L5/S1	Static Abdominal pressure force
	stooped back 0 - 13.6 kg	1107 - 2381	343 - 490	Muscle: Erector spinae (5 cm moment arm)
	Vertical back 0 - 13.6 kg	1539 - 3244	265 - 372.4	
	Lifting - 76 cm from ankle to 25 cm of floor			
	stooped back 0 - 6.8 kg	1421 - 1460	353 - 431	
	Vertical back 0 - 13.6 kg	1558 - 2695	314 - 382	
affin 75 erage man	Lifting - 38 cm from ankle to 38 cm of floor	L5/S1	L5/S1	Static Muscle: Erector spinae (5 cm moment arm)
	stooped back 0 - 13.6 kg	970 1796	340 - 486	
	Vertical back 0 - 13.6 kg	1264 - 2729	239 - 342	
sh 76 = 4	Lifting - 20 kg Straight back\bent knees	L5/S1 2675 - 4684	-	
	Flexed back\straight knees	2881 - 4253		
	Lifting - 0.2 kg Flexed back\straight knees	1137 - 2685		

le 2 (Continued).

/ESTIGATORS	ACTIVITY	MAXIMUM COMPRESSION FORCE RANGE(N)	MAXIMUM SHEAR FORCE RANGE (N)	MODEL DESCRIPTION
:g & Herrin /9 :rage Man	Lifting Stoop 0 - 70 kg Squat 0 - 70 kg	L5/S1 2313 - 6096 2215 - 7605	L5/S1 353 - 1029 78 - 235	Static Abdominal pressure force Muscle: Erector spinae (5 cm moment arm)
)up et al.)3 : 10	Lifting & Lowering 15 kg Load Kinetic lift Trunk Kinetic lift Back lift Leg lift	L5/S1 6647 6815 6534 5867	 -	Dynamic Muscle: Erector spinae (5 cm moment arm)
kinen et al. 3a 20	Lifting 15 kg box Load kinetic lift Trunk kinetic lift Back lift leg lift	L5/S1 6042 6629 6365 5866	 -	Dynamic Muscle: Erector spinae (5 cm moment arm)
kinen 5 20	Lifting 15 kg box Load kinetic lift Trunk kinetic lift Back lift leg lift	L5/S1 4548 4650 3989 4033	 -	Static Muscle: Erector spinae (5 cm moment arm)
kinen et al. 5 10	Lifting, lowering 15 kg Back lift Leg lift Shelf lift Back lowering Leglowering Shelf lowering	L5/S1 5765 6039 3304 5290 5317 2938	 -	Dynamic Muscle: Erector spinae (5 cm moment arm)
.11 & Norman ; 4	Lifting 18 kg Static Dynamic	L4/L5 5218 6391	 -	Static & Dynamic Muscle: Erector spinal (5 cm arm)
.11 & Norman ; 3	Lifting 27.3 - 90.9 kg	L4/L5 5571 - 8921	L4/L5 39 - 1272	Dynamic 48 Muscular group component & 7 ligamentous components
rson and fin 1986 1	Lifting - 5 different techniques Load 44 N 222 N 444 N	L5/S1 1361 - 2887 2179 - 4932 3092 - 7814	L5/S1 Max = 1735	Static 5 ligaments
odihardjo tal 4	Lifting acceptable weight Symmetrical & Asymmetrical	L5/S1 2651 - 4706	L5/S1 251 - 529	Dynamic 10 muscular components

ble 2 (Continued).

NVESTIGATORS	ACTIVITY	MAXIMUM COMPRESSION FORCE RANGE(N)	MAXIMUM SHEAR FORCE RANGE(N)	MODEL DESCRIPTION
ager and uttmann 1989 0th percentile erman male	Load lifting 0-50 kg Humpback & Hollow back	L5/S1 400 - 10000	L5/S1 200 - 900	Dynamic 8 Muscular components
ndersson et al 980 = 10	Submaximal arm lifts while seated Weight 0, 40 & 80 N	L3/L4 140 - 720	-	Static 3 Muscular components
chultz et al. 982b = 10	Isommetric weight holding whilst standing with twists and bends 4 - 20 kg	L3/L4 470 - 2610	L3/L4 0 - 200	Static 10 Muscular components
cchipinti t al. 985 = 11	Sitting (varied trunk posture) Unsupported upper limbs Supported upper limbs	L3/L4 1161 - 2127 959 - 1398	-	Static Muscle: Erector spinae (5 cm moment arm)
agnon et al. 986 = 6	3 different methods of raising a patient (Hands, forearm & belt)	L5/S1 5744 - 7951	L5/S1 968 - 1039	Static Abdominal pressure force 1 extensor force
agnon et al. 987 = 15	Nurses turning patients in bed 8 tasks	L5/S1 2479 - 3526	L5/S1 554 - 661	Dynamic 1 extensor force

proves to be a most sensitive variable in force calculations (ie., small changes in its value can produce large changes in calculated stresses). For many years, the force vector created by such a muscle group was assumed to act at a distance of 50 mm from the centre of the lumbar intervertebral disc (Morris et al., 1961; Leskinen, 1985). Recent advancements in technology (computer tomography, magnetic reasonance imaging) have enabled more accurate *in-vivo* measurements of this muscle moment arm to be made. This work would suggest the value is in excess of 50 mm (Németh and Ohlsén, 1986; McGill et al., 1988; Kumar, 1988; Tracy, 1988). Table 3 provides estimates of selected trunk muscle moment arm lengths as determined by various authors using the forementioned techniques.

The influence of these findings on the predicted forces from biomechanical models has been investigated by McGill and Norman (1986, 1987). They found that a more realistic lever arm length of 75 mm reduces compression force estimates by up to 35% when compared to a model incorporating a 50 mm moment arm. It was stated that this reduction was appreciable, decreasing compression values to levels well below disc failure measurements.

The data reported in Table 3 have, in the majority of cases, been obtained from subjects in a supine and relaxed position. Consequently, they take no account of the possible influences posture or muscle loading have on moment arm lengths. Kumar (1988) did not find any relationship between height, body mass or height-body mass product and abdominal muscle moment arm lengths.

The influence of inertial forces and torques induced by the acceleration of body segments and external loads on spinal compression loading was often felt to be negligible. Consequently, a majority of lifting models were static and considered only the forces due to gravity (Morris et al., 1961; Chaffin, 1975; Garg and Herrin, 1979). More recently, such parameters have been found to contribute significantly towards spinal stress during certain lifting activities. Leskinen (1985) compared the peak lumbar compression loading for a static and dynamic modelling approach and found the latter to increase estimated loads by 33 to 66%, depending on the lift.

able 3. Selected trunk muscle moment arm lengths as determined by CT scans and MRI.

NVESTIGATOR	SUBJECTS	METHOD AND DISC LEVEL	ANTERIOR-POSTERIOR DISTANCE FROM DISC CENTRE (mm) (-ve sign denotes posterior to disc) (muscle group)		
			ERECTOR SPINAE	RECTUS ABDOMINIS	EXTERNAL OBLIQUES
EID AND OSTIGAN 1985)	28 (M*-10) (F*-8) Mean Age Ht+ Wt+ (M) 55 1.73 74.7 (F) 52 1.59 63.8	CT Scans 15 Transverse scans Xiphoid process to symphysis pubis	-0.293° (M) -0.280° (F)	0.500° (M) 0.403° (F)	- -
EMETH AND ILSÉN (1986)	21 (M-11) (F-10) Mean Age Ht Wt (M) 70 1.76 75.1 (F) 63 1.66 60.1	CT Scans L5-S1	-71 (M) -65 (F)	109 (M) 80 (F)	64 (M) 52 (F)
:Gill et al. 988)	13 (M) Mean Age Ht Wt (M) 40 1.74 89.1	CT Scans L4-L5	-59 (M)	102.8 (M)	59.4 (M)
MAR (1988)	13 (M-8) (F-5) Mean Age Ht Wt (M) 58 1.78 82.6 (F) 55 1.62 65.5	CT Scans L5	-60.0 -61.7	104.7 (M) 106.3 (F)	50.6 (M) 59.5 (F)
ACY (1988)	26 (M) Mean Age (M) 29	MRI L4-L5 L5-S1	-60.0 -61.7	73.9 81.3	- -
AFFIN et al. 990)	96 (F) Mean Age Ht Wt (F) 50 1.63 67.6	CT Scans L3-L4	-53 (right) -52 (left)	72 (right) 70 (left)	20 (right) 23 (left)

- males F - females
t - Height (m) Wt - weight (kg)
atio value of trunk depth
scans - Computer Tomography
- Magnetic Resonance Imaging

Similarly, McGill and Norman (1985) noted that a dynamic model of lifting resulted in an average peak L4/L5 moment 19% higher (maximum difference of 52%) than a static model. This was further substantiated by Bush-Joseph et al. (1988) who compared a similar lift at three different speeds:- slow, normal and fast. They concluded that inertial forces should be considered when analysing lifting tasks biomechanically and that excessive speed, including jerking, should be avoided.

3.4.5.7 The Modelling of Impact Loads

Rigid-link segment modelling has provided a method by which the stresses placed on joints can be quantified. A variety of human movements, from simple walking (Winter, 1979) through to complex gymnastic events (Watkins and Nicol, 1986) have been studied. The majority of these models, particularly those associated with assessing the human body's reaction to impact forces (large forces over a short duration), have tended to focus their attention on the lower extremities (Smith, 1975; Harrison et al., 1986; Bobbert et al., 1987a, 1987b; Nisell and Mizrahi, 1988). Few have considered the forces imposed on the spine (Cappozzo, 1983; Berme et al., 1984). Of these studies, the work by Smith (1975), Nisell and Mizrahi (1988), Bobbert et al. (1987a, 1987b) and Berme et al. (1984) are of particular interest as they considered the impact forces resulting from drop landing and drop jumping.

Smith (1975), inspired by the high incidences of 'fatigue' failure in the knee joints of athletes taking part in exercises involving impacts of a moderate and apparently non-injurious nature, sought to model the joint forces about the knee and ankle. A quasi-static rigid-link segment model estimated peak vertical compression forces of 7.4 and 18.0 body mass units at the ankle and knee joints, respectively, when drop landing from 105 cm. Not surprisingly, these were considered to be high and provide a possible explanation for the observed fatigue failures.

Nisell and Mizrahi (1988) calculated the ankle and knee joint forces incurred during various step down activities from 0.2 and 0.43 m. Using a sagittal plane, semi-dynamic biomechanical model, they observed peak ankle joint compression forces of 4 to 6 body mass units and peak knee compression forces between 4 to 7 body mass units. It was noted that the latter forces were much more influenced by step height than the former.

Bobbert et al. (1987b) undertook a study of the joint reaction forces encountered during the first ground contact of a drop jump performed from three different heights (0.2, 0.4 and 0.6 m). A semi-dynamic link-segment model approach estimated reaction forces at the ankle, knee and hip joints, along with actual compression forces at the ankle. As the authors pointed out, the actual forces encountered by joint surfaces are greater than the net joint reaction forces. In addition, because of the nature of the muscle model the estimated ankle forces were only an indication of the minimum possible compression force.

Estimated net reaction forces showed increases with corresponding increases in drop height. From a height of 60 cm a sharp peak in the time histories of these variables occurred. This was deemed to be the direct result of the subjects being unable to control the impact and prevent a heel strike. The peak ankle forces were similar to those calculated by Smith (1975), almost eight times body mass. The maximum drop height was some 0.4 m less than that implemented by Smith (1975), which would imply that drop jumping may well be more detrimental in terms of inducing injury.

Berme et al. (1984) provided some indication as to spinal compression forces during impact landing. Utilising a biomechanical model, as described by Cappozzo and Gazzani (1982), these authors determined peak compression values for 0.33 m drop lands. Maximum load values (2.3 to 6.2), normalised with respect to body weight, were comparable with those estimated during running (2.7 to 6.3; Cappozzo, 1983). As Berme et al. (1984) stated, it would be wrong to assume that both these activities were equally strenuous as the rate of loading is considerably higher during landing.

All these studies raise interesting and somewhat worrying indications as to the forces with which joints must contend. However, some authors have questioned the validity of such studies, particularly with regard to the use of a rigid-link segment modelling approach to investigate highly dynamic loading. The argument raised is that if a dynamic load is applied to rigid links joined with pins then the load will be instantaneously transmitted through the linkage system, with no distortion of the applied force-time curve. Isolated investigations on

body segments reveal that they are composed of a number of assorted deformable soft tissues with varying viscoelastic properties (Nordin and Frankel, 1989). Consequently, these may well play a significant role in the attenuation of dynamic loads through the dissipation of energy by motion and defomation of tissues.

Pratt (1989) outlined two possible mechanisms by which the human body seeks to absorb impact forces, active and passive. The active mechanism refers to such factors as proprioceptors, joint positions and muscle tone, and the passive mechanism, elasticity of bone, cartilage, synovial fluid and soft tissues (eg. capsule and intervertebral discs).

Radin et al. (1970) and Paul et al. (1978) both examined the passive shock attenuating properties of body tissues. When conducting *in-vitro* studies to compare both cancellous bone and cartilage, Radin et al. (1970) found that although cancellous bone is considerably stiffer than cartilage, it is still able to absorb shock. It was also shown that removal of synovial fluid and cartilage from a joint has only a minimal reduction in shock attenuating capabilities. Paul et al. (1978) impacted a range of frequencies on the splinted hind legs of rabbits, the splint being used as a means to eliminate any possible shock attenuation through joint movement. At frequencies of between 3 and 18 Hz there was minimal shock attenuation. However, shock attenuation increased gradually with increasing frequency, with almost total attenuation over a range of 500-3000 Hz. When the heel pad was removed, the shock attenuation over the full range of frequencies was decreased by 20-28%. They concluded that active shock attenuation occurs at low frequencies and passive mechanisms are responsible for high frequency attenuation.

Voloshin et al. (1981) performed studies which suggested that the intervertebral discs act as a shock absorbing mechanism and can be made less efficient by degenerative changes. A comparison of a group of healthy and LBP patients (Voloshin and Wosk, 1982) identified a correlation between LBP and reduced shock attenuation capacity. It is possible to raise some criticisms of the method Voloshin and Wosk (1982) used to record and measure shock attenuation. One such criticism is the use of skin-mounted accelerometers to measure bone vibration. The soft

tissue between the accelerometer and bone must surely have introduced some noise component.

When examining the transmissibility of the human spine during walking and running, Smeathers (1989) did attempt to take account of both skin movement and differences in the inclination of the spine. Experiments were performed on a group of healthy subjects and a group of subjects with diagnosed ankylosing spondylitis. It was observed that the former were able to attenuate frequency components above 20 Hz, whereas the spines of the latter group behaved in the form of a rigid strut.

An investigation of the trunk's ability to transmit dynamic axial loads was performed by McGill et al. (1989). Their intention was to examine whether the trunk could be represented as a rigid structure when subjected to impact loads. Loads were manually applied at either the hands or shoulders, and the resulting force transmitted to the base of the pelvis was measured. Three input force magnitudes (<251 N, 251-500 N and >500 N) and three load durations (<80 ms, 80-500 ms and >500 ms) were investigated. They concluded that a rigid body segment assumption of the trunk is reasonable when the rise time of the applied force is greater than approximately 300 ms. However, for rise times of shorter duration the trunk did not behave rigidly. Interestingly, for an applied load with a rise time of approximately 80 ms there appeared to be an augmentation of the applied force rather than a damped response. This was felt to stem from the upper body mass rebounding as a result of either the active (recoil due to stretch reflex) or passive mechanisms (elastic properties of the tissues).

Support for a non-rigid approach to the human modelling of impact loading was provided by Gruber et al. (1987). They developed a 'wobbling mass' model which took account of soft tissue movements. Although, the three-link model (trunk and upper body, thighs and lower legs) was simplistic they found good agreement between calculated motion and that derived from high-speed film, when subjects performed drop landings.

3.5 SUMMARY AND CONCLUSIONS

Reported incidences of back pain suggest that it is one of the major musculoskeletal disorders afflicting a large proportion of the population. Often, causes are uncertain and indeterminate, although risk factors have been identified. Epidemiological studies have shown clear links between a mechanical origin, in the form of increased loading on the spinal column and back pain symptoms. More recent data would also imply that prolonged sedentary type occupations are as important a risk factor as heavy physical tasks.

The extent to which sports participation increases the risk of incurring back problems is unclear, as in many cases symptoms manifest over a prolonged period and cannot be distinguished from factors within the normal daily routine. Limited statistical data do suggest that certain sporting pursuits can carry a higher risk of spinal injury than others and it is often the nature of the force or stress, inherent to the activity, which is the important predeterminant of the type of injury the spine will incur.

Mechanisms leading to pain appear varied and complex but a mechanical derangement of structural components seems likely. Many implicate disc degeneration as a direct or indirect cause of most mechanically induced back pain. Resultant structural changes within the spinal motion segment lead to irritation of nerves within close proximity of the spinal structures. As to the most likely site of disturbance, many are agreed that the lower or lumber spinal region is more susceptible to injury because of the increased loading and physical stress placed on it.

Plyometric exercises can be regarded as a relatively new form of training employed by those athletes seeking to increase muscular power. The very nature of bounding and drop jumping exercises (a group of plyometric exercises), which involves forceful impact landings as a means to invoke the necessary muscle actions (eccentric followed closely by a concentric muscle action), may subject the athlete's spine to forces which are detrimental. However, the literature is clearly devoid of such information.

Establishing the nature of the forces incurred, how they are transmitted to the spine and the effects they have upon the spine, are issues which need to be addressed. As a result, it will then be possible to identify steps by which to reduce any detrimental effects. As part of this evaluation, there is a clear need to assess the long term spinal responses to recovery exercise involving spinal unloading postures. For example, will short-term recovery be sustained or will subsequent loading quickly negate any possible benefits? Although studies on cadaveric specimens have provided important data on response characteristics of spinal structures, the need to investigate spinal loading *in-vivo* is important when addressing the problem from an ergonomics standpoint.

The literature identified several *in-vivo* measuring techniques - intra-discal pressure (IDP), intra-abdominal pressure (IAP), electromyography (EMG), biomechanical modelling and changes in stature. Each was found to possess limitations which need to be fully understood prior to making recommendations based on their findings. Both IDP and IAP were considered to be either ethically or practically difficult to apply. The latter procedure, measurements of alterations in stature, has only recently been established as a reliable technique. Further work was considered necessary to enhance and validate the procedure.

The thesis will now seek to undertake the objectives laid down in Section 2, drawing on information gathered during the literature search.

4. DEVELOPMENT OF STADIOMETER

4. DEVELOPMENT OF THE STADIOMETER

4.1 EFFECTS OF BOUNDING ACTIVITIES ON SPINAL SHRINKAGE: A PILOT STUDY

A poster presentation of this work was presented at the one day symposium on the effects of work activity on the spine, 24th October 1986, University of Nottingham. The abstract submitted for this conference can be found in appendix 1.

4.1.1 Introduction

A suitably reliable and repeatable method for measuring small changes in a person's stature was not developed until the early 1980's (Eklund and Corlett, 1984), since which time little work has been undertaken to enhance the technique or its equipment. As stadiometry is a potential research tool for recording loads imposed on the human spine and a possible screening procedure for potential back pain sufferers, improvements to both the equipment and technique were seen as important steps towards establishing the procedure validity.

To appreciate the limitations of the stadiometer employed by previous researchers (Wilby et al., 1987; Foreman, 1988) to record changes in stature, it was felt that a pilot study should be conducted using their original apparatus. Aspects of the apparatus have been outlined in the literature review (section 3.4.3) and in greater detail by Foreman (1988). It was hoped that in this way a more efficient, reliable and accurate method to record small changes in stature could be developed. In addition, it was seen as a means to provide preliminary information on spinal responses to plyometric exercises.

4.1.2 Method

The activity under consideration consisted of 20 repetitions of 5 continuous double footed bounds from a stationary position. This is a common activity often incorporated into the training programme of many jumpers and sprinters. Four males aged 23 - 30 (mean 25.25) yr acted as subjects. Mean values for height were 1.74 m (range = 1.70

- 1.76 m) and corresponding mean body mass was 69.6 kg, ranging between 67-75 kg. None of the subjects had any history of back problems and none had regularly participated in this form of exercise.

Prior to experimentation, all subjects undertook a training period using the stadiometer to familiarize themselves with both the apparatus and technique, and ensure that they could achieve a reproducible, relaxed standing posture. A subject was deemed to be suitably trained once he could undertake ten consecutive measures with a standard deviation of 0.5 mm or less. All subjects participating in this study achieved this criterion. Between stature measurements and whilst on the stadiometer, subjects were instructed to place their hands against the frame and ease their body forward of the postural control switches, thereby breaking the microswitch contacts. They were then asked to move their feet slightly before repositioning their heels against the back plate. Subjects did not dismount the stadiometer as it was felt this would impose additional and unwanted load upon the spine.

Prior to exercise, subjects maintained the Fowler position for 10 min in an attempt to standardise stature and control for possible abnormal spinal loading prior to the experimentation. The Fowler position is a standard orthopaedic posture recommended to patients as a means of alleviating spinal loading. It involves subjects lying supine, with their feet supported such that the angle formed at the knee joint is approximately 135 degrees. Post-exercise recovery involved 40 min sitting in a relaxed posture with the back supported. The chair was composed of a fixed back rest set at an angle of 90 degrees to the horizontal seat pan. Only the lower portion of the thoracic spine was supported by the back rest.

Employing the same stadiometer used by both Wilby et al. (1987) and Foreman (1988), measurements of stature were taken on entering the laboratory, after 10 min in the Fowler position, at the end of the exercise period, and after 20 and 40 min of the recovery period. The same routine was repeated at two specified times of day (09:00 and 18:30 h) and on two separate days. At least three days separated the two days of testing to accommodate for the possible effects of muscle

soreness, the order of each session being randomised. Subjects performed the exercise in bare feet on standard gymnastic tumbling mats laid on a concrete floor. During each repetition of the bounding exercise the subjects were allowed a 30 s recovery. In total, the average exercise period was 11.46 min. Prior to and during the exercise, each subject was encouraged to achieve a maximum horizontal distance for each bound.

4.1.3 Results

Stature alterations following each activity and for each subject are shown in Table 4. Figure 7 shows the mean alterations in stature during both the morning and evening sessions. Mean stature alterations resulting from the pre-exercise and exercise behaviour at both times of day were similar. The recovery trend was less consistent. The Fowler position caused all subjects to increase in stature by an average of 1.33 and 1.25 mm, at 09:00 and 18:30 h, respectively. The bounding regimen caused similar losses in stature of 2.19 and 2.32 mm, at 09:00 and 18:30 h respectively. At both times of day there was evidence of spinal recovery during the first 20 min sitting; recovery was greater in the evening than in the morning, 1.64 and 0.36 mm respectively. The second 20 min period of sitting caused a decrease in stature at both times of day. Again, the evening period caused the greater loss. A mean shrinkage of 0.50 and 0.04 mm occurred in the evening and morning, respectively.

Table 4. Changes in stature for each subject during the experimental period.

SUBJECT	ACTIVITY							
	Fowler position		Bounding exercise		Initial 20 min recovery		Final 20 min recovery	
	Morning	Evening	Morning	Evening	Morning	Evening	Morning	Evening
A	+1.25	+0.05	-2.40	-2.45	-0.72	+2.13	-0.30	-0.53
B	+0.47	+2.09	-1.40	-2.52	-0.46	+3.45	+0.72	-1.00
C	+2.22	+2.10	-3.04	-2.33	+1.54	+0.36	-0.49	-0.16
D	+1.38	+0.75	-1.93	-2.00	+1.09	+0.61	-0.07	-0.29
- x	+1.33	+1.25	-2.19	-2.32	+0.36	+1.64	-0.04	-0.5

NB: + denotes increase in stature

- denotes decrease in stature

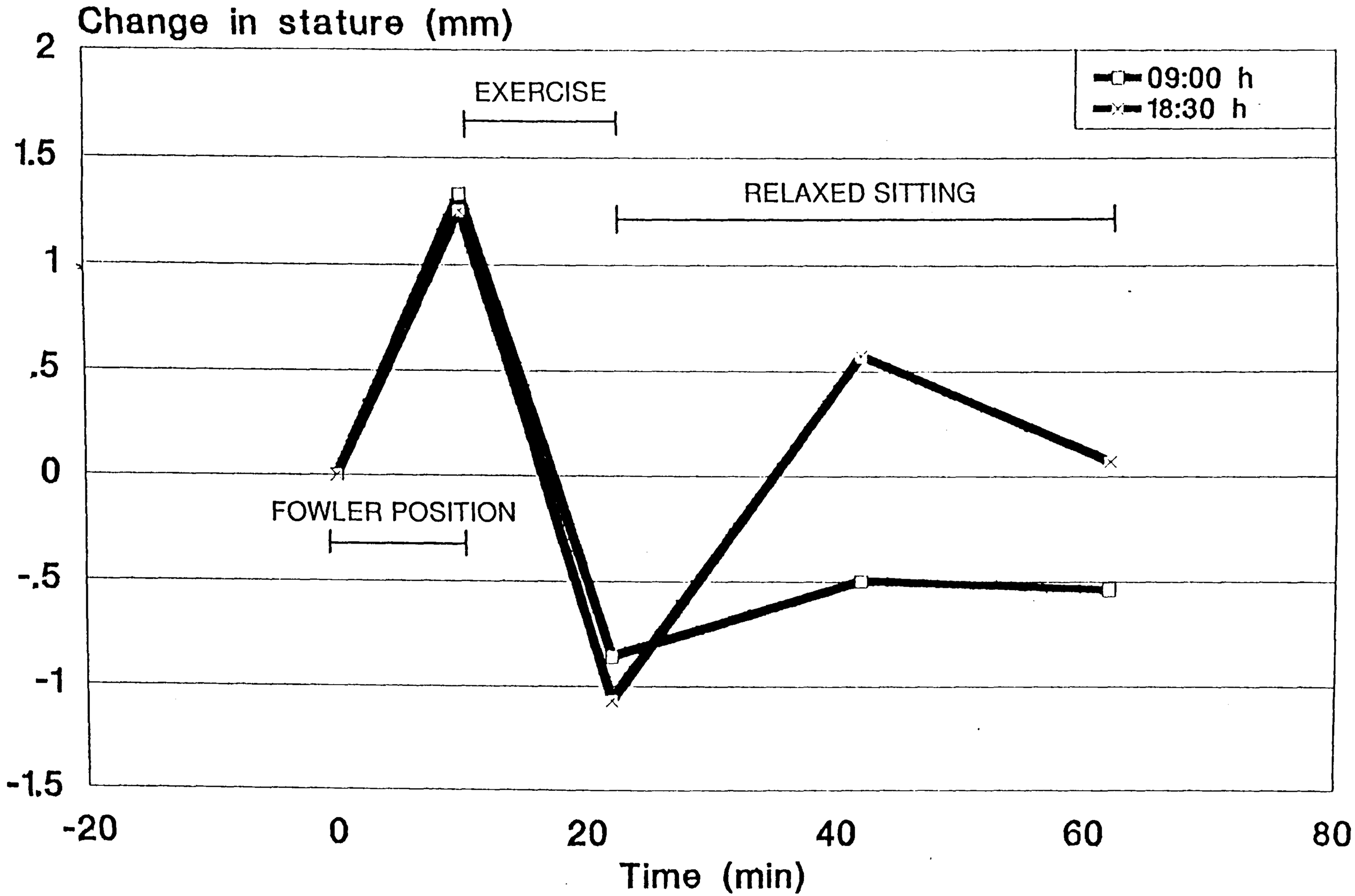


Figure 7. Mean stature alterations during the morning and evening sessions.

Although the sample size was small, making statistical comparison difficult, it was noticeable that intersubject variations were high for all measurement sessions, with the exception of the shrinkage occurring following the exercise regimen (refer to Table 4).

4.1.4 Discussion

Inferences relating to the findings of this initial pilot study possess the normal limitations associated with data derived from small samples. The primary purpose of the study was to gain experience of the apparatus and technique used to record small changes in stature as an index of spinal loading. However, it served as an initial footing towards formulating hypotheses about human spinal responses to plyometric activities and unloading procedures.

The short duration bounding regimen studied during this experiment did cause a loss in stature, results suggesting that exercises of this nature increase the physical load acting on the spine. Stature gains during an initial period of unloading were shown to be quickly negated as a result of the exercise.

Time of day was not shown to influence shrinkage patterns during the initial pre-exercise and exercise periods. However, post-exercise seated recovery was shown to differ, with stature regains being greater in the evening. These discrepancies may well be the consequence of the small sample size and large intersubject variability recorded, the nature of the seated task and its effectiveness as a unloading procedure, or possibly the mechanisms responsible for the intervertebral discs' creep and recovery. Although subjects were asked to relax against the back rest, it is unlikely that they were able to maintain a constant, relaxed posture throughout the seated period. Slight variations in posture will have affected the rate at which stature was regained.

Both the pre- and post-exercise periods were of a relatively short duration in comparison to the seated recovery. Brinckmann (1988) stated that the visoelasticity of the disc may have a time constant

amounting to several minutes while the time constant for fluid exchange may be in the order of hours. The differences noted in this experiment may reflect these mechanisms and their prominence.

The potential of bounding regimens to increase the rate of spinal shrinkage suggests that the intervertebral discs' shock absorbing capabilities are decreased and vulnerability towards injury is increased the more disc height is lost. Clearly, exercises performed later in the day would seem to present the greater risk. Further work needs to assess both pre- and post-exercise behaviour as means to minimise the risk of incurring injury to those athletes taking part in bounding regimens as part of their exercise programme.

4.1.5 Summary

1. The results indicated that a bounding regimen of short duration does induce spinal loads which can be measured using spinal shrinkage as the criterion.

2. The Fowler position caused increases in stature which support its use as an unloading procedure.

3. There was no difference in the pattern of spinal loading and unloading at the times of day studied. However, the magnitude of stature alteration did differ in the post-exercise recovery period at the different times of testing. Discrepancies may be attributable to the mechanisms underlying disc creep and recovery.

4.2 LIMITATIONS ASSOCIATED WITH THE STADIOMETER

Based on the experience gained during the forementioned pilot study, the limitations associated with the stadiometer and experimental techniques (refer to Foreman, 1988) were as follows.

1. The methods of data collection, analysis and storage were felt to possess inherent limitations. Data collection was presently achieved via a chart recorder with graph paper and pen. Hand measurements were taken from the trace and resultant calculations made. Head disc displacement was monitored over a three second period and its displacement after 1.5 s was taken as a measure of height. During a testing session, three measurements were taken and a mean calculation made. The format of the trace (line thickness, etc.), the short sampling duration and hand calculating procedure could be improved in order to decrease the likelihood of errors, and increase reliability.

   The time consuming nature of data analysis prevented an immediate feedback of performance which could serve an important role during subject familiarisation. The system possessed no time base which was deemed important considering the time dependent nature of stature alterations.

2. The calibration procedure was difficult, time consuming and, again relied on measurements taken from the chart recorder and dial gauge. Calibration at regular intervals ensures that any alterations in strain gauge resistance which may come about through changes in either environment or physical conditions is accommodated. Thus, a simpler and more accurate procedure was deemed necessary.

3. The stadiometer, and consequently the weighing scales, were fixed at a 13 degree angle. Obtaining a true value for a person's weight in such a position was not possible. A variable angle of inclination would accommodate subject variability when trying to obtain a relaxed posture.

4. Fixation of the cervical, thoracic and lumbar control relay was achieved using a nut and bolt arrangement. Difficulty was often encountered when trying to locate and make fine adjustments to their position, critical in controlling the upright posture.

5. The head control relay was moveable only in the vertical plane and some subjects complained that the head position was uncomfortable leading to neck ache, particularly after a training period.

6. The chart recorder was unable to cater for the full 50 mm movement of the measuring disc; thus, repositioning of the height displacement unit was often undertaken. In these instances adjustments had to be made to the calculations, which were felt could induce additional errors.

7. The stadiometer could only accommodate subjects between the ranges of 155 to 188.5 cm, thus limiting the range of the population that could be studied.

8. Although the stadiometer could be dismantled for transportation, the tubular steel proved to be heavy and cumbersome, restricting the speed and ease with which it could be relocated.

9. The operator was often required to view a small display panel directly in front of the subject when locating and making fine adjustment to the postural control relays. The present arrangement hindered such an operation.

10. On a number of occasions, external light sources and insufficient displacement between infra-red emitter and photodiode sensor were noted to influence the precision of the head alignment system.

11. Whilst the subject was in the stadiometer, two aspects of a subject's postural requirements were found to present difficulties; leaving the initiation of the weight control relay to the latest possible moment and folding the arms across the chest. The 'weight-control' switch would often be initiated last as it was the easiest switch by which to control the activation of the pen-recorder. However, this could lead to difficulties with some subjects complaining of discomfort in the calf muscles through their prolonged attempts to offset their body weight away from the normal, relaxed position. In addition

some subjects were observed to slightly extend their body upwards, which appeared to stem from the increased muscular tension within their lower limbs. This directly influenced their ability to activate all postural control relays which had been positioned for relaxed standing. Varying degrees of shoulder rotation were noted when the arms were folded across the chest. This influenced the effectiveness of the thoracic postural switches' ability to control upper body rotation.

4.3 ADAPTATIONS TO THE STADIOMETER

The limitations identified previously led to the following developments.

1. Major changes were made to the procedure used to collect, process, analyse and store data. Like the previous stadiometer, two back-to-back, 2 mm steel foil strain gauges (RS ref: 632-124) mounted on a spring riding directly above the head disc, provided, via a strain gauge amplifier (RS ref: 308-815), a linear voltage relating to the head disc displacement. The voltage, instead of being feed to a chart recorder, was passed directly to one channel of the analogue-to-digital convertor of a BBC microcomputer (Master Series). The output from the strain gauge amplifier was such that it gave a maximum value of 1.5 V for full head disc displacement (50 mm), a voltage within the maximum (1.8 V) capabilities of the BBC's analogue input. A software package was then developed to process, analyse and store the digital signal. Figure 8 shows a schematic diagram of the collection, transfer and processing route used by the stadiometer. Program flowcharts for the software are shown in Figures 9 to 15.

   The software incorporated storage on floppy disc and hard copies of the position of each postural control switch for each subject, collection, storage and visual display of head disc displacements during a training and test period, and a visual display and hard copies of training and test data. A feature unique to the training display routine was a graph of continually changing standard deviation values for consecutive

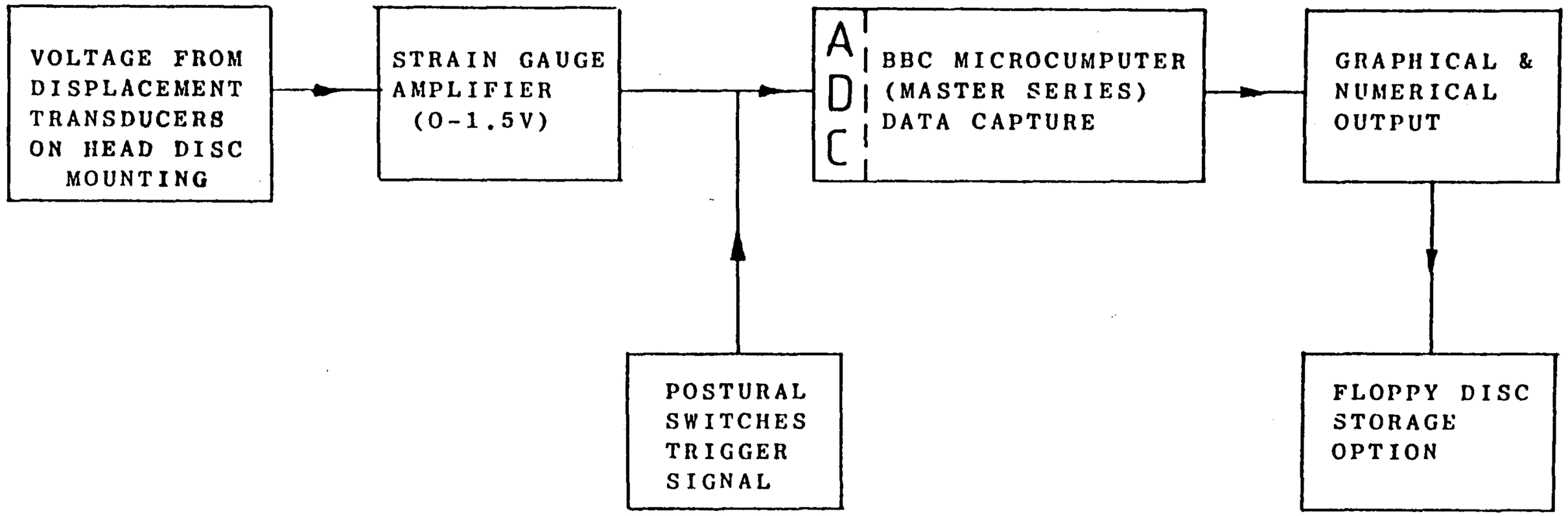


Figure 8. Stadiometer: Data collection, transfer and processing route.

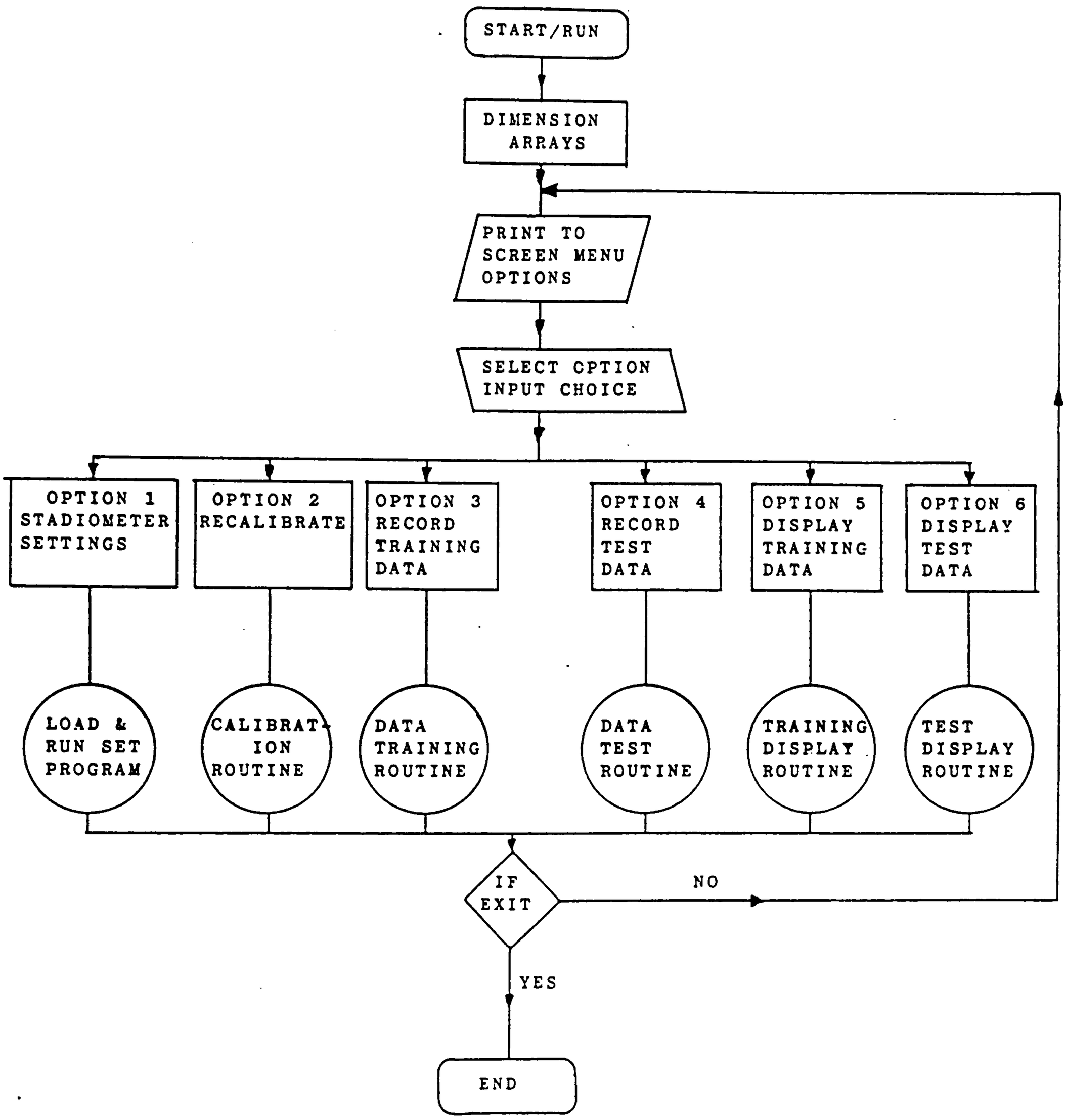


Figure 9. Stadiometer program flowchart: Main menu program.

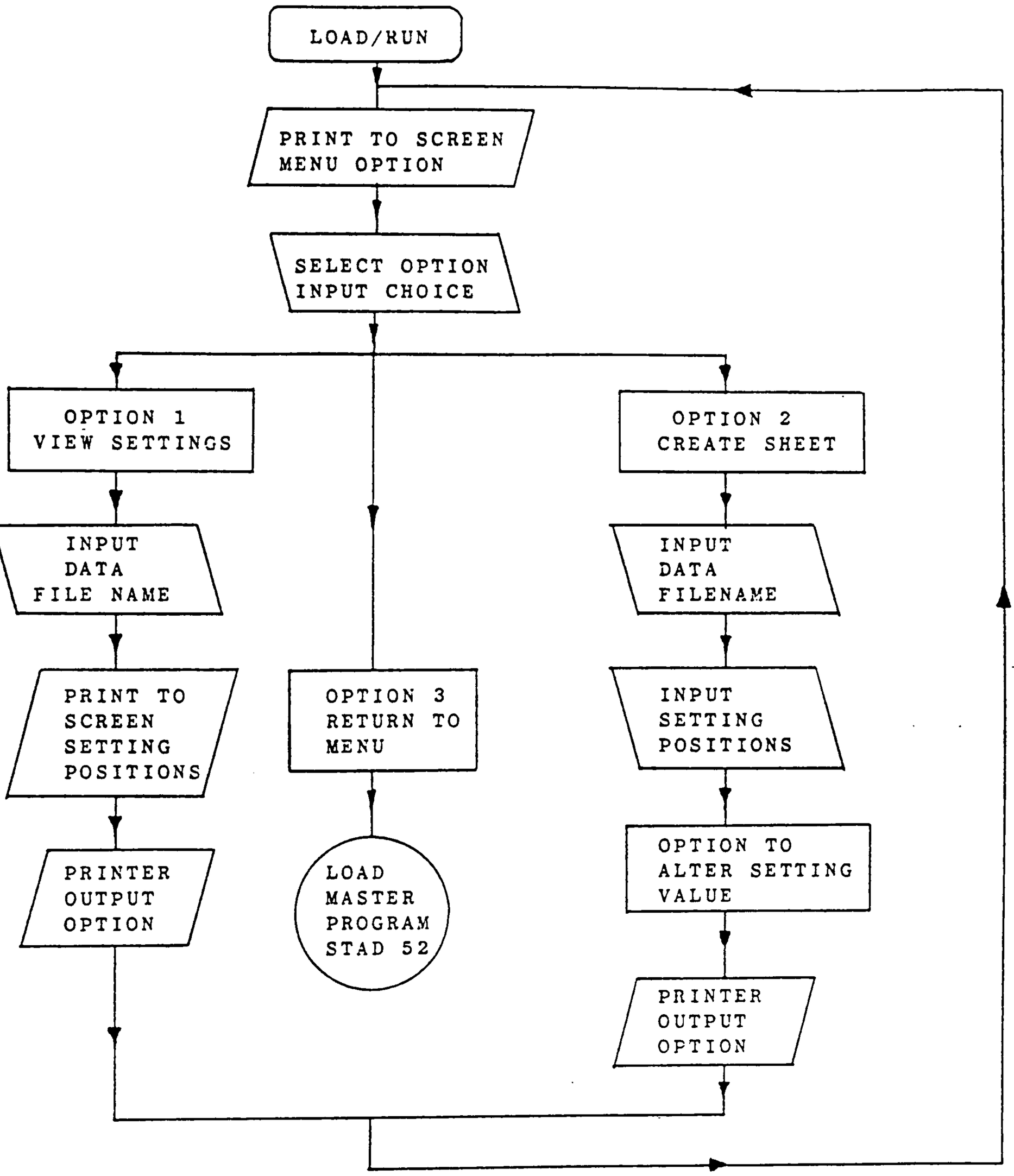


Figure 10. Option 1: Stadiometer settings program.

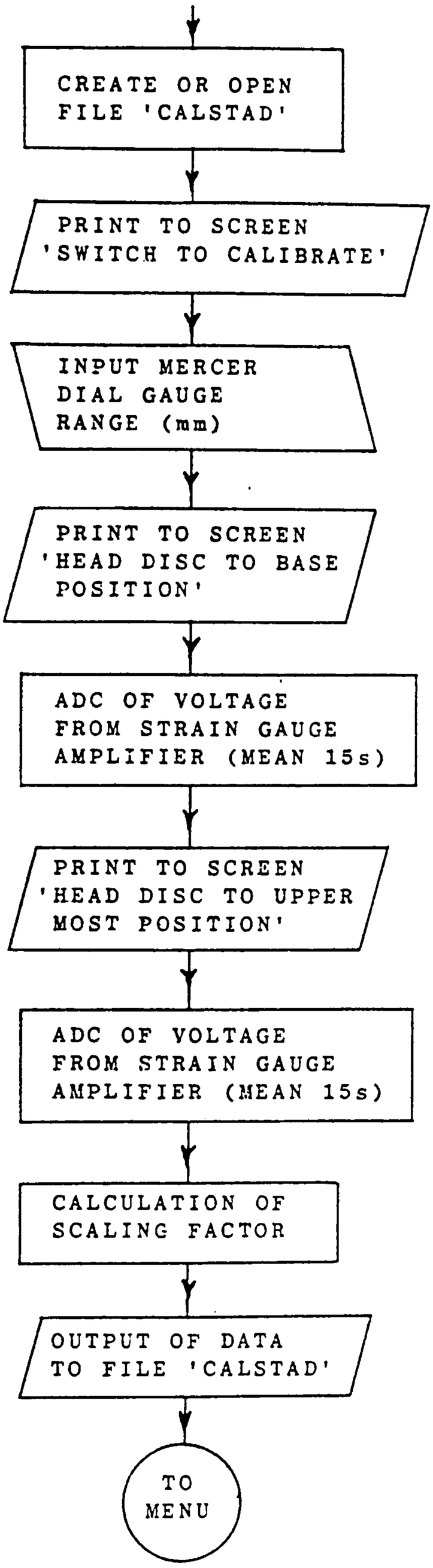


Figure 11. Option 2: Recalibration routine.

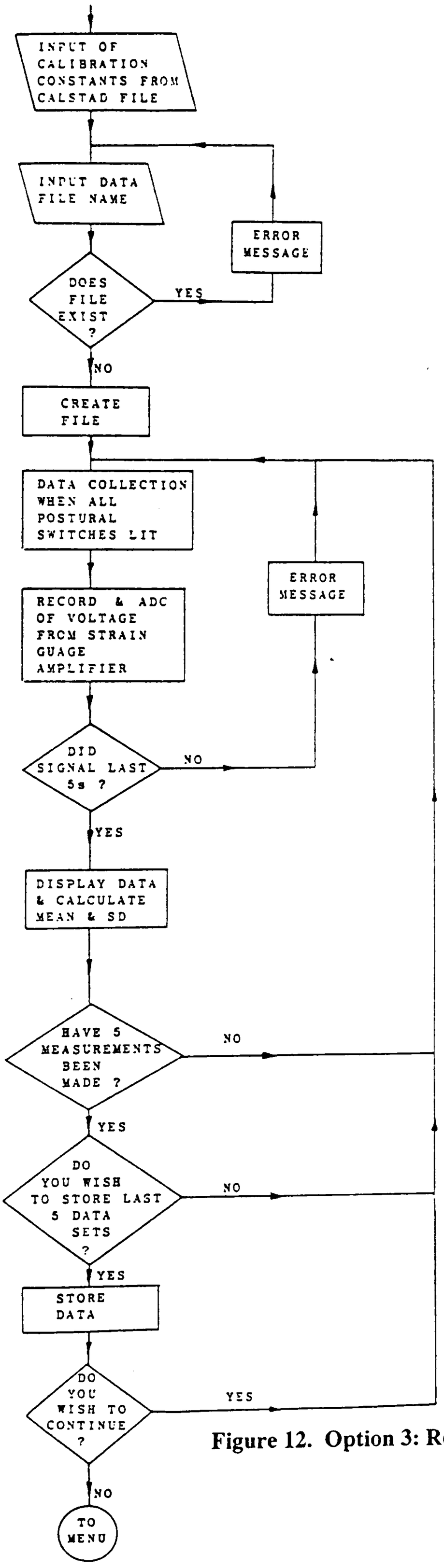


Figure 12. Option 3: Record training data.

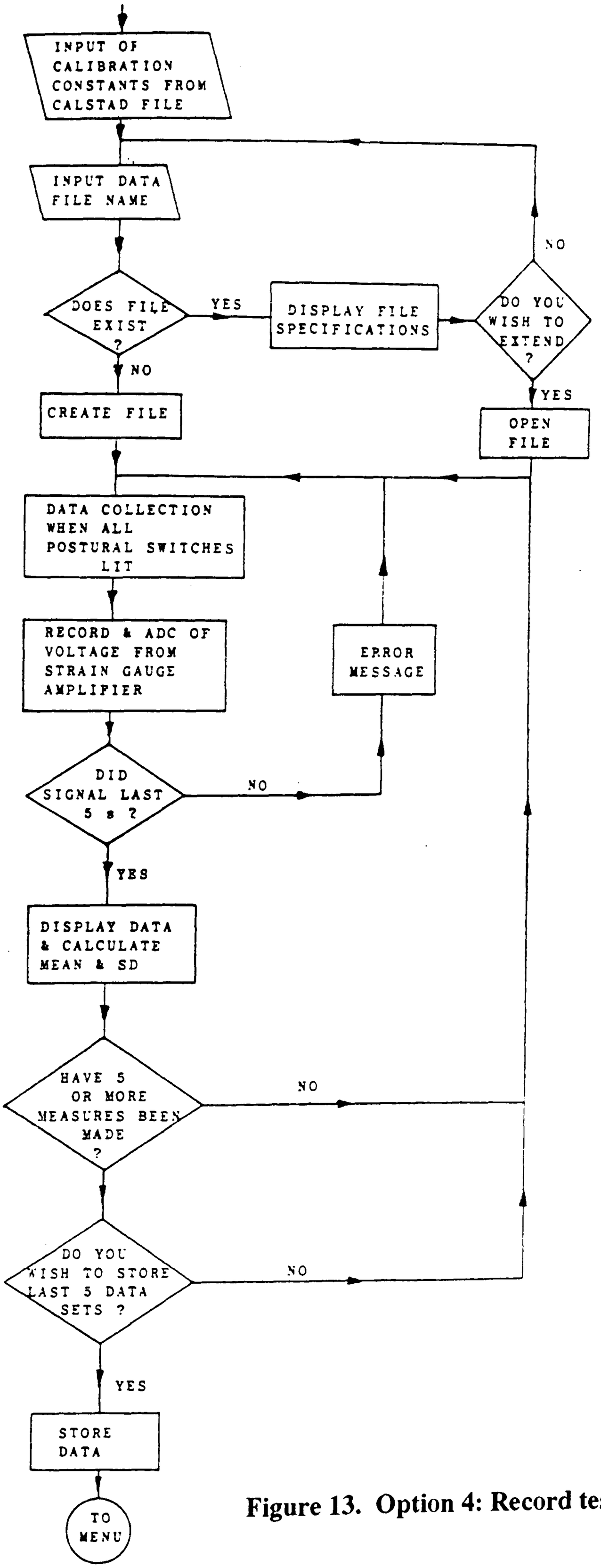


Figure 13. Option 4: Record test data.

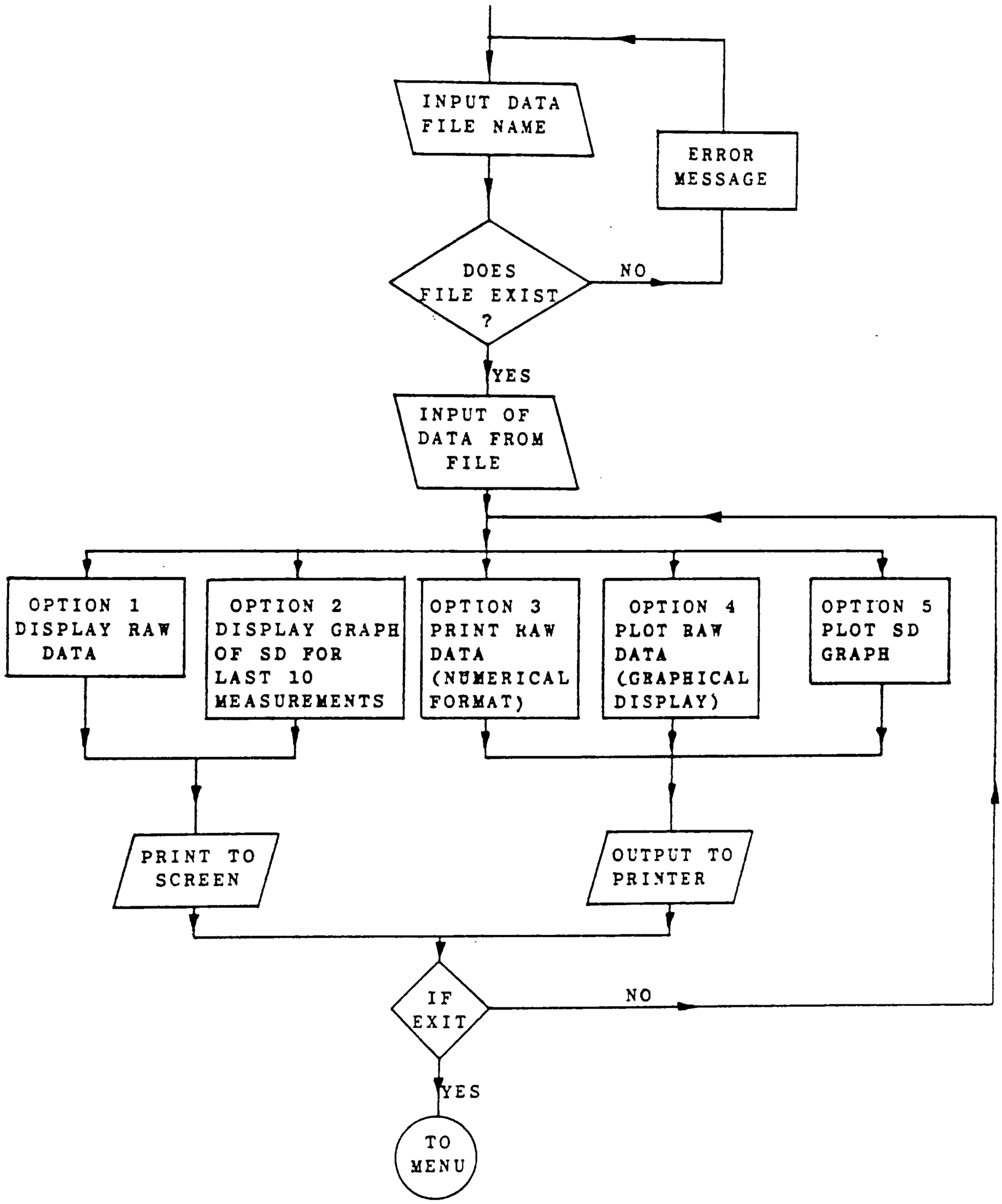


Figure 14. Option 5: Display training data.

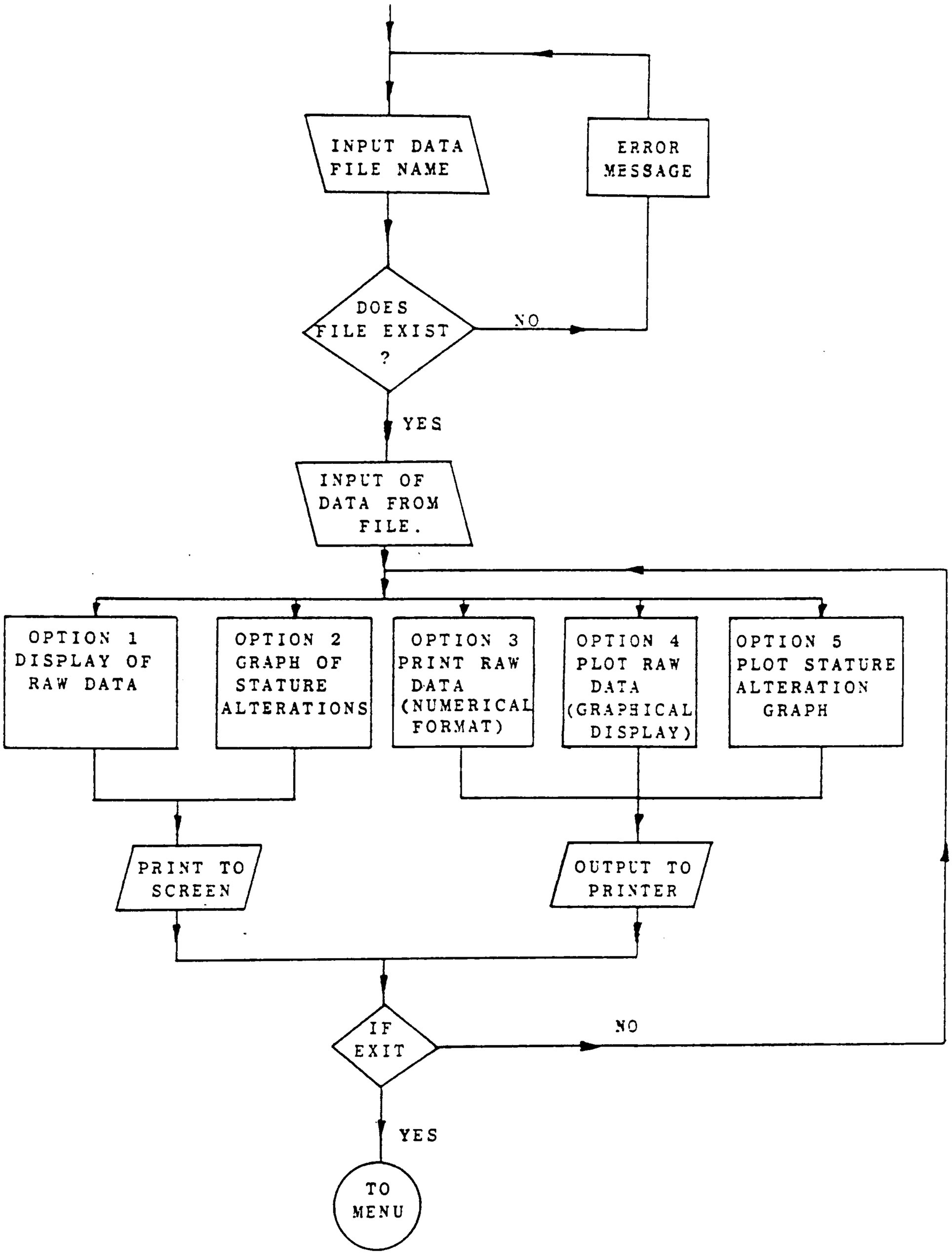


Figure 15. Option 6: Display test data.

measurements. This served to provide instantaneous feedback of a subject's ability to attain a repeatable posture. The testing display routine allowed a graph of mean changes in stature for each measurement period of the test session. The built in time clock of the computer provided a continuous time base.

During data collection, the signals from the strain gauge amplifiers were sampled every second, for a period of 5 s, once the required posture had been achieved and all control switches activated. Each data point was instantaneously displayed on a visual display unit in a bar-chart format. A mean value from the five data points was calculated and displayed, along with mean and standard deviations for all previous measures. During the collection of test data, five consecutive measurements were necessary, two more than had previously been used. This was intended to increase the reliability of the height measurement and provide sufficient time for heel compression to take place (Foreman, 1988). If noticeable heel compression was observed during the first two measurements they were disregarded and two further recordings taken. Storage of information was possible using the floppy disc capabilities of the computer system. The advanced disc filing system (ADFS) of the BBC microcomputer was used to facilitate an updating of data files.

2. The software included a simple calibration routine which allowed a regular calibration of the strain gauge voltage output and digital conversion. By positioning the head disc at the extremes of its movement and sampling for a period of 20 s, the digital range could be calibrated against the displacement recorded by the Mercer dial gauge. The software derived a calibration factor to convert digital signals into displacement measures. This meant that calibration could be achieved within approximately 1.5 min.

3. To allow the apparatus to be positioned vertically and at various angles of inclination, the stadiometer was composed of two main structures - a central support frame, carrying a vertical pillar with postural control switches, on which subjects stood, and a base frame. A central pivot rod passed

through both the base frame and central support, enabling the latter to pivot. Support struts on the central frame were used to fix the angle of inclination. Thus, the central body of the apparatus could be positioned either in the vertical or at an angle between 10 and 15 degrees from the vertical.

4. The eight postural control relays were maintained but adaptations were made to the mechanisms by which certain ones were adjusted to suit individual body contours. Head, cervical, thoracic and lumbar switches were vertically located using a spring locking device, with 1 mm spaced teeth on the rear of the central pillar and within the body of each switch. Fine adjustments within the sagittal plane were possible due to a screw mechanism, which could be easily rotated by hand.

5. The head control relay was altered so that it could be positioned within the sagittal plane, allowing the head to be supported away from the central pillar. An anterior linear travel capability of 7.5 cm, away from the central pillar, was possible.

6. Adaptation to data collecting equipment (BBC microcomputer) meant that full range displacement of the head measuring disc could be catered for within the hardware of the amplification system and software of the computer. This, eliminated the need to relocate the position of the height displacement measuring unit once it had been set.

7. Vertical positioning of the head measuring unit was increased to allow the measurement of taller individuals. The new stadiometer could accommodate subjects between 155 and 210 cm.

8. The steel framework was replaced with tubular aluminium, thereby, maintaining the necessary rigidity but decreasing the weight of the device and enhancing the ease of transportation.

9. All necessary electrical circuitry was housed within an amplification unit, 50 cm in length, 15 cm in height and 32 cm in depth. This was separate from the stadiometer and positioned alongside the computer system. The front of the amplification unit possessed power supply indication lights, the weight balance control switch, a display of the analogue output from the strain gauges, a switch for calibration purposes and a body diagram with 8 light emitting diodes (LEDs), corresponding to each postural control relay. The latter enhanced the ease with which the operator could assess and adjust the position of each postural control relay.

10. The infra-red head alignment was changed to incorporate a high powered infra-red emitter (RS ref: 635-296) on the spectacle frames and photo-diode sensor (RS ref: 307-913) on the adjustable control panel in front of the subjects. These were coupled to a 351 operational amplifier comparator. An adjustment was made to compensate for ambient lighting conditions so as to give maximum gain, allowing a distance of 15 cm to be attained between emitter and sensor, improving the precision of the system.

11. Prior to measurement of stature, subjects were asked to activate all postural control relays. They were then instructed to break the infra-red head alignment by very slightly rotating the head to either the right or left. Maintaining the relaxed posture, they were then instructed to re-activate the head alignment relay and maintain the posture until a buzzer had stopped sounding. Such a procedure was found to enhance a subject's ability to relax within the apparatus.

Subjects were asked gently to clasp the hands in front of the body below the waist. This was found to reduce and control the variation in upper body rotation found when folding the arms across the chest.

The newly constructed stadiometer is shown in plates 1 to 7. It was built at the Bioengineering Department of the Royal Liverpool Hospital by Mr K Linge. I was responsible for the development of software and testing protocoles.

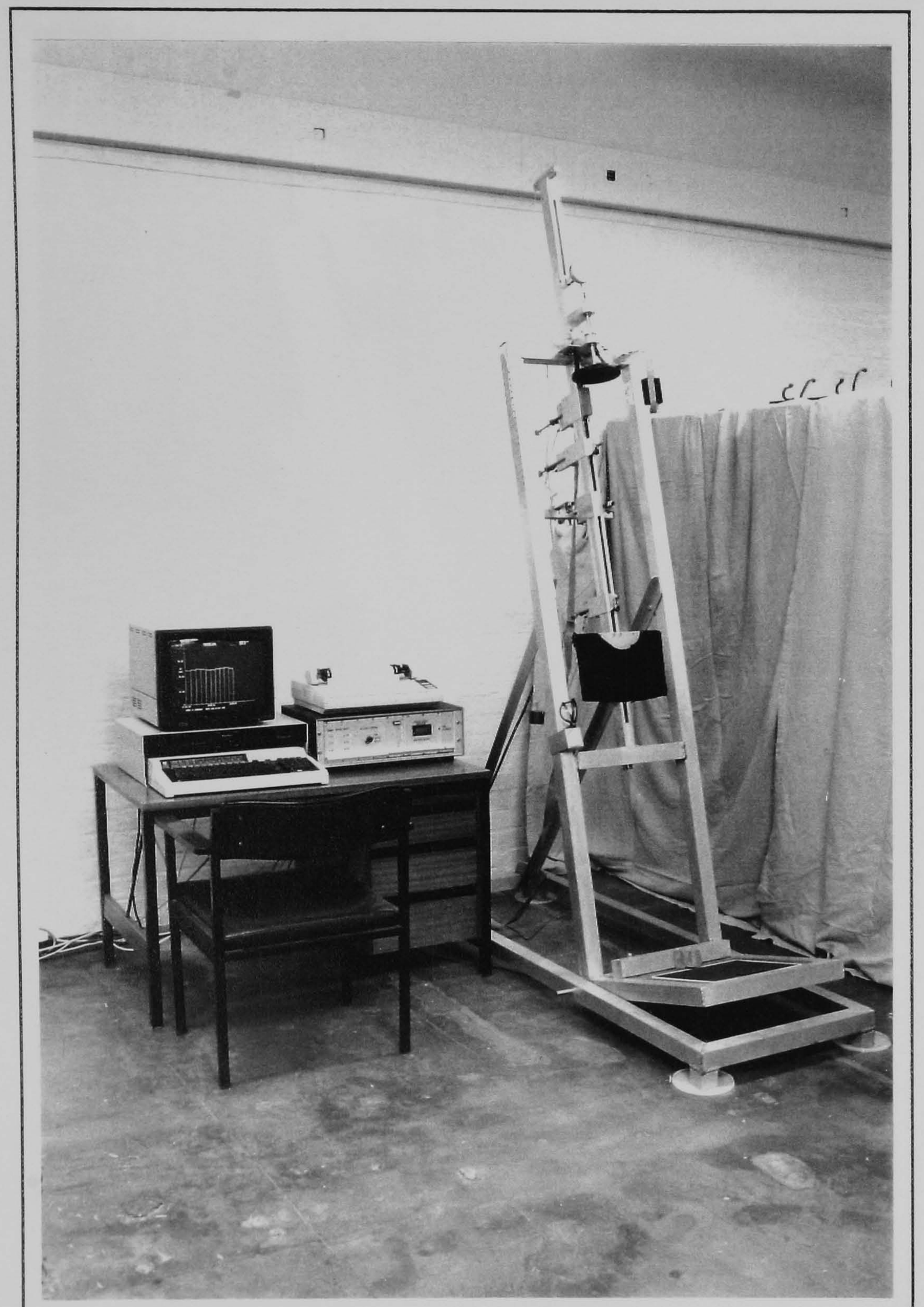


Plate 1 - The newly constructed stadiometer with accompanying control and computer system

Plate 2 - A subject positioned within the newly constructed stadiometer

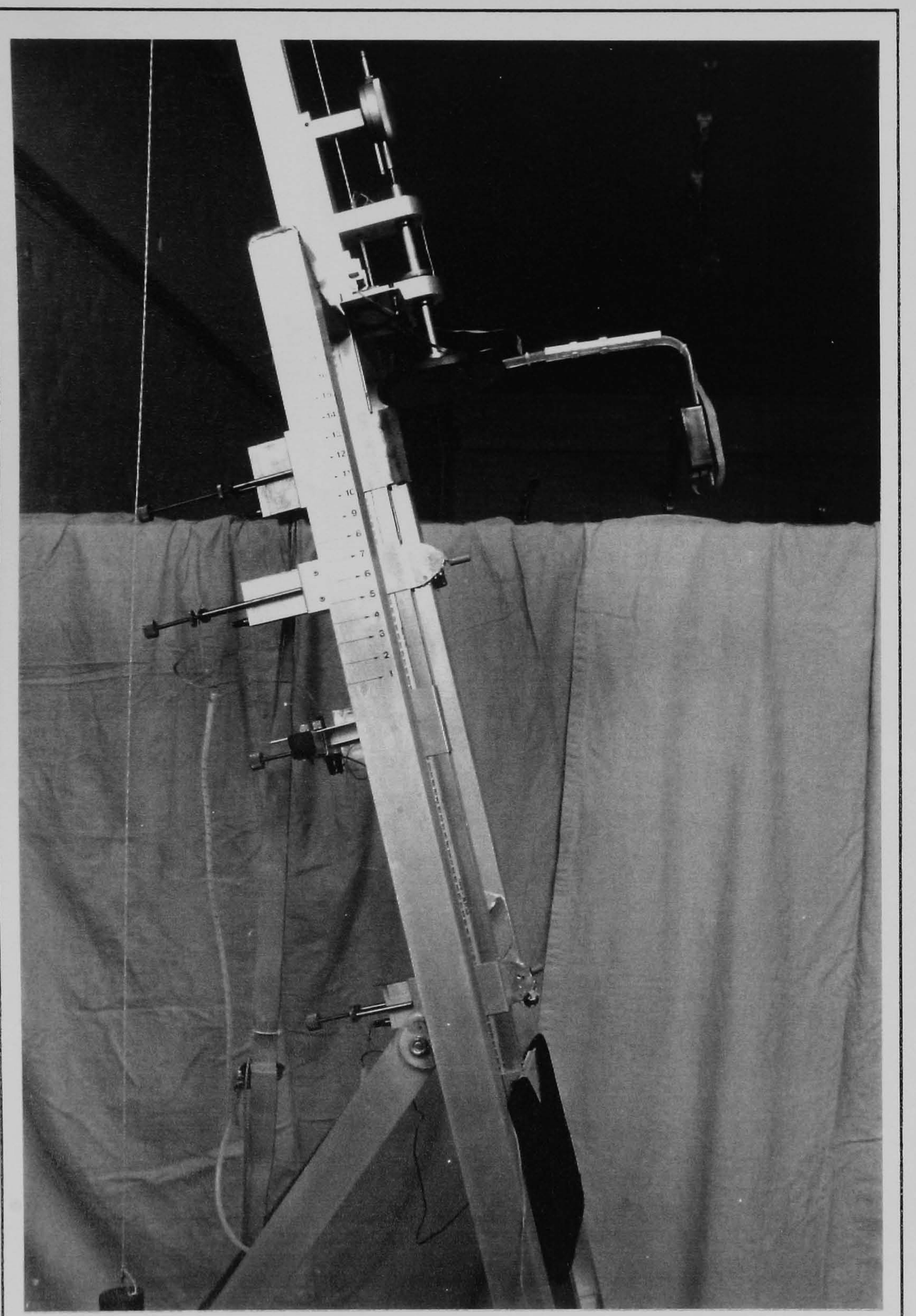


Plate 3 - A side view of the stadiometer showing postural control relays

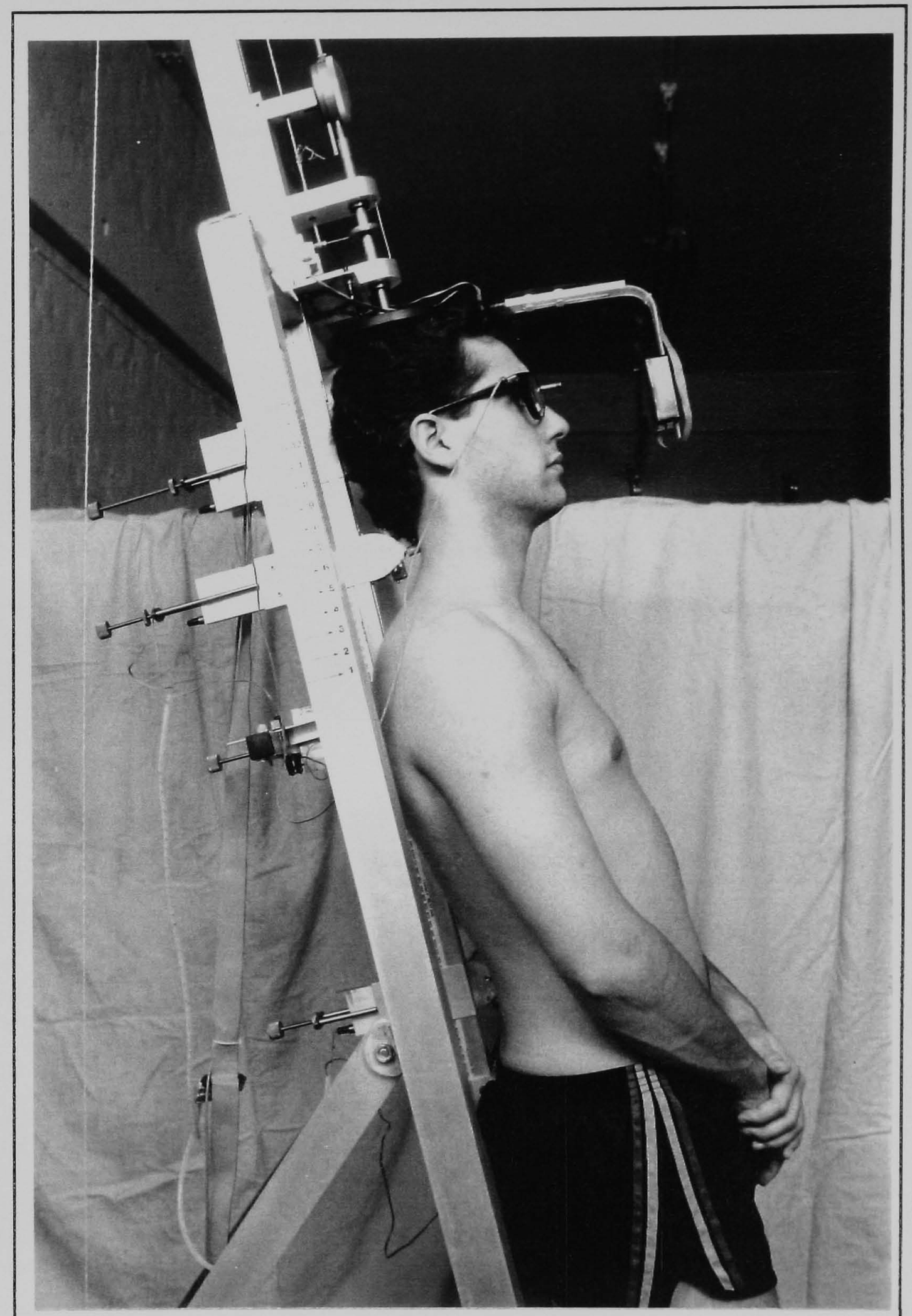


Plate 4 - A side view of the stadiometer with a subject standing in the apparatus

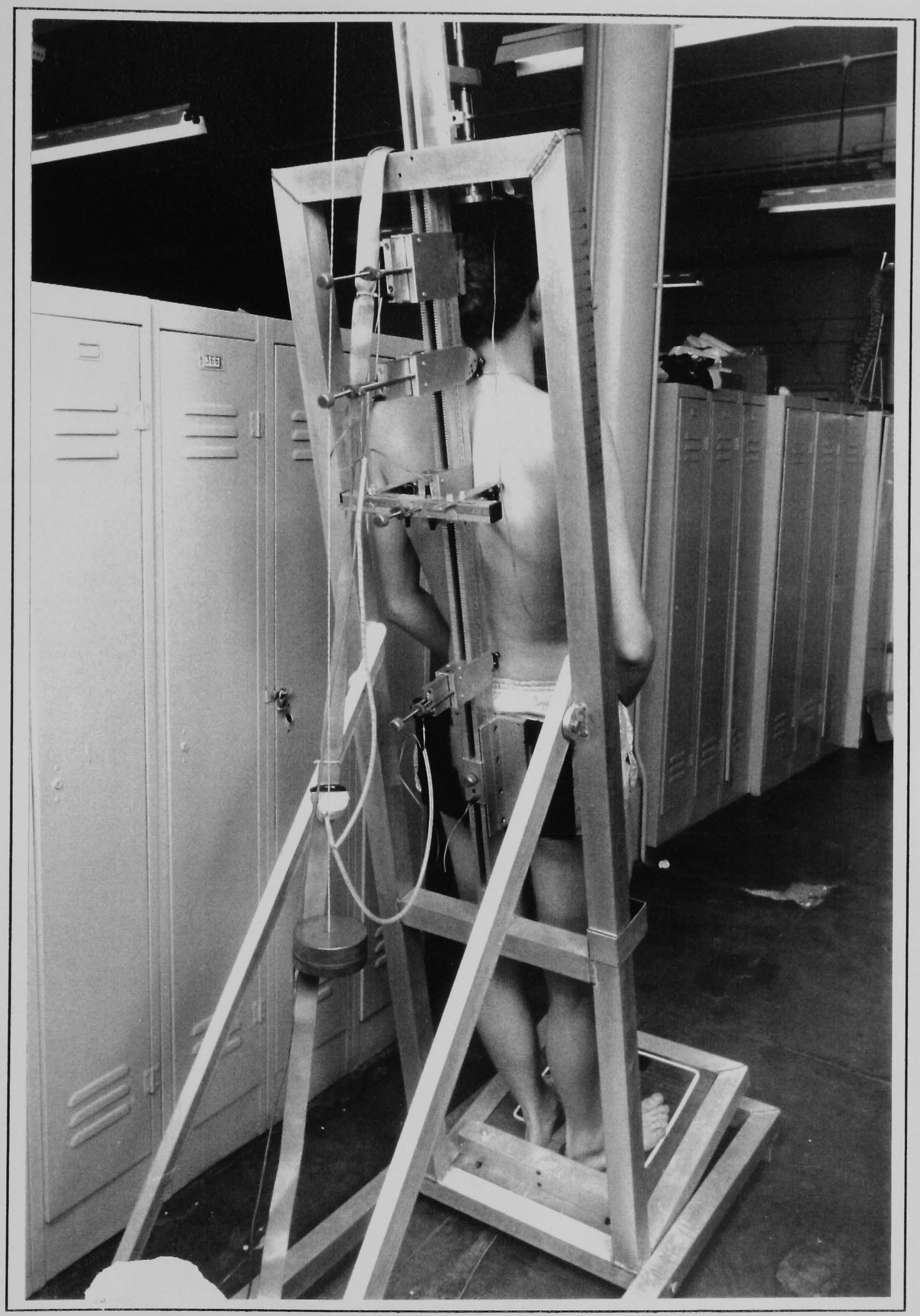


Plate 5 - A posterior view of the stadiometer

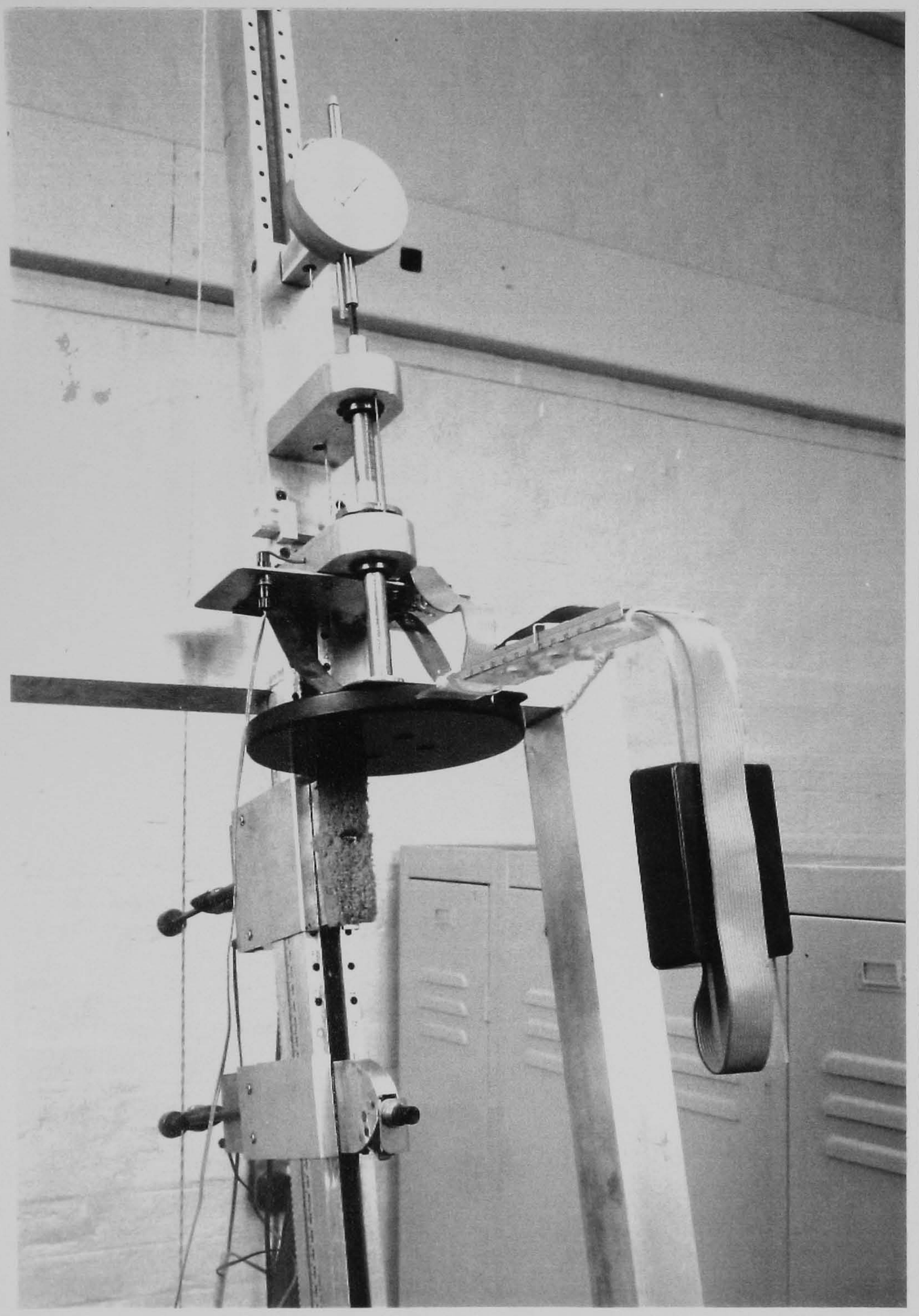


Plate 6 - A close- up of the head gantry

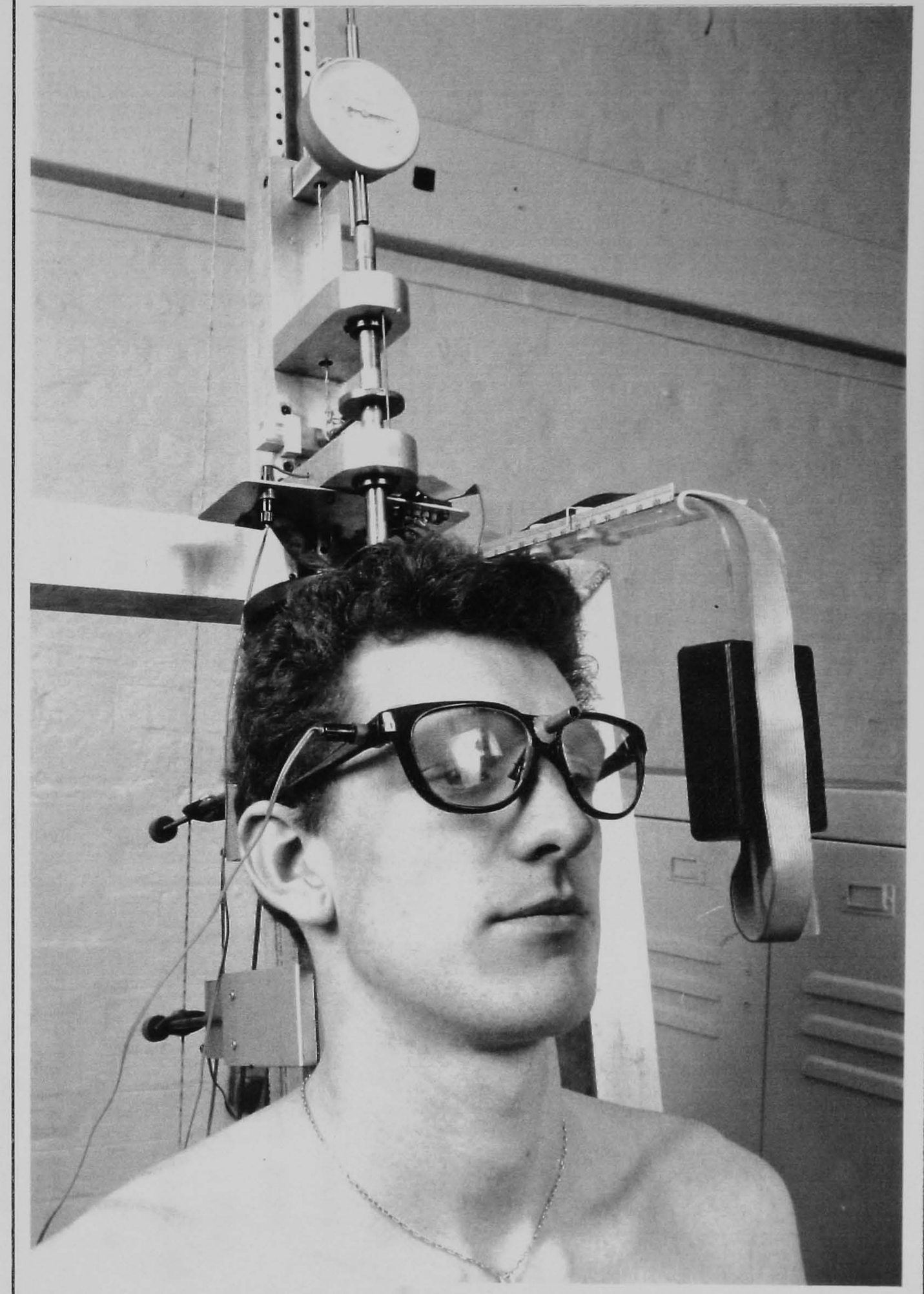


Plate 7 - A close-up of the head gantry with a subject wearing the head alignment spectacles

4.4 ASSESSMENT OF THE HEAD MEASURING UNIT

4.4.1 Introduction

To assess the validity and reliability of the strain gauge system for measuring head disc displacement, a number of repeated tests were conducted. These sought to examine the relationship between the strain gauges derived measures of displacement and those recorded directly from the Mercer dial gauge.

4.4.2 Method

Five second samples of the digital signal were obtained for each 1 mm movement of the head disc, as recorded by the Mercer dial gauge (sensitive to 0.01 mm). Fixed 1 mm increments in the head disc displacement were achieved using a clamp and fine adjustor. The experiment protocol was repeated on ten occasions over a period of ten weeks, during which time the stadiometer was being used on a regular basis. Prior to any measurements, the stadiometer was switched on for at least one hour and the standard calibration procedure undertaken.

Statistical analysis assessed the linear relationship between measured displacement from the Mercer dial gauge against the digital signals from the strain gauges. For each 1 mm displacement, mean and standard deviations for the ten trials were compared.

4.4.3 Results

All correlation coefficients for each of the ten tests were highly significant ($r = 0.99$, $p<0.001$). A comparison of 1 mm displacements showed that for only the initial 4 mm of travel, when in the uppermost position, and during the last 3 mm of travel, towards the bottom of its movement range, did mean values exceed 1.1 mm or were less than 0.9 mm, respectively. Mean values for head disc displacement at the uppermost point of travel suggest the strain gauges slightly over-predicted movement and towards the bottom of

travel, under predicted movement. Standard deviations were always less than 0.1 mm for the middle 30 mm of travel.

4.4.4 Conclusion

The strain gauge system for recording movement of the head disc was found to be sensitive for measuring small displacements and was linear over its full range of movement. During the middle 30 mm of the head disc travel displacement measures were found to be reliable to 0.1 mm. It was concluded that when setting the position of the head measuring unit, the initial displacement of the head disc should be within the middle 30 mm of its total travel capabilities.

5. EFFECTS OF GRAVITY INVERSION ON EXERCISED-INDUCED SPINAL LOADING

5. EFFECTS OF GRAVITY INVERSION ON EXERCISE-INDUCED SPINAL LOADING

Aspects of this work were published in the journal Ergonomics, 1988, vol 31, No 11, pp1631-1637, a copy of which can be found in appendix 2.

5.1 INTRODUCTION

Static and dynamic loading of the upper torso with weights have been shown to accelerate the normal rate of stature loss (Corlett and Eklund, 1986; Tyrrell et al., 1985). Few studies have considered the transmission and ultimate effects on the spine of forces encountered during ground impacts. Combining these forces with the intense muscular exertions common to many sporting activities, such as plyometric exercises, would suggest that these activities have a high risk of incurring lower limb and back disorders. It was hypothesised that:

Hypothesis 1: A short duration bounding regimen places a mechanical load on the spine and leads to an incurrence of spinal shrinkage.

Wilby et al. (1987) identified time of day as an important variable governing the rate of stature loss following an exercise period; greater shrinkage occurred shortly after rising than in the evening. It was hypothesised that as stature decreases over the course of the day due to constant loading of the intervertebral discs, the discs' stiffness characteristics increase inhibiting any future disc height losses. In accordance with the mechanisms reported to control disc height creep and recovery (Koeller et al., 1984), a greater disc load will increase the osmotic pressure and annular stress within the disc. Consequently, on removal of the load, a more compressed disc will undergo a greater rate of disc height recovery.

Hypothesis 2: During a period of recovery, regains in stature are greater for those subjects who have undergone a greater initial shrinkage.

Experience in the initial pilot study (Section 4.1) suggested that intervertebral disc creep and recovery may be dependent not only on the magnitude of the pre-load but also on the time duration of the pre-load conditions. Koeller et al. (1984) have shown that two mechanisms are responsible for disc height changes:- a viscoelastic and a fluid exchange mechanism, the former possessing a much larger time constant. On this basis, the primary disc changes in response to short-duration unloading procedures should reflect alteration in annular elasticity. It would seem reasonable to hypothesise that an immediate period of loading will quickly negate any disc height increases previously observed. It was hypothesised that:

Hypothesis 3: Spinal shrinkage resulting from a short duration bounding regimen is greater after a period of unloading.

The relevance of such findings will be of significance to athletes seeking to structure training sessions to reduce the risk of back injury. In addition, it will serve to aid those employing unloading procedures during the rehabilitation from back pain symptoms.

Broad jumps (standing jumps for maximum horizontal distance) have been recommended as a form of plyometric exercises as eccentric muscle actions are reported to be prominent during their execution. Consequently, in accordance with reported post-exercise symptoms associated with unaccustomed eccentric exercise, it can be predicted that subjects would experience muscle soreness between 24 to 48 h following completion of a broad jump regimen, which peaks between 24 and 72 h and subside within five to seven days post-exercise. Additionally, a repeated bout of exercise should reduce muscle soreness symptoms.

Hypothesis 4: Following a bounding regimen, reported muscle soreness is most prominent between 24 and 72 hours post-exercise.

Hypothesis 5: A bounding regimen repeated within a week of the first execution shows a reduced severity of muscle soreness ratings after the second performance.

5.2 METHOD

Eight males aged 20-26 (mean = 21.75, SD = ± 2.25) years acted as subjects. Mean (± SD) values for height were 178 (± 7.1) cm, corresponding values for body mass being 68.75 (± 5.0) kg. All were physically active without any history of back pain or high blood pressure and none of the subjects had regularly participated in plyometrics as a form of training. Measurements of changes in stature were undertaken using the updated stadiometer set at an angle of 13 degrees. All subjects were trained with the measuring procedure and had on average, four 20 minute sessions, entailing 48 measurements. Prior to conducting any experimental work, each subject achieved a standard deviation for ten consecutive measures of less than 0.5 mm (mean = 0.44 mm, range = 0.42 to 0.47 mm). Subjects were all instructed to refrain from any sporting and stressful physical exercise and keep to as normal a daily routine as possible on the day of testing. A normal daily routine was also specified for the three days post-exercise.

Subjects underwent two experimental sessions on two separate days, with at least five days separating each. This was to control for the effects of residual muscle soreness. Both sessions were conducted at 14:00 hours to control for circadian variation in stature, differing in only one respect - the 10 min pre-exercise. On entering the laboratory subjects were required to stand, with weight evenly distributed, for a period of 20 min in an attempt to standardise and control for any abnormal spinal loading or unloading that may have directly preceded testing. Following this subjects underwent a 10 min pre-exercise procedure. During the first session of testing, this involved standing with weight evenly distributed. On the second occasion subjects were subjected to gravity inversion using the Nissen Sprecklift (Plate 8) set at an angle of 50 degrees to the vertical, the angle reported to cause the greatest rate of unloading (Leatt et al., 1985). Subjects were then required to perform 50

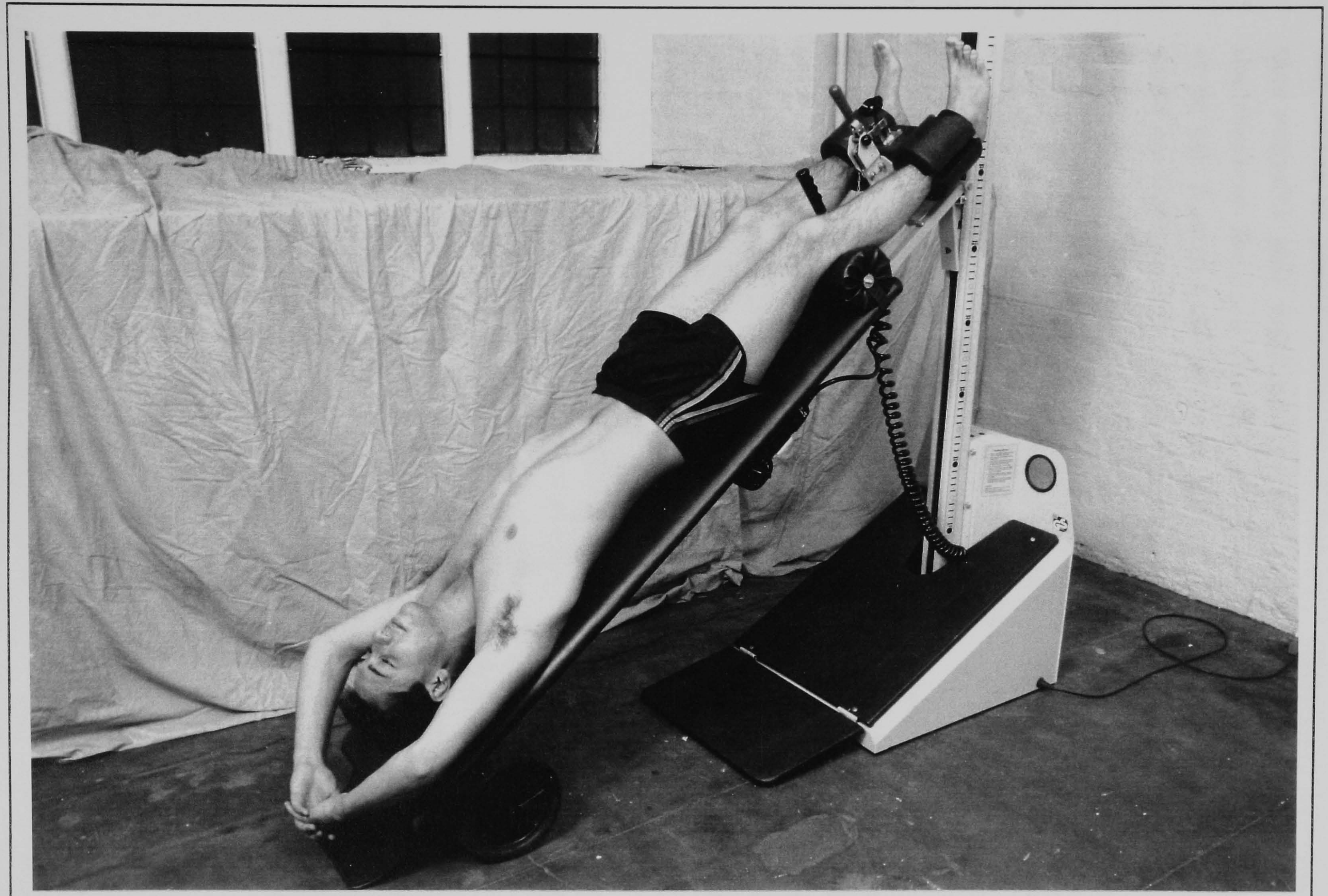


Plate 8 - The Nissen Specklift gravity invertor

standing broad jumps, in sets of five, with 15 seconds recovery between each set. The exercise was performed in bare feet on standard gymnastic tumbling mats laid on a concrete floor. The average length of the total exercise period was 6.7 min. Subjects' breathing had sufficiently normalised within 2 min of exercise completion to permit a measure of stature. Throughout this period of time subjects were relaxing within the frame of the stadiometer, in the required inclined standing position. A standing recovery period of 20 min followed the exercise period. Measurements of stature were taken on entering the laboratory, prior to the pre-exercise period, prior to the exercise period, at the end of the exercise period, and at 5 min intervals during the recovery period. Each measurement period lasted on average 3.2 min providing sufficient time for any heel compression to have stabilised (Foreman, 1988).

During the first and last set of five jumps, three measurements were taken of broad jump performance, with the mean value being recorded. Perceived exertion (RPE) was rated at the end of the exercise period using Borg's (1970) scale, shown in Appendix 3. Subjects were also asked to rate the intensity of any lower back pain using an eleven point scale, shown in Appendix 4. Lower back pain was defined as pain experienced between the mid-back and buttocks.

Following each session of testing subjects were given a questionnaire in order to monitor post-exercise muscle soreness and low-back pain. The low-back pain rating scale was identical to that previously mentioned. Muscle soreness rating involved the use of a visual analogue scale and was defined as the soreness experienced in the upper legs. The extremes of the scale were described as no soreness and severe soreness, a 10 cm line separated each extreme. These two questions were required to be completed at 21:00 and 10:00 hours on the three days following exercise. The questionnaire can be found in Appendix 5.

5.3 ANALYSIS OF DATA

A number of statistical tests were performed using the MINITAB statistical programme on the DEC 20 computer system. Descriptive statistics were obtained for age, weight and stature

before testing commenced, jumping performance, RPE, muscle soreness rating, back pain rating and stature alterations for each regimen of the two sessions of testing. The same variables were subjected to a cross-correlation using the Pearson Product Moment method.

A t-test was performed on the difference in stature alterations for the different regimens during both test sessions and between the regimens for the same test session. Also subjected to the t-test were jumping performances during and between the two test sessions, and RPE for the two sessions.

A two-way analysis of variance was also employed. This incorporated terms to account for interaction between subjects and regimens. In this analysis, overall recovery was considered as opposed to stature alterations for each 5 min recovery period.

A regression analysis was used to determine the predictability of the change in height from easily obtainable data. The predictors used included weight and stature measured before testing commenced, and age. These three variables have been reported in the literature to influence the amount of shrinkage.

5.4 RESULTS

Mean alterations in stature over the entire test period for the two test sessions are depicted in Figure 16. Table 5 displays these same stature alterations, along with standard deviation values, and shows whether differences existed between test regimens or between successive measurements, within the same test regimen. Positive values indicate increases in stature and negative values, decreases in stature. For discussion purposes, session 1 and 2 refer to the test regimens which possessed standing and gravity inversion as the pre-exercise treatment, respectively.

For the two 20 min sessions prior to pre-exercise only small mean increases in stature of 0.1 and 0.35 mm were recorded, the difference being non-significant ($p>0.05$). A further 10 min standing, during the pre-exercise period, caused non-significant decrease in stature of 0.03 mm. In contrast, the 10 min gravity inversion pre-exercise

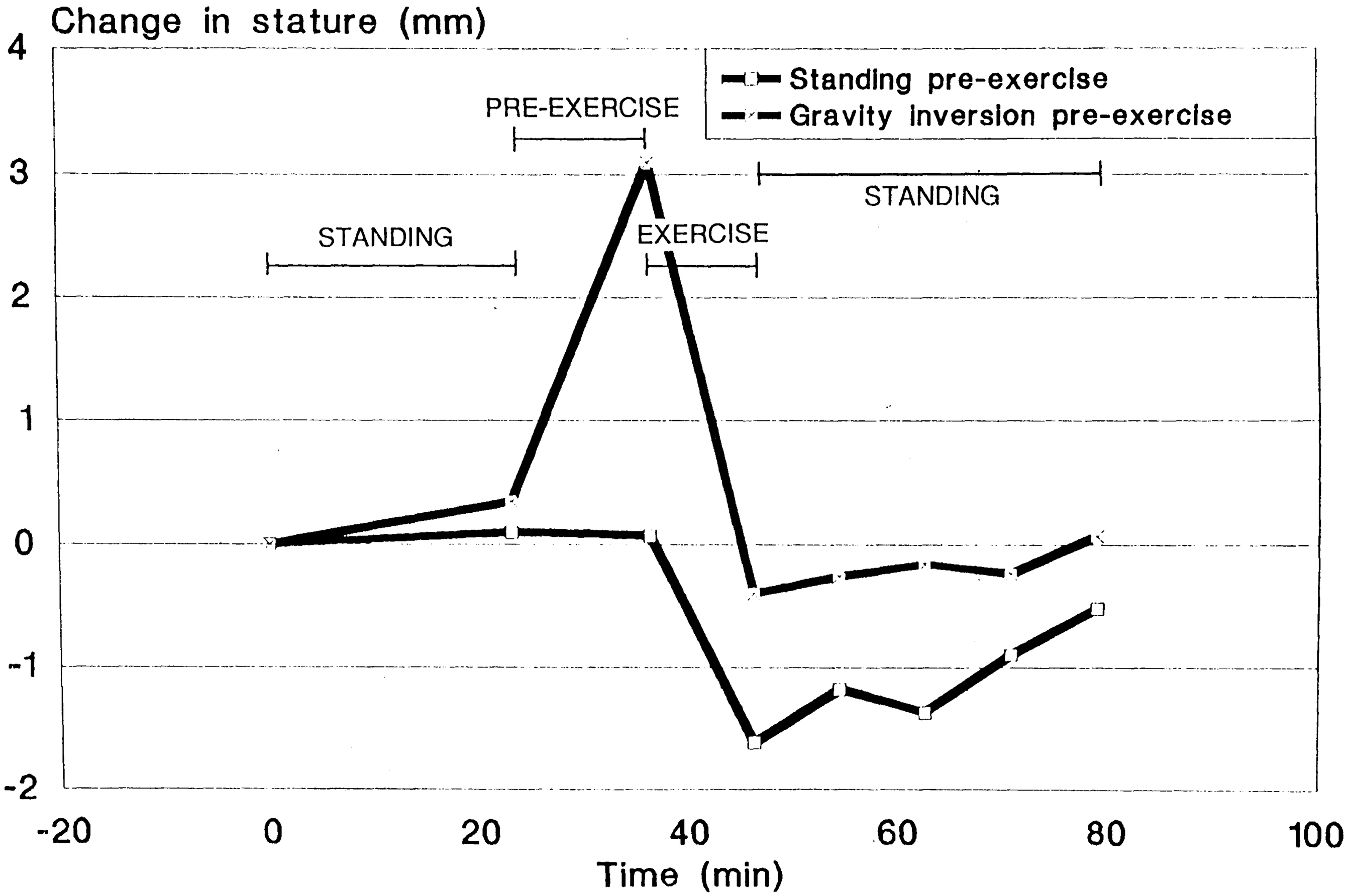


Figure 16. Mean stature alterations for the test regimens involving pre-exercise standing and gravity inversion.

Table 5. Mean (± SD) stature alterations (mm) for the regimens involving two different pre-exercise behaviours prior to a bounding regimen.

Pre-exercise behaviour session	20 min standing		10 min pre-exercise behaviour		Exercise		Recovery (standing)						
							5 min		10 min		15 min		20 min
1. Standing	+0.1 (± 0.7)	*NS	-0.03 (± 0.6)	S^b	-1.69 (± 1.13)	S^b	+0.44 (± 0.94)	NS	-0.19 (± 0.90)	NS	+0.47 (± 0.47)	NS	+0.38 (± 1.04)
2. Gravity inversion	NS		S^a		S^b		NS		NS		S^b		NS
	+0.35 (± 0.58)	+S^a	+2.74 (± 1.28)	S^a	-3.49 (± 1.87)	S^a	+0.14 (± 0.66)	NS	+0.10 (± 0.70)	NS	-0.08 (± 0.70)	NS	+0.31 (± 0.97)

* NS: non-significant
\+ S : significant: a - $p < 0.01$
b - $p < 0.05$

NB: + denotes increase in stature
- denotes decrease in stature

was found to increase stature by 2.74 mm, the difference between regimens being significant ($p<0.01$). When the exercise period directly followed standing, a mean loss in stature of 1.69 mm was found compared to 3.49 mm for the period following inversion. Again, the difference between regimens was significant ($p<0.05$). During the overall 20 min standing recovery there occurred mean increases in stature of 1.1 and 0.47 mm for the control and gravity inversion periods, respectively.

The two-way analysis of variance revealed that there was no significant difference between subjects for stature alterations, and the interaction between subject and regimen was not significant ($p>0.05$). However, during recovery, there was a significant difference in stature alterations between regimens ($p<0.01$).

An examination of the contribution covariates (age, stature and body weight) made to the changes in height, using the multiple regression analysis, failed to produce any significant results ($p>0.05$). Neither age, stature or body weight was related to the amount of shrinkage.

Table 6 provides data on shrinkage, perceived exertion, back pain rating and jumping performance for the exercise period. For the two different days of testing, perceived exertion and jumping distance recorded at the start and end of the exercise period showed no significant difference ($p>0.05$). Jumping performance, on both occasions, was found to increase between the first and final set of jumps ($p<0.05$). The mean increase on each occasion was 0.1 m.

No significant correlations between shrinkage and perceived exertion at the end of either exercise regimen were found ($p>0.05$). A significant correlation, consistent for both sessions, was found for jumping distance and subjects' body weight during the initial and final set of jumps ($r = 0.86$ to 0.92, $p<0.01$). None of the subjects experienced any back pain symptoms following completion of the exercise period, on either occasion.

Post-exercise measures of muscle soreness for the days following exercise are shown in Figure 17. For the first period of exercise,

Table 6. Shrinkage, perceived exertion, back pain rating and jumping performance for the exercise period on the two different days of testing.

	1. Session including standing pre-exercise		2. Session including gravity inversion pre-exercise	
Shrinkage (mm)	1.69 (± 1.13)		3.49 (± 1.87)	
Perceived exertion	13.1 (± 1.5)		13.2 (± 1.7)	
Back pain rating	0		0	
Jumping distance (m)	First set	Final set	First set	Final set
	2.04 (± 0.26)	2.14 (± 0.19)	2.07 (± 0.24)	2.17 (± 0.19)

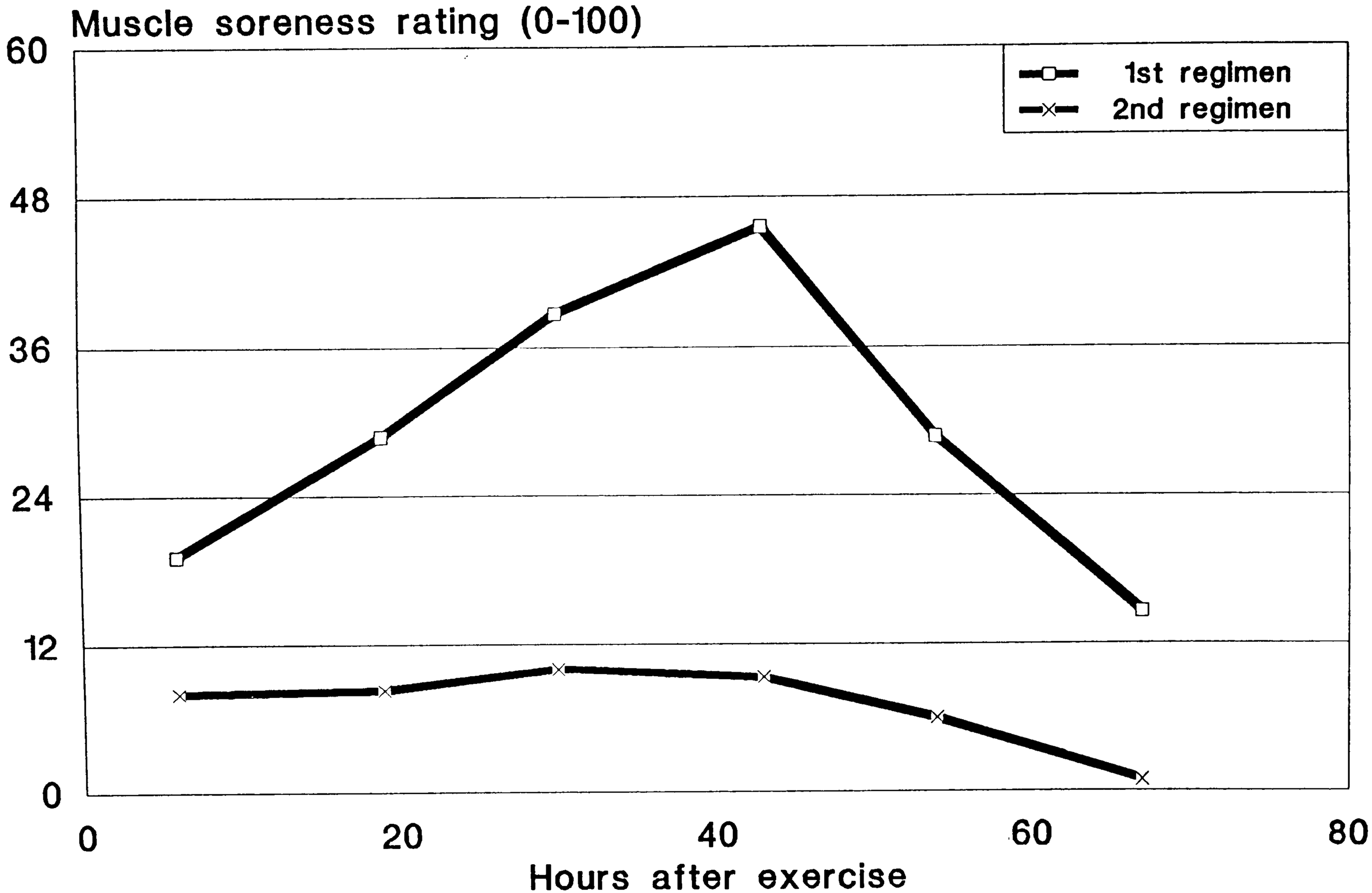


Figure 17. Mean post-exercise muscle soreness ratings scores.

ating scores were found to increase steadily and peak 43 hours after ompletion of the exercise, following which, there was a sharp drop n the muscle soreness experienced. On the second occasion, some ive days later, post-exercise muscle soreness rating values were onsiderably lower, with only the values recorded on the evening of he exercise showing no significant difference. Mean muscle soreness ating values for the second day of testing remained fairly constant or the first 43 hours following exercise, after which time values ere found to drop and approach zero.

our subjects complained of low-back pain during the days following he first exercise regimen and only one following the second regimen. or those subjects, the maximum pain experienced was rated as mild. correlation was noted between mean muscle soreness and back pain atings post-exercise, $r = 0.70$ and 0.74 ($p<0.05$).

.5 DISCUSSION

or both exercise regimens, a 20 min standing period prior to the re-exercise period did not effect height changes, an increase in tature of less than 0.4 mm being recorded. This suggested that ubjects had approached the trough in their normal circadian ariation in body height which has been reported elsewhere (Reilly et l., 1984). A further 10 min of standing had little influence on tature measurements, implying that at the time of testing (14:00 h) nd for the time duration studied, standing neither loaded nor nloaded the spine. In addition, it appeared that the initial 20 min eriod of standing had served to standardise posture.

10 min period of gravity inversion pre-exercise caused a hange in measured height (2.74 mm increase), in line with values eported by Leatt et al. (1985) for the same angle and period of nversion. This provided additional support for its use as a rocedure to unload the spine.

ıe short duration bounding regimen (6.7 min) caused spinal shrinkage ı both days of testing, supporting hypothesis 1 and suggesting that ıe muscular exertions, combined with the impact forces on landing, ıcreased the mechanical load acting on the spine. Pre-exercise

conditions were found to significantly influence the rate of stature loss during the exercise period; twice the magnitude of stature loss occurred when the exercise period followed spinal unloading, leading to an acceptance of hypothesis 3.

Alterations in stature are a reflection of disc height changes. The short duration of the activities and the noticeable changes in stature recorded during this experiment, imply that the mechanism underlying disc height creep and recovery has a similar time response. Two mechanisms have been identified, an elastic and osmotic system; the former is thought to have the much shorter time constant (Koeller et al., 1984). Consequently , it seems likely that the elastic machanism is the primary system underlying the disc height changes recorded in this experiment. Although, there is little in the form of experimental data to explain what occurs within the fibres of the disc, it is postulated that the sudden increase in load causes a re-orientation and stretching of annular fibres, with the subsequent narrowing of the disc. A removal of the load permits the fibres to contract and return to their original state, resulting in an increase in disc height. Damage may ensue if the fibres are taken beyond their elastic limit.

The short-lived benefits of an unloading procedure prior to exercise clearly illustrate the limitations of such procedures as preventative activities against low back injury. It also has implications for those undertaking unloading procedures during low-back pain rehabilitation, suggesting that for any benefits to be long lasting the exposure time may need to last for several hours.

The pattern of recovery post-exercise is in line with the second hypothesis, the greater rate of recovery occurring when subjects' overall stature had reached its lowest point in this study. This would seem logical considering the increased osmotic pressure and annular stress that intervertebral discs possess when subjected to greater compressional forces (Koeller et al., 1984). However, it is not possible to accept hypothesis 2, given that the difference between recovery periods was not significant. This probably reflects the relatively small difference in overall shrinkage recorded at the end of each exercise regimen.

No significant differences were reported for perceived exertion and jumping distance towards the beginning and end of the exercise period, for each day of testing, suggesting bounding regimens were alike and imposed a similar degree of spinal loading. Differences were significant when the initial and final set of jumping distances were compared, the latter increases probably reflecting a warm-up effect leading to an increase in muscle contractile efficiency. The lack of a relationship between shrinkage and perceived exertion following exercise is consistent with the findings of Wilby et al. (1987) and may be a reflection of the type of exercise studied. Both RPE and shrinkage have been correlated for novice runners undertaking a 30 min continuous run (Leatt et al., 1986). Borg's (1970) scale was principally designed for application to aerobic work so RPE and shrinkage may be unrelated in high intensity, anaerobic exercises, such as plyometrics.

A distinctive rise and peak in post-exercise muscle soreness ratings was noted following the first exercise regimen. The peak occurred 43 hours after exercise after which time the symptoms were found to subside. These findings lead to an acceptance of the fourth hypothesis and imply that the bounding regimen incorporated muscle actions that were eccentric in nature. A significant fall in muscle soreness on the second occasion suggested that a training effect had taken place. Thus, the fifth hypothesis could be accepted.

Although LBP was recorded on the days following exercise, it would seem reasonable to suggest that it was associated with muscle soreness, given the close correlation and similar pattern displayed by each.

Whilst, the order of treatment was not randomized, it is unlikely that the training effect noted with muscle soreness would occur with shrinkage following a single exercise session, since effects of a common exercise regimen on shrinkage differ little between novices and experienced athletes (Leatt et al., 1986). The similarity between jumping distance and perceived exertion for each day implies that alteration to physical loading, resulting from changes in muscle co-ordination, did not take place.

5.6 SUMMARY

1. A short period (10 min) of gravity inversion caused an increase in stature (mean of 2.7 mm).

2. The benefits gained by unloading the spine prior to exercise were quickly negated during the execution of a bounding regimen. Twice the rate of shrinkage took place when the exercise regimen followed the period of unloading. This brings into question the use of short periods of gravity inversion as a pre-exercise preventative activity for athletes with low-back pain. Post-exercise gravity inversion may have more benefits than a pre-exercise treatment.

3. The short duration bounding regimen caused delayed onset muscle soreness which peaked 43 hours after exercise. Repeated exercise reduced the severity of post-exercise symptoms.

6. LONGITUDINAL REPEATABILITY OF FINE STATURE MEASUREMENTS AND THE ASSESSMENT OF A PRE-EXERCISE WARM-UP ROUTINE

6. LONGITUDINAL REPEATABILITY OF FINE STATURE MEASUREMENTS AND THE ASSESSMENT OF A PRE-EXERCISE WARM-UP ROUTINE

6.1 INTRODUCTION

The repeatability of stature measurements has only been studied over a short period, often 10 consecutive measures on the same day (Tyrrell et al., 1985; Leatt et al., 1986). Considering the apparent high subject variations in response to spinal loading and unloading, as evident from the large standard deviations recorded in the literature, repeatability is of major concern to future studies implementing stature changes as a reliable measuring procedure. It was hypothesized that:

Hypothesis 6: On separate days, mean fluctuations in stature for a specified regimen are repeatable.

Warm-up exercises prior to more strenuous exercise is a practice uniformly supported by practitioners (Reilly, 1981b). The benefits include a reduction in the risk of injury in the exercise to follow. No work has considered the implications of a commonly practiced warm-up routine on the loads imposed on the spine. Considering the nature and postures adopted during such routines it was hypothesized:

Hypothesis 7: A short duration warm-up routine increases the normal rate of stature loss and consequently, spinal loading.

Again, this has implications for the scheduling and structuring of exercise regimens.

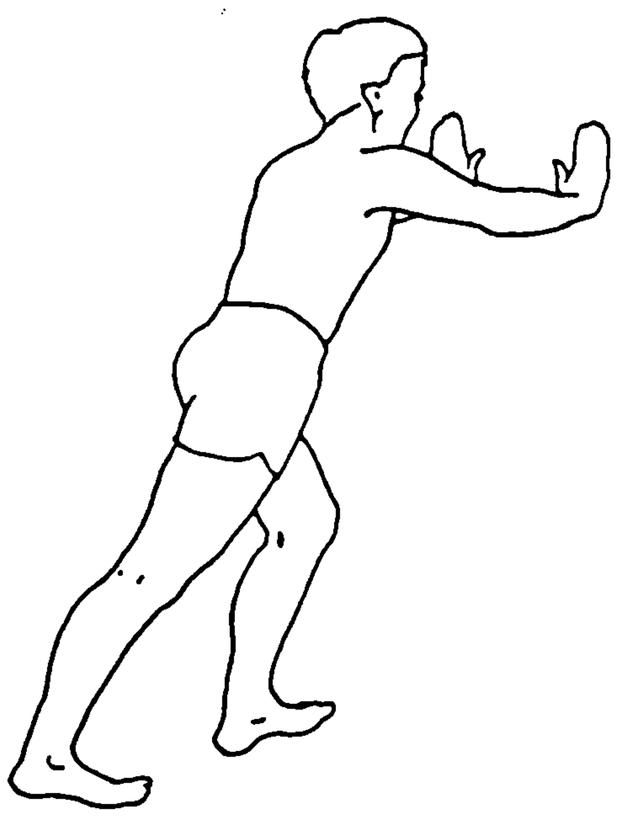
6.2 METHOD

Eight male subjects, who had a mean (±SD) age of 23.2 (±3) yr and height of 1.79 (±0.05) m were recruited into the study. All were physically active with no history of back complaints, did not regularly engage in plyometric training, had not taken part in any previous study and gave prior informed consent.

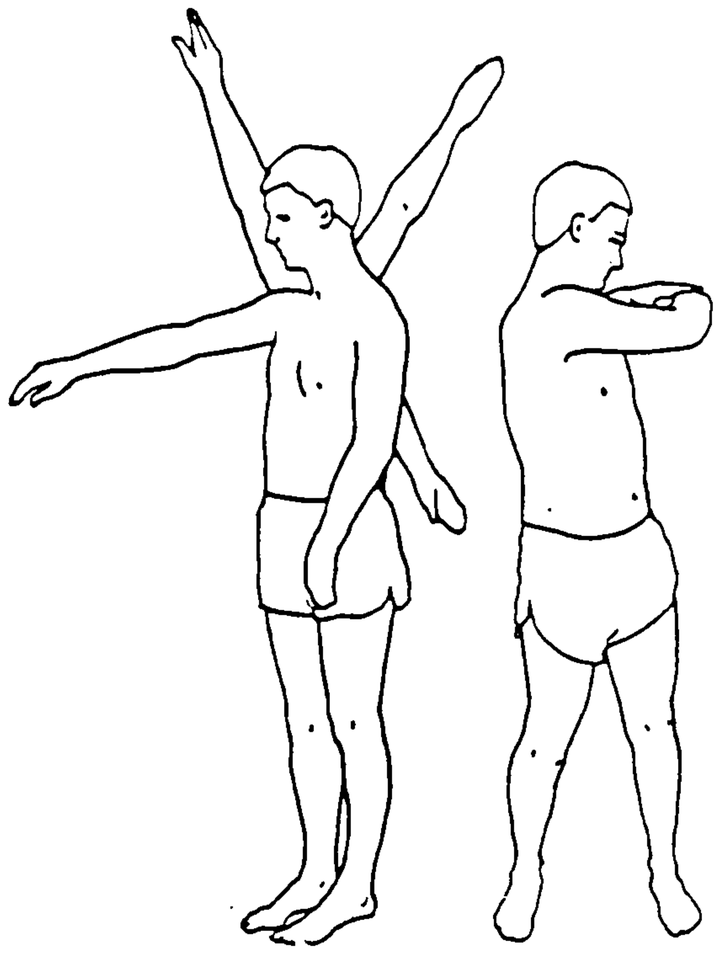
As in the previous experiment (section 5), the modified stadiometer was used to record changes in stature over closely monitored exercise regimens. The stadiometer was inclined at 13° and subjects required to undertake an average of 48 min (54 measures) to familiarise themselves with the apparatus and meet the training requirements of 10 consecutive measures with a standard deviation of 0.5 mm or less. All subjects who volunteered achieved this training requirement, with standard deviations ranging between 0.26 and 0.49 mm (mean = 0.41 mm), for the last ten consecutive measures.

Prior to testing, subjects were instructed to refrain from any sporting and stressful physical exercise and maintain as normal a daily routine as possible on the day of exercise. To monitor their daily routine, subjects listed activities from the time of wakening to the period of testing. Throughout the entire test sessions and for five days after the last exercise regimen, subjects were asked to refrain from any unfamiliar sporting activities in order to eliminate any external training effects.

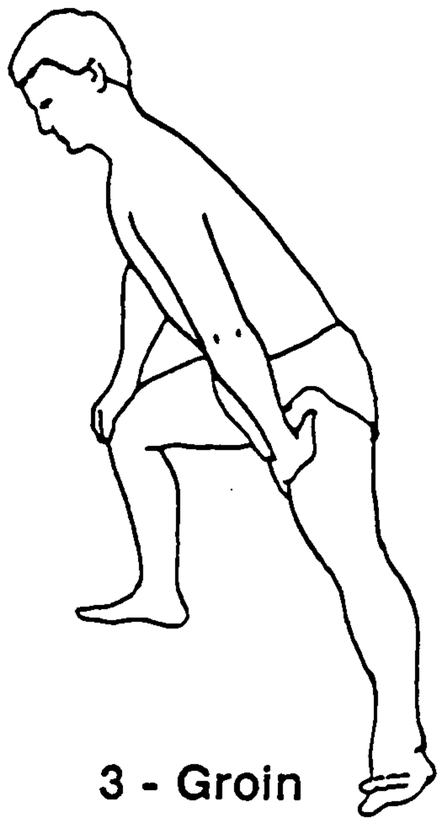
The format of the test regimen and time at which stature measurements were taken were identical to that described earlier (section 5), with alterations only being made to the pre-exercise behaviour, thereby maintaining uniformity between experiments. The exercise period again consisted of 50 standing broad jumps, performed in sets of five, with 15 s recovery between each set. The mean duration of the exercise regimen was 6.2 min and on average stature measurement periods lasted 4.1 min. In total, four regimens were conducted over a period of four weeks, with each regimen being undertaken on the same day, at the same time (14:00 h). The ten minute pre-exercise periods involved standing on the first two occasions, gravity inversion at an angle of 50° to the vertical on the third and a warm-up routine on the fourth: these will be referred to throughout as regimens 1, 2, 3 and 4, respectively. Twelve exercises were selected for the warm-up routine which was comprised of both static stretching and dynamic movements. These were felt to be common exercises often undertaken by athletes prior to more strenuous activities. The nature and sequence of exercises are depicted in Figure 18. Static stretches were conducted three times and held for 10 s, with 10 s recovery. For alternate limbs, 10 s recovery for one limb



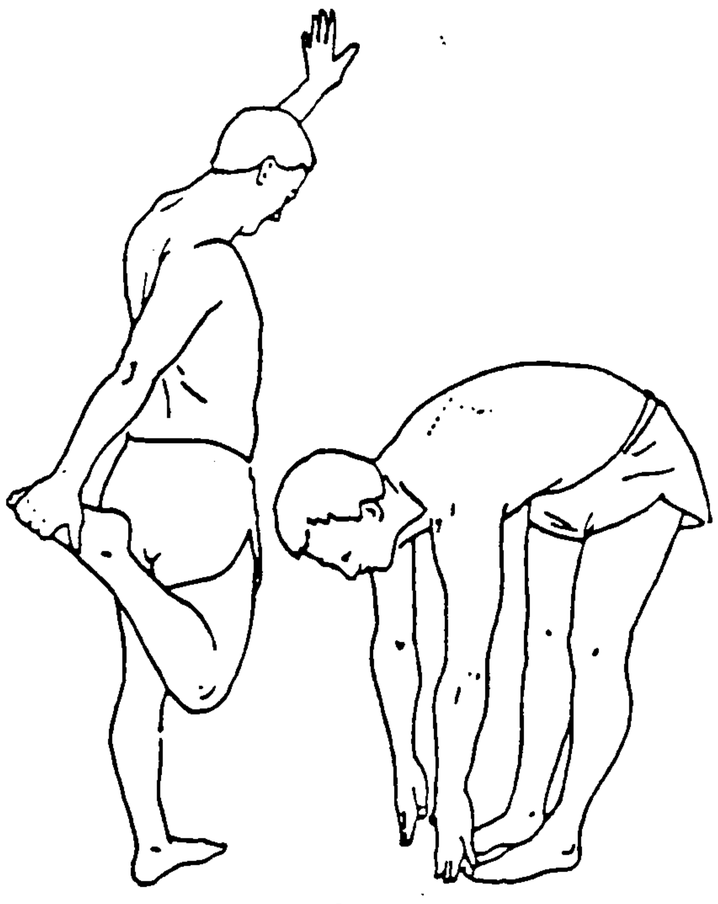
1 - Calves



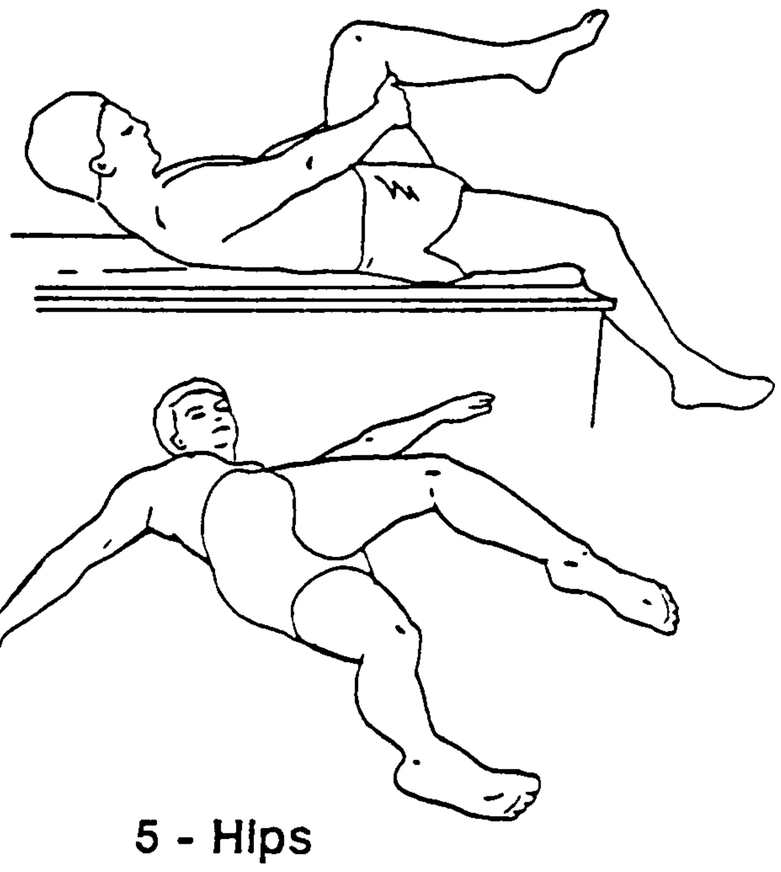
2 - Shoulders / upper body



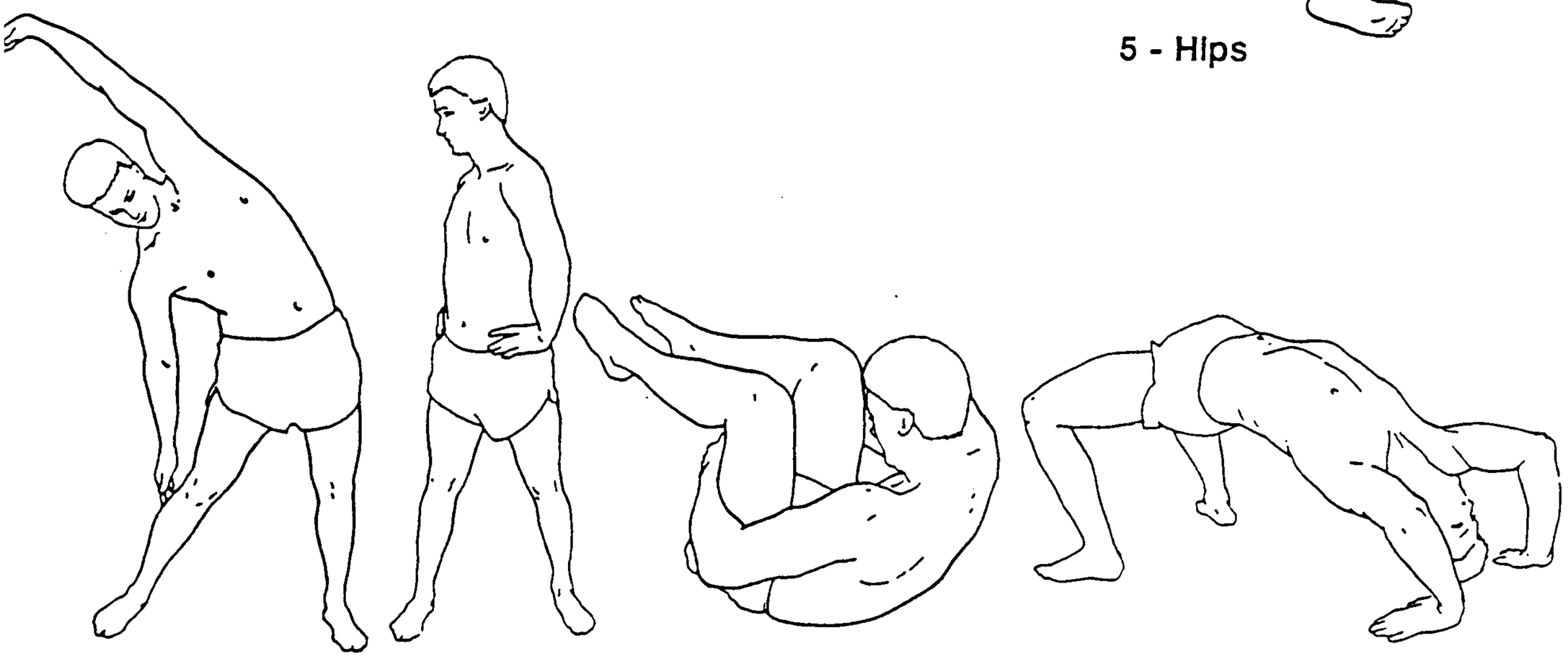
3 - Groin



4 - Legs



5 - Hips



6 - Back

Figure 18. The 10 min pre-exercise stretching and warm-up routine.

represented stretching of the other. Dynamic movements lasted for a period 30 s.

In addition to measures of perceived exertion, jumping performance for the first and last set of broad jumps, back pain ratings and post-exercise muscle soreness (described for the previous experiment), a number of anthropometric, strength, flexibility and muscular power measurements were made. On completion of each test regimen, measures were taken of height and weight, lumbar flexibility, isometric lifting ability for the back and legs (referred to as back and leg strength, respectively), hip flexibility and anaerobic power.

A standard anthropometer, accurate to 0.5 cm, was used to measure body height. Lumbar flexibility was measured with a pendulum goniometer in accordance with the technique described by Griffin et al. (1984). Back and leg strength were measured with a back and leg dynamometer (Takei Kiki Kogyo Co Ltd., Tokyo). During back strength measurements subjects were instructed to maintain straight legs and look directly forward. The posture during leg strength measures involved subjects maintaining a straight back, a raised head and flexing at the knees. For both back and leg strength measures the handle grip was placed slightly above the knees, subjects were instructed to apply a force for 3 s, and the peak value was recorded. Three trials were allowed with the maximum value being used during analysis.

A sit-and-reach box (Clarke, 1967) provided a procedure by which to record hip flexibility. The Margaria stair test (Margaria et al., 1966) was used as a measure of anaerobic power, with three warm-up runs being allowed, followed directly by three trials, the quickest time being recorded.

6.3 ANALYSIS OF DATA

A series of t-tests was used to examine whether differences were observable for sequential stature changes during each exercise regimens and between each different regimen. Similarly, these alterations were subjected to a two-way analysis of variance to

account for the interaction between subject and regimen during each time period. Application of the Pearson Product Moment method allowed a cross-correlation between stature alterations and the anthropometric and functional performance measures (body weight, height, back and leg strength, lumbar flexibility, hip flexibility, anaerobic power, and perceived exertion). An assessment of changes in the anthropometric and functional measures over the four weeks of testing was undertaken by subjecting these parameters to a t-test. For t-tests and variance analysis, probability values less than 0.05 were deemed significant.

6.4 RESULTS

Table 7 and Figure 19 display the alterations in stature recorded during the measurement periods of the four separate exercise regimens. Comparing each activity period of regimens 1 and 2, it was found there was no significant differences ($p>0.05$) between the measured stature changes. The initial 20 min standardization period caused almost identical mean decreases in stature, 0.29 and 0.34 mm for regimens 1 and 2, respectively. Slightly differing mean spinal responses occurred during the 10 min standing pre-exercises - an increase (0.24 mm) and decrease (0.15 mm) in stature for regimens 1 and 2, respectively. Overall, stature loss since the initiation of the experiment reached similar levels on completion of the bounding regimens, 2.19 and 2.22 mm respectively. A more marked difference, although again not significant ($p>0.05$), was noted during the 20 min standing recovery phase, where an overall regain in height of 1.28 mm was recorded on the first occasion as opposed to 0.17 mm on the second occasion.

Regimens 3 and 4 also showed negligible changes in stature for the initial 20 min standing, differences between all four regimens were non-significant ($p>0.05$). Pre-exercise gravity inversion caused a 2.85 (± 1.23) mm gain in stature, whereas the warm-up routine decreased stature by 0.89 (± 0.4) mm, the difference being significant ($p<0.01$). On comparing these pre-exercise stature changes with those during regimens 1 and 2, it was shown that different rates of stature alteration had taken place ($p<0.01$), the

Table 7. Mean (± SD) stature alterations (mm) for the four exercise regimens involving repeated and different pre-exercise activities prior to a bounding routine.

Pre-exercise regimen	20 min standing	10 min pre-exercise behaviour	Exercise period	Standing Recovery			
				5 min	10 min	15 min	20 min
1. Standing	- 0.29 (± 0.98)	+ 0.24 (± 0.5)	- 2.14 (± 0.9)	+ 0.64 (± 1.26)	+ 0.34 (± 1.06)	- 0.02 (± 0.5)	+ 0.32 (± 0.54)
2. Standing	- 0.34 (± 1.28)	- 0.15 (± 1.05)	- 1.73 (± 0.51)	+ 0.11 (± 1.22)	+ 0.11 (± 0.52)	- 0.10 (± 0.98)	+ 0.05 (± 0.5)
3. Gravity inversion	- 0.03 (± 1.32)	+ 2.85 (± 1.23)	- 3.54 (± 1.53)	+ 0.36 (± 0.97)	- 0.24 (± 0.82)	- 0.65 (± 1.12)	+ 0.34 (± 0.57)
4. Warm-up routine	+ 0.18 (± 0.5)	- 0.89 (± 0.4)	- 1.31 (± 0.77)	+ 0.47 (± 0.66)	+ 0.03 (± 0.6)	- 0.01 (± 0.35)	+ 0.29 (± 0.56)

NB: + denotes increases in stature
- denotes decreases in stature

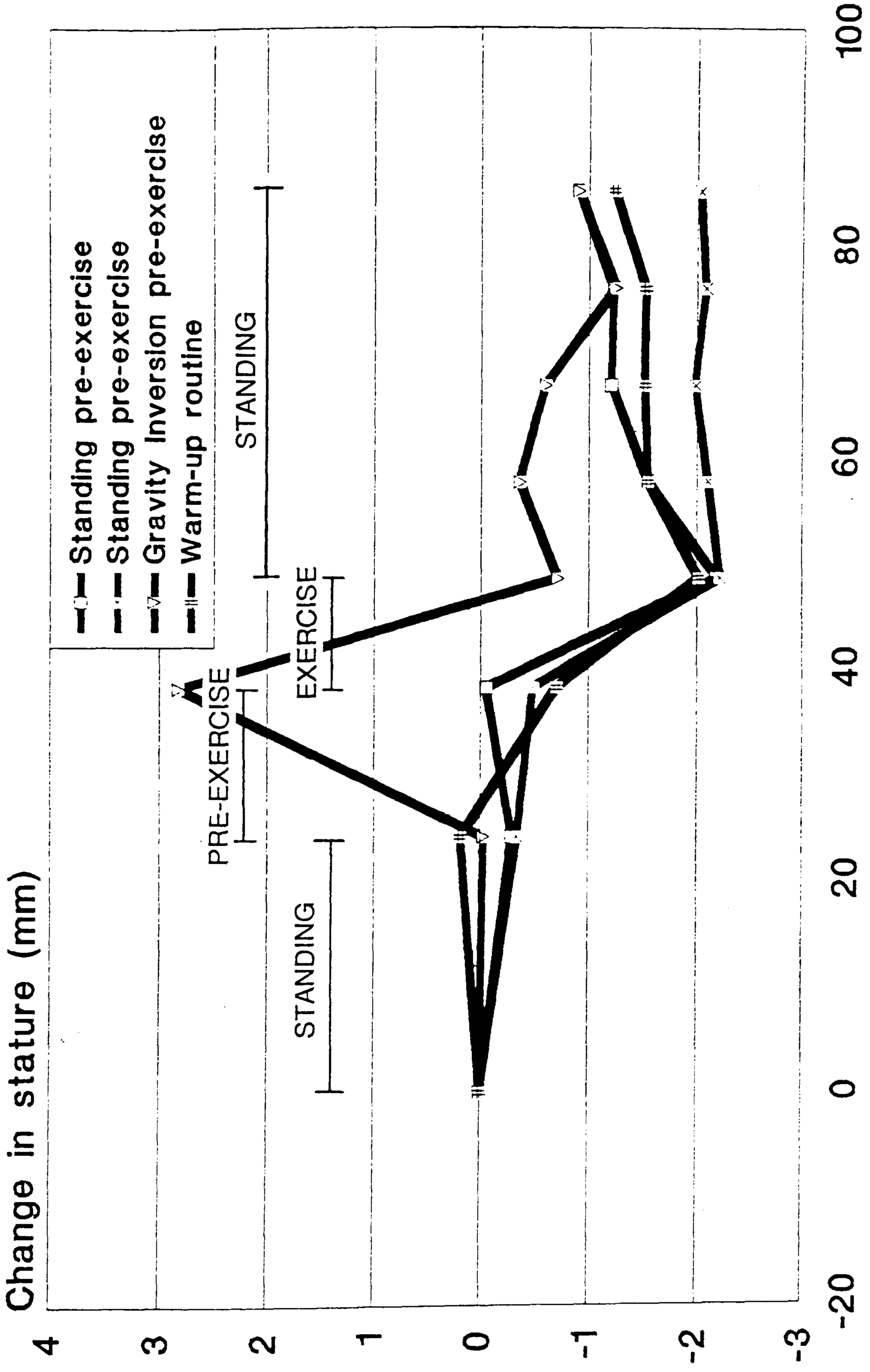


Figure 19. Mean stature alterations for the 4 exercise regimens.

only exception occurring for comparisons between regimens 2 and 4 ($p>0.05$).

The bounding routine caused 3.54 (± 1.53) mm and 1.31 (± 0.77) mm shrinkage for regimens 3 and 4 ($p<0.01$), respectively. For regimen 4, overall height changes from initiation of the experiment reached a similar level to that recorded during regimens 1 and 2 (2.02 mm). Following exercise, 20 min standing caused an overall height loss of 0.19 mm for regimen 3, whereas, an overall increase in stature (0.78 mm) occurred during regimen 4.

The two-way analysis of variance revealed that only during the initial 5 min recovery period was there a significant difference for subject variance ($F=3.75$; $p<0.01$).

Perceived exertion and back pain rating scores following the bounding regimen are presented in Table 8. Mean perceived exertion for the four exercise regimens varied only slightly between 14.4 (±1.5) and 13.6 (±1.1). Only two subjects complained of lower back pain following the exercise periods of regimens 2 and 3, and only one subject following the fourth regimen. On no occasions did low-back pain ratings exceed a value of 2 (very mild pain).

Table 9 displays the anthropometric and performance measures undertaken at the end of each test session. Although body weight, back strength, leg strength and lumbar flexibility for week 4 were higher than those recorded during week 1, the increases were non-significant ($p>0.05$). Mean back strength values showed the largest increases of 6 kp (50%) from week 1 to week 4.

A drop in mean leg strength and lumbar flexibility was noted for week 2, the change in leg strength being significant at the 0.05 level. The slowest mean time for the Margaria stair test also occurred during week 2. A training effect was noted with standing jump performance; a significant difference occurred between initial and final broad jumps for week 1 and week 4 ($p<0.01$), and week 2 and week 4 ($p<0.01$).

Table 8. Mean rating scores (± SD) following four bounding regimens repeated over a four week period and prior to different pre-exercise activities.

	Pre-exercise regimen			
	1. Standing	2. Standing	3. Gravity inversion	4. Warm-up routine
Perceived exertion	14.4 (± 1.5)	14.5 (± 1.8)	14.0 (± 1.6)	13.6 (± 1.1)
Back-pain rating	0	0.3 (± 0.6)	0.4 (± 0.8)	0.2 (± 0.5)

Table 9. Mean (± SD) anthropometric and performance measures following the activity regimens for the four week tests.

	Body mass (kg)	Back strength (kp)	Leg strength (kp)	Lumbar flexibility (°)	Hip flexibility (cm)	Margaria stair test (s)	Jumping distance (first set) (m)	Jumping distance (final set) (m)
Week 1	70.8 (± 4.9)	126 (± 21)	147 (± 23)	73 (± 13)	7 (± 5)	0.413 (± 0.04)	2.18 (± 0.18)	2.25 (± 0.18)
Week 2	70.9 (± 4.7)	127 (± 17)	140 (± 25)	67 (± 14)	8 (± 7)	0.418 (± 0.043)	2.18 (± 0.17)	2.27 (± 0.16)
Week 3	71.3 (± 4.6)	129 (± 19)	147 (± 25)	74 (± 13)	8 (± 6)	0.414 (± 0.043)	2.21 (± 0.16)	2.27 (± 0.17)
Week 4	71.5 (± 4.9)	132 (± 14)	150 (± 29)	74 (± 13)	7 (± 6)	0.412 (± 0.055)	2.26 (± 6.21)	2.35 (± 0.20)

A cross correlation of all functional and anthropometric measures showed that back strength and body weight were positively correlated across all four weeks of testing ($r=0.71 - 0.93$; $p<0.05$). Similarly, across all four days both the initial and final mean standing broad jump showed significant correlations with the time for the Margaria stair test ($r=0.63 - 0.88$; $p<0.05$), the relationship being inverse. A linear regression analysis produced two regression equations with similar coefficients (refer to Figure 20)

$Y = -2.89x + 3.36$ (1st set)

$Y = -2.86x + 3.44$ (10th set)

where: Y = Standing broad jump distance (m)

x = Time to complete Margaria stair test (s)

Correlation coefficients for the first and tenth set of broad jumps was -0.70 and -0.71, respectively. These correlations were found to improve when both body weight and the time for Margaria stair test were used to predict short term (anaerobic) power (refer to Figure 21);

$$\text{Power} = \frac{\text{body weight x vertical distance x gravity}}{\text{time for Margaria stair test}}$$

Correlation coefficients of 0.80 and 0.79 were found between the calculated power output and the initial and final set of broad jumps respectively. Regression equations for estimating power output from standing broad jump performance were;

$Y = 803.3x - 211.6$ (1st set)

$Y = 805.2x - 286$ (10th set)

where: Y = power (w)

x = Standing broad jump distance (m)

Leg strength was found to correlate with body weight ($r=0.66 - 0.85$, $p<0.05$) during weeks 1 and 2, whereas, height significantly correlated with back flexibility ($r=0.65 - 0.77$; $p<0.05$) during weeks 2 and 3. On a week to week basis, back strength was found to significantly correlate with leg strength ($r=0.64 - 0.81$; $p<0.05$).

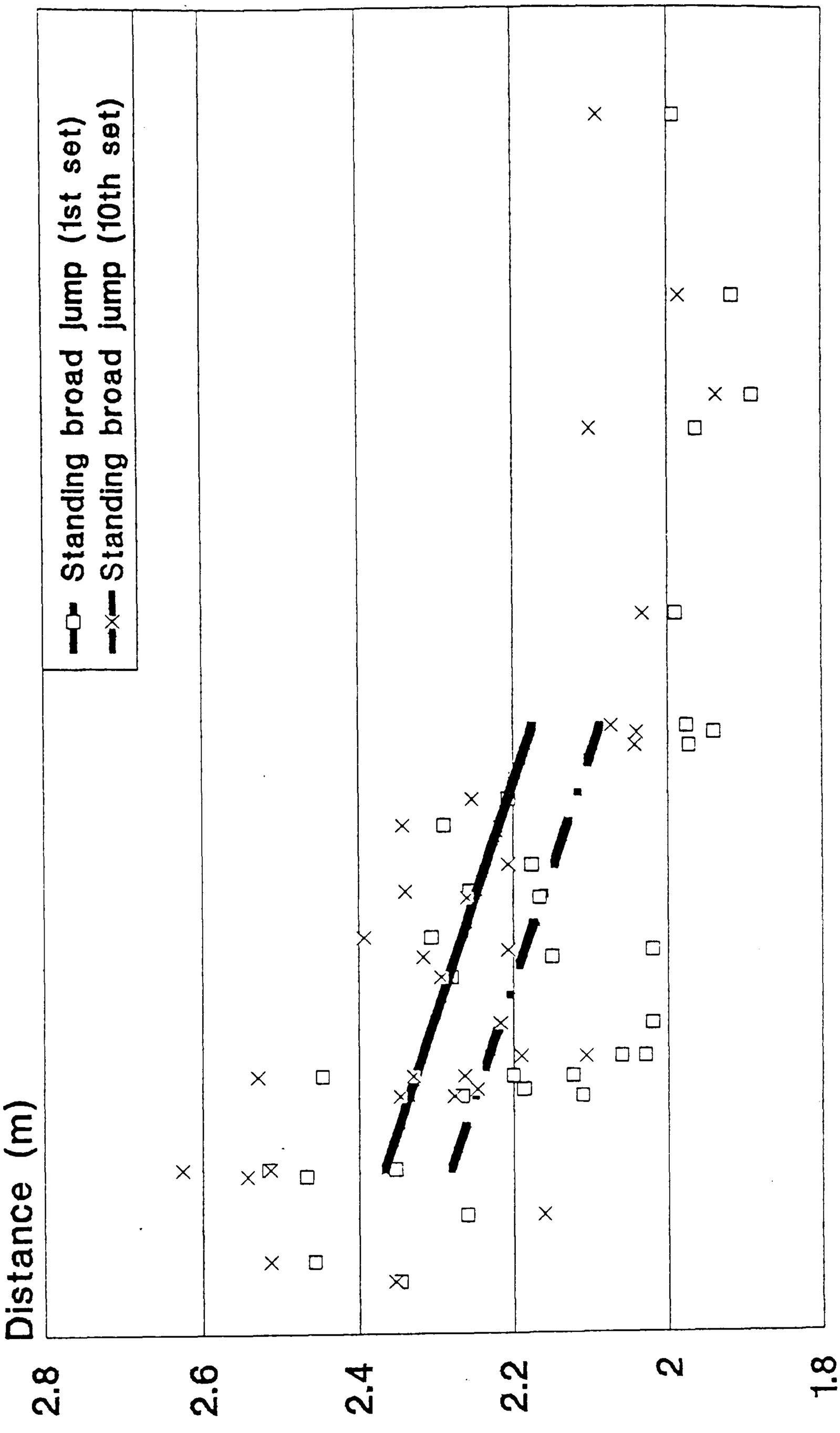


Figure 20. Standing broad jump distance against time for the Margaria stair test.

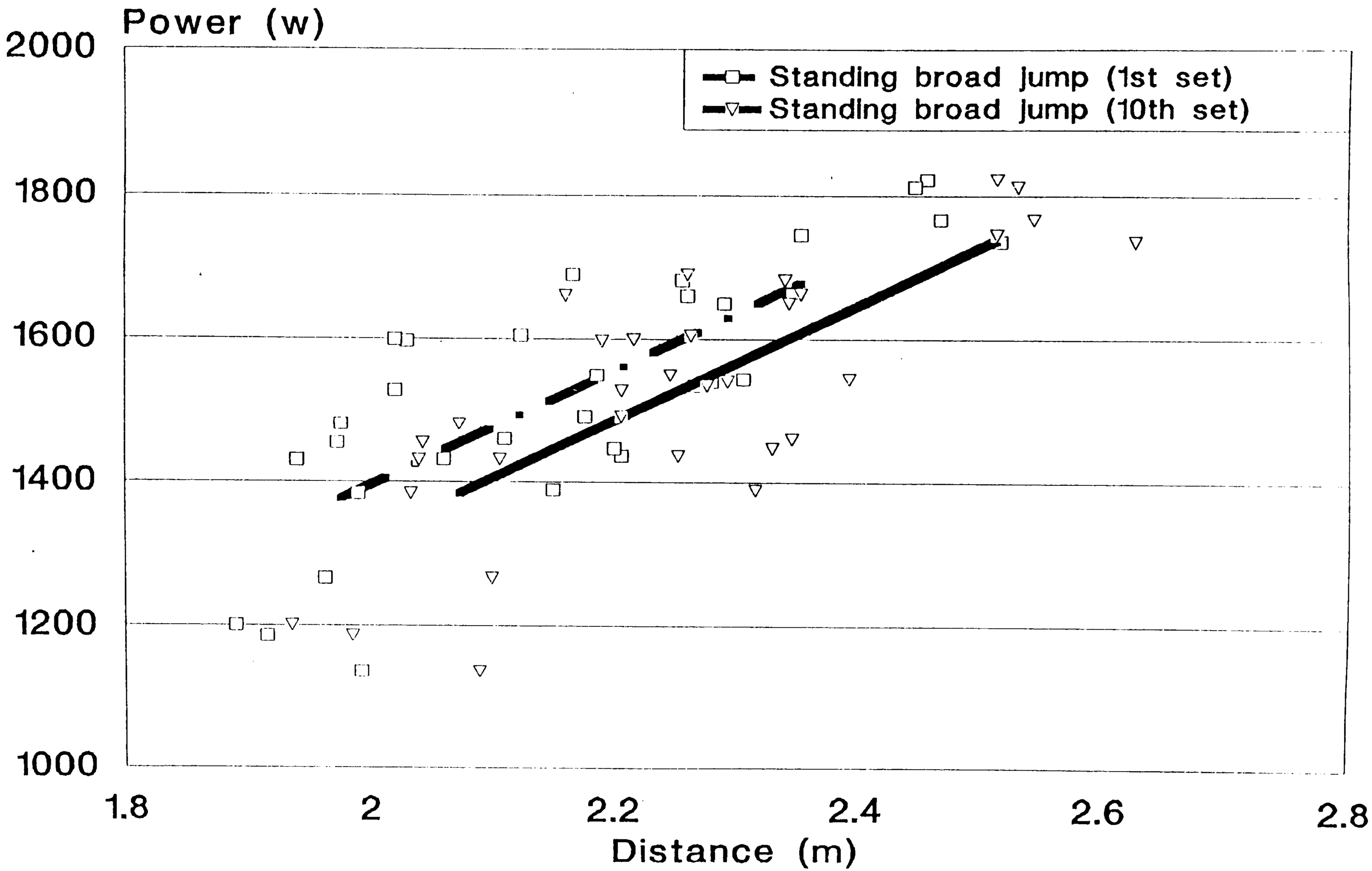


Figure 21. Standing broad jump distance against estimated power output.

When stature changes were correlated with the same functional and anthropometric measures it was found that shrinkage occurring during exercise was unrelated to both RPE and measures of back strength. In addition, only during week 1 was leg strength related to the shrinkage that occurred during the exercise ($r=0.75$, $p<0.05$). Stature changes post-exercise were also unrelated to back or leg strength. A positive correlation was found between the stature changes which took place following exercise during regimen 3 and subject height ($r=0.82$, $p<0.01$). For regimen 2, hip flexibility showed a positive relationship with the shrinkage occurring from the exercise period ($r=0.77$, $p<0.01$). However, this same relationship became negative during regimen 4 ($r=-0.78$, $p<0.01$).

Mean post-exercise muscle soreness ratings for the four activity regimens are displayed in Figure 22. A significant variation in subject response was found following exercise on the first occasion with those reported on the further three occasions. For week 1, muscle soreness rating increased and peaked approximately 43 h post-exercise, after which time they decreased to approach zero. During the hours after exercise, for the further three regimens, mean post-exercise soreness ratings were found to remain constant with symptoms never exceeding 4 on the 0 - 10 linear analogue scale (0 referring to no soreness). A comparison of scores for each period of recording showed that for weeks 2, 3 and 4, mean rating values did not significantly differ ($p<0.01$). When week 1 measures were compared with the remaining three weeks significant differences ($p<0.05$) were noted, on all occasions, from 19 hours through to 54 hours post-exercise. When comparing muscle soreness rating scores at approximately 6 hours after exercise, there was no significant ($p>0.05$) difference for each week of testing, with the exception of scores for week 1 and week 3 ($p<0.05$). At 67 hours into the post-exercise phase, only values recorded between weeks 1 and 4 were found to be significant ($p<0.05$).

6.5 DISCUSSION

The mean changes in stature recorded during two identical activity regimens, one week apart, were not found to differ. A short duration

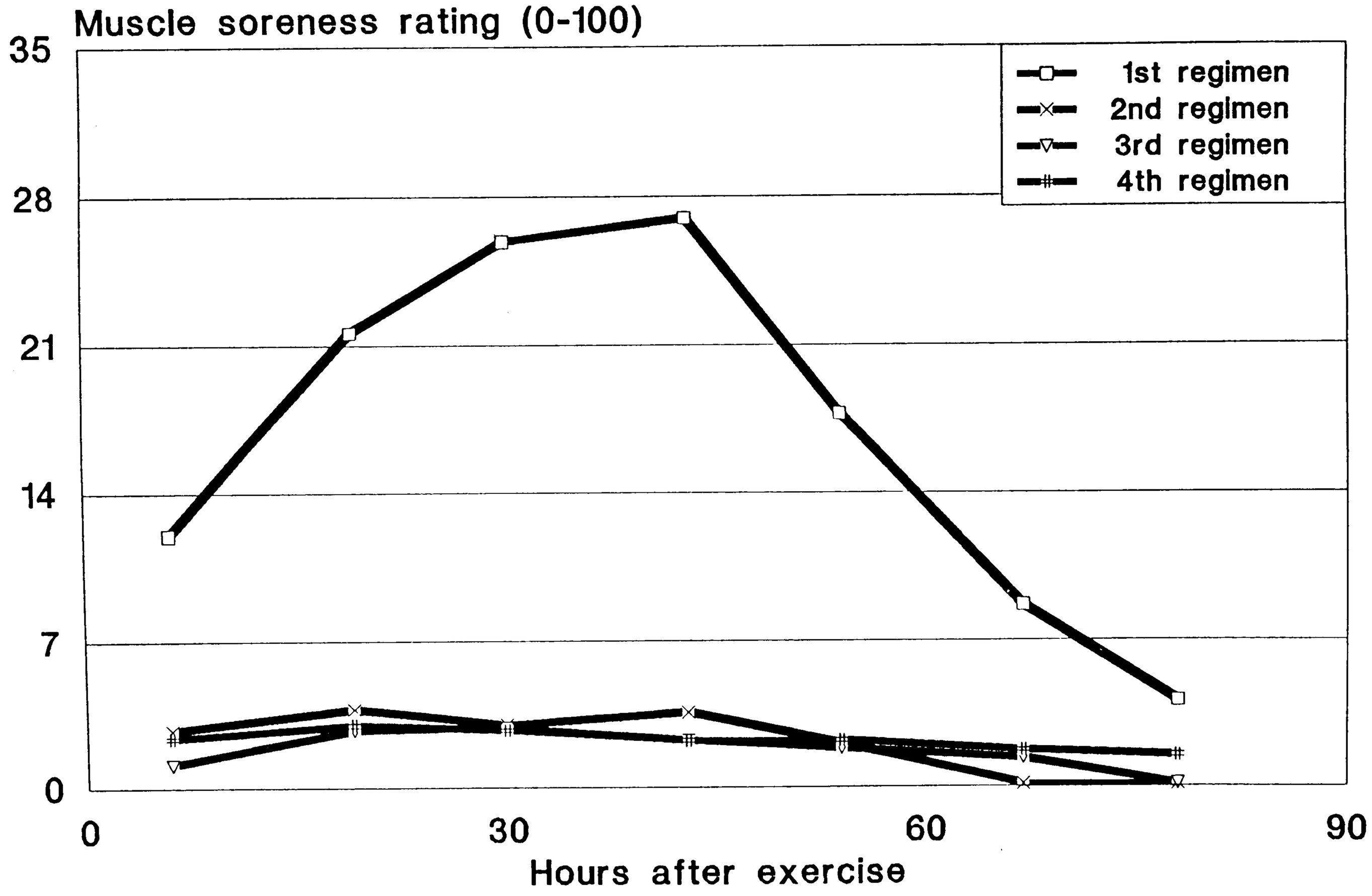


Figure 22. Mean post-exercise muscle soreness rating scores.

bounding regimen interspersed between periods of relaxed standing was found to accelerate shrinkage, the changes not only being comparable (2.14 and 1.73 mm, $p>0.05$) on this occasion but also with the previous (1.69 mm) study (section 5) under identical circumstances. Further support for the repeatability of the technique is evident when stature changes occurring during the period of gravity inversion and exercise are again compared with the earlier study (section 5). Gravity inversion increased stature by 2.85 and 2.74 mm, and subsequent exercise caused shrinkage of 3.54 and 3.49 mm, for regimen 3 and the previous investigation, respectively. The repeatability reported here leads to an initial acceptance of hypothesis 6. However, it could be questioned whether mean changes in stature are an acceptable method to study repeatability. Stature changes are very much subject dependent and therefore, analysis may need to be restricted to individuals.

Although differences in the rate of recovery post-exercise were non-significant during regimens 1 and 2, a marked regain in height was not noted following exercise on the second occasion (0.17 mm as opposed 1.28 mm). It is difficult to find a plausible explanation although leg strength and lumbar flexibility were found to be lower on this occasion, lumbar flexibility being significantly different to pre-and post-measures ($p<0.05$). However, stature changes during recovery did not appear to correlate with leg strength or lumbar flexibility.

A 10 min pre-exercise warm-up routine comprised of a series of static stretching and dynamic movements was found to increase shrinkage when compared with an equivalent period of standing, 0.89 mm against an increase of 0.24 mm and decrease of 0.15 mm for regimens 4, 1 and 2, respectively. However, the differences between regimens 4 and 2 were found to be non-significant, preventing an acceptance of hypothesis 7.

Standard deviations recorded for stature measurements (Table 7) provide an indication of subject variability. These were found to range between ±0.35 and ±1.53 mm. However, the analysis of variance revealed that only on one occasion, the initial 5 min recovery period, was subject variance significant ($p<0.01$), implying that the

variance in recorded shrinkage for the activity periods could not be attributed to subjects.

The activities selected during the warm-up regimen were chosen to represent 'typical' exercises that may be deployed prior to more strenuous work and promote an increase in flexibility and enhance muscle function. Although small increases in broad jump distance were found between weeks 3 and 4, these changes were not significant ($p>0.05$), suggesting that the warm-up routine had not significantly enhanced performance. However, the regularly observed increases between the initial and the final set of broad jumps do suggest that more specific exercise will enhance performance.

The cross-correlation of anthropometric and functional measures with stature alterations did not identify any consistent trends. Similar to previous findings (section 5.5), RPE and shrinkage during exercise were unrelated; a possible explanation is provided in section 5.5. Wilby et al. (1987) did report a relationship between back strength measurements and shrinkage; such an association was not apparant in this study and again may be explained by the difference in nature between the two exercises. The circuit weight training used by Wilby et al. (1987) entailed a continuous period of aerobic exercise with sustained back muscle exertions. Bounding exercises are unlikely to be composed of a similar aerobic component, nor include the same degree of sustained back muscle activity.

Consistent correlations were observed between back strength and body weight and, standing broad jump distance and time for the Margaria stair run. The Margaria stair test has become an accepted measure of anaerobic power; hence, the linear correlation between predicted power output and standing broad jump performance reported here serves to underline how the latter may be used as a simple measure of anaerobic power.

As in the previous study (section 5), post-exercise muscle soreness was shown to be substantially reduced when the same bounding regimen was repeated one week later. A peak in muscle soreness was found approximately 43 hours after exercise, findings consistent with the previous study and literature governing unaccustomed eccentric

exercise (Ebbeling and Clarkson, 1989). Further repeated bouts of exercise failed to induce a marked increase in post-exercise muscle soreness from that which had occurred following week 1. Thus, there is evidence to support the conclusion reached by Schwane et al. (1987) that delayed muscle soreness may be difficult to avoid totally but reduced symptoms are more likely when training encompasses a substantial eccentric component.

6.6 SUMMARY

1. Mean changes in stature during an exercise regimen proved repeatable.

2. Differences in stature alterations for each activity period, with the exception of the first 5 minutes of recovery post-exercise, could not be attributed to subject variance.

3. A short-duration warm-up routine (10 min) did not significantly increase the loads imposed on the spine when compared with relaxed standing. Consequently, the benefits of a pre-exercise warm-up routine would not seem to be offset by an increased risk of spinal injury.

4. Standing broad jump performance correlated with predicted anaerobic power output using the Margaria stair test, supporting its use as a simple measure of anaerobic power.

5. Findings support the use of repeated bouts of eccentric exercise during training to reduce the severity of post-exercise muscle soreness.

7. CHANGES IN STATURE FOLLOWING DROP JUMPING AND POST-EXERCISE GRAVITY INVERSION

7. CHANGES IN STATURE FOLLOWING DROP JUMPING AND POST-EXERCISE GRAVITY INVERSION

Aspects of this work were published in Medicine and Science in Sports and Exercise, vol 22, no 3, pp 385-390, a copy of which is presented in Appendix 6.

7.1 INTRODUCTION

Drop jumping has been advocated by many as a form of plyometric training most appropriate to enhancing leg muscle power. The side-effects or potential risks of incurring injury, particularly when undertaken by novices, are not fully understood. The aim of this study was to investigate whether, as with bounding exercises, short duration drop jumping regimens impose a cumulative load upon the spine which can be observed by measuring small changes in stature.

Hypothesis 8: A short duration drop jumping regimen increases the rate of spinal shrinkage when compared to an equivalent period of standing.

Short duration, pre-exercise spinal unloading was found to have only short term benefits in facilitating a restoration of intervertebral disc height (section 5 and 6). Consequently, adopting such postures post-exercise may be of greater benefit in quickly negating the effects of exercised induced spinal loading. Thus, the hypothesis put forward was:

Hypothesis 9: Gravity inversion quickly promotes spinal recovery following a period of exercise - induced loading.

7.2 METHODS

Eight male subjects took part in the study and had a mean age of 25.1 yr (SD = ± 3.5 yr), mean height of 181 cm (SD = ± 5.6 cm) and mean body mass of 74.8 kg (SD = ± 4.7 kg). All were physically active, had not experienced any back problems, did not take part in any

previous studies and did not regularly participate in drop jumping or plyometric activities as a form of training.

The modified stadiometer and familiarisation procedure was implemented prior to any experimental sessions. Stature measurements were conducted in a manner identical to that outlined earlier (Section 5.2). Subjects required an average of four training periods lasting a total time of approximately 50 min. In all, ten subjects were recruited into the study but two failed to meet the necessary training requirements after a period of 90 min and were, therefore, eliminated from the study. The eight subjects who achieved the necessary criteria attained an average standard deviation for the last ten consecutive measures of 0.42 mm (range = 0.3 - 0.5 mm).

All subjects participating in the experiment were required to undertake two regimens which varied only in the nature of the post-exercise activity. These were 20 min standing (regimen 1) and 20 min of gravity inversion at a 50 degree angle to the vertical (regimen 2), the order of treatment being randomised. Exercise entailed 25 drop jumps, performed in sets of five, with 20 s recovery between each set and lasted an average 6.01 min. Drop jumps were performed in bare feet, from a gymnastic vaulting box 1 m in height, on to standard gymnastic tumbling mats laid on a concrete floor. On making contact with the ground, subjects immediately initiated a vertical jump, sufficient to clear a 0.5 m high hurdle placed 1 m away from the base of the box.

To prevent injury, the hurdle was constructed out of lightweight aluminium mounted on wooden blocks and designed to topple if struck. A stairway was built alongside the vaulting box which allowed subjects easy access to the top of the box, thereby, minimising the physical exertions normally entailed. The instructions issued to subjects were that they should step out from the box, bring both feet together for a two footed landing, and immediately perform a rebound jump. At the end of the exercise period subjects were asked to rate their perceived exertion using Borg's scale (Borg, 1970) and the intensity of any back pain symptoms with the rating scale outlined earlier (Section 5.2).

As in previous experiments, the test sessions were conducted at the same time of day (14:00 h), on two separate days (1 week apart) and with an initial standing period of 20 min. All of which served to control for the effects of circadian variations in stature, post-exercise muscle soreness and abnormal loading prior to testing. To maintain compatibility with previous experiments, subjects undertook a further 10 min standing prior to the exercise periods. Directly following the post-exercise treatments, stature changes were monitored for 40 min while subjects stood. In total, nine measurements of stature were taken:- at the start of the experiment, after 20 and 10 min pre-exercise standing, after the exercise and post-exercise sessions, and every 10 min of the 40 min standing recovery. Each measurement period lasted on average 3.98 min, providing sufficient time to stabilise heel compression.

To prevent unaccustomed exercise having an influence on stature changes or jumping performance, all subjects were instructed to refrain from any unfamiliar sporting activities for the week prior to and during the two sessions of testing, plus the four days post-exercise. Similarly, on the day of testing, they were asked to maintain as normal a daily routine as possible and not participate in any unfamiliar sporting activities.

As in previous investigations, muscle soreness and lower back pain rating scales were administered to subjects for completion in the four days following exercise. Measures of body height and weight, lumbar flexibility, isometric lifting ability of the back and legs, hip flexibility, and anaerobic power were also conducted using the procedures described in Section 6.

7.3 ANALYSIS OF DATA

Cross-correlations of stature alterations, anthropometric data and functional measures were undertaken using Pearson Product Moment method. A series of t-tests was used to examine whether differences existed between stature alterations recorded between regimens and the anthropometric and functional measures recorded for both weeks. A two-way analysis of variance was used on stature alterations occurring from the initiation of the experiment to examine variance

due to subject and regimen. Probability values of less than 0.05 were deemed significant when using both the t-test and analysis of variance.

7.4 RESULTS

The mean alterations in stature recorded for both regimens are shown in Figure 23 and Table 10. The pre-exercise activities were identical for regimens 1 and 2 and caused similar spinal shrinkage. An initial 20 min period of relaxed standing caused 0.44 and 0.35 mm decrease in stature for regimens 1 and 2, respectively, the difference being non-significant ($p>0.05$). Similarly, a non-significant ($p>0.05$) stature change occurred for the further 10 min relaxed standing, 0.14 (SD = ±0.38) and 0.01 (SD = ±0.91) mm decrease for regimens 1 and 2, respectively.

Increased rates of spinal shrinkage occurred when the drop jumping exercise periods were undertaken, 1.68 mm and 1.81 mm for regimens 1 and 2 respectively, the difference between regimens being non-significant ($p>0.05$), but significant when compared to the standing period prior to exercise ($p<0.01$). Mean RPE values for the two drop jumping regimens were also similar, 9.9 (SD = ±1.2) and 10.0 (SD = ±1.1) (very light/fairly light) for regimens 1 and 2, respectively. No subjects complained of any lower back discomfort directly following the two exercise periods.

Twenty minutes of gravity inversion post-exercise caused subjects' stature to increase by an average of 5.18 mm (range = 2.65 to 8.76 mm; SD = ± 2.24) compared to 0.76 mm (range = 0.59 to 1.72 mm; SD = ± 0.81) for the identical standing duration ($p<0.01$). The 40 min standing following each of these periods also caused significant variations in spinal response, a gain in stature of 0.04 mm and a loss of 4.07 mm for regimens 1 and 2, respectively. Changes in stature for each 10 min period of the standing recovery were significantly different when comparing regimens ($p<0.01$), with the exception of the second 10 min period ($p>0.05$). A comparison of overall stature changes for each regimen showed that there were no differences after 30 min of the 40 min recovery period ($p>0.05$).

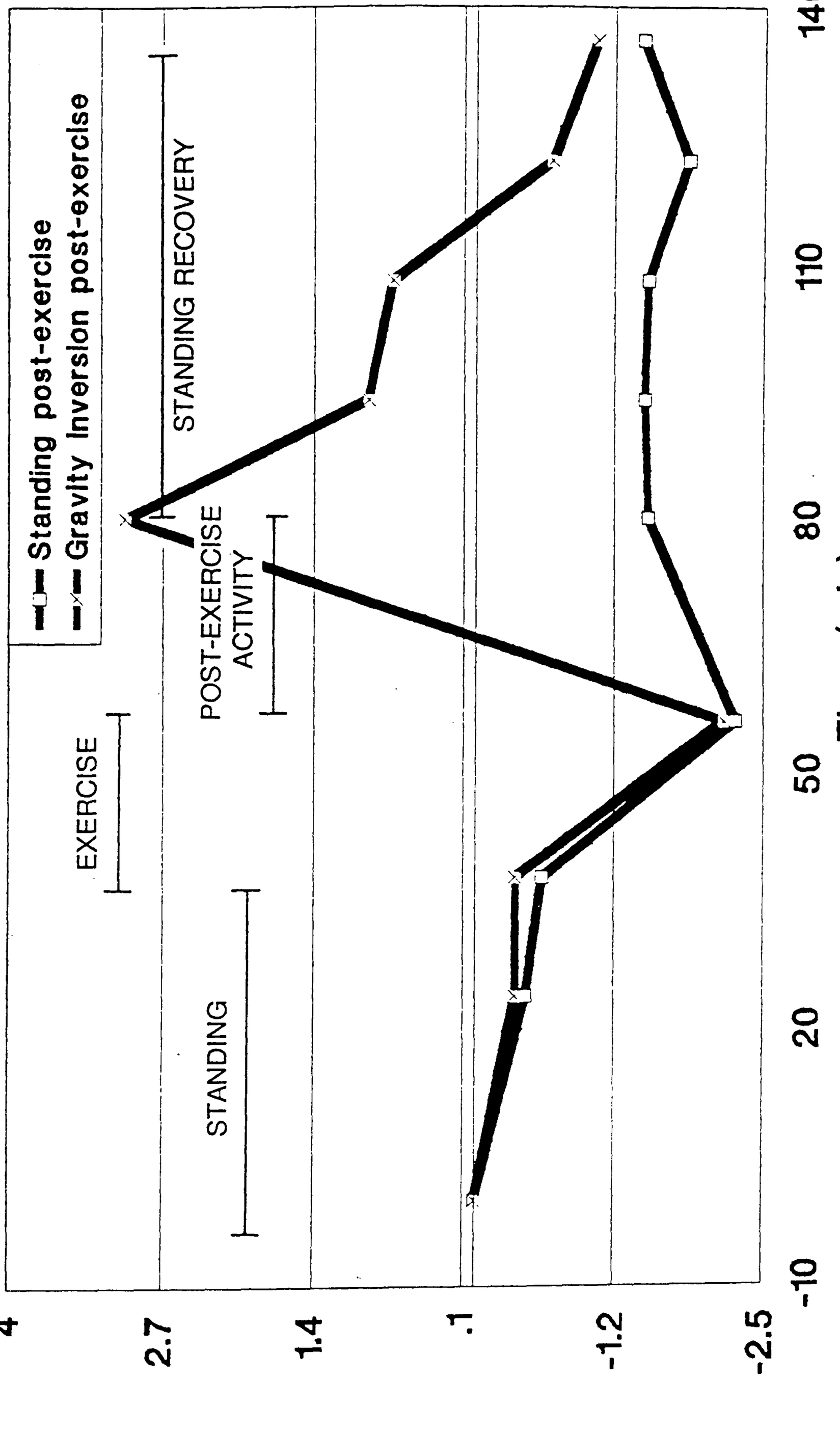


Figure 23. Mean stature alterations for the test regimens involving post-exercise standing and gravity inversion.

Table 10. Mean (± SD) stature alterations (mm) for two drop jumping regimens involving differring post-exercise activities.

	20 min standing	10 min standing	Exercise	20 min Post-exercise recovery	Standing Recovery			
					10 min	10 min	10 min	10 min
Standing post-exercise	- 0.44 (± 0.65)	- 0.14 (± 0.38)	- 1.68 (± 0.48)	+ 0.76 (± 0.81)	+ 0.03 (± 0.76)	- 0.03 (± 0.39)	- 0.35 (± 0.68)	+ 0.04 (± 0.56)
Gravity inversion post-exercise	- 0.35 (± 0.34)	- 0.01 (± 0.91)	- 1.81 (± 0.69)	+ 5.18 (± 2.24)	- 2.10 (± 0.93)	- 0.21 (± 0.94)	- 1.37 (± 0.82)	- 0.39 (± 0.53)

NB: + denotes increases in stature
- denotes decreases in stature

Table 10 provides data on subject variance (SD) for the stature changes recorded for each individual period during the two regimens. With the exception of the 20 min gravity inversion (SD = ± 2.24 mm), standard deviations did not exceed ± 1 mm. The two-way analysis of variance revealed that inter-subject variability was only significant for the second and third 10 min periods of the final 40 min standing recovery.

A cross-correlation matrix of the anthropometric and functional measures for week 1 and 2 is shown in Table 11. Variables which were found to significantly correlate for each week of testing included hip flexibility and body height ($r = -0.709$, $r = -0.659$, $p<0.05$), back strength and leg strength ($r = 0.823$, $r = 0.847$, $p<0.01$), back strength and the time for the Margaria stair test ($r = -0.821$, $p<0.01$, $r = -0.699$, $p<0.05$), leg strength and hip flexibility ($r = 0.832$, $r = 0.849$, $p<0.01$), leg strength and the time for the Margaria stair test ($r = -0.706$, $p<0.05$, $r = -0.886$, $p<0.01$,), and lumbar flexibility and hip flexibility ($r = 0.813$, $r = 0.836$, $p<0.01$). During week 2 lumbar flexibility was found to correlate with both back and leg strength ($r = 0.698$, $r = 0.682$, $p<0.05$), along with hip flexibility and the time for the Margaria stair test ($r = -0.644$, $p<0.05$). No significant changes in anthropometric and functional measures were found between week 1 and week 2 with the exception of body mass ($p<0.01$), whose mean value decreased by 0.262 (SD = ± 0.9) kg.

The cross-correlation of functional and anthropometric measures with changes in stature produced a significant negative relationship between increases in stature during inversion (with respect to initial height) and body weight for week 1 ($r = -0.673$, $p<0.05$) and week 2 ($r = -0.704$, $p<0.05$). During the 40 min standing recovery, back strength (week 1 and 2) was found to correlate with stature alterations during post-gravity inversion ($r = 0.668$, $r = 0.716$, $p<0.05$). No significant correlation was found between RPE and shrinkage following exercise.

Mean muscle soreness rating values for the hours after exercise are shown in Figure 24. Values for each week were found to significantly differ between 19 and 67 hours after exercise ($p<0.05$). Soreness values increased gradually during week 1 to peak approximately

Table 11. Cross-correlation matrix of anthropometric and functional measures for both weeks of testing.

	Body height		Back strength		Leg strength		Lumbar flexibility		Hip flexibility		Time for the Margaria stair test	
	Week 1	Week 2	Week 1	Week 2	Week 1	Week 2	Week 1	Week 2	Week 1	Week 2	Week 1	Week 2
Body mass	0.514	0.523	0.210	0.387	0.016	0.072	0.269	-0.114	0.393	0.300	-0.192	-0.121
Body height			-0.499	-0.201	-0.608	0.550	-0.289	-0.280	-0.709[a]	-0.659[a]	0.397	0.446
Back strength					0.823[b]	0.847[b]	0.467	0.698[a]	0.595	0.587	-0.821[b]	-0.699[a]
Leg strength							0.564	0.682[a]	0.832[b]	0.849[b]	-0.706[a]	-0.886[b]
Lumbar flexibility									0.813[b]	0.836[b]	-0.232	-0.382
Hip flexibility											-0.494	-0.644[a]

NB: a denotes $p<0.05$
b denotes $p<0.01$

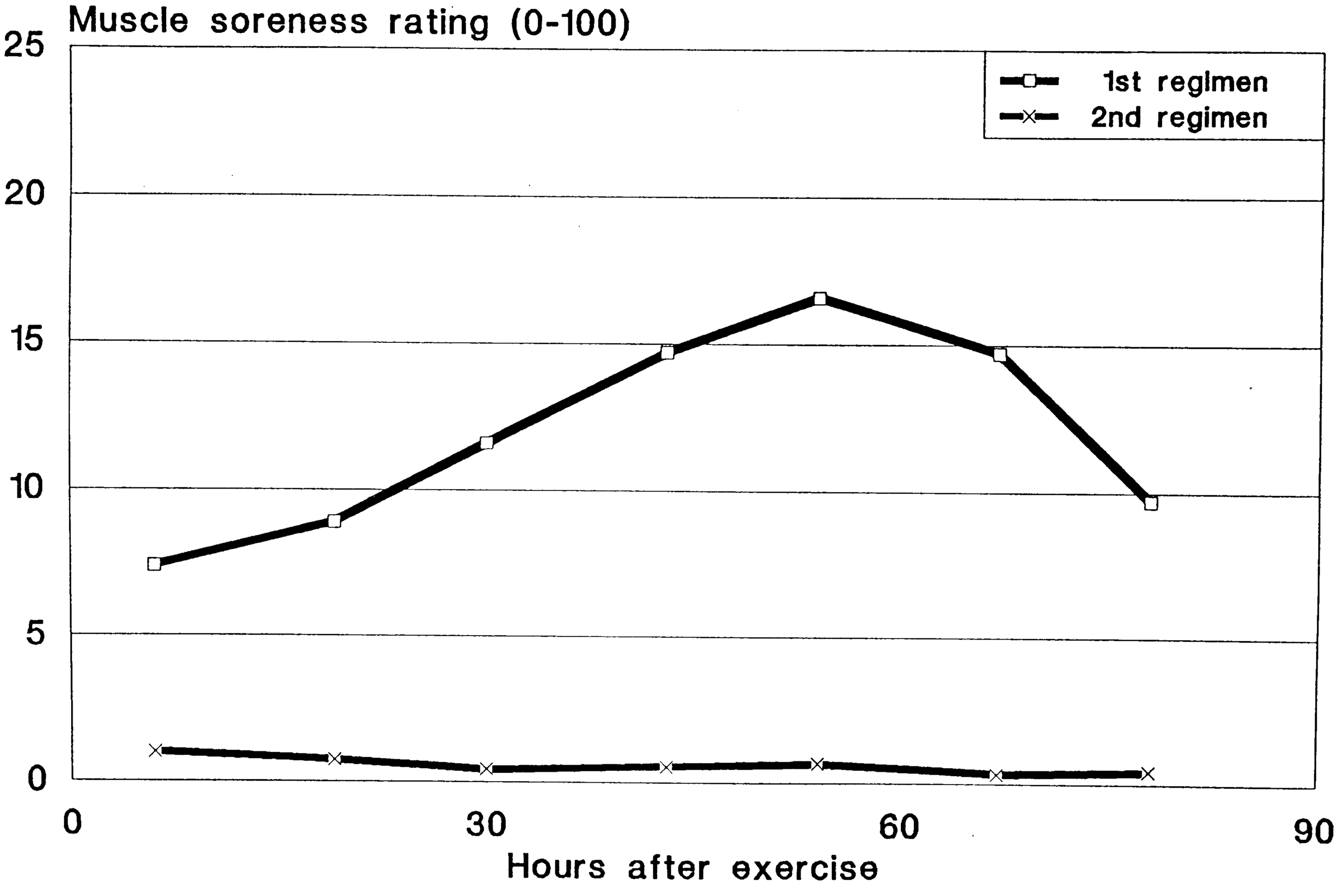


Figure 24. Mean post-exercise muscle soreness rating scores.

54 hours after exercise, whereas they remained at non-significant levels throughout the days following the second drop jumping regimen. Three subjects did report lower-back discomfort in the days post-exercise, the maximum value being 4 (fairly mild pain).

7.5 DISCUSSION

As in previous experiments, not only did pre-exercise standing lead to stature changes that were comparable between weeks but so too did the drop jumping regimen. Statistical analysis showed no significant difference between regimens when comparing changes in stature during this phase of the experiment. Nor did it suggest that variance could be accountable to subjects. A decrease in stature of only 0.14 mm and 0.01 mm for the 10 min prior to exercise implied that any abnormal spinal loading had been controlled and subjects were approaching the trough in their circadian variation in stature (Reilly et al., 1984).

Increased rates of shrinkage following the 6 min drop jumping regimen on both occasions were significant, a mean loss in stature of 1.74 mm being recorded. Thus, hypothesis 8 could be accepted. The drop jumping regimen undertaken in this experiment had been chosen to be comparable, in terms of its duration and landing impacts, to the bounding regimen executed in previous investigations. The exercise-induced shrinkage reported on that occasion (1.7 mm) was found to be comparable.

Gravity-inversion post-exercise induced a rapid increase in stature, significantly in excess of that occurring during a similar period of standing. If it is assumed that regains in stature observed during this small time period approach linearity, given that it is only of short duration and the disc height changes are likely to emanate from elastic responses within the disc, then it is possible to conclude that exercise-induced shrinkage was negated within approximately 7 min. Therefore, it seems reasonable to accept hypothesis 9. The increase in stature found during inversion exceeded values reported by Leatt et al., (1985) by approximately 1 mm, for the same time period and time of day. This apparent discrepancy may be explained by the prior spinal loading induced during exercise which will have

led to an overall decrease in disc height. Consequently, both the annular stress and osmotic pressure within the discs (Koeller et al., 1984) will have been greater and on removing the load, during inversion, an increased rate of disc height recovery will have occurred.

In this and the previous experiment, no measure of spinal blood-flow was recorded, either pre- or post- exercise. Such a factor may have an influence on the rate of fluid imbibition by the disc. This should be a variable considered in future studies using contemporary imaging techniques.

What appeared to be of major significance following the period of inversion was the apparent rapid loss in stature during the 40 min standing recovery. Within 30 min of this standing period there was no significant difference in overall stature change for the two test regimens. The 60 min standing post-exercise during regimen 1 was found to invoke a net regain in stature of only 0.45 mm. Both these results, along with stature loss noted during the initial 30 min of the experiment, suggest a return to the normal diurnal shrinkage curve, findings in line with those put forward by Klingenstierna and Pope (1987). Consequently, for those athletes seeking to structure their training programmes to lessen the risks of incurring back injury, it might seem prudent for them to alternate brief bouts of exercise with unloading postures during a single training session.

The rate of shrinkage following inversion was investigated by fitting both a linear and exponential curve to the data which produced the following equations:-

$$H = 2.58 - 9.72 \times 10^{-2}t \quad \text{(linear regression equation)}$$

$$H = -5.55 + 5.43\, e^{-0.024t} \quad \text{(exponential curve fit)}$$

where:-

H = stature loss (mm)

t = time period (min)

A correlation coefficient of 0.96 was obtained from the linear regression analysis. Substituting a time factor of 90 min into both these equations it is estimated that losses in stature would be between 6.17 mm and 4.93 mm for the regression and curve fit equations, respectively. A comparison of these values with those reported by both Leatt et al. (1986) (5.53 mm) and Wilby et al. (1987) (6.93 mm) for the first 90 min after rising, would suggest that stature loss following inversion is similar in magnitude to that occurring first thing in the morning, directly after rising. Consequently, it may be that the mechanisms underlying the disc height responses following inversion are similar to those governing shrinkage immediately after rising.

Wilby et al. (1987) identified an inverse relationship between back strength and shrinkage, proposing that the former could well compensate for disc height losses. During this experiment there appeared to be a positive correlation between post-inversion shrinkage and back strength measurements (week 1 and week 2). This apparent contradiction may stem from the nature of the exercises studied. Wilby et al. (1987) employed a continuous period of aerobic exercises in the form of circuit weight training, entailing sustained back muscle activity. Such back muscle exertions are unlikely to be as extensive during drop jumping. In addition, back strength and shrinkage during the present study were not a function of body size since body weight and shrinkage were unrelated.

During the period of inversion, a negative correlation was found between body weight and increases in stature with respect to initial height. This too would seem surprising considering that inversion relies on the subject's own body weight to act as an unloading force. Thus, heavier subjects would be expected to undergo greater rates of stature recovery. Disagreement between theory and observations may be apparent if consideration is given to the findings of Leatt et al. (1985). They showed that 50° inversion was more effective than 90° inversion when unloading the spine. Ratings of discomfort suggested that the latter posture was more uncomfortable, resulting in increased truncal stresses opposing the unloading forces. Similarly, in this study the heavier subjects may have found it more difficult

to relax during periods of inversion, although no measures of discomfort were undertaken.

The cross-correlation of anthropometric and functional measures revealed a significant relationship between back strength and leg strength. At first this would not seem unreasonable as a muscular physique would contribute to the strength of both back and leg muscles. However, it may raise questions about the measuring procedure's ability to discriminate between the two parameters. The two measures were conducted with the same piece of equipment, with only slight adjustments made to the posture. Adopting and ensuring correct muscle group involvement is subject dependent and should be investigated further before drawing any conclusions. The significant negative correlations between both these strength measures and the time for the Margaria stair test is in line with the commonly held belief that strength is a pre-requisite for high power output (Reilly, 1981a).

Body height was found to be inversely correlated with the measure for hip flexibility implying that the taller individuals were less flexible. A similar significant correlation was not found when examining lumbar flexibility and body height.

Measures of delayed onset muscle soreness were found to be less severe for drop jumping exercise than previously reported bounding routines, although, a similar drop in symptom severity was found following repeated exercise. The peak in delayed muscle soreness was found to occur some 54 hours after exercise which was approximately 11 hours later than the peak reported following a bounding regimen.

These findings would appear to agree with Ebbeling and Clarkson (1989) that the severity and delay in muscle soreness symptoms are dependent on the type of exercise.

7.6 SUMMARY

1. When loading conditions were identical, measures of mean changes in stature were found to be repeatable.

2. A short duration drop jumping regimen (6 min) increased the rates of stature loss and caused mean shrinkage of 1.74 mm. This was comparable with shrinkage occurring following a broad jumping regimen of similar duration and impact landings.

3. A 20 min period of gravity inversion post-exercise resulted in a rapid regain in stature (5.18 mm). This increase in stature was greater (1 mm) than that reported by Leatt et al. (1986) for the same duration and time of day. However, the stature change reported in this experiment had followed a period of spinal loading. These findings would imply that the more loaded discs have the greater affinity for height restoration, an assumption in line with the increased annular stress and osmotic pressure known to exist within discs subjected to greater compression (Koeller et al., 1984).

4. A rapid loss in stature occurred following the 20 min gravity inversion. Within 30 min of this standing period, differences in stature compared to a control period were found to be non-significant. The rapid return to the 'normal' diurnal stature change would appear to be in agreement with Klingenstierna and Pope (1987).

5. The implication of this quick restoration of normal disc response would seem to be relevant for those athletes seeking to structure training programmes to lessen the risks of incurring back injury. It seems prudent that they should alternate unloading postures and brief bouts of exercise within a single training session.

6. Negative correlations were found when stature regains (with respect to initial height) during inversion were compared with subjects' body mass. It would appear that heavier individuals may find it more difficult to relax when undergoing gravity inversion than lighter individuals.

7. When compared with a bounding regimen, delayed muscle soreness was found to be less severe and peak at a later stage following the drop jumping regimen.

8. BIOMECHANICAL LOADING OF THE HUMAN SPINE DURING DROP JUMPING

8 BIOMECHANICAL LOADING OF THE HUMAN SPINE DURING DROP JUMPING

8.1 INTRODUCTION

Only a few biomechanical studies have investigated forces occurring within the human body during drop landing and drop jumping (Smith, 1975; Bobbert et al., 1987a, 1987b). The majority of these have been restricted to forces acting about the ankle, knee and hip joints. One study is known to have examined lower back compression forces (Berme et al., 1984) during drop landings, but these were confined to heights of 0.33 m. Athletes participating in plyometric exercises are known to execute drop jumps from heights in excess of this value.

A common trend running throughout these studies is their use of rigid-link segment modelling to quantify joint loads. In such instances peak forces are often presented as a measure of risk. Considering the potential shortcomings of such a technique to modelling highly dynamic impacts (refer to section 3.4.5.7), a number of questions may be raised as to the validity of conclusion based solely on peak forces. It was the intention of this study to examine rigid-link segment modelling of lower back compression and shear forces during drop jumping. Analysis included a number of salient features of the force-time curves. In addition, it sought to investigate whether biomechanically derived measures of loading were comparable with the shrinkage observed following a drop jumping regimen.

8.2 AIMS

The aim of the experiment was to:

i) Estimate the *in-vivo* loads placed on the spine during the execution of drop jumping exercises. Three methods of investigation will be implemented; biomechanical modelling, EMG and changes in stature.

The approaches adopted will serve to:

i) Assess each technique's ability to measure loads imposed on the spine during impact landing.

ii) Establish the loads with which the spine must contend while performing drop jumping activities and make assumptions as to the potentially hazardous nature of such activities.

8.3 OBJECTIVES

The investigation set out the following objectives:

i) To develop a biomechanical model to estimate musculoskeletal forces about the lower back during the landing phases of a drop jumping exercises from two different heights - 0.75 m and 0.30 m.

ii) In conjunction with i), to record the electrical activity of three major muscle groups, erector spinae, external obliques and rectus abdominis, while performing drop jumps from heights of 0.75 and 0.30 m.

iii) In conjunction with i) and ii), to record changes in stature following two drop jumping regimens using 0.75 and 0.30 m as the drop heights.

8.4 HYPOTHESES

An important factor governing maximum power output during plyometric based exercises is the speed during the eccentric phase of the muscle action. During drop jumping, the greater the drop height the greater the deceleration of the body mass which in turn influences the speed of the eccentric muscle action. By increasing the body's impact velocity at ground contact there must inevitably be an increase in the forces transmitted to and ultimately absorbed by the body. The hypotheses put forward are:

Hypothesis 10: Drop jumping activities from a height of 0.75 m induce higher spinal compression and shear forces on the L5/S1 intervertebral disc, as measured using a biomechanical model, than those performed from a height of 0.30 m.

Hypothesis 11: When undertaking 0.75 m drop jumps there is an increased muscle activity of the erector spinae, external obliques and rectus abdominis, above that recorded during 0.30 m drop jumps.

Hypothesis 12: Dropping from a height of 0.75 m during the execution of drop jumps causes greater spinal shrinkage than when performing the same jumps from a height of 0.30 m.

8.5 EXPERIMENTAL PROCEDURES

The measurement of stature changes, muscle activity and the data necessary for a biomechanical model required a considerable amount of technical apparatus. In addition, certain pieces of equipment were restricted to specific locations; for example, a force plate, necessary for obtaining ground reaction forces for input into the biomechanical model could not be relocated from its specially designed pit. Consequently, it was necessary to conduct two separate experimental sessions, one concerned purely with a biomechanical approach to modelling forces and the second with the measurement of muscle activity and stature alterations. On both occasions high speed film was taken of the drop jumping regimens, thus, providing information as to the repeatability of performance. All subjects took part in both sessions, the order of which was randomised.

The following sections will discuss each test session, beginning with a description of the biomechanical model, describe how data were collected for the three methods of investigation, and conclude by drawing together the findings.

8.6 DEVELOPMENT OF THE BIOMECHANICAL MODEL

8.6.1 The Rigid-Link Segment Modelling Approach - Assumptions and Simplifications

i) Drop jumping can be viewed as a two-dimensional activity with the athlete restricting gross movements to the sagittal plane. Consequently, ground reaction forces and muscle moments acting in the frontal plane were considered negligible. Therefore, to reduce the complexity of the modelling approach the analysis was restricted to two dimensions. This is in line with studies performed by other authors (Smith, 1975; Bobbert et al., 1987a, 1987b).

ii) Body segments were modelled as simple rigid-links articulating about hinge type joints (Figure 25). These joints were non-complex articulating surfaces with point load contact. Although

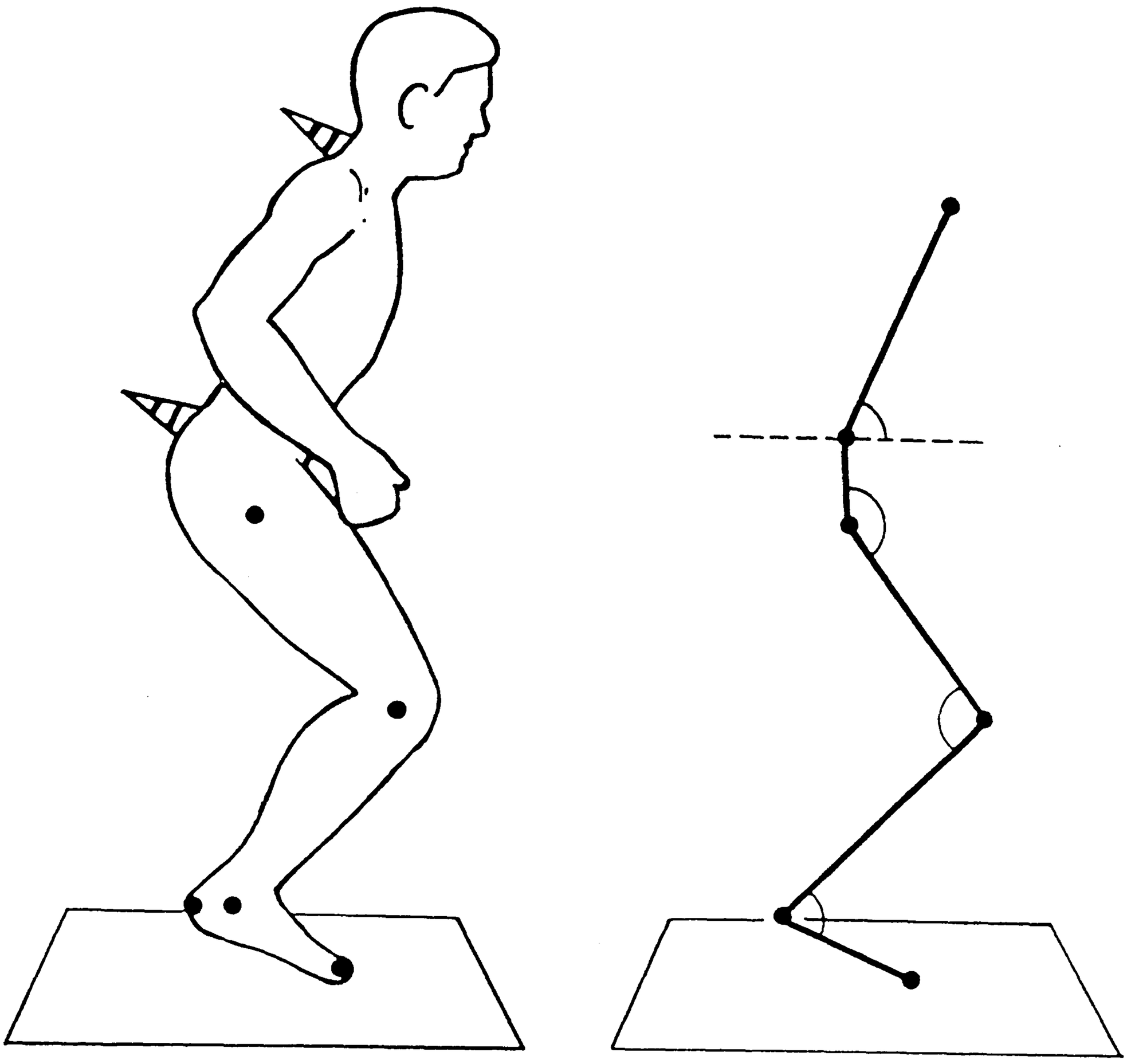


Figure 25. Position of body landmarks and spinal fins in order to construct the five segment link model.

this makes gross simplifications as to the structure and mechanical functioning of the joint, it has been a well accepted practice throughout the biomechanical literature (Plagenhoef, 1971; Winter, 1979).

iii) As the approach sought to measure the external forces acting at the feet, by implementing a force platform, it was only necessary to consider the lower limbs. Thus, only the feet, shank, thigh and a lower section of the trunk, referred to as the pelvis, were modelled as rigid-link segments. The pelvis was considered to be a rigid-link joining the hip joint centre with the centre of the L5/S1 intervertebral disc. Forces acting about this disc centre have been the focus of attention for a majority of mechanical studies of the spine as it represents the point at which the greatest forces are experienced. Epidemiological studies have provided further support for this assumption by identifying it as the most likely site of spinal injury. An additional link segment was considered, joining L5/S1 to C7/T1, which only served to provide a measure of trunk angle.

iv) To enable ground reaction forces, and consequently centre of pressure to be located with respects to the foot, subjects landed with only one foot on the force platform. It was assumed, due to the symmetrical nature of the activity, that equivalent forces would be transmitted to both feet.

v) The rigid-link segments were considered to have a fixed mass located at their mass centre, which did not alter during movement. Each segment's moment of inertia was also considered to remain constant during movement.

These values were principally derived from Dempster's (1955) work on cadavers to maintain compatibility with other studies on spinal loading (Leskinen et al., 1985; Freivalds et al., 1984). Unfortunately, Dempster's (1955) work did not consider separate sections of the trunk and it was necessary to obtain segmental parameters for the pelvis using data derived by Zatsiorsky and Seluyanov (1983). A computer program was developed which provided a number of segmental parameters using equations derived

by other researchers (Clauser et al., 1969; Chandler et al., 1975; Widule 1976; McConville et al., 1980) and which could also be selectively input into the biomechanical model. These studies formulated regression equations based on anthropometric dimensions to facilitate more personalised segmental parameters. In all, 20 dimensions were required by the computer program.

vi) The internal force distribution used to calculate spinal compression and shear forces consisted of a simple muscle model of the spine. It assumed that all back extensor moments were created by the action of the main extensor muscle group, the erector spinae. This has been standard practice throughout many biomechanical studies concerned with symmetrical sagittal plane lifting (Morris et al., 1961; Leskinen et al., 1983; McGill et al., 1985). The control of flexor moments was assumed to be the sole responsibility of the external obliques. The rectus abdominis, although possessing a greater moment arm, was not included in the muscle model. Preliminary investigations had found it to be inactive during the majority of such activities. In all instances, the muscle forces were always assumed to act parallel to the inclination of the trunk. Erector spinae muscle moment arms of 50 and 70 mm were used during analysis as these were felt to represent the extremes identified in the literature (Table 2). For the external obliques a muscle moment arm of 64 mm was implemented in accordance with the findings of Németh and Ohlsén (1986).

vii) The muscle model assumed that no antagonistic muscle actions took place. Although this is unlikely, it was necessary to reduce the number of unknowns and facilitate solvable equations. By undertaking such an approach it would inevitably lead to an under-, rather than over-estimation of the compressive force. Although more complex muscle models have been developed which have considered up to 10 separate muscles (Bean et al., 1988), they require complex linear programming techniques to estimate the likely muscle force contributions. Preliminary studies suggested that such an approach would not significantly improve

the predictive power of the present model due to the symmetrical nature of the activity being studied and the inherent limitations associated with such techniques.

viii) An intra-abdominal pressure (IAP) force was not included in the muscle model of the lower back. Controversy exists as to the beneficial role IAP plays in relieving load on the spine. Several modellers (Troup et al., 1983; McGill and Norman, 1985; Gagnon et al., 1987; Kromodihardjo and Mital, 1982) of lifting activities have sought to ignore such a force component due to its unpredictable nature, the difficulty in measuring and quantifying the force, and as McGill and Norman (1986) pointed out, in cases where there is an increase in abdominal pressure there is often an increase in the activity of the abdominal muscle which will in itself increase the load on the spine. Whether the abdominal pressure moment always exceeds the increased muscle moment is unclear; the work by McGill and Norman (1986) on lifting exercises would suggest not.

During drop jump landings, the situation may be somewhat different to lifting. McGill et al. (1988) measured both IAP changes and abdominal muscle activities on landing from 40 cm high platforms and concluded that the changes in IAP were more likely a result of hydrostatic compression of the decelerating viscera, rather than actual muscle activity. Thus, if IAP was to play a role in reducing spinal loading during landing its beneficial effects should be more noticeable during maximum body deceleration. It is also likely that the viscera would be forced into oscillation, causing rapid changes in IAP.

Provision of an accurate and reliable measure of IAP was not possible and it was concluded that in order to prevent misleading assumptions the model should exclude an IAP force. However, when drawing conclusions its likely effects should not be ignored as IAP may well reduce or increase load on the spine.

On the basis of these assumptions a five-link segment, dynamic, symmetrical, two-dimensional, sagittal plane model was constructed to calculate compressive and shear forces between the L5 and sacrum. By

considering each segment independently, beginning with the foot, and knowing all the forces acting on the segment, it was possible to conduct an inverse dynamics approach based on Newtonian mechanics. Thus, for each segment:-

$$\Sigma \text{ forces} = \text{mass} \times \text{acceleration}$$

$$\text{thus: } \Sigma \text{ forces} - (\text{mass} \times \text{acceleration}) = 0$$

This gives rise to 12 equations for each major segment producing both net horizontal and vertical joint reaction forces, and net muscle moments at the ankle, knee and hip joint centres, and L5/S1 intervertebral disc. The implementation of a simple muscle model across a transverse section through the L5/S1 disc centre facilitated the calculation of compression and shear forces. A discussion of each segment's equilibrium constraints and associated mathematical equations will follow.

8.6.2 Biomechanical Model Analysis

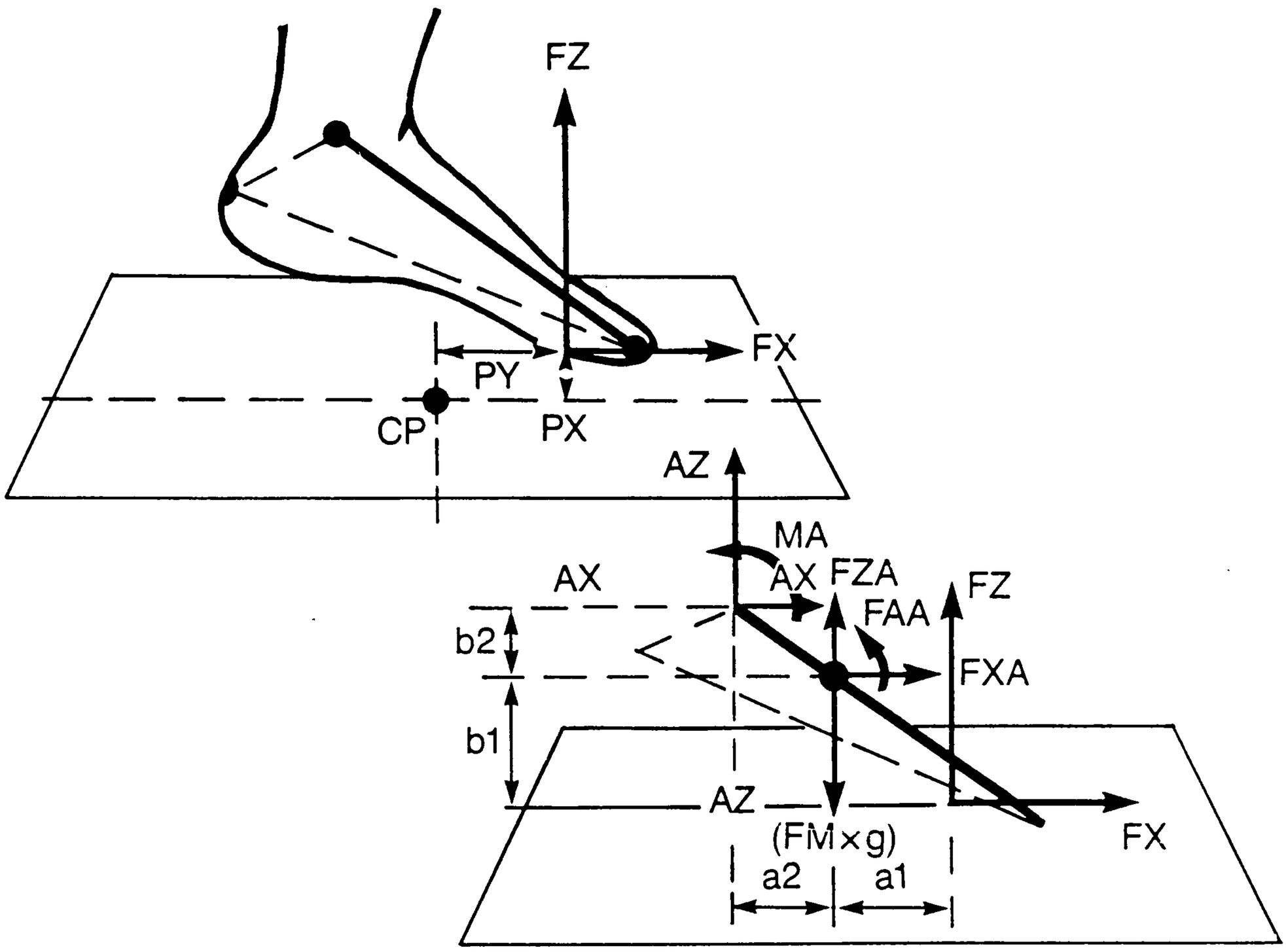
In the following mathematical model the normal sign convention is used where all forces acting to the right and vertically upwards, along with anti-clockwise moments are viewed as positive. Likewise, linear and angular kinematics possess the same sign convention.

i) **Foot Segment** (refer to Figure 26)

Externally derived ground contact forces and their relative positions with respect to the foot were recorded directly from a force platform and used to calculate ankle joint reaction forces and muscle moments. To facilitate a fully dynamic approach, it was necessary to consider inertial characteristics of the foot segment, which included foot segment mass, moment of inertia and linear accelerations.

Resolving for forces in the vertical and horizontal plane Equations (1) and (2) were established.

FOOT SEGMENT



Where:

FZ = Vertical ground reaction force
FX = Horizontal ground reaction force
AZ = Vertical reaction force at the ankle joint
AX = Horizontal reaction force at the ankle joint
FZA = Vertical acceleration of the foot's centre of mass
FXA = Horizontal acceleration of the foot's centre of mass
FAA = Angular acceleration of the foot about its centre of mass
MA = Net muscle moment acting at the ankle joint
FM = Mass of the foot
FI = Moment of inertia of the foot
g = Acceleration due to gravity ($9.81 ms^{-2}$)
CP = Centre of the force platform
PX = Anterior/posterior co-ordinates of the centre of pressure
PY = Lateral co-ordinates of the centre of pressure
a1 = Perpendicular distance between vertical ground reaction force and the foot's centre of mass
a2 = Perpendicular distance between vertical ankle joint reaction force and the foot's centre of mass
b1 = Perpendicular distance between horizontal ground reaction force and the foot's centre of mass
b2 = Perpendicular distance between horizontal ankle joint reaction force and the foot's centre of mass

Figure 26. Modelling assumptions and nomenclature used at the foot.

Σ Vertical forces acting on the foot = 0

$$FZ + AZ - (FM \times g) - (FM \times FZA) = 0$$

Thus : $$AZ = (FM \times g) + (FM \times FZA) - FZ \qquad (1)$$

Σ Horizontal forces acting on the foot = 0

$$FX + AX - (FM \times FXA) = 0$$

Thus : $$AX = (FM \times FXA) - FX \qquad (2)$$

Thus, the ankle joint reaction forces could be determined by experimentally deriving the remaining unknowns.

To determine the net muscle moment acting about the ankle joint, moments were taken about the foot's centre of gravity which lead to Equation (3).

Σ Moments about the centre of gravity of the foot = 0

$$MA + (FZ \times a1) + (FX \times b1) - (AX \times b2) - (AZ \times a2) - (FI \times FAA) = 0$$

Thus :

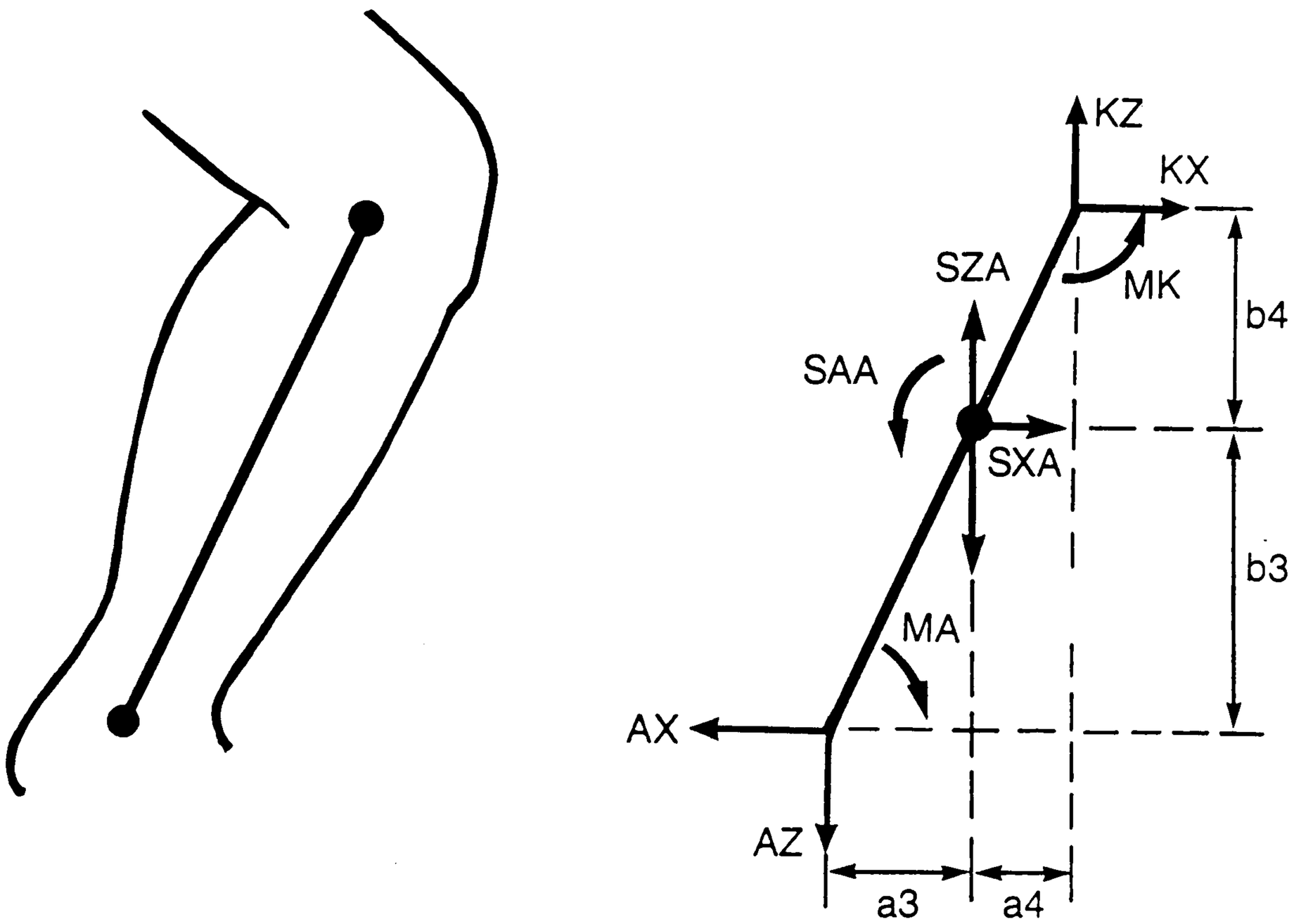
$$MA = (FI \times FAA) - (FZ \times a1) - (FX \times b1) + (AX \times b2) + (AZ \times a2) \qquad (3)$$

The values derived from equation (1), (2) and (3) were all used to determine the forces acting about the knee joint. It was important to know the direction in which these forces were acting so they could be reversed when considering the next adjoining segment; the shank (i.e. equal and opposite forces must act to maintain equilibrium about the ankle joint).

ii) **Shank Segment** (refer to Figure 27)

The same procedure was implemented to derive joint reaction forces and muscle moments about the knee. Referring to Figure 27 and resolving in the vertical and horizontal planes

SHANK SEGMENT



Where:

AZ = Vertical reaction force at the ankle joint
AX = Horizontal reaction force at the ankle joint
KZ = Vertical reaction force at the knee joint
KX = Horizontal reaction force at the knee joint
SZA = Vertical acceleration of the shank's centre of mass
SXA = Horizontal acceleration of the shank's centre of mass
SAA = Angular acceleration of the shank about its centre of mass
MA = Net muscle moment acting at the ankle joint
MK = Net muscle moment acting at the knee joint
SM = Mass of the shank
SI = Moment of inertia of the shank
g = Acceleration due to gravity (9.81ms^{-2})
a3 = Perpendicular distance between vertical ankle joint reaction force and the shank's centre of mass
a4 = Perpendicular distance between vertical knee joint reaction force and the shank's centre of mass
b3 = Perpendicular distance between horizontal ankle joint reaction force and the shank's centre of mass
b4 = Perpendicular distance between horizontal knee joint reaction force and the shank's centre of mass

Figure 27. Modelling assumptions and nomenclature used at the shank.

Σ Vertical forces acting on the shank = 0

$$KZ - AZ - (SM \times g) - (SM \times SZA) = 0$$

Thus : $$KZ = AZ + (SM \times g) + (SM \times SZA) \quad (4)$$

Σ Horizontal forces acting on the shank = 0

$$KX - AX - (SM \times SXA) = 0$$

Thus : $$KX = AX + (SM \times SXA) \quad (5)$$

Taking moments about the shank's centre of gravity.

Σ Moments about the shank's centre of gravity = 0

$$MK - MA - (AX \times b3) - (KX \times b4) + (AZ \times a3) + (KZ \times a4) - (SI \times SAA) = 0$$

Thus :

$$MK = MA + (AX \times b3) + (KX \times b4) - (AZ \times a3) - (KZ \times a4) + (SI \times SAA) \quad (6)$$

iii) Thigh Segment (refer to Figure 28)

Similarly, the knee joint reaction forces and muscle moments derived from Equations (4), (5) and (6) were used to calculate hip joint reaction forces and the net muscle moment.

Resolving both vertically and horizontally,

Σ Vertical forces acting on the thigh = 0

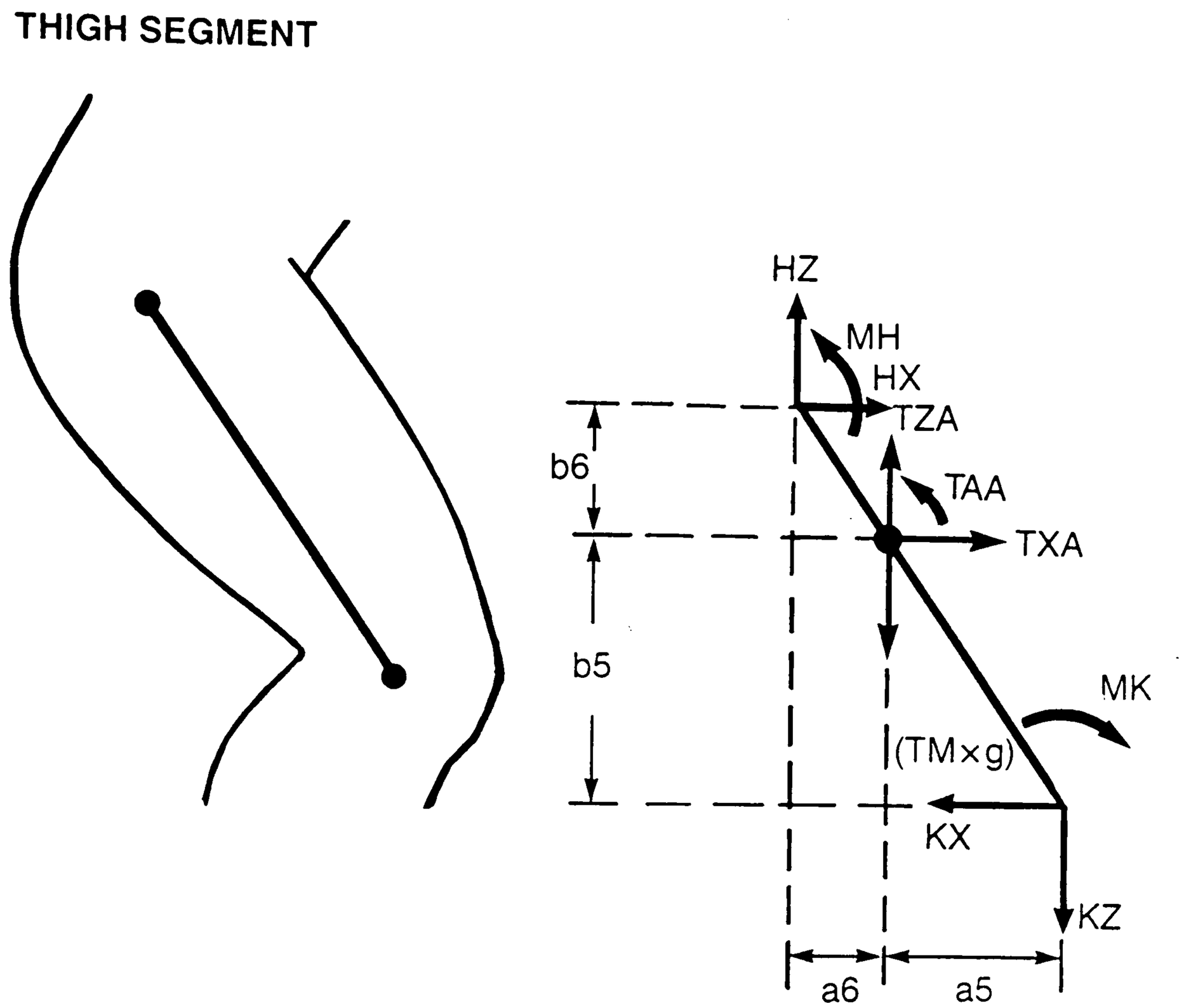
$$HZ - KZ - (TM \times g) - (TM \times TZA) = 0$$

Thus : $$HZ = KZ - (TM \times g) + (TM \times TZA) \quad (7)$$

Σ Horizontal forces acting on the thigh = 0

$$HX - KX - (TM \times TXA) = 0$$

Thus : $$HX = KX + (TM \times TXA) \quad (8)$$



Where:

KZ = Vertical reaction force at the knee joint
KX = Horizontal reaction force at the knee joint
HZ = Vertical reaction force at the hip joint
HX = Horizontal reaction force at the hip joint
TZA = Vertical acceleration of the thigh's centre of mass
TXA = Horizontal acceleration of the thigh's centre of mass
TAA = Angular acceleration of the thigh about its centre of mass
MK = Net muscle moment acting at the knee joint
MH = Net muscle moment acting at the hip joint
TM = Mass of the thigh
TI = Moment of inertia of the thigh
g = Acceleration due to gravity (9.81ms^{-2})
a5 = Perpendicular distance between vertical knee joint reaction force and the thigh's centre of mass
a6 = Perpendicular distance between vertical hip joint reaction force and the thigh's centre of mass
b5 = Perpendicular distance between horizontal knee joint reaction force and the thigh's centre of mass
b6 = Perpendicular distance between horizontal hip joint reaction force and the thigh's centre of mass

Figure 28. Modelling assumptions and nomenclature used at the thigh.

Taking moments about the thigh's centre of gravity.

Σ Moments about the thigh's centre of gravity = 0

$$MH - MK - (KX \times b5) - (KZ \times a5) - (HX \times b6) - (HZ \times a6) - (TI \times TAA) = 0$$

Thus : $$MH = MK + (KX \times b5) + (KZ \times a5) + (HX \times b6) + (HZ \times a6) + (TI \times TAA) \quad (9)$$

iv) Pelvis Segment (refer to Figure 29)

Hip joint forces and net muscle moment calculated from equations (7), (8) and (9) were used to estimate the reaction forces and net muscle moment acting at the L5/S1 intervertebral disc.

Resolving vertically and horizontally,

Σ Vertical forces acting on the pelvis = 0

$$LZ - HZ - (PM \times g) - (PM \times PZA) = 0$$

Thus : $$LZ = HZ + (PM \times g) + (PM \times PZA) \quad (10)$$

Σ Horizontal forces acting on the pelvis = 0

$$LX - HX - (PM \times PXA) = 0$$

Thus : $$LX = HX + (PM \times PXA) \quad (11)$$

Taking moments about the pelvis centre of gravity.

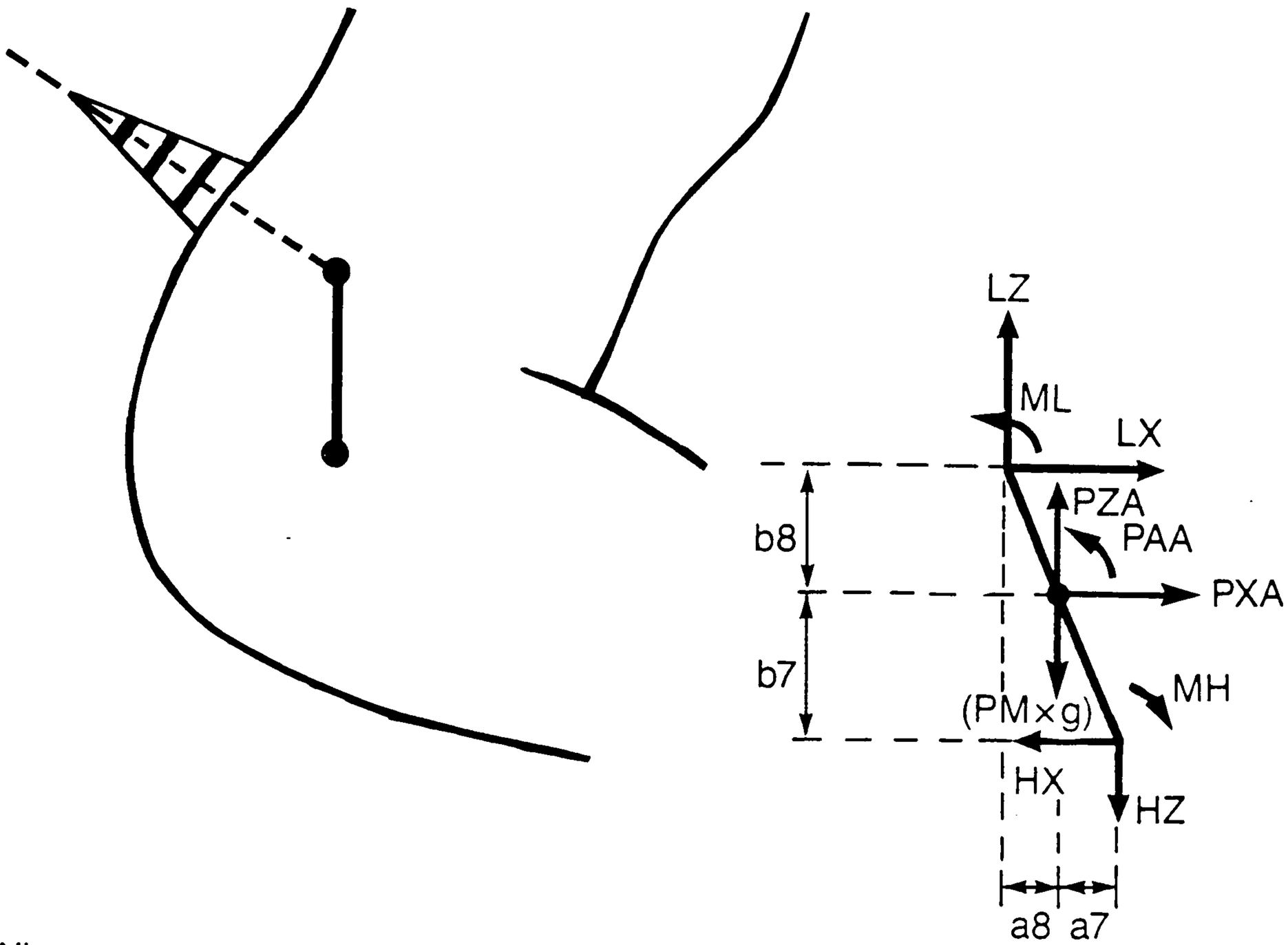
Σ Moments about the pelvis centre of gravity = 0

$$ML - MH - (HX \times b7) - (HZ \times a7) - (LX \times b8) - (LZ \times a8) - (PI \times PAA) = 0$$

Thus :

$$ML = MH + (HX \times b7) + (HZ \times a7) + (LX \times b8) + (LZ \times a8) + (PI \times PAA) \quad (12)$$

PELVIS SEGMENT



Where:

HZ = Vertical reaction force at the hip joint
HX = Horizontal reaction force at the hip joint
LZ = Vertical reaction force at the L5/S1 intervertebral disc
LX = Horizontal reaction force at the L5/S1 intervertebral disc
PZA = Vertical acceleration of the pelvis centre of mass
PXA = Horizontal acceleration of the pelvis centre of mass
PAA = Angular acceleration of the pelvis about its centre of mass
MH = Net muscle moment acting at the hip joint
ML = Net muscle moment acting at the L5/S1 intervertebral disc
PM = Mass of the pelvis
PI = Moment of inertia of the pelvis
g = Acceleration due to gravity ($9.81 ms^{-2}$)
a7 = Perpendicular distance between vertical hip joint reaction force and pelvis centre of mass
a8 = Perpendicular distance between vertical L5/S1 joint reaction force and pelvis centre of mass
b7 = Perpendicular distance between horizontal hip joint reaction force and pelvis centre of mass
b8 = Perpendicular distance between horizontal L5/S1 joint reaction force and pelvis centre of mass

Figure 29. Modelling assumptions and nomenclature used at the pelvis.

Equations (10), (11) and (12) provide the joint reaction forces and net muscle moment acting at the L5/S1 intervertebral disc. To determine the actual compression and shear forces acting on this disc it was necessary to take account of the active forces created by the back muscles. The muscle model proposed earlier assumed all muscle activity to act parallel to the inclination of the trunk. Therefore, the shear force would not be influenced by muscle activity.

v) L5/S1 Compression and Shear Forces (refer to Figure 30)

The muscle constraints assumed that a positive net muscle moment would be created by the action of the erector spinae muscles and a negative moment by action of the external obliques.

Thus,

If ML $<$ 0 then ES = 0

If ML $\geq$ 0 then EO = 0

Therefore:

ML $<$ 0, EO = ML/OA = MF

Therefore:

ML $\geq$ 0, ES = ML/EA = MF

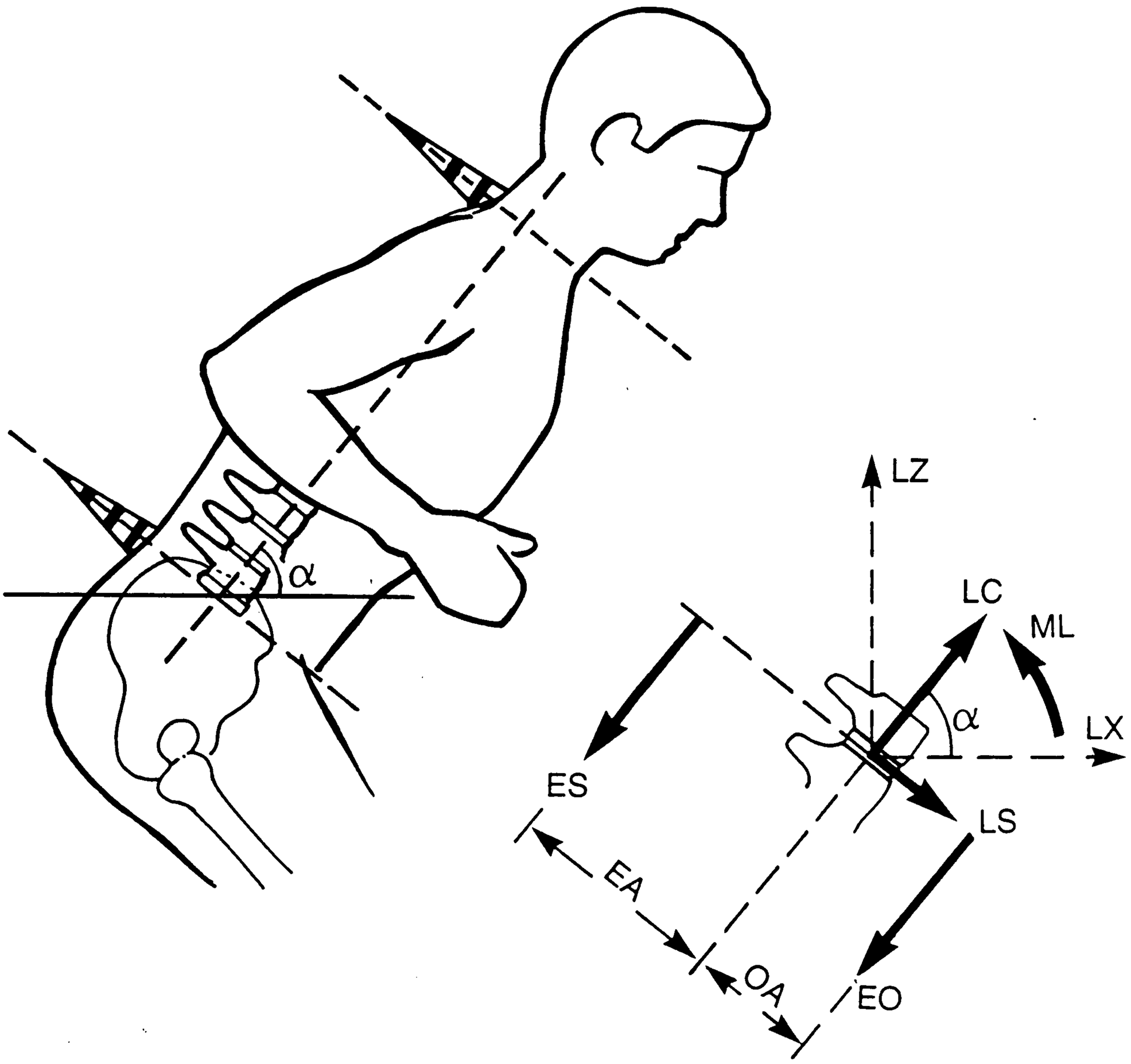
Consequently,

Total compressive force = Muscle force activity + joint reaction force parallel to trunk angle

$$LC = MF + LZ \sin \alpha + LX \cos \alpha \qquad (13)$$

Total shear force = joint reaction force perpendicular to trunk angle

$$LS = LX \sin \alpha - LZ \cos \alpha \qquad (14)$$



Where:

LZ = Vertical reaction force at L5/S1
LX = Horizontal reaction force at L5/S1
ML = Net lumbar muscle moment
LC = Compressive force at L5/S1
LS = Shear force at L5/S1
ES = Erector spinae muscle force
EO = External oblique muscle force
EA = Erector spinae muscle moment arm
DA = External oblique muscle moment arm
α = Trunk angle

Figure 30. Modelling assumptions and nomenclature used at the L5/S1 intervertebral disc.

8.7 METHODOLOGY

8.7.1 Subjects and Experimental Protocol

Six healthy male subjects (mean age = 24 ± 3 yr; mean height = 1.84 ± 0.05 m; mean body mass = 74.8 ± 6.7 kg) who regularly participated in some form of physical exercise, but did not use drop jumping exercises as a method of training, took part in the experiment. None of the subjects reported having any back or lower limb ailments and all gave informed consent prior to testing. Each subject underwent four separate test sessions during which data were collected to enable two experimental approaches to be undertaken - biomechanical modelling (Test 1) and combined analysis of electromyographical and stature changes (Test 2). The independent variable examined during both the experimental approaches was drop jump height (0.75 and 0.30 m). Each test session was carried out on a separate day and only one subject was studied per day, commencing at 14:00 hours, to control for any time of day effects. The time between a single subject's test session was never less than one week, thus eliminating any possible effects that muscle soreness might have on performance, and never exceeded two weeks so that subjects remained familiar with the experimental procedures. The exercise regimen and method of collecting kinematic data were identical on every occasion.

All subjects had had a preliminary introduction to both the exercise regimen and the method of measuring stature. When using the stadiometer all subjects attained a standard deviation of 0.5 mm or less for ten consecutive measures. The exercise regimen consisted of 25 drop jumps from a gymnastic vaulting box, the height of which was varied by removing sections of the box. Just the top section of the box constituted a height of 0.30 m, three sections, a height of 0.75 m. A gradual inclined staircase enabled subjects to mount the box without undue stress. On reaching the edge of the vaulting box, subjects were instructed to step out with their right foot and then immediately bring the left foot alongside the right. Once subjects were clear of the box they were encouraged to look forward and not at their feet, thereby, reducing the likelihood of over-balancing on landing. As soon as subjects made contact with the ground they were instructed that they must immediately perform a maximum vertical jump. On making ground contact for the second time, subjects were not

allowed to proceed until they had fully controlled the second landing and obtained an upright stance, with the knees fully extended.

The 25 drop jumps were executed in sets of 5 with 45 s separating each. Prior to the exercise regimen, all subjects stood with their weight evenly distributed for a period of 30 minutes. This was in line with previous experiments which used this time to control for any abnormal activity that might have preceeded testing and which might have caused a loading or unloading of the spine.

During the biomechanical investigation, subjects performed the drop jump with their right foot on a force platform and their left foot on the concrete floor, at the side of the force plate, the surfaces of which were level. This enabled centre of pressure to be located with respect to the foot. For the investigation into muscle activity and stature change, drop jumps were undertaken with both feet on a concrete floor, although an outline of the force platform, along with a pressure sensitive mat was placed on the floor to ensure subjects contacted the ground at a similar distance from the vaulting box (0.50 m).

8.7.2 Acquisition of Kinematic Data

During both the biomechanical investigation and the study into trunk muscle activity and stature alterations, kinematic data were taken from selected drop jumps. This provided relevant data for input into the biomechanical model and facilitated a method by which to monitor repeatability, particularly as the two test sessions were being conducted under slightly differing laboratory conditions.

Kinematic data were recorded using a 16 mm Locam high speed cinematographic camera filming at a rate of 100 frames per second (100 Hz). An initial pilot study had implemented a sampling frequency of 200 Hz but this was found to generate excessive data and produce a high level of noise during analysis. Halving the sampling frequency considerably reduced both the noise and time required to perform an analysis.

The optical axis of the camera was perpendicular to the plane of motion (sagittal plane) and was set at a distance of 6 m from the

centre of the force platform. The centre of the camera lens was at a height of 0.90 m from the ground. Prior to any film recording, the Locam camera speed was verified by filming a calibrated 2 s sweep clock.

Prior to experimentation, a 1 m vertical ruler, aligned with plumb line, was filmed directly over a point on the force platform, or ground, where subjects had been instructed to land. This served as a method by which the film data could be scaled. During experimentation this was moved to a point within the field of view, but away from the performer, where it served as a reference point and vertical axis during data analysis.

Subjects dressed only in a pair of shorts and a pair of white track suit bottoms, devoid of one leg. This, along with a white back cloth, helped in the identification of certain body landmarks. On the side of the subject facing the camera (right side), small circle dots were drawn directly on the skin, using a non-permanent black marker pen. Points of identification were the greater trochanter, knee joint (on the lateral collateral ligament at the height of the joint cleft), lateral malleolus (0.5 cm anterior to the tip), most posterior point on the heel (acropodion) and the tip of the most anteriorly projecting toe (pternion). Two lightweight aluminium fins helped locate the centre of the L5/S1 and C7/T1 intervertebral joints (Plate 9). The fins measured 10 cm in length and 3.0 cm wide at their base, with 1.25 cm wide black and white bands marked alternatively along its length. A base plate, which helped secure the fins against the back measured 30 x 37 x 2 mm. Each fin was attached to the subjects' back using double side adhesive and strong elastic tape. A cloth band, with velcro fasteners, was also attached to the fins and past around the subject's body to aid stability. These fins were placed at the lowest point of the L5/S1 and C7/T1 spinous process, located using the palpation procedure described by Burton (1986).

Once all joint centres had been marked and the fins attached, a Harpenden anthropometer, tape measure and skinfold calipers were used to record 20 anthropometric dimensions. These served as input parameters to formulate body segment masses, centre of mass location and segmental moments of inertia. The 20 measures were:-

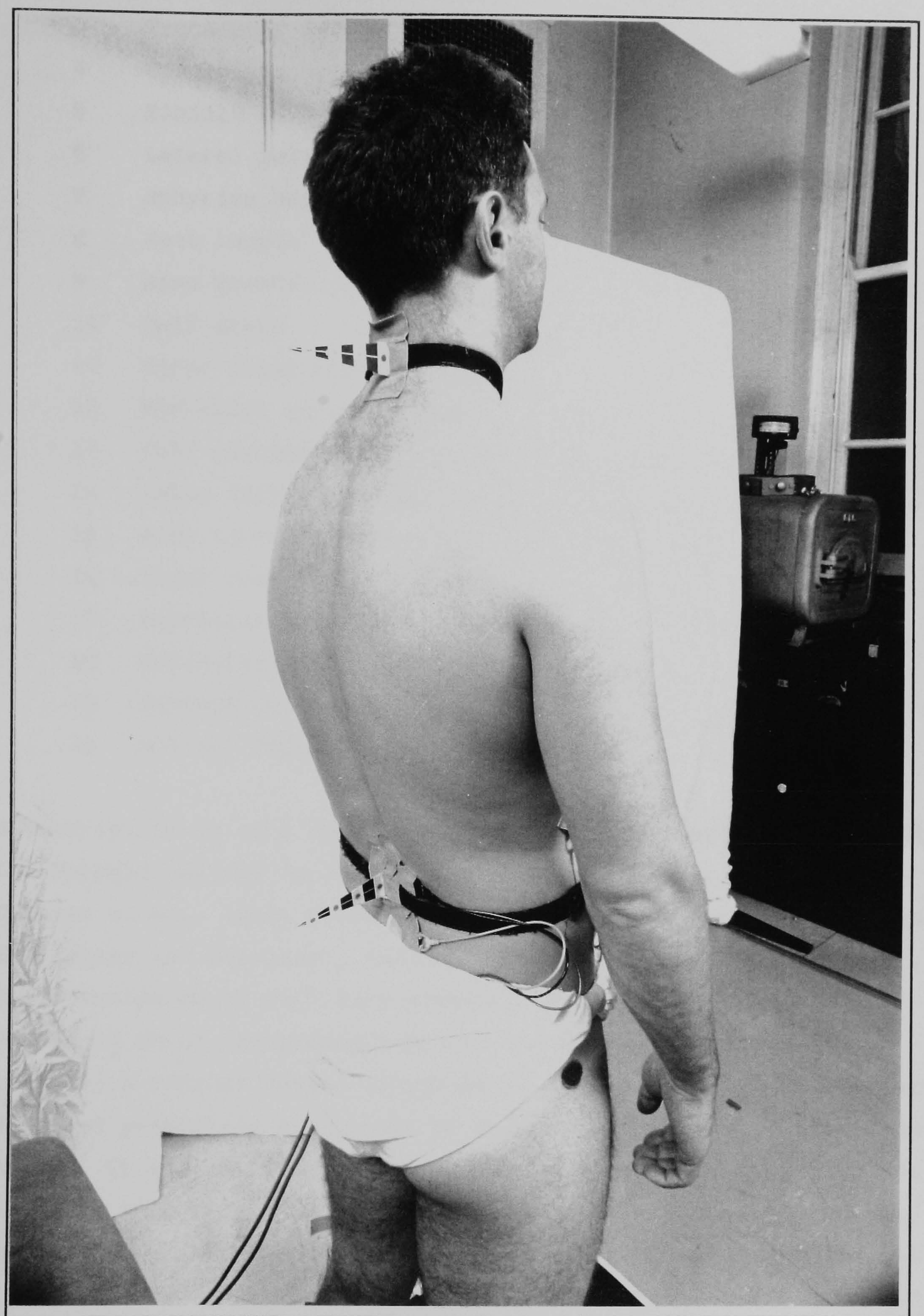


Plate 9 - Spinal fins used to locate and track intervertebral disc centres

1	Body weight
2	Stature
3	Trochanter height
4	Epicondyle height
5	Tibiale height
6	Lateral malleolus height
7	Sphyrion height
8	Foot length
9	Knee breadth
10	Calf depth
11	Upper thigh circumference
12	Mid-thigh circumference
13	Calf circumference
14	Ankle circumference
15	Arch circumference
16	Iliac crest fat
17	Omphalion height
18	Subischial height
19	Buttock circumference
20	Buttock depth

A description of each measure can be found in Appendix 7, along with the formula derived by six authors (Dempster, 1955; Widule, 1976; Clauser et al., 1969; Chandler et al., 1975; McConville et al., 1980; Zatsiorsky and Seluyanov, 1983) to estimate the mass and inertia characteristics of each body segment. A computer program was developed which, following input of the anthropometric measures, provided a tabular output (Appendix 7) of the calculated parameters. Selected parameters could then be implemented into the biomechanical model. It was anticipated that this would enable an investigation into the influence of such measures on the calculated compression forces. Dempster's (1955) data, along with Zatsiorsky and Seluyanov's (1983) data on the pelvis, were used in the initial model as these had been most frequently used throughout the literature.

Film data were collected from two jumps during each set of five, constituting a total of 10 recorded jumps. These were randomly selected and filmed without subjects' prior knowledge.

8.7.3 Acquisition of Kinetic Data

During the two ground contacts of the drop jump, external forces acting on the body were recorded using a Kistler force platform (Type 9281B, Kistler Instrument Corp). This was linked via a CIL-16 (analogue to digital convertor) to a BBC microcomputer (Master series), capable of sampling six channels at a frequency of 101 Hz. The sampling frequency was governed by the analogue to digital convertor and the time required to complete the activity. The six channels provided information about force and moment components about the force platform's three principle axes.

Previous work by Harrison (1989) had identified an inadequacy of the analogue divider belonging to the electronic unit of the force platform system. This divider was found to be unreliable in calculating the coordinates of the centre of pressure, as it only became operational above a threshold value of 10% of the sensitivity setting of the charge amplifiers. Thus, the position of the force vector at the beginning and end of ground contact was given as zero. Consequently, software was written to independently calculate the force vector location using the force data and equations given by Kistler (1984).

The force platform was calibrated for point of force application, centre of platform and load accuracy prior to experimentation, using the technique described by Harrison (1989). For the identical force platform arrangement he established a point of load response overall error of 2.0% and a load response mean error of 0.77% (regression slope = 0.997), which were deemed acceptable. For all experimental test sessions the force platform system was switched on at least one hour in advance to enable an ambient temperature to be reached.

Synchronisation of the film and force data was achieved by the use of light sources, a standard camera flash light and a number of high intensity light emitting diodes. The power used to drive these was simultaneously fed to a free channel of the ClL-16. As subjects initiated the drop jump, the camera was activated first, followed closely by the force platform and synchronising pulse. The experimental set up is shown in Plate 10.

Plate 10 - The experimental set-up during the collection of biomechanical data

8.7.4 Acquisition of Trunk Muscle Activity and Measurement of Stature Changes

Surface electrodes were used to monitor the muscle activity of three major lower trunk muscle groups - the erector spinae, external obliques and the rectus abdominis. Prior to experimentation, bipolar arranged silver/silver chloride surface electrodes, centre-to-centre spacing of 6 cm, were applied to prepared skin. These were placed on the right side of the body, 2.5 cm lateral to the umbilicus (rectus abdominis), between the anterior superior iliac spine and caudal border of the rib cage (external obliques), and 3 cm lateral to the mid-line at the level of L4 (erector spinae).

Electrode arrangement was such that the ground electrode was in between the active and reference electrodes, an arrangement providing a balance input for the differential amplifier, which minimises 60 Hz artifacts and maximises the signal's common mode rejection ratio (Winter, 1979). The electrodes were linked directly to a Polygraph (Type 381, NEC San-ei Instruments Ltd, Tokyo, Japan) possessing three bioelectric amplifiers with filters set to record a bandwidth of 10 to 1000 Hz. These in turn were linked to integrators which provided a fully integrated signal, resetting every 5 s. Both the raw EMG and integrated signals were stored using a chart recorder running at a speed of 25 $mm.s^{-1}$

Synchronisation of the EMG trace and film data was achieved using a power supply connected to a DC amplifier of the Polygraph. During ground contact, the pressure sensitive switch mat provided a synchronising pulse that was stored on the chart recorder alongside the EMG traces.

Once the electrodes and leads had been firmly attached to the subject, the body landmarks were applied and spinal fins were positioned as previously discussed. Again, the 20 anthropometric measures were taken. Before commencing the exercise regimen a measure of stature was recorded using the stadiometer located approximately 5 m from the vaulting box. Subjects then had to stand for their 30 minute control period during which measures of stature were taken after 20 and 30 minutes. Immediately following this period, the subjects made their

way to the vaulting box where the EMG leads were connected to the Polygraph. They then embarked upon the exercise regimen previously outlined. Again, randomly throughout the regimen two drop jumps were filmed from each set of five without the subject's prior knowledge (Plate 11). Immediately the exercise period was completed, the EMG leads were disconnected from the Polygraph and subjects walked the short distance to the stadiometer where a measure of stature was taken.

Following the final measurement of stature, subjects were reconnected to the Polygraph and undertook maximum isometric contractions. Contraction of the erector spinae muscle group was achieved using a back and leg dynamometer. Subjects were required to perform a maximum back extension with the legs straight, the hand grip being placed just above the knees. A recording of rectus abdominis and external oblique isometric muscle activity was accomplished using a resisted sit up, with and without trunk rotation. Resistance was provided by applying a force to the shoulders which prevented subjects flexing beyond a trunk angle of approximately 45°. The legs were straight and restrained at the ankles. On all occasions the Polygraph chart recorder was activated 1 s after initiation of the exercise and recorded for a period of 5 s.

8.8 ANALYSIS OF DATA

8.8.1 Processing of Kinematic and Kinetic Data

The developed cine film was digitised using a TDS (HR48) magnetorestrictive digitising tablet (active area of 1.2 m x 0.9 m, resolution 0.025 mm in both X and Y directions) linked to a BBC microcomputer (Master series). A front-on projection of the film data, using a NAC high speed film projector, proved to be the most practical arrangement for the equipment used and the laboratory location of the tablet. From the 10 jumps recorded, per person, per session, the landing phases of two drop jumps were analysed. These were selected on the basis of the symmetrical and balanced nature of the jumps and central positioning of the right foot in relation to the force platform. Each individual frame was digitised for both the initial ground contact phase (land 1) and final ground contact phase (land 2).

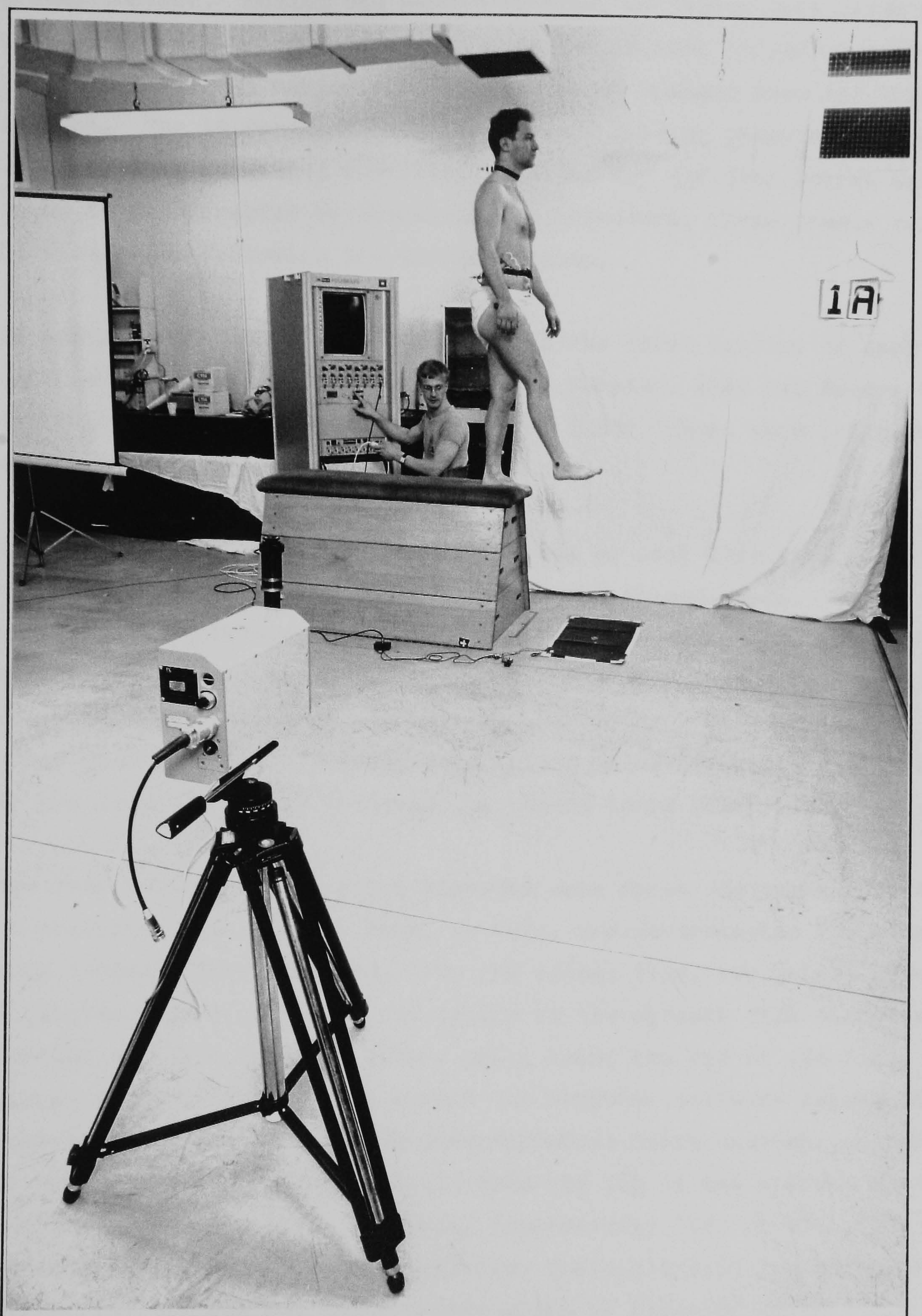


Plate 11 - The experimental set-up during the collection of EMG data

Besides the digitisation of each individual frame of the first landing, 10 additional frames were recorded prior to, and after, ground contact. During the second landing, 10 frames were taken prior to ground contact and 50 frames thereafter, a time period found to be sufficient for all subjects to have attained maximum knee and trunk flexion. The 10 frames either side of the landing phase were required when applying smoothing routines. Initial and end data points were found to be corrupted by smoothing and therefore, these frames could be eliminated following its implementation.

On average, 60 frames were recorded for the first landing of each drop jump, which meant for each subject, per session, some 240 frames were digitised. Consequently, in total some 2,880 frames were individually digitised.

The photographic image was reduced to 33% of real life size by using the centrally located vertical ruler on the filmed image. Digitisation proceeded by firstly recording an origin and vertical reference from the vertical ruler which had been relocated within the field of view. These points then became the axis about which all other points could be referenced. This enabled any erroneous movement of the film within the projector or camera to be controlled.

The four corners of the force platform were first digitised in order to calculate its centre. Then, in turn, the co-ordinates for each body landmark were recorded. For the spinal fins, two points were digitised, the first being the centre of the closest dark shaded area nearest the skin, and the latter point being the end of the fin. A simple mathematical routine within the computer software estimated the location of the corresponding intervertebral joint centres. Distances of 8.5 cm and 4.3 cm (real scale) from the tip of the spinous process, for the L5/S1 and C7/T1 junctures, respectively (Liu et al., 1971) were used in the analysis. Similarly, the digitising program made a correction for the hip joint centre which is reported to be approximately 3 cm above and 1 cm anterior to the most lateral bony prominence of the greater trochanter (Plagenhoef, 1971; Williams and Lissner, 1977).

Once a single frame had been digitised, a visual display of that frame and all previous frames was provided, allowing a visual check for any digitising errors. A completed sequence of coordinate data for each joint centre, for each frame was stored on floppy disc. The memory capacity of the BBC (Master series) microcomputer (128K) prevented any subsequent analysis and thereafter, independent sub-analysis programs were developed. All preceding data analysis procedures were performed on the BBC (Archimedes series) microcomputer.

The stored joint co-ordinate data were fed directly into the program developed to calculate mass and inertia characteristics of individual body segments. Following calculation, selected values were added to the data array and stored on disc. The new data array provided the necessary information for deriving kinematic parameters.

Raw displacement data for joint centres and limb segments centre of gravity were first smoothed using a Butterworth second-order zero lag filter with a cut-off frequency of 16 Hz (Challis, 1988). The filter and cut-off frequency were chosen to comply with those used by Bobbert et al. (1987). In addition, all smoothed data were visually checked to establish whether the cut-off frequencies had dramatically affected each wave form. From the smooth displacement data, limb segment angles, angular velocities, angular accelerations and linear accelerations of the segmental centres of gravity were all calculated in accordance with the method described by Winter (1979). Again, the data array was updated and sent to floppy disc for storage.

The raw force platform data were fed into a data display program which enabled each individual landing trace to be isolated. Combining traces with the relevant kinematic data was achieved using the synchronising pulse and a piecewise polynominal cubic spline interpolation routine (Conte and DeBoor, 1980), applied to the force platform data. This produced a force trace output in line with the slightly lower frequency of the kinematic data (100 Hz). As there was only a slight difference between both sampling frequencies there was little effect on the contour and peaks of the force trace.

The final data array possessed all relevant kinematic and force platform data necessary for calculation of net joint reaction forces, muscle moments, lumbar compression and shear forces, throughout the landing phase of each jump. The final software program possessed an analysis routine which facilitated an assessment of salient features within each curve. It was also capable of assessing the relative contribution dynamic components and erector spinae muscle moment arm (ESMMA) length had on estimating spinal compression force. To eliminate dynamic components and therefore, adopt a 'quasi-static' model, all linear and angular accelerations were excluded from the calculations. The ESMMA length could be selected as either 50 mm or 70 mm, the extremes of the values identified in the literature (refer to Table 3). Schematic representation of the data collection, transfer and processing routines, along with the computer program flowcharts, are illustrated in Figures 31 to 39.

8.8.2 Processing of EMG Data

The corresponding electromyographic traces from the two jumps, during each exercise regimen, which had been subjected to a kinematic analysis were processed. Synchronising pulses from the pressure mat were used to separate EMG data during each landing phase. The second landing trace extended for a time period of 0.4 s from the initial point of ground contact, in accordance with the number of film frame analysed. Thus, muscle activity of the erector spinae, external obliques and rectus abdominus was isolated for each landing phase.

A photograph enlargement (3 x original scale) was taken of each trace to aid the processing procedure. In addition, consideration was given only to the portion of the signal above the baseline (i.e. positive amplitudes), thereby reducing the complexities of analysis induced by the stored nature of the signal.

The EMG processing sought to normalise the EMG signal with respect to maximum amplitudes present during the activity, for the same day of testing and for the individual muscle group. Once each phase of the drop land had been isolated, from ground contact to the point either when the subject broke ground contact or 0.4 s after the second land, an envelope was hand drawn around the enlarged EMG trace. This attempted to connect individual positive amplitudes across the trace.

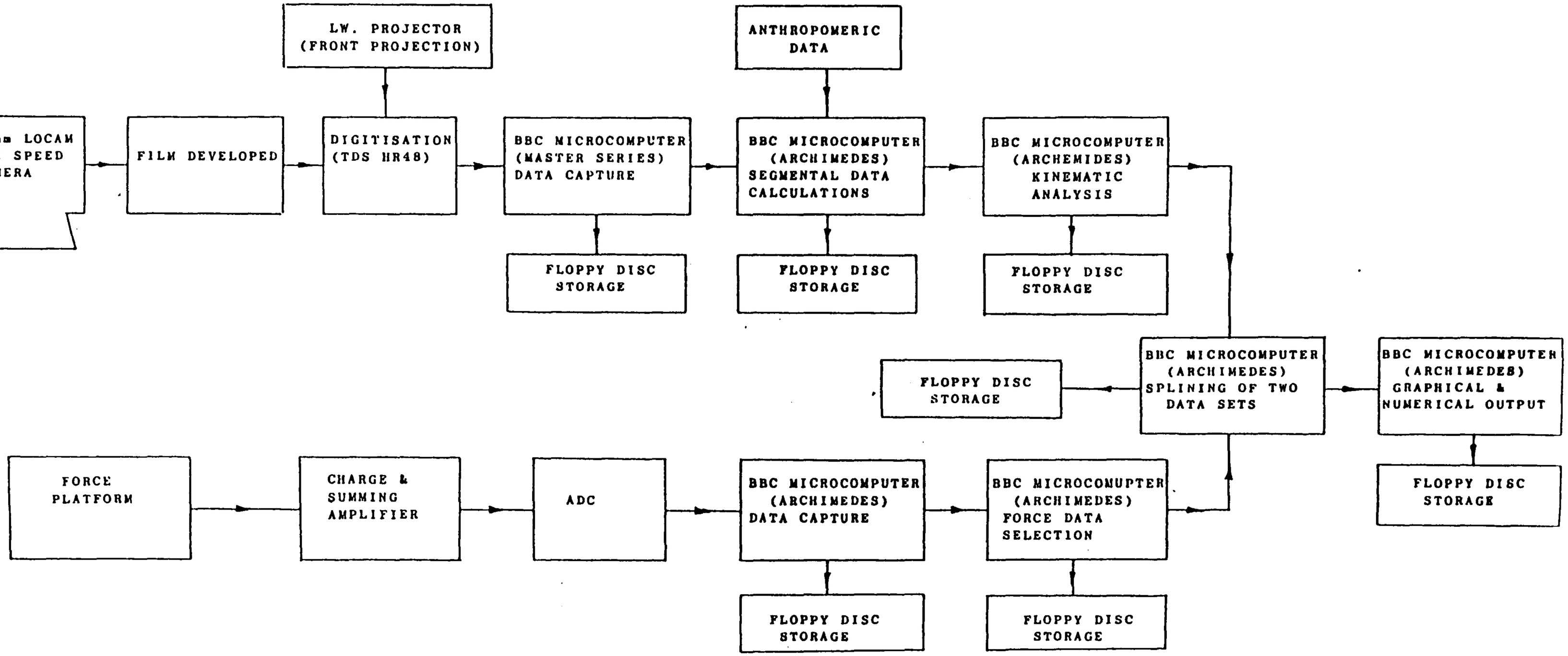


Figure 31. Biomechanical model: Data collection, transfer and processing route.

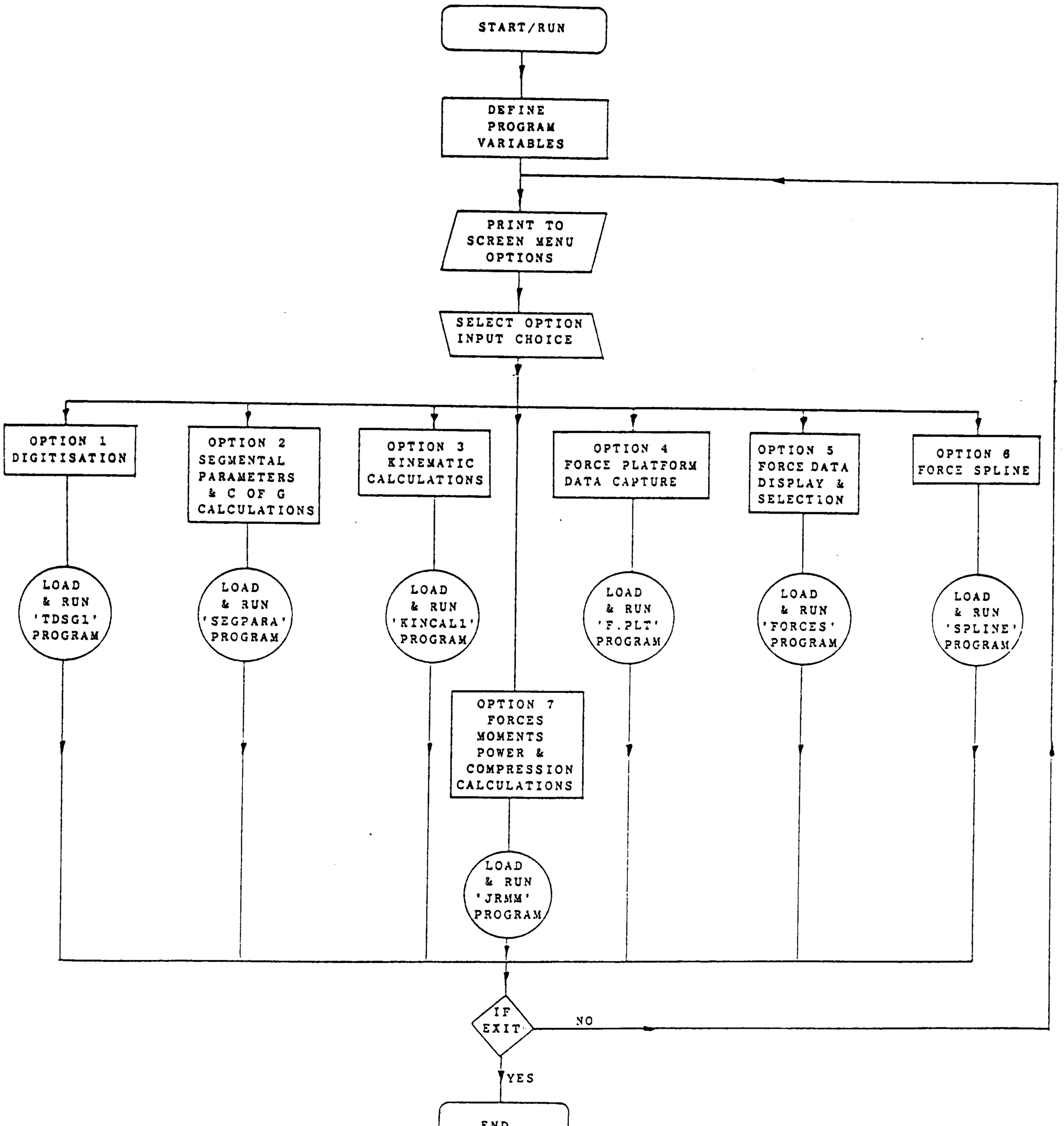


Figure 32. Biomechanical modelling flowcharts: Main menu program.

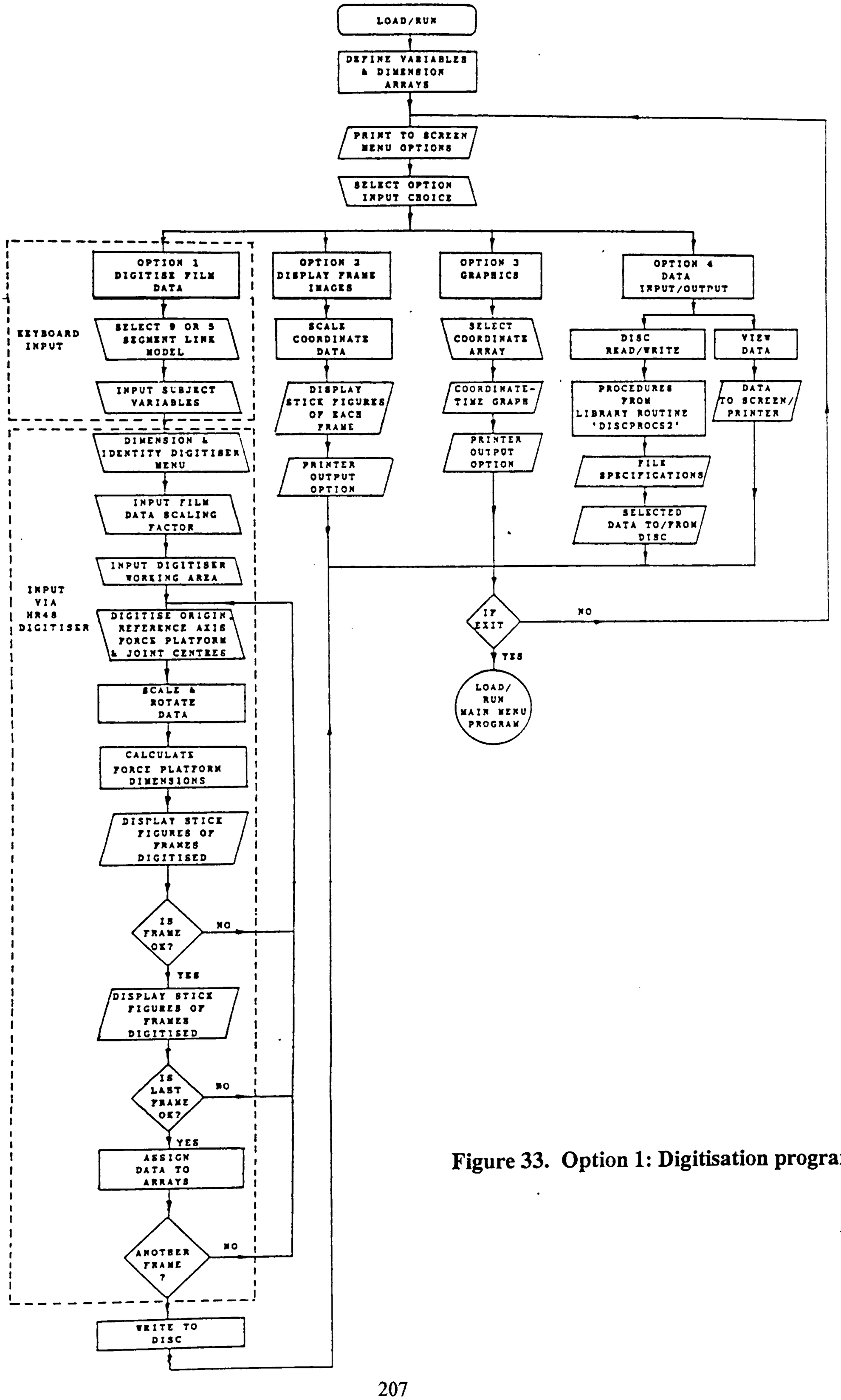


Figure 33. Option 1: Digitisation program.

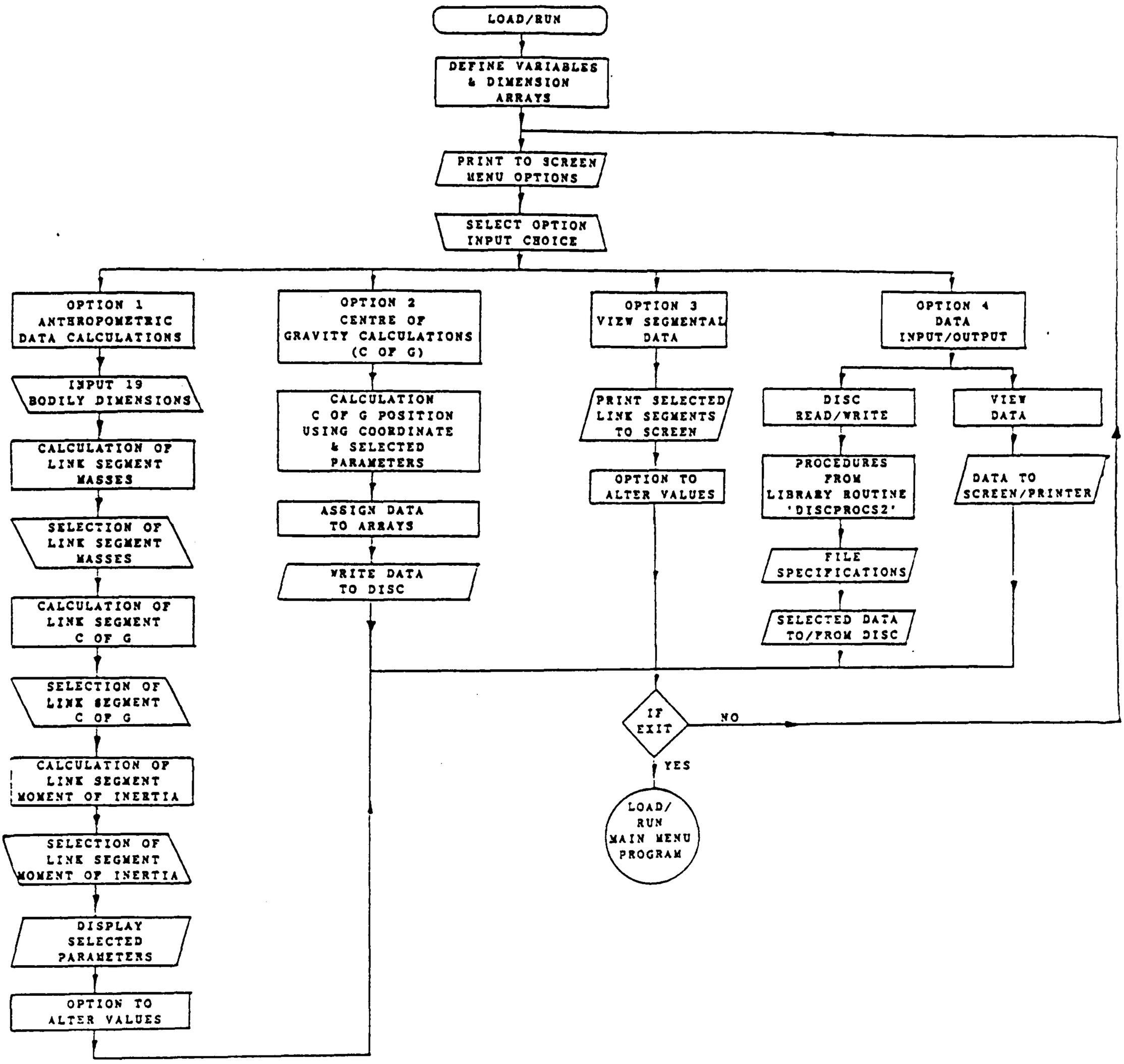


Figure 34. Option 2: Segmental parameters.

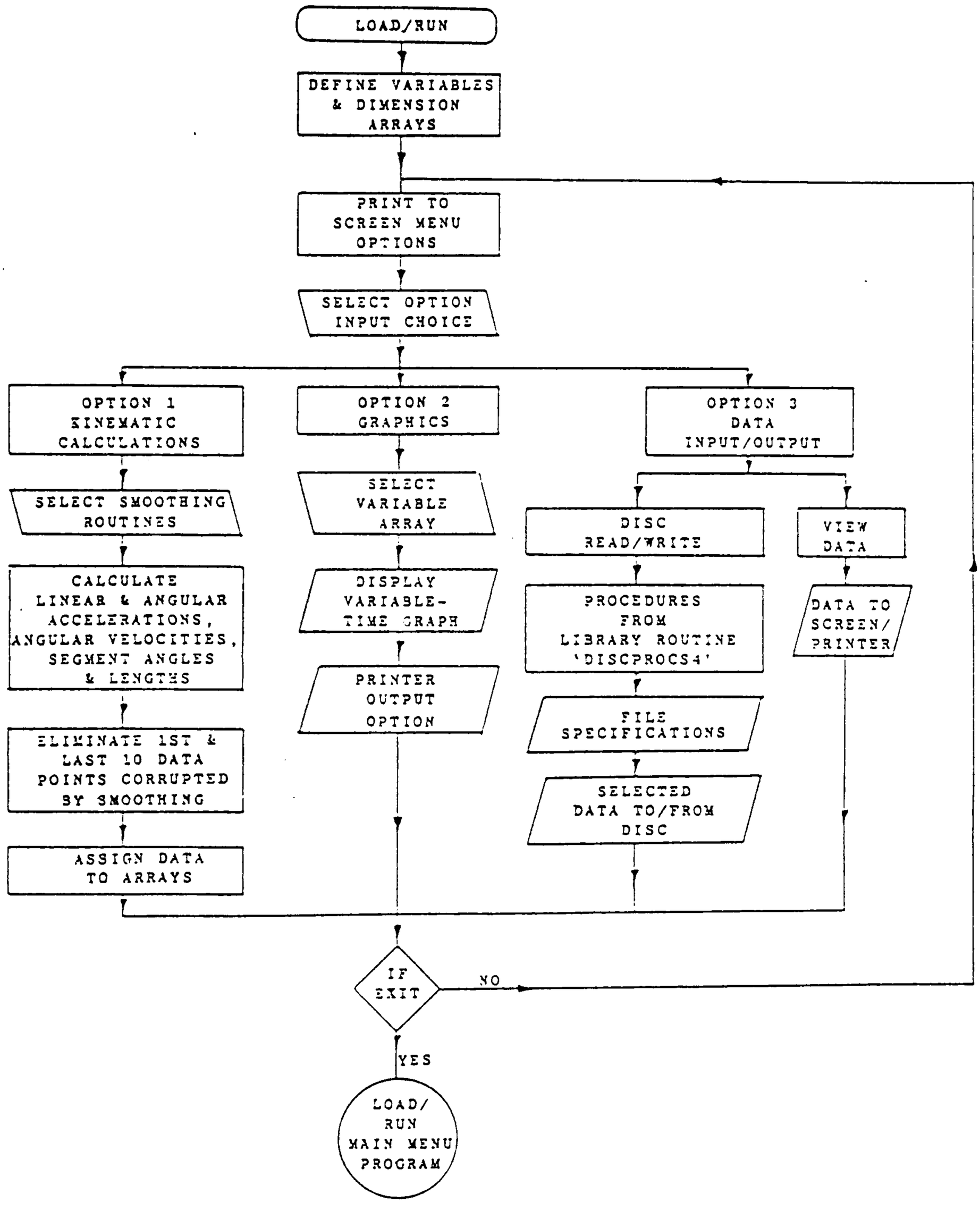


Figure 35. Option 3: Kinematic parameters.

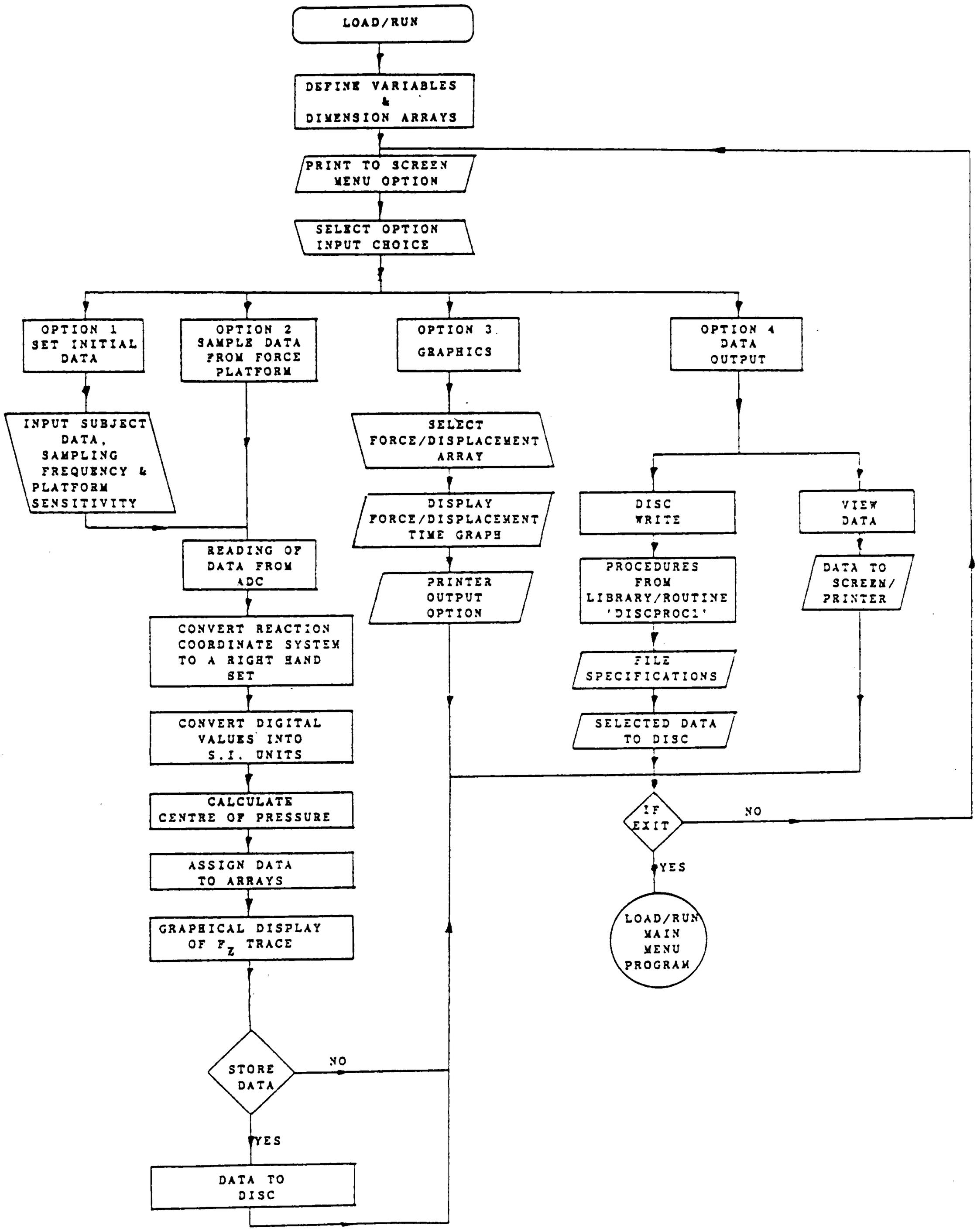


Figure 36. Option 4: Force data collection.

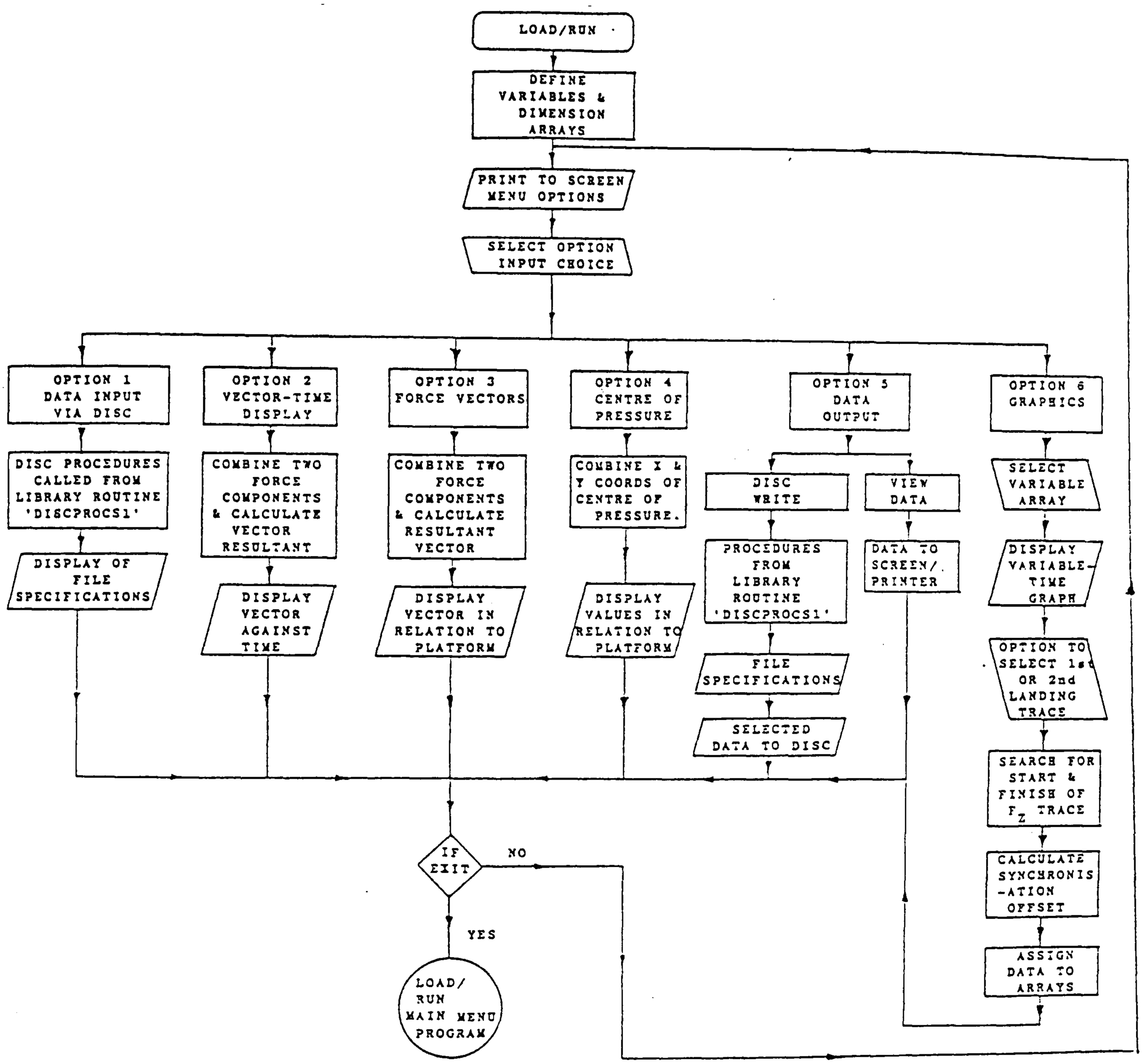


Figure 37. Option 5: Force data display and selection.

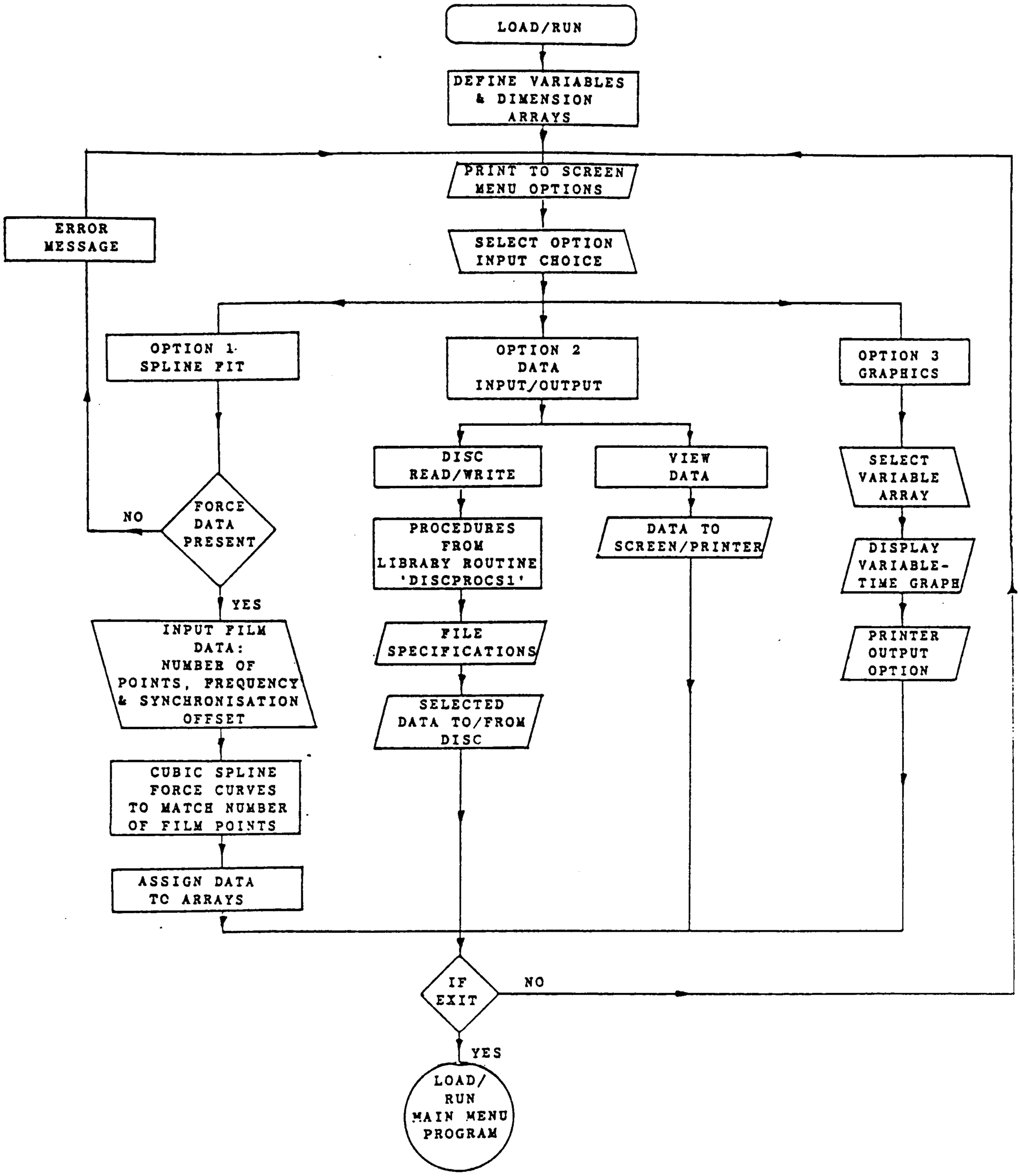


Figure 38. Option 6: Force spline program.

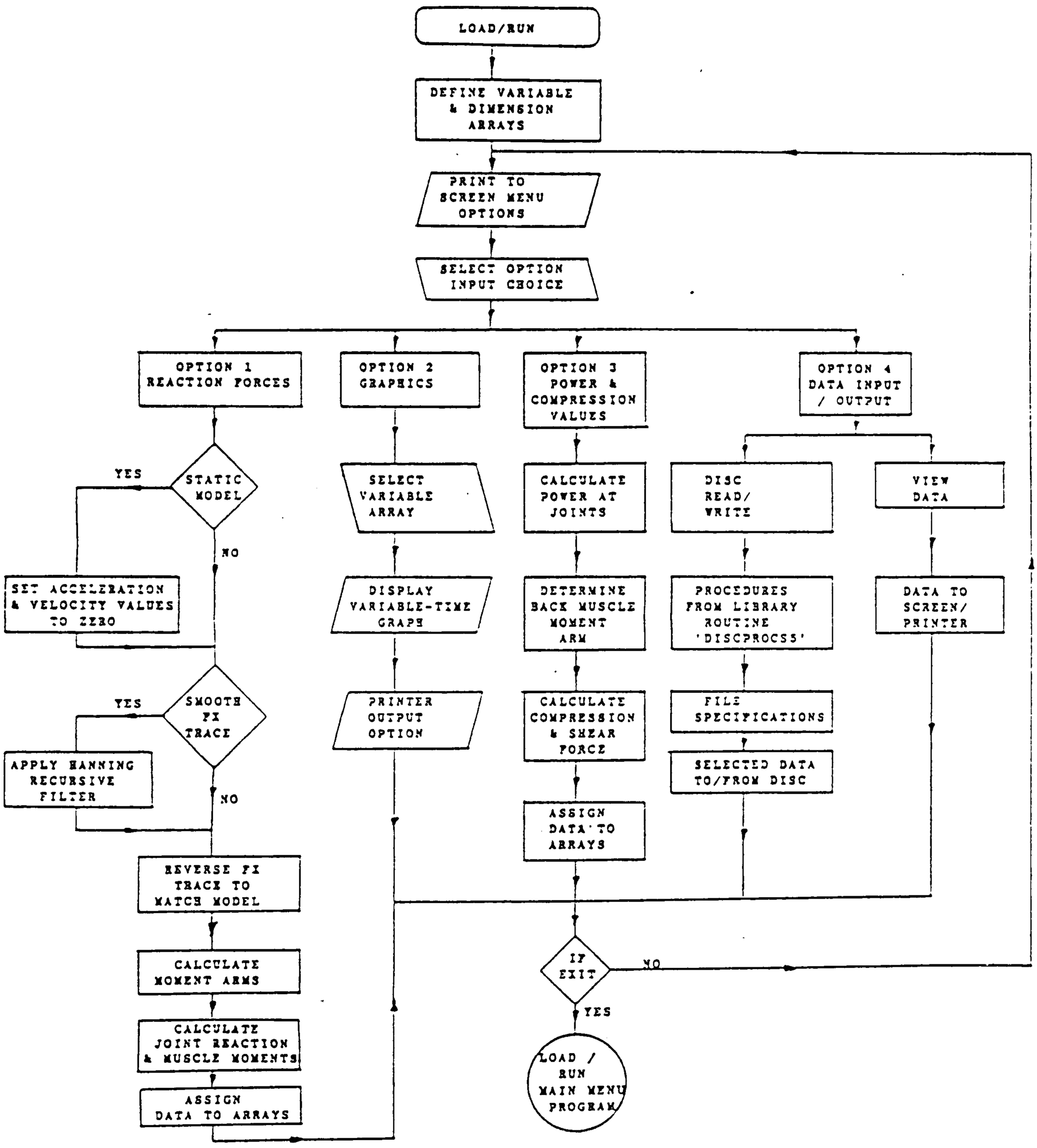


Figure 39. Option 7: Forces and moments calculation program.

The hand drawn procedure was adopted because of the nature of the stored signal. It sought to simulate the mathematical moving average technique, a standard procedure adopted during EMG analysis (Winter et al., 1980). A digitiser and software package (Cabri, 1989) calculated the area encompassed by the envelope and expressed it as a percentage of both the maximum amplitude and the time duration (Figure 40). The whole process of enveloping and digitising individual muscle traces was carried out on two separate days to examine the method's repeatability.

In addition to this normalising procedure, a comparison was also made between the peak positive EMG amplitudes occurring during the individual landing phases and those recorded during the 5 s isometric test, carried out on the same day. The drop jump peak amplitude was then expressed as a percentage of the isometric peak amplitude. Thus, this latter normalising procedure did not take account of any time factor.

Digitisation and analysis of the EMG traces were carried out at Vrije Universiteit Brussel using a digitising tablet and software routines developed by Prof. J. P. Clarys and Dr. J. Cabri.

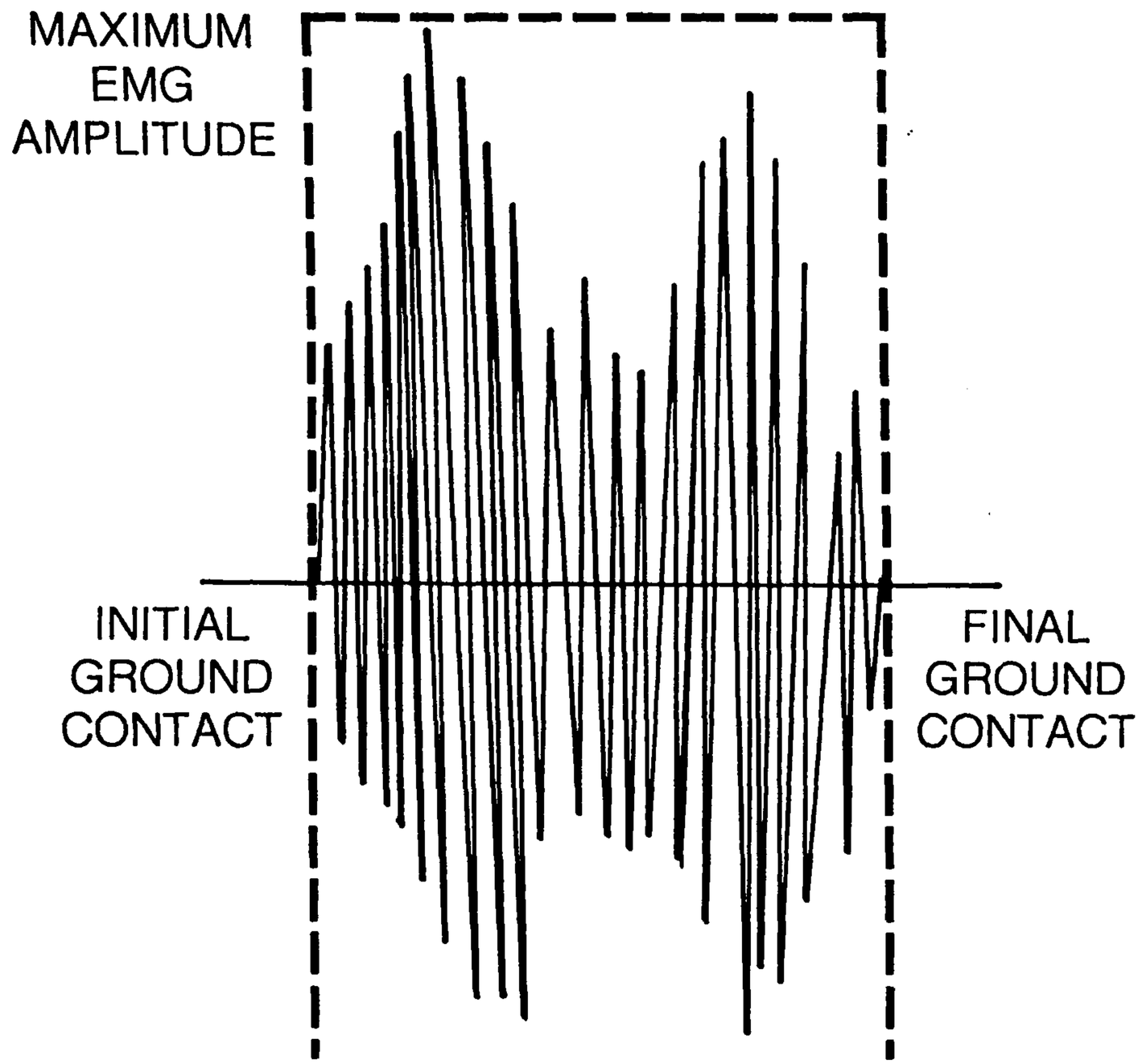
8.9 RESULTS

8.9.1 Kinematic Parameters

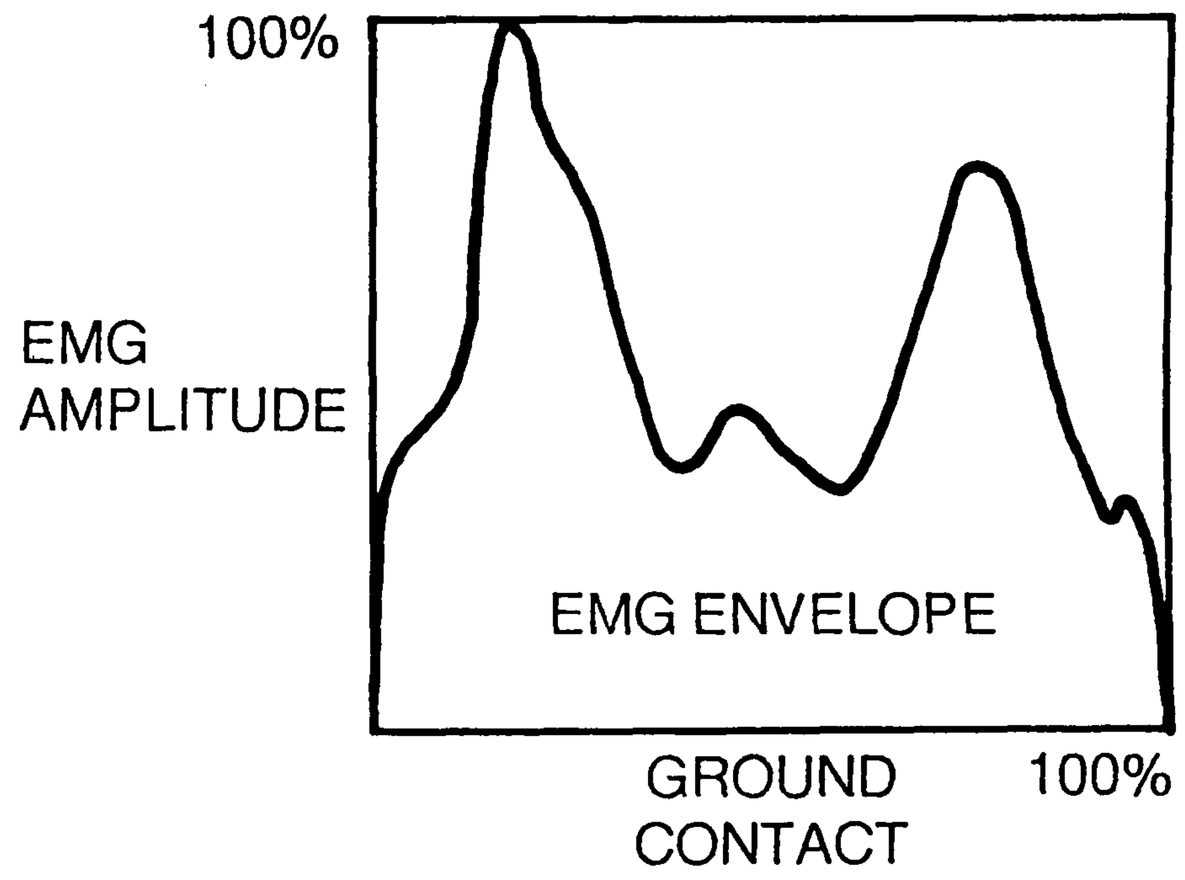
An analysis of drop jump performance and an evaluation of repeatability were undertaken by examining a number of salient features inherent to selected kinematic parameters. T-tests were used to compare the similarities and differences between the two test sessions (test 1 and 2) and between the two drop jump heights (0.75 and 0.30 m). Parameters selected for analysis included:-

i) initial, minimum and final included angles for the ankle, knee and hip joints (Tables 12 to 17);

ii) initial, minimum and final angle formed by the upper trunk with the horizontal (Tables 18 and 19);



Raw EMG trace



Processed EMG trace

Figure 40. Normalisation of the raw EMG signal.

iii) maximum angular velocities during flexion and extension at the ankle, knee and hip joints (Tables 20 to 25);

iv) time taken from the point of foot contact for the ankle, knee, hip and trunk angular velocities to reach zero (Tables 20 to 27);

v) ground contact time during the initial land and the time in flight between the first and second land (Table 28).

These parameters represented clearly identifiable phases within the drop jump. The included angles used during this analysis are depicted in Figure 25. Initial angle refer to the joint angles at the time subjects first made ground contact. Final angle occurs when subjects either break ground contact during the first land or is the last recorded angle during the second land. A minimum angle is the smallest angle attained at each identified hinge or joint; thus a reflection of maximum flexion. Angular velocities refer to rates of change in the associated angle. Mean parameters were derived for each subject based on the two analysed jumps belonging to each test session. Overall means for both tests were determined using subject means. In Tables 12-25 and Table 28 these values are displayed within brackets. T-tests were used to compare intra-test variability and differences between drop heights. A probability value of less than 0.05 was deemed significant.

8.9.2 Repeatability and Variations Resulting from Drop Height.

8.9.2.1 Ankle Joint Angle None of the selected parameters for the mean angle formed at the ankle joint (Tables 12 and 13) showed any significant intra-test variability. Differences between drop jump height were also found to be non-significant although, initial and final ankle angles following the drop from 0.75 m were 5° greater on both occasions. A larger initial ankle angle would not seem unreasonable given the longer time period over which the joint could plantarflex in anticipation of ground contact. This would also serve to lengthen the period of time over which the body could absorb the increased momentum it possessed.

Table 12. Ankle angles during the first land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	108±6 (108±5)	76±4 (75±4)	125±17 (124±10)
	test 2	108±5	75±4	122±5
0.30	test 1	103±9 (103±7)	76±3 (76±4)	119±11 (119±7)
	test 2	103±6	75±4	119±4

Table 13. Ankle angles during the second land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	105±9 (104±9)	80±6 (76±7)	88±11 (87±11)
	test 2	102±10	74±9	86±12
0.30	test 1	104±10 (104±9)	76±6 (75±7)	88±11 (88±11)
	test 2	104±8	74±7	89±10

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 14. Knee angles during the first land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	148±5 (148±5)	96±4 (94±5)	165±6 (167±3)
	test 2	147±6	93±8 (*)	168±2
0.30	test 1	147±5 (147±5)	100±7 (100±7)	163±3 (165±1)
	test 2	147±5	100±10	166±2

* Significant Difference ($p<0.05$)

Table 15. Knee angles during the second land.

Drop jump height(m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	155±7 (155±7)	105±19 (107±20)	138±38(136±38)
	test 2	154±9	107±22	135±39
0.30	test 1	156±8 (154±9)	105±21 (107±19)	141±46(145±39)
	test 2	153±11	108±18	149±31

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 16. Hip angle during the first land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD) Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	151±6 (152±12)	106±19 (108±24)	170±6 (168±6)
	test 2	152±17	110±31	167±6
0.30	test 1	146±8 (145±9)	112±18 (114±17)	167±11(166±6)
	test 2	145±11	115±18	166±10

Table 17. Hip angles during the second land.

Drop jump height(m)	Analysis session	Mean angle (°) during the land (±SD) Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	161±13 (161±9)	125±35 (125±30)	139±37(139±31)
	test 2	161±7	125±28	139±28
0.30	test 1	158±9 (161±8)	124±31 (127±25)	145±40(149±32)
	test 2	163±7	131±20	149±25

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 18. Trunk angles during the first land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	72±5 (73±5)	49±14 (50±13)	78±7 (79±6)
	test 2	73±5	51±13 (*)	79±7 (*)
0.30	test 1	68±4 (70±7)	55±13 (57±11)	80±8)(83±6)
	test 2	71±8	60±9	85±5)*

* Significant Difference ($p<0.05$)

Table 19. Trunk angles during the second land.

Drop jump height (m)	Analysis session	Mean angle (°) during the land (±SD)		
		Initial (mean)	Minimum (mean)	Final (mean)
0.75	test 1	80±8 (82±9)	58±17 (60±15)	65±20 (66±19)
	test 2	83±11	62±14	67±19
0.30	test 1	79±11)(83±9)	59±20 (63±16)	68±23 (71±20)
	test 2	86±9)*	66±13	75±17

* Significant Difference ($p<0.05$)

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 20. Ankle angular velocities and time taken to reach zero during the first land.

Drop jump height (m)	Analysis session	Maximum angular velocity (rad/s) for flexion/extension at a period during the land and time(s) taken to reach zero (±SD) Flexion (mean)		Time (mean)		Extension (mean)	
0.75	test 1	-9.6±1.8	(-9.6±1.3)	0.21±0.05	(0.20±0.5)	13.4±6.4	(12.7±3.6)
	test 2	-9.6±1.3	(**)	0.20±0.07	(**)	12.0±1.5	
0.30	test 1	-6.5±2.6	(-6.4±1.7)	0.24±0.05	(0.23±0.05)	12.0±1.5	(11.7±2.4)
	test 2	-6.2±1.2		0.21±0.06		11.4±1.0	

** Significant Difference ($p<0.01$)

Table 21. Ankle angular velocities and time taken to reach zero during the second land.

Drop jump height (m)	Analysis session	Maximum angular velocity (rad/s) for flexion/extension at a period during the land and time(s) taken to reach zero (±SD) Flexion (mean)		Time (mean)		Extension (mean)	
0.75	test 1	-8.5±2.0	(-8.4±1.6)	0.12±0.03	(0.12±0.03)	1.3±0.6	(1.2±0.5)
	test 2	-8.3±1.4		0.12±0.03		1.2±0.5	(*)
0.30	test 1	-8.1±2.5	(-8.5±2.4)	0.12±0.03	(0.12±0.03)	1.4±0.7	(1.5±0.5)
	test 2	-8.8±2.5		0.12±0.02		1.6±0.3	

* Significant Difference ($p<0.05$)

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between

Table 22. Knee angular velocities and time taken to reach zero during the first land.

Drop jump height (m)	Analysis session	Maximum angular velocity (rad/s) for flexion/extension at a period during the land and time(s) taken to reach zero (±SD) Flexion (mean)	Time (mean)	Extension (mean)
0.75	test 1	-9.7±1.6 (-9.7±1.3)	0.22±0.03 (0.22±0.05)	11.9±1.2 (12.3±0.9)
	test 2	-9.6±1.1	0.21±0.06	12.6±0.9
0.30	test 1	-8.2±1.2 (-8.0±1.2)	0.20±0.06 (0.20±0.05)	12.2±1.3 (12.2±1.1)
	test 2	-7.9±1.4	0.20±0.05	12.2±0.9

Table 23. Knee angular velocities and time to reach zero during the second land.

Drop jump height (m)	Analysis	Maximum angular velocity (rad/s) for flexion/extension at a period during the land and time(s) taken to reach zero (±SD) Flexion (mean)	Time (mean)	Extension (mean)
0.75	test 1	-8.9±1.6 (-9.0±1.1)	0.19±0.07 (0.18±0.07)	4.0±2.2 (3.9±1.7)
	test 2	-9.0±1.1	0.17±0.07 (*)	3.9±1.4
0.30	test 1	-9.4±1.7 (-9.0±1.5)	0.18±0.09 (0.17±0.06)	4.5±2.85 (4.7±2.5)
	test 2	-8.7±2.2	0.16±0.04	4.8±2.1

* Significant Difference ($p<0.05$)

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 24. Hip angular velocities during the first land.

Drop jump height (m)	Analysis session	Maximum angular velocity (rad/s) for flexion /extension at a period during the land (±SD)			
		Flexion (mean)		Extension (mean)	
0.75	test 1	-7.8 ±1.9	(7.6 ±1.7)	8.4 ±1.3	(7.9 ±1.4)
	test 2	-7.3 ±1.6	(**)	7.3 ±1.6	
0.30	test 1	-6.5 ±1.7	(6.0 ±1.3)	10.3±6.0	(9.4 ±3.8)
	test 2	-5.5 ±1.6		8.6 ±3.6	

** Significant Differences ($p<0.01$)

Table 25. Hip angular velocities during the second land.

Drop jump height (m)	Analysis session	Maximum angular velocity (rad\s) for flexion /extension at a period during the land (±SD)			
		Flexion (mean)		Extension (mean)	
0.75	test 1	-6.7 ±1.4	(-6.3 ±1.6)	4.6 ±3.5	(4.0 ±2.0)
	test 2	-5.9 ±2.5		3.4 ±1.1	
0.30	test 1	-6.6 ±1.3	(-7.1 ±1.7)	3.6 ±1.4	(4.1 ±1.7)
	test 2	-7.6 ±2.6		4.6 ±2.7	

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Table 26. Time taken to reach zero trunk angular velocity during the first land.

Drop jump height (m)	Analysis session	Time(s) taken to reach zero trunk velocity	
		Time	Mean
0.75	test 1	0.20±0.05	0.20±0.05)
	test 2	0.20±0.05	)**
0.30	test 1	0.16±0.06	0.16±0.05)
	test 2	0.15±0.05	

** Significant Difference ($p<0.01$)

Table 27. Time taken to reach zero trunk angular velocity during the second land.

Drop jump height (m)	Analysis session	Time(s) taken to reach zero trunk velocity	
		Time(s)	Mean
0.75	test 1	0.23±0.07	0.22±0.06
	test 2	0.21±0.06	
0.30	test 1	0.20±0.07	0.19±0.08
	test 2	0.19±0.08	

Table 28. Ground contact time during the first land and the time in flight between the first and second land.

Drop jump height(m)	Analysis session	Mean contact and flight time (s) (±SD) contact (mean)		flight (mean)	
0.75	TEST 1	0.459±0.10	(0.463±0.09)	0.557±0.05	(0.543±0.05)
	TEST 2	0.468±0.08	(*)	0.530±0.05	
0.30	TEST 1	0.401±0.10	(0.399±0.09)	0.541±0.04	(0.549±0.05)
	TEST 2	0.398±0.08		0.557±0.04	

* Significant Difference ($p<0.05$)

NB: Values contained in brackets refer to the overall mean for test 1 and 2.

An asterisk within a bracket refers to differences between means for each drop height.

An asterisk not enclosed refers to differences between tests at the same drop height.

Difficulties were often encountered when locating and digitising the pternion. This, coupled with the small sample size, may have inhibited any observable variations in ankle angle.

8.9.2.2 <u>Knee Joint Angle</u> The mean angle formed by the long axis of both the thigh and shank (knee angle, Tables 14 and 15) did not significantly vary when drop jumps were performed from the same height, on separate occasions. With the exception of the minimum knee angle, during the initial drop land, drop height was also found to have a non-significant influence on the selected parameters ($p<0.05$). Thus, the higher drop height forced subjects to increase knee flexion in order to cushion the landing. This is not surprising considering the greater velocity attained by the body's mass when falling from the higher drop height.

8.9.2.3 <u>Hip Joint Angle</u> Tables 16 and 17 provide data on the mean angles formed by the thigh and pelvis (hip angle). No significant differences were reported for any of the selected parameters when drop jumps were compared for the same height and between heights. However, it should be noted that all parameters did show large variations in reported angle, evident from the high SD values. This could imply that a marked difference occurred with respect to drop landing technique or, as is more probable, variations were a result of the inherent errors associated with the technique used to track the intervertebral disc centres. For example, in some instances it was necessary to estimate spinal fin location when the fins were obscured by the arms. In addition, the use of surface mounted probes inevitably means they are highly susceptible to skin vibration.

8.9.2.4 <u>Trunk Angle to the Horizontal</u> Final trunk angle (Tables 18 and 19) for the initial 0.30 m drop land and the initial trunk angle during the second 0.30 m drop land showed significant inter-test variability ($p<0.05$). This again may reflect deficiencies in either the disc centre detection procedure or the subject's ability to replicate performance. Drop jump height was only found to have an influence on mean minimum and final trunk angles during the first drop land. As with knee flexion, subjects tended to increase the amount of trunk flexion (7°) following the 0.75 m drop land ($p<0.05$). Subjects

were also more upright (4°) when initiating the vertical jump associated with the 0.30 m drop jump ($p<0.01$).

8.9.2.5 <u>Angular Velocities at the Ankle Joint</u> Mean angular velocities and the time taken to reach maximum flexion at the ankle joint (Tables 20 and 21) showed significant variations when comparing drop heights. Inter-test variability was not found for any of the selected parameters. During the initial drop land, the maximum rate of ankle dorsiflexion was 3.2 rad/s greater following the 0.75 m drop land ($p<0.01$). For the same land and drop height, the time taken to reach maximum ankle dorsiflexion occurred 0.03 s earlier ($p<0.01$). No variations were evident for the maximum rates of ankle plantarflexion prior to the vertical jump ($p>0.05$). The maximum rate at which subjects extended the ankle joint following the second land did show a significant variation ($p<0.05$), values for the 0.30 m drop jump being 0.3 rad/s greater.

8.9.2.6 <u>Angular Velocities at the Knee Joint</u> Mean angular velocities recorded for the knee joint revealed no significant differences either between test sessions or drop jump heights (Tables 22 and 23). Only when drop heights were compared did the mean time to reach maximum knee flexion during the second land show any significant variation ($p<0.05$). Subjects on this occassion reached maximum knee flexion 0.01 s earlier following the 0.30 m drop jump.

8.9.2.7 <u>Angular Velocities at the Hip Joint</u> The only significant variation ($p<0.01$) in mean maximum hip velocities occurred for flexion during the initial land (Table 24 and 25). The 0.75 m drop jump induced a mean hip angular velocity that was 1.6 rad/s larger than that found during the 0.30 m drop jump. No inter-test variability was found for maximum rates of flexion or extension during either of the two lands.

8.9.2.8 <u>Time Taken for Trunk Angular Velocity to Reach Zero</u> No inter-test variability was found for the time taken to reach maximum trunk flexion (Tables 26 and 27). When mean values were compared for drop heights, maximum trunk flexion occurred 0.04 s later following the 0.75 m drop jump ($p<0.01$).

8.9.2.9 Ground Contact and Flight Times As indicated in Table 28, there was no significant variation in mean times for either ground contact or flight time when comparisons were made of test sessions performed at the same drop height. When comparing mean values for drop heights a significant difference existed for ground contact times ($p<0.05$). The 0.75 m drop jumps caused subjects to be in contact with the ground for 0.064 s longer than when executing the 0.30 m drop jump.

Based on the assumption that flight time provides an indication of jumping performance, the lack of a significant variation between drop heights suggests that the 0.75 m drop height did not enable subjects to increase their vertical jump.

8.9.3 Ground Reaction Force Data

Ground reaction force traces for individual subjects are shown in Figures 41 through to 52, a zero point in time referring to the period at which the feet first made contact with the ground. For ease of comparison, only one force trace per drop height is shown, although the two traces recorded at each drop height were essentially similar. During the initial drop jump, two force peaks were identified:- the maximum force during knee flexion (drop peak) and knee extension (push-off peak). Mean force values for each subject and overall means for each drop height and force component (vertical (FZ) and anterior-posterior (FX)) are displayed in Table 29. During the second land consideration was only given to the former peak as marked differences were observed in the post-impact force curves (Table 30). This appeared to emanate from variations in the technique adopted during the second landing; some subjects preferred a more controlled and forceful knee flexion as opposed to some remaining more upright and returning quickly to a standing posture. T-tests and a significance of less than 0.05 were again used to compare peak forces.

Force curves during the first land were characterised by one or two sharp peaks in both the vertical and anterior-posterior ground reaction force components. On average these occurred within the first 0.102 s and coincided with forefoot and heel impact. Mean vertical peak forces during this phase of the drop land achieved values of 1301 ± 291 N (1.78 ± 0.44 body weight units (BW)) and 2336 ± 399 N (3.19 ±

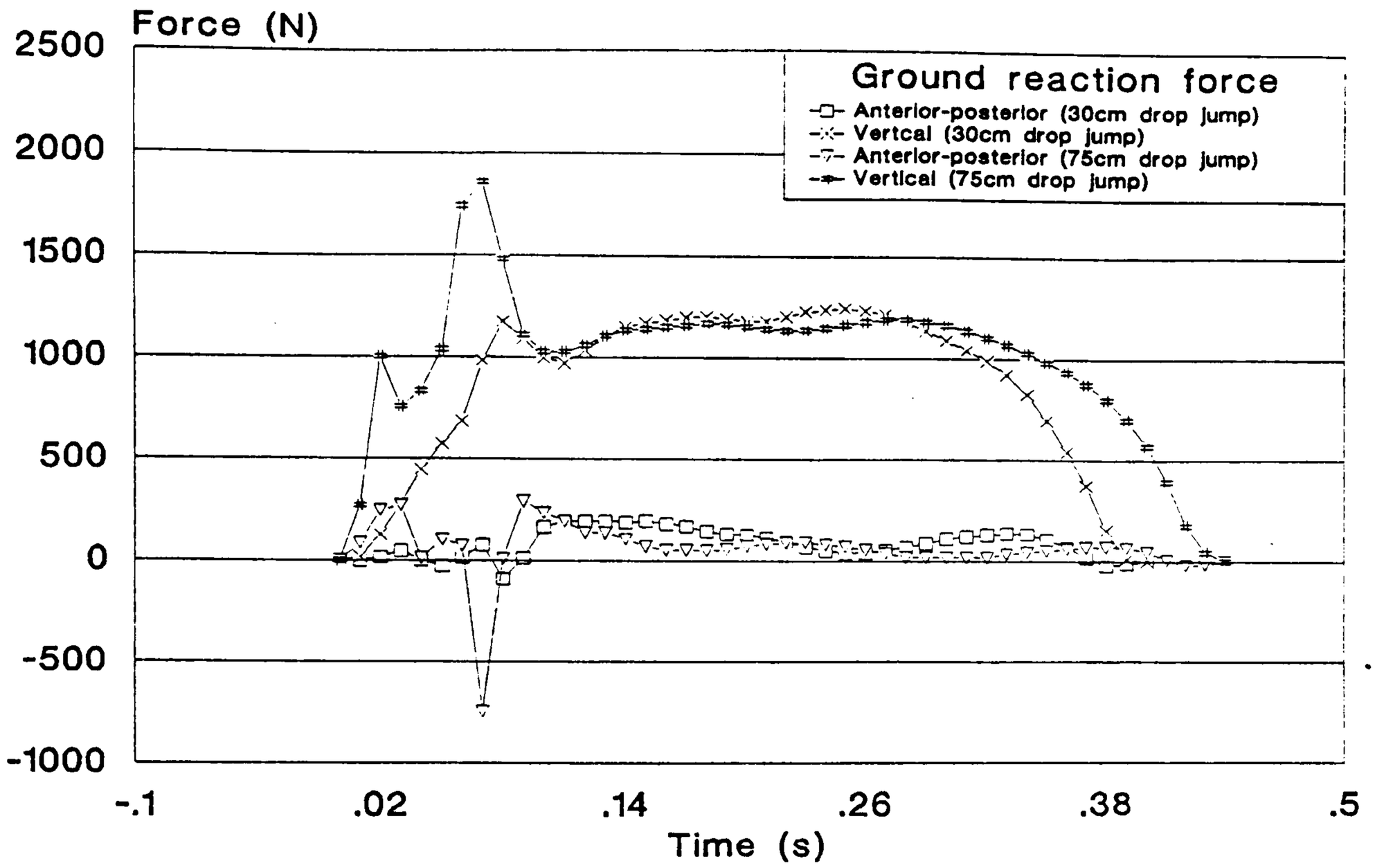


Figure 41. Subject 1: Ground reaction force curves during the first drop land.

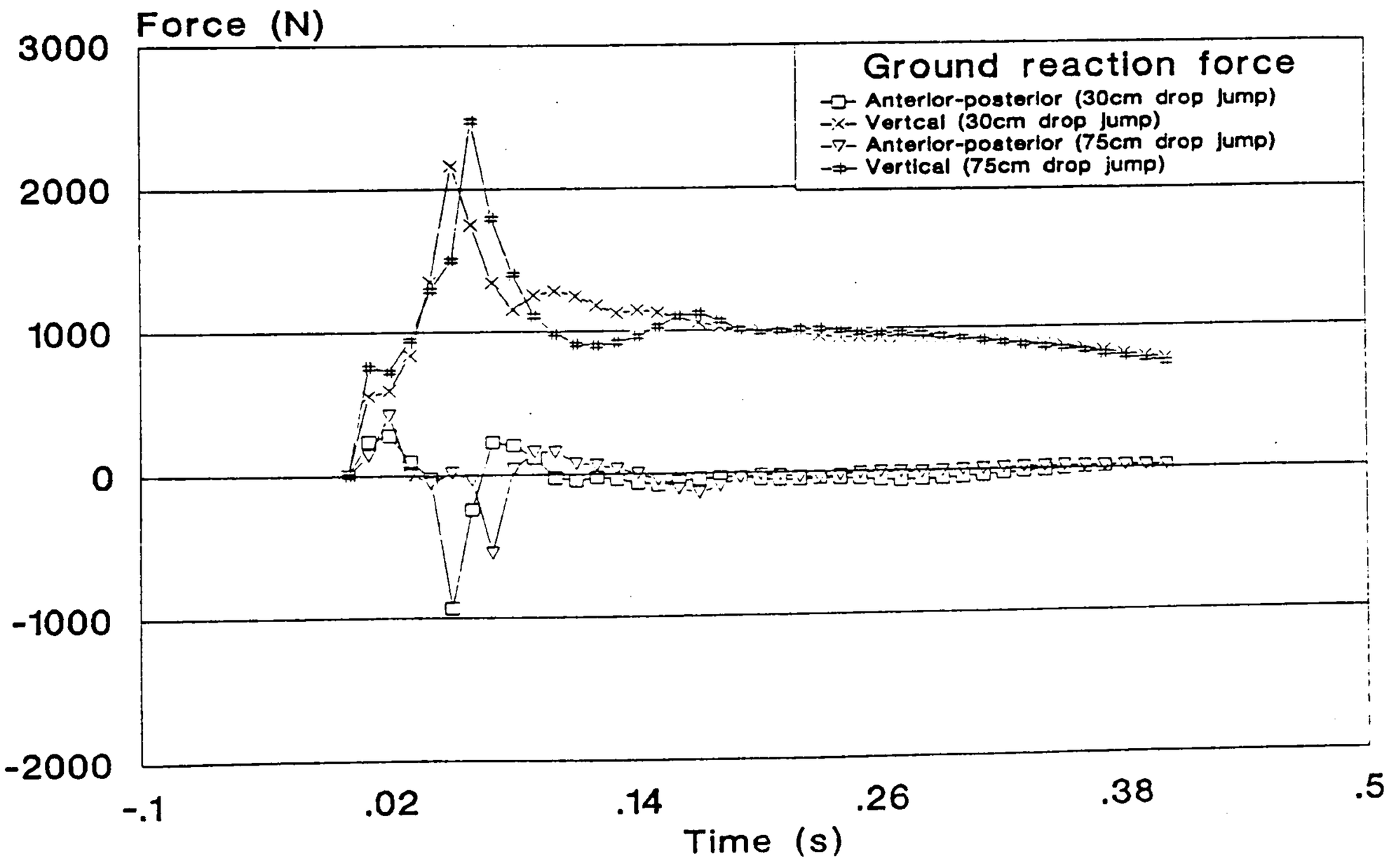


Figure 42. Subject 1: Ground reaction force curves during the second drop land.

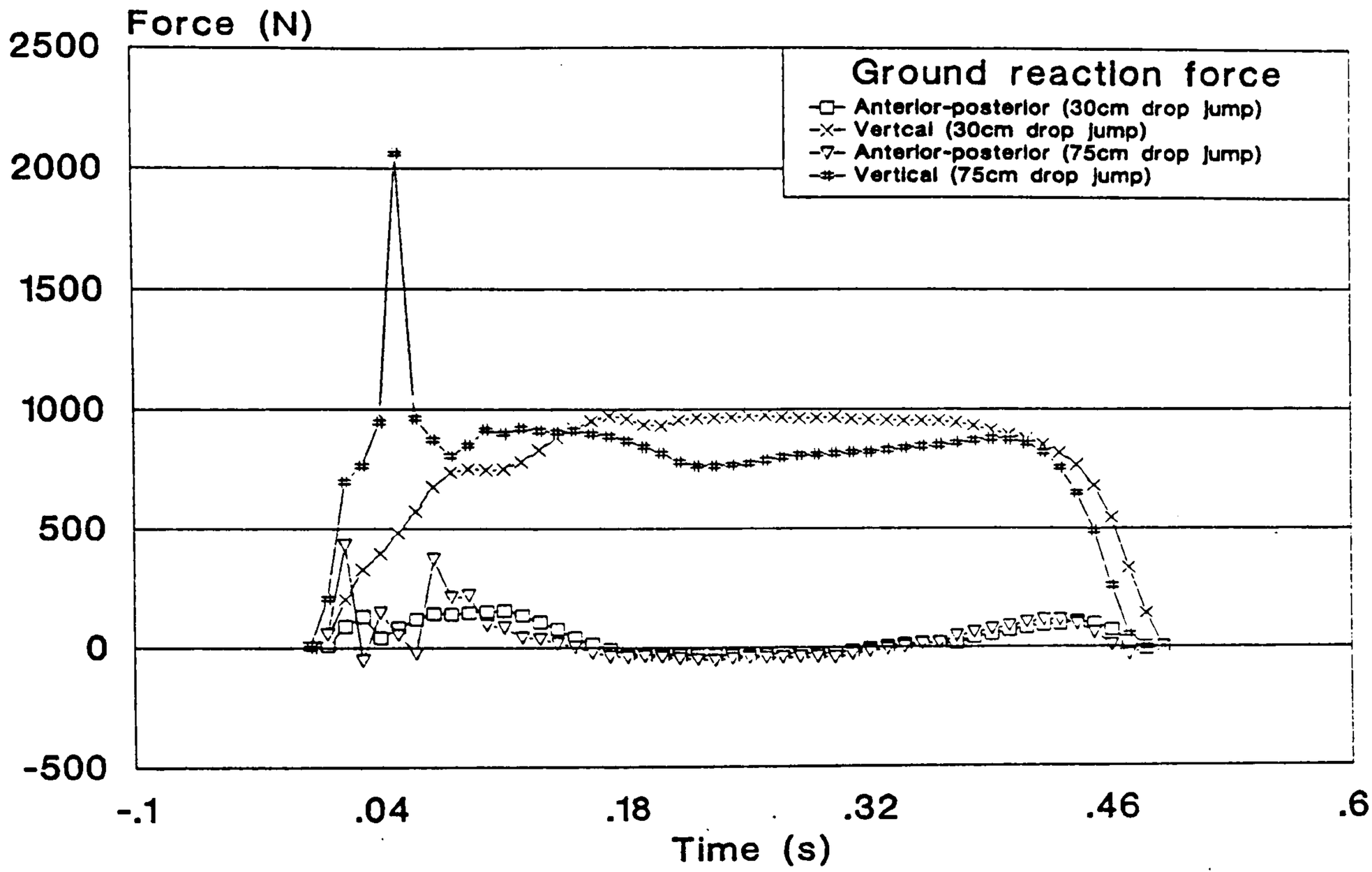


Figure 43. Subject 2: Ground reaction force curves during the first drop land.

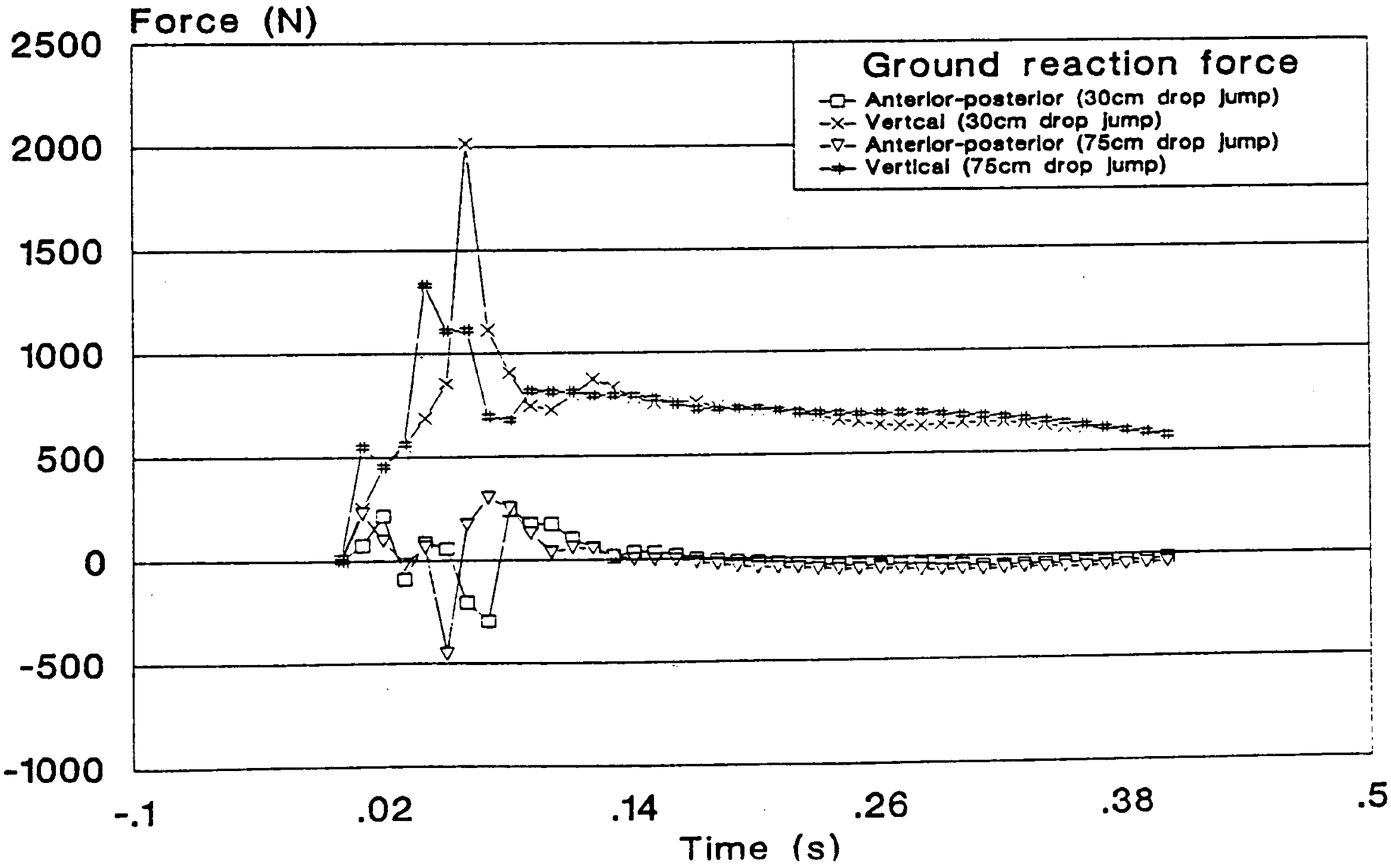


Figure 44. Subject 2: Ground reaction force curves during the second drop land.

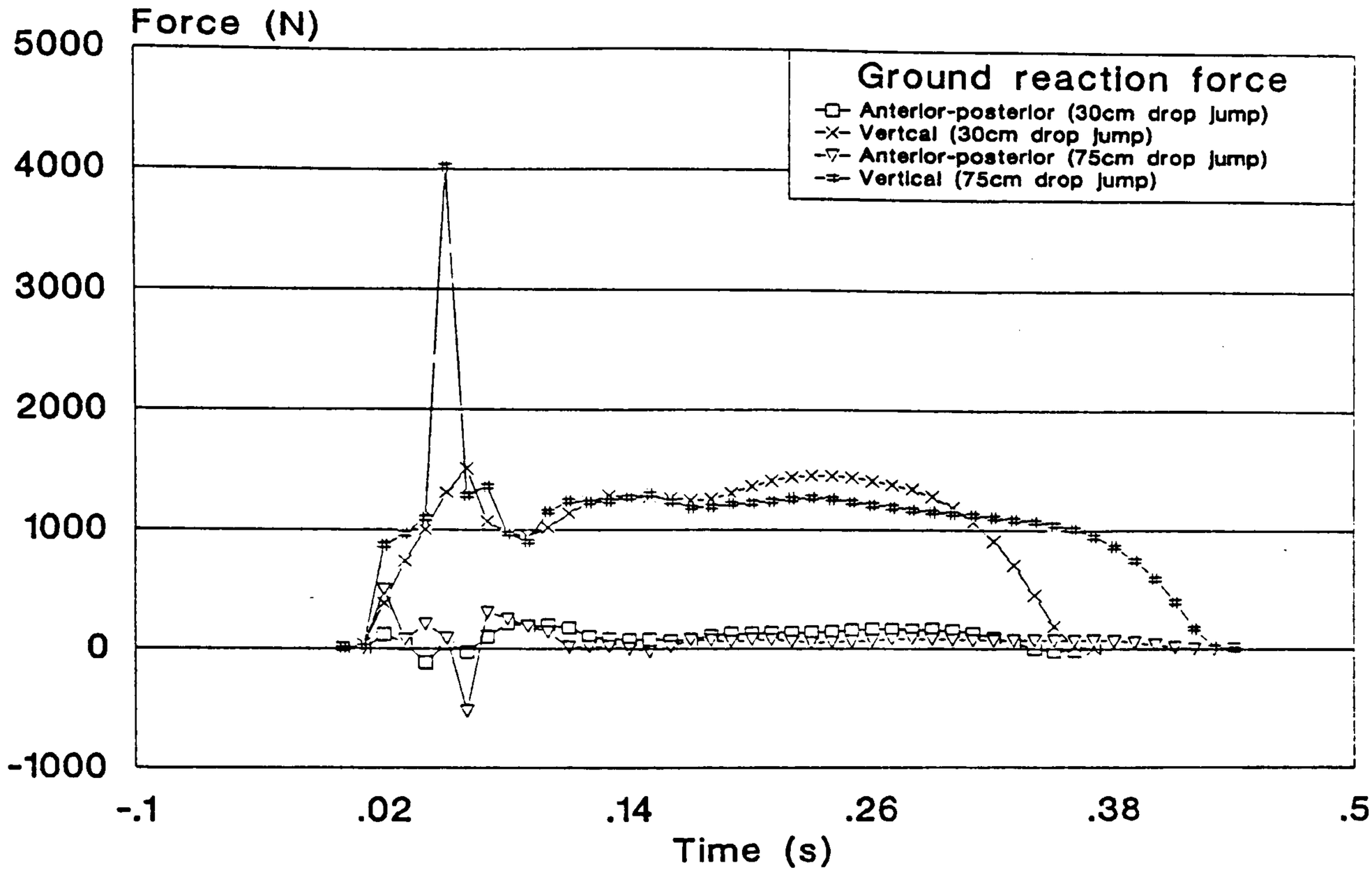


Figure 45. Subject 3: Ground reaction force curves during the first drop land.

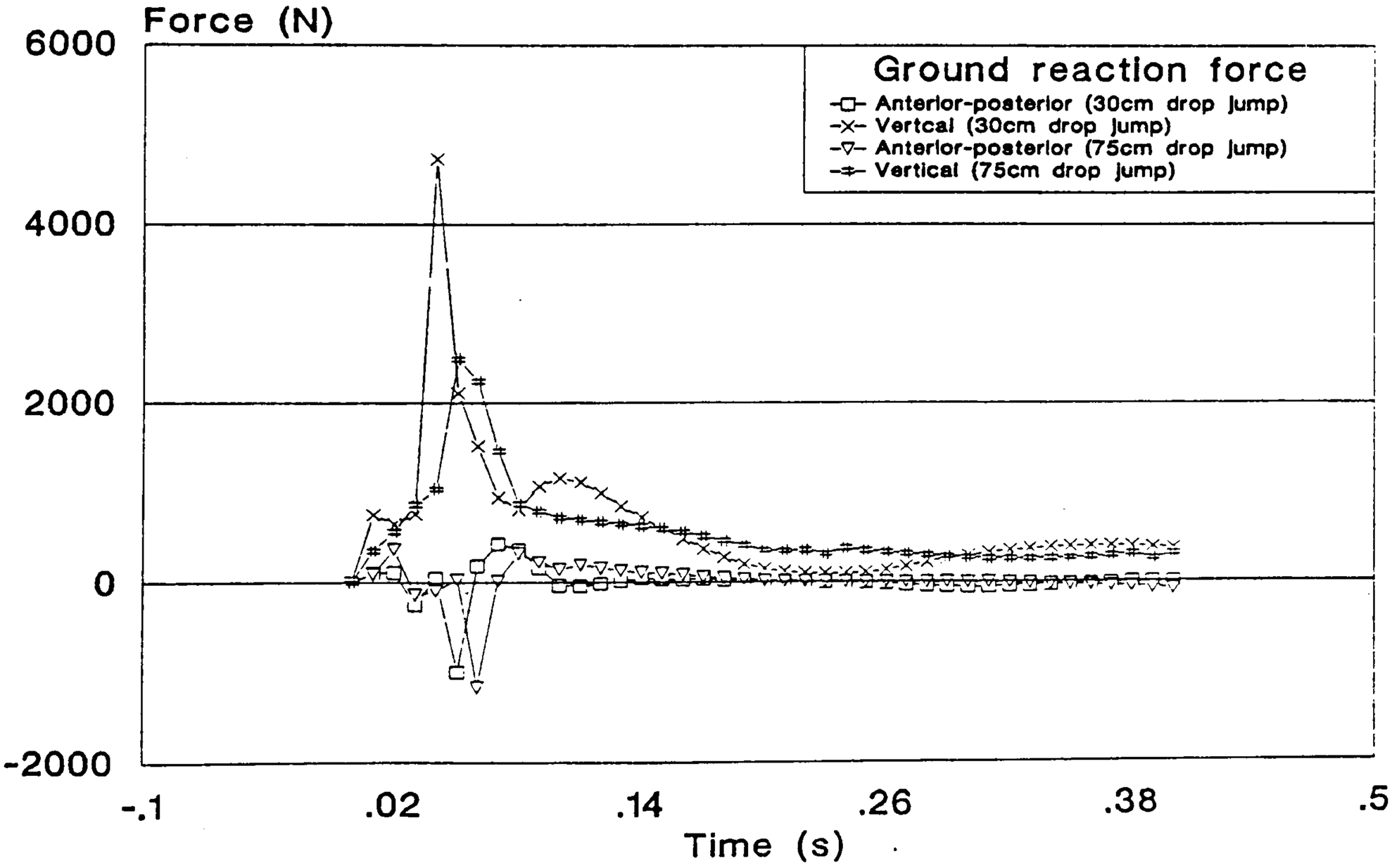


Figure 46. Subject 3: Ground reaction force curves during the second drop land.

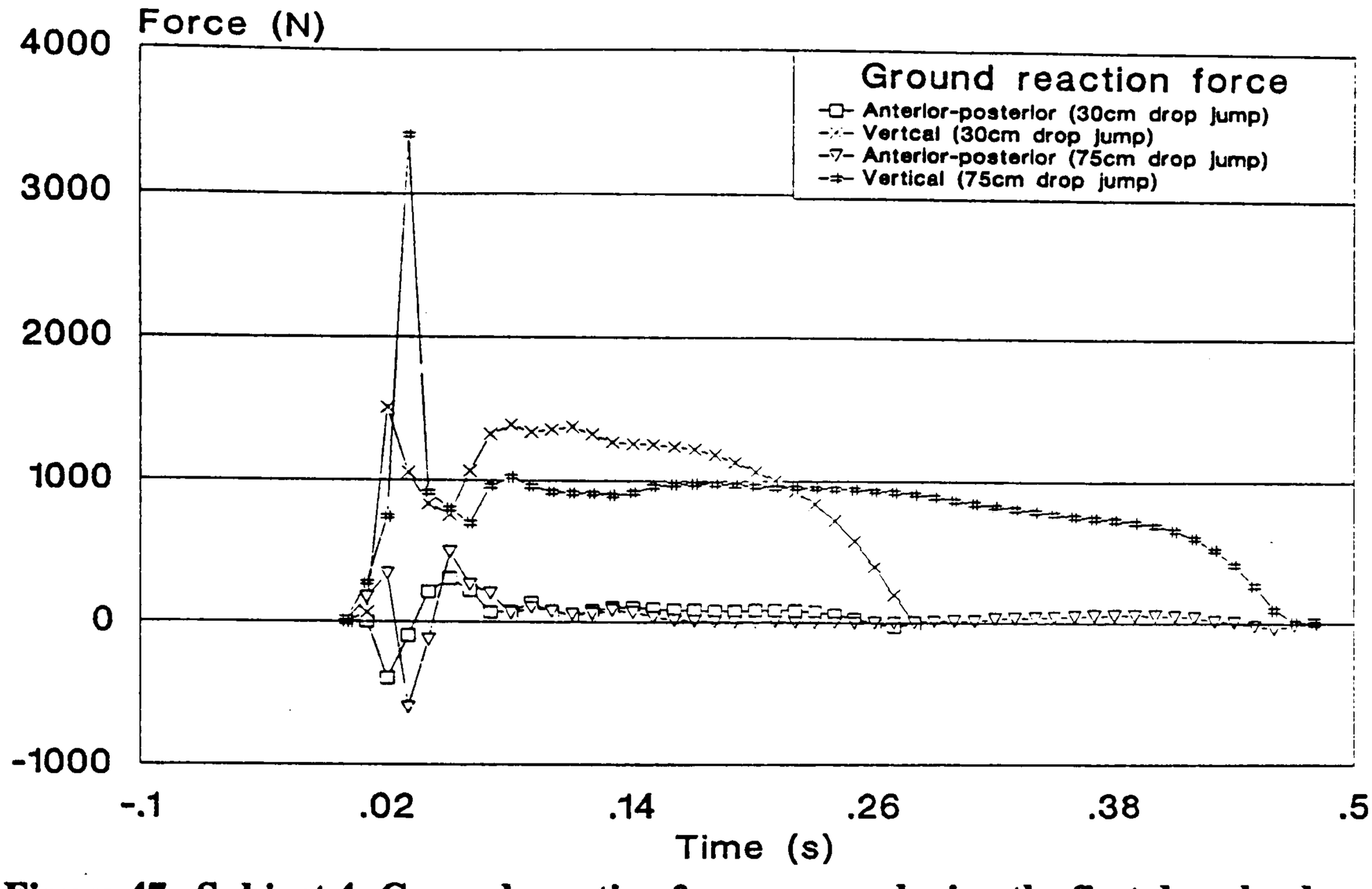


Figure 47. Subject 4: Ground reaction force curves during the first drop land.

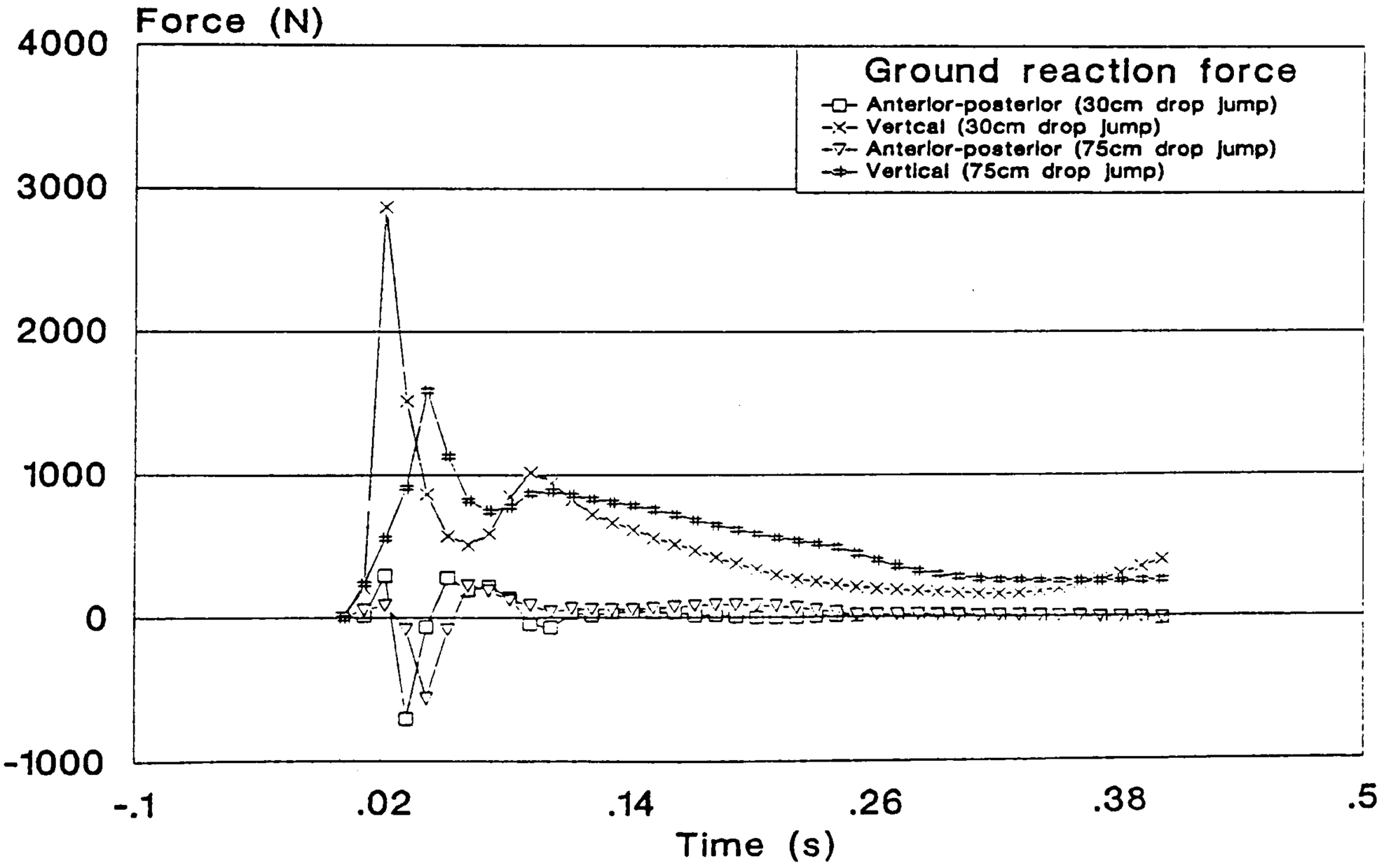


Figure 48. Subject 4: Ground reaction force curves during the second drop land.

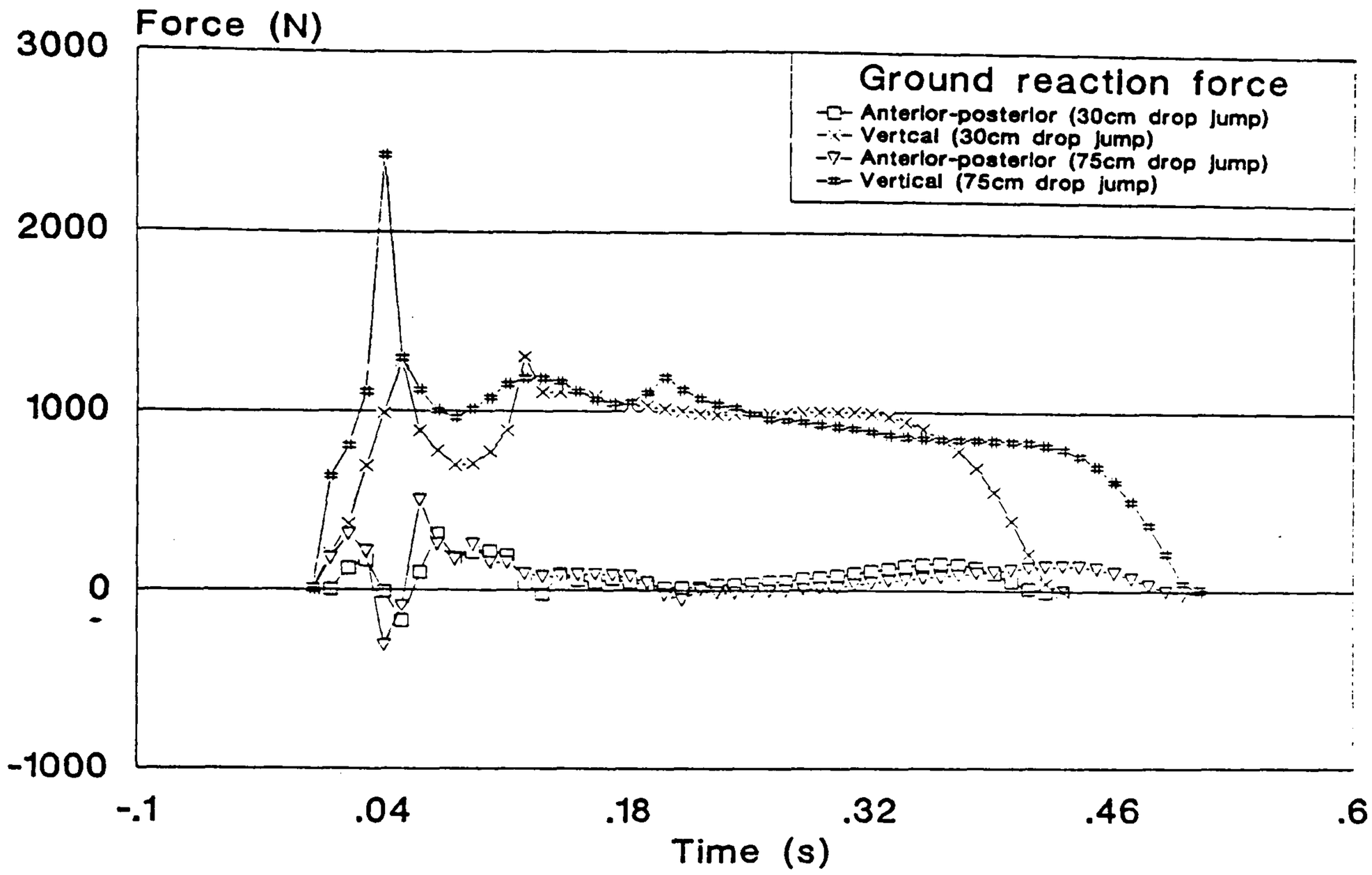


Figure 49. Subject 5: Ground reaction force curves during the first drop land.

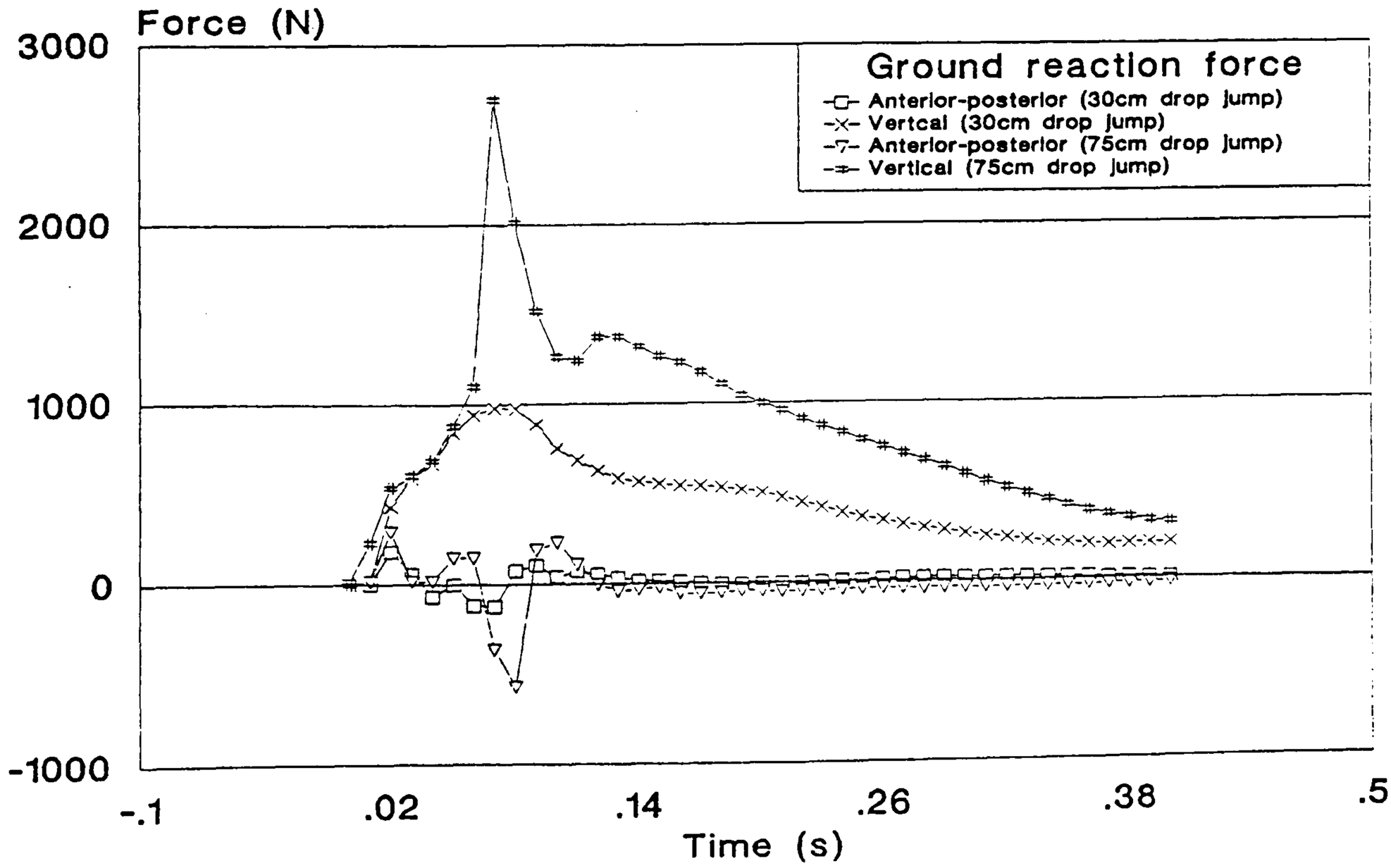


Figure 50. Subject 5: Ground reaction force curves during the second drop land.

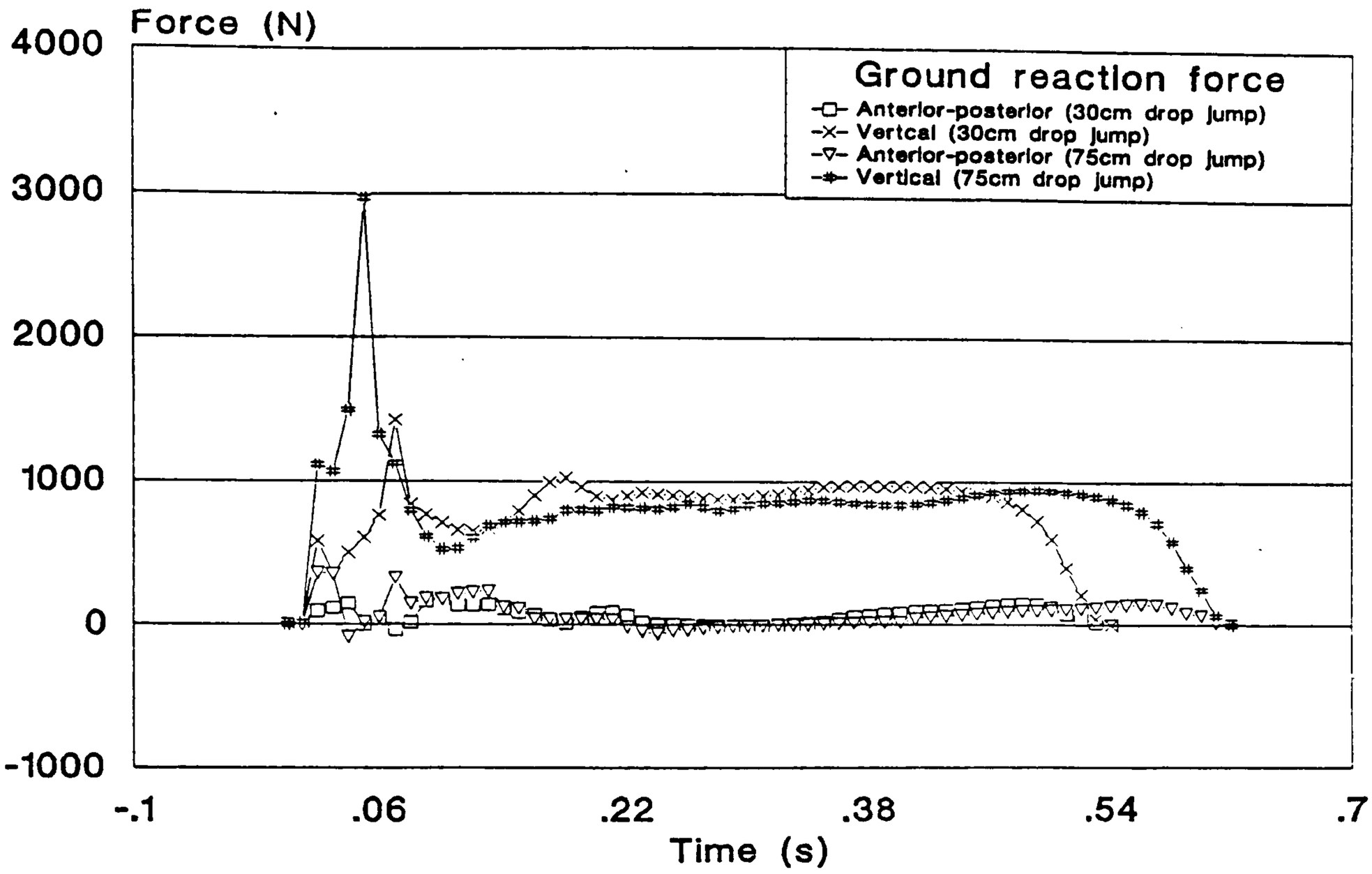


Figure 51. Subject 6: Ground reaction force curves during the first drop land.

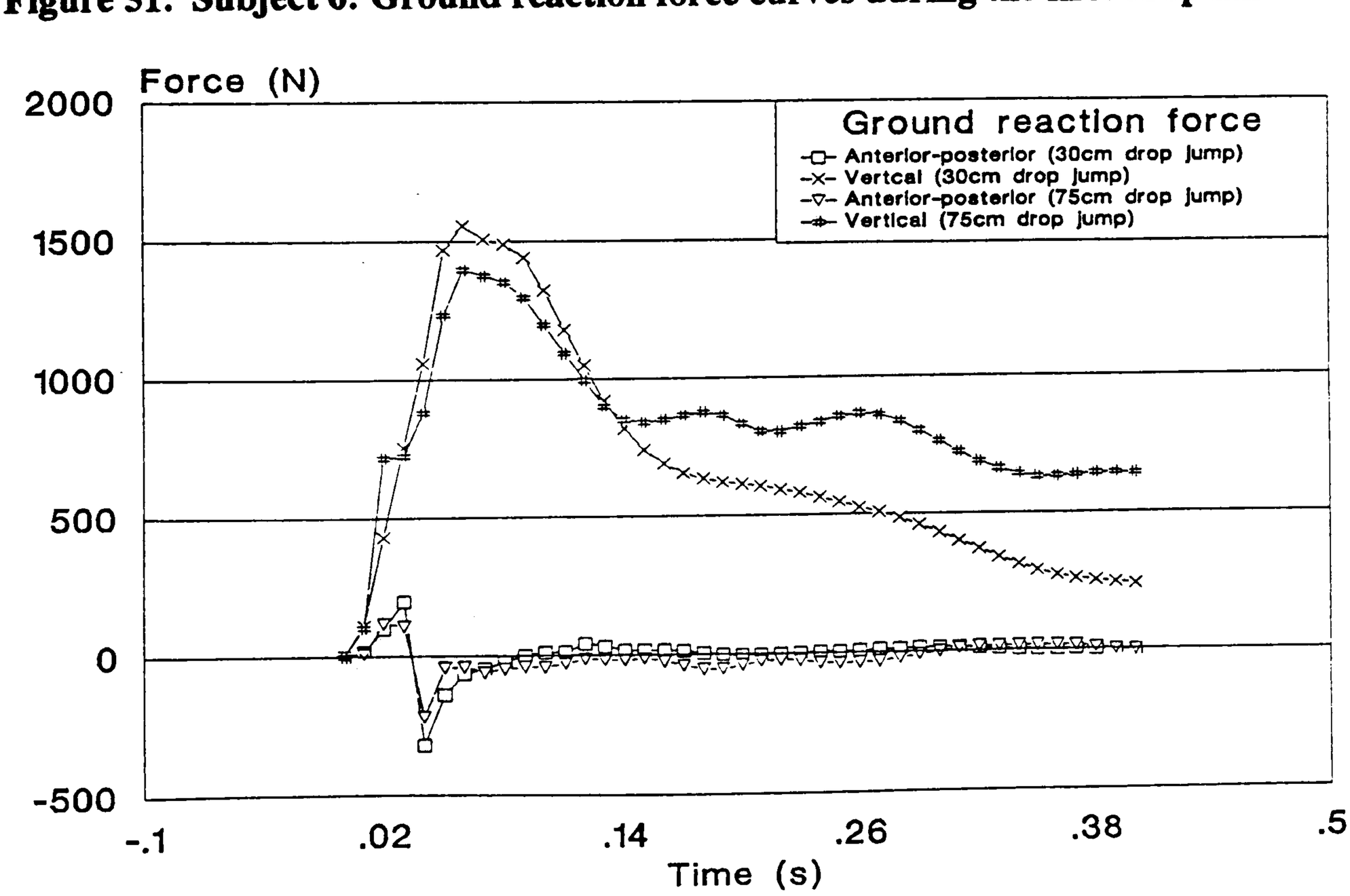


Figure 52. Subject 6: Ground reaction force curves during the second drop land.

Table 29. Peak vertical (FZ) and anterior-posterior (FX) ground reaction forces (N) during knee flexion (drop peak) and knee extension (push-off peak) for the initial land of the 0.30 and 0.75 m drop jumps.

Subject	Drop Peak				Push-off Peak			
	FZ		FX		FZ		FX	
	0.30 m	0.75 m	0.30 m	0.75 m	0.30 m	0.75 m	0.30 m	0.75 m
1	1167 (1.45)	2225 (2.77)	-177* (0.22)	-533 (0.66)	1309 (1.63)	1186 (1.47)	127 (0.16)+	81 (0.10)
2	800 (1.12)	1828 (2.55)	-169 (0.24)	-405 (0.57)	943 (1.32)	903 (1.26)	111 (0.15)	120 (0.17)
3	1525 (1.87)	2874 (3.53)	-210 (0.26)	-420 (0.52)	1627 (2.00)	1297 (1.59)	171 (0.21)	94 (0.12)
4	1615 (2.35)	2709 (3.95)	-334 (0.49)	-517 (0.75)	1278 (1.86)	891 (1.30)	98 (0.14)	71 (0.10)
5	1353 (2.09)	2027 (3.13)	-301 (0.46)	-599 (0.92)	997 (1.54)	1034 (1.60)	142 (0.22)	121 (0.19)
6	1348 (1.83)	2352 (3.20)	-189 (0.26)	-453 (0.62)	965 (1.31)	940 (1.28)	132 (0.18)	170 (0.23)
Mean	1301 (1.78)	2336 (3.19)	-230 (0.32)	-488 (0.67)	1186 (1.61)	1042 (1.42)	130 (0.18)	109 (0.15)
SD ±	291 (0.44)	399 (0.51)	70 (0.12)	75 (0.14)	269 (0.28)	166 (0.16)	25 (0.03)	36 (0.05)

* Negative FX values denote a posterior force component
\+ Positive FX values denote an anterior force component

NB:- Bracketed values refer to forces as a function of body weight.

Table 30. Peak vertical (FZ) and anterior-posterior (FX) ground reaction forces (N) during knee flexion (drop peak) for the second land of the 0.30 and 0.75 m drop jumps.

Subject	Drop peak			
	FZ		FX	
	0.30 m	0.75 m	0.30 m	0.75 m
1	1857 (2.31)	2374 (2.95)	-637* (0.79)	-438 (0.54)
2	1650 (2.30)	1624 (2.27)	-495 (0.69)	-419 (0.59)
3	4656 (5.73)	2405 (2.96)	-773 (0.95)	-1028 (1.26)
4	2711 (3.95)	1608 (2.34)	-805 (1.17)	-655 (0.95)
5	1655 (2.56)	2348 (3.63)	-508 (0.78)	-399 (0.62)
6	1709 (2.32)	1309 (1.78)	-304 (0.41)	-204 (0.28)
Mean	2373 (3.19)	1945 (2.65)	487 (0.80)	524 (0.71)
SD ±	1189 (1.40)	486 (0.65)	290 (0.25)	285 (0.34)

* Negative FX values denote a posterior force component

NB:- Bracketed values refer to forces as a function of body weight.

0.51 BW) for the 0.30 and 0.75 drop jumps, respectively ($p<0.01$). The mean peak anterior-posterior force components accompanied these vertical forces and were directed posteriorly. Mean values of 230 ± 70 N (0.32 ± 0.12 BW) and 488 ± 75 N (0.67 ± 0.14 BW), respectively ($p<0.01$), were attained. On one occasion, whilst performing the 30 cm drop jump, one subject (subject 2) did not produce any initial sharp peaks in the recorded force traces (Figure 43). Closer examination of the film data showed that on this occasion the subject had prevented a heel strike.

Once past this initial impact phase, the vertical force components tended to 'level' out. During knee extension (push-off phase) mean peak values of 1186 ± 75 N (1.61 ± 0.28 BW) and 1042 ± 166 (1.42 ± 0.16 BW) N were found for the 0.30 and 0.75 m drop jumps respectively, the difference being non-significant ($p>0.05$). The FX force component was generally observed to be directed anteriorly after initial impact, approach zero or become slightly negative towards the peak in knee flexion, and return to a positive peak during the push-off phase. The latter peak was found to attain mean values of 130 ± 25 N (0.18 ± 0.03 BW) and 109 ± 36 N (0.15 ± 0.05 BW) for the 0.30 and 0.75 m drop jumps respectively; again the difference was not significant.

The second landing of the drop jumps was similar with respect to initial sharp peaks in the FZ and FX force components. On average they were found to occur within the first 0.09 s of ground contact. Mean peak forces attained similar values to those found during the initial landing of the 0.75 m drop jump (Table 30). On this occasion there were no significant differences in either the FZ or FX force components following drop jumps from 0.30 or 0.75 m.

Once the initial impact phase of the second drop land had taken place, FZ force curves either dropped rapidly to attain values approaching half body weight or they declined gradually. Subjects who displayed the former were found to undertake less knee flexion than the latter and return quickly to an upright posture. On all occasions the FX force fluctuated about the base line.

8.9.4 Ankle and Knee Moments

Consideration was given to ankle and knee moments during the push-off phase of the initial land of each drop jump (Table 31). These, in conjunction with the maximum rates of ankle and knee angular extension velocities (Table 22) provide an indication as to the power output at both joints.

Mean ankle moments were found to be larger during the execution of the 0.30 m drop jump, 133.2 Nm as opposed to 123.4 Nm for the 0.75 m drop jump. The reverse was true for mean knee moments; 171.4 and 191.1 Nm for the 0.30 and 0.75 m drop jumps, respectively. These differences were non-significant ($p>0.05$).

8.9.5 Spinal Compression Force-Time Curves

Individual spinal compression force-time curves for each subject and for both lands during 0.30 and 0.75 m drop jumps are shown in Figures 53 through to 64. Again, for ease of comparison only one curve per drop height is shown. The compression curves are for a model with a ESMMA of 70 mm, a moment arm more realistic of the findings from present day literature (Table 3). To determine the influence of drop height on compression forces and investigate how forces changed with respect to periods during each land, a number of salient features within each force curve were subjected to analysis.

The compression curves for the initial drop land appeared to be characterised by the three distinct periods (Figure 65). In accordance with the vertical ground reaction force curves, the spinal compression curves displayed sharp peaks within the first 0.1 s of ground contact. Again, these represented the impact forces transmitted through the lower limbs during forefoot and heel strike. For the purpose of analysis this was described as period 1 of the land. Period 2 of the compression curve represented a point in time directly after heel impact, extending throughout knee flexion and extension to a point where a final distinct rise in the force curve occurred. This represented, with the exception of one subject (subject 4) performing the 0.30 m drop jump, a shift from a net positive lumbar muscle moment to a net negative lumbar muscle moment. The final portion of the compression curve extended from the point at

Table 31. Maximum mean net ankle and knee moments during the push-off phase of the first land.

Drop jump height(m)	Analysis session	Net ankle moments (Nm)	Net knee moments (Nm)
0.75	Test 1	123.4 ± 24	191.1 ± 55
0.30	Test 1	133.2 ± 32	171.4 ± 43

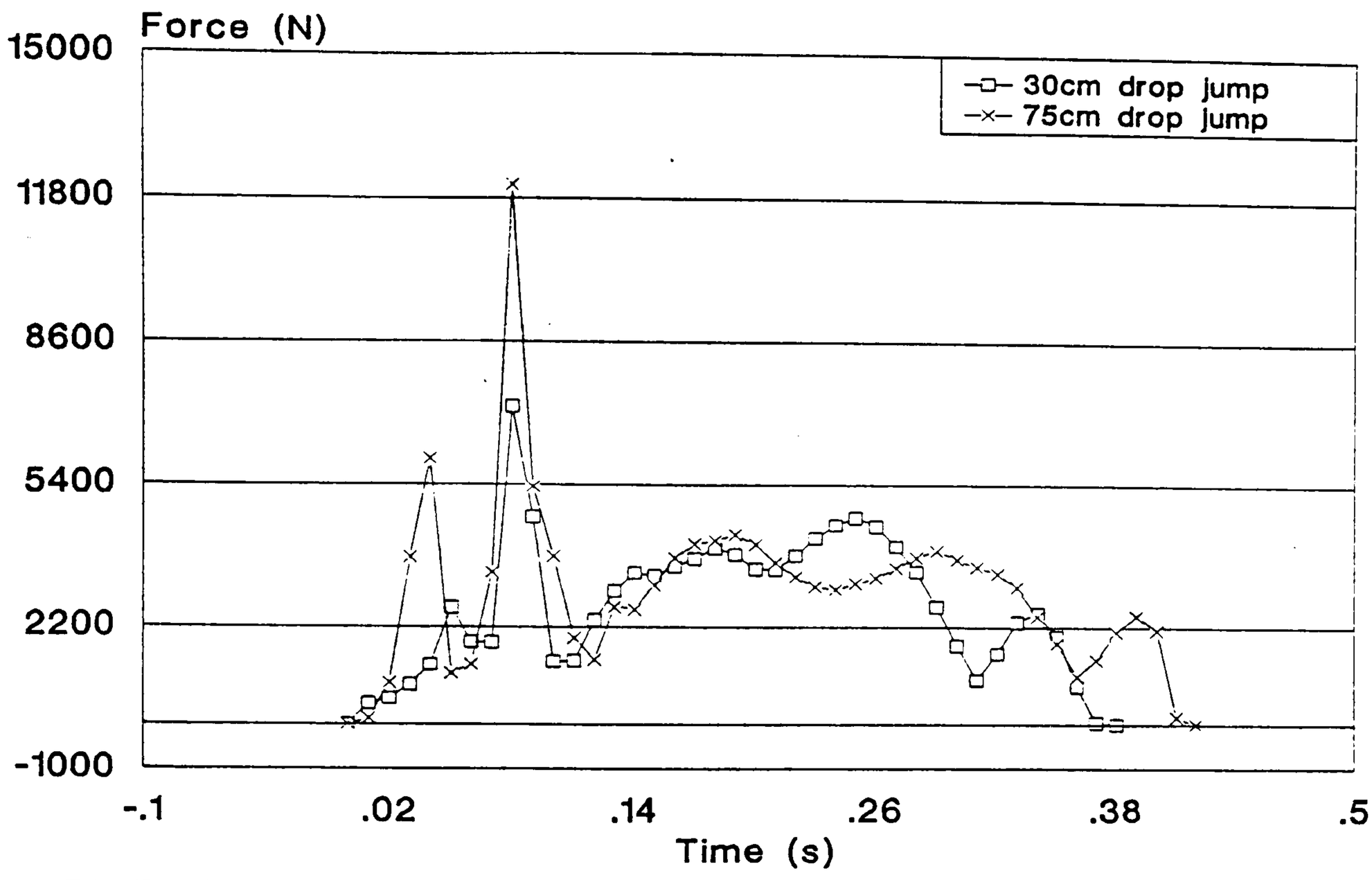


Figure 53. Subject 1: Spinal normal compression force curves during the first drop land.

Figure 54. Subject 1: Spinal normal compression force curves during the second drop land.

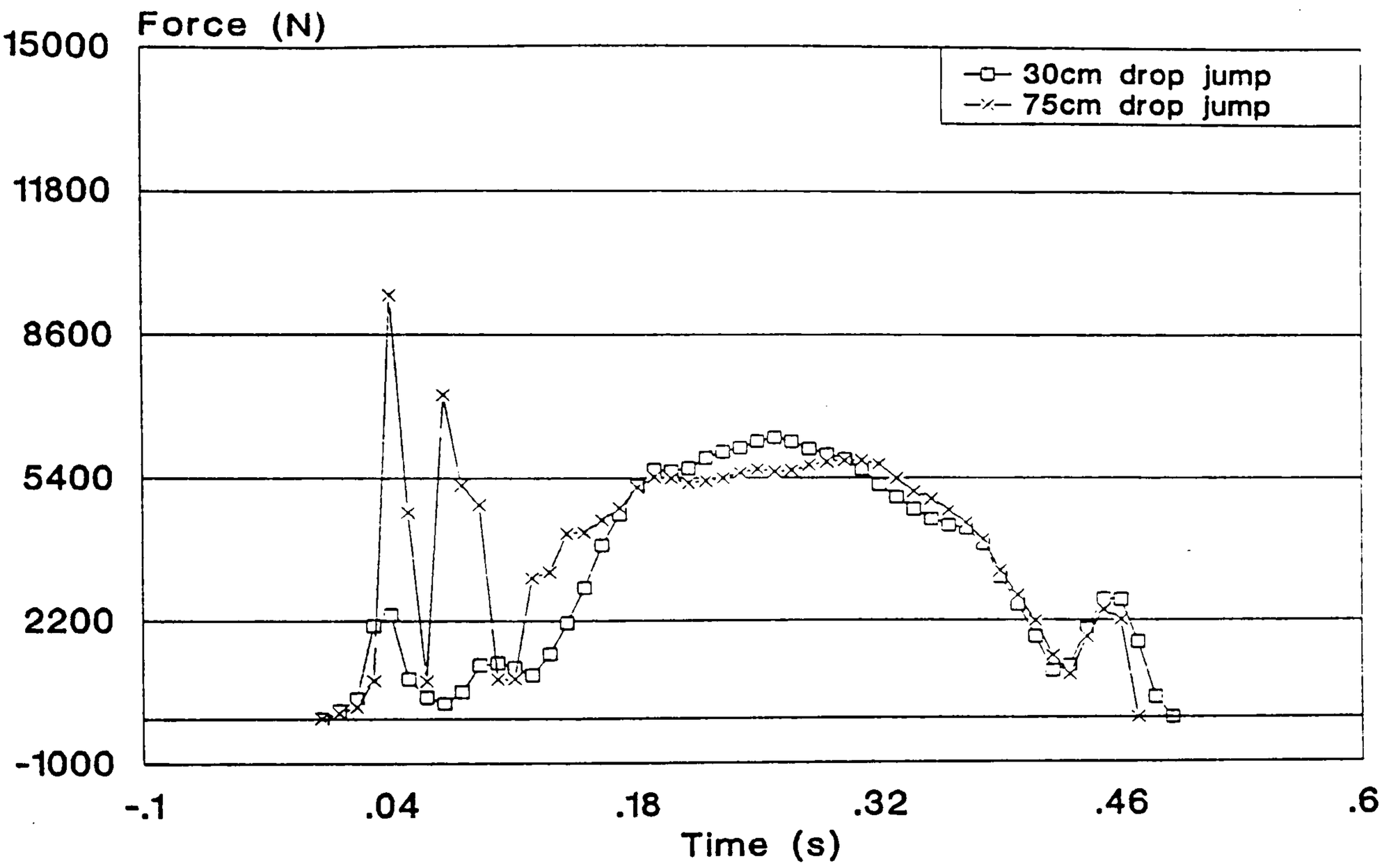


Figure 55. Subject 2: Spinal normal compression force curves during the first drop land.

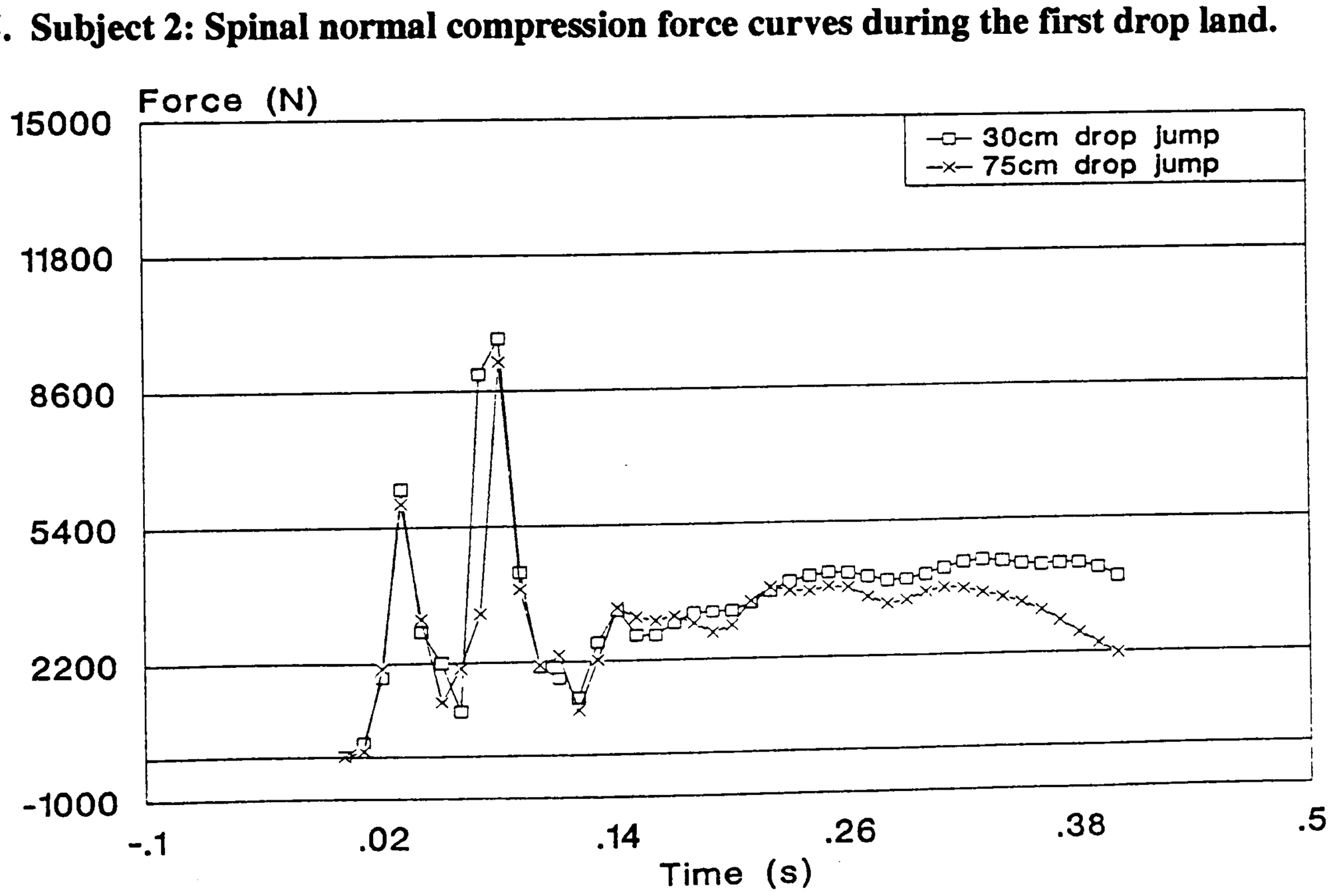


Figure 56. Subject 2: Spinal normal compression force curves during the second drop land.

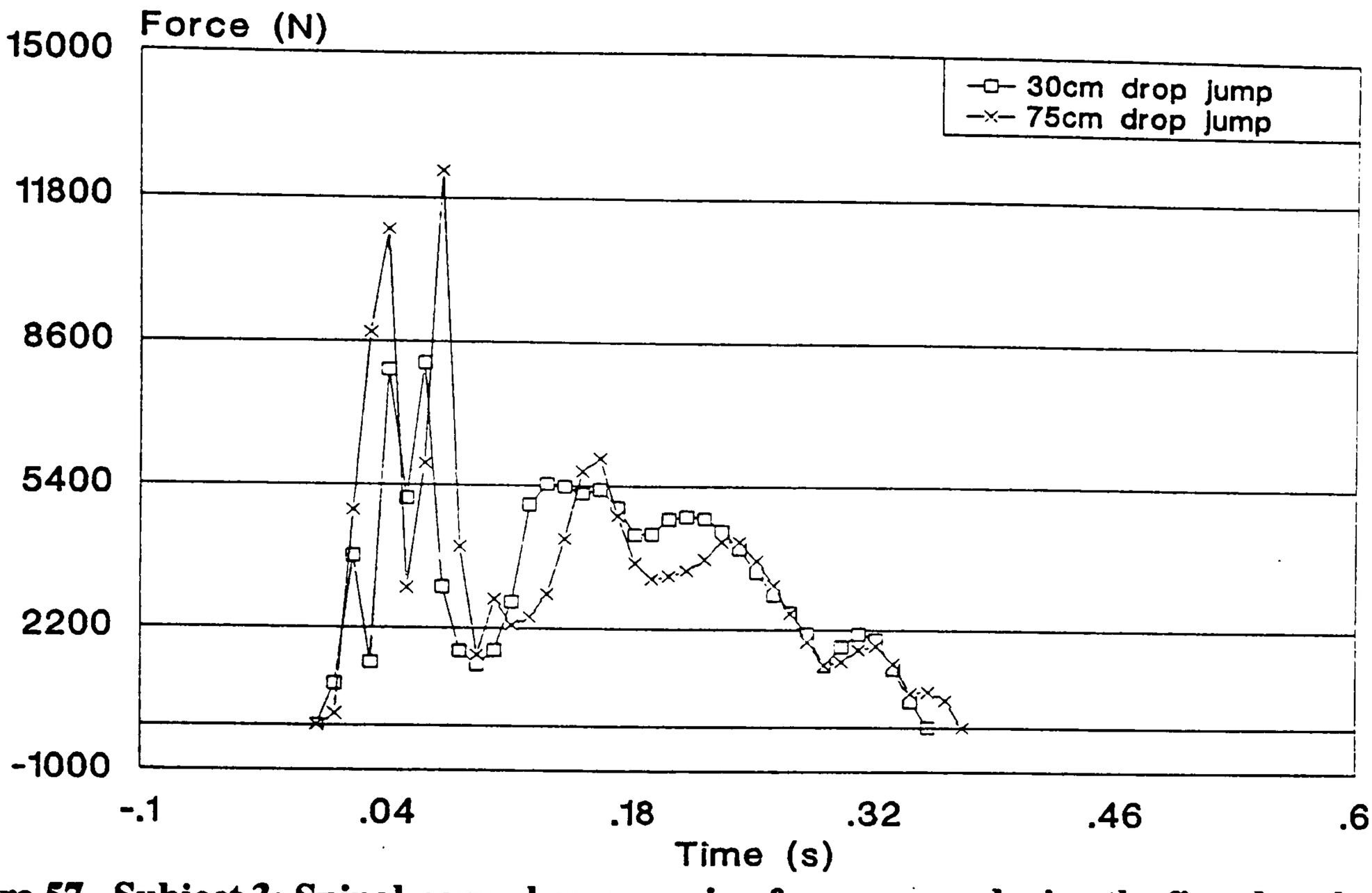


Figure 57. Subject 3: Spinal normal compression force curves during the first drop land.

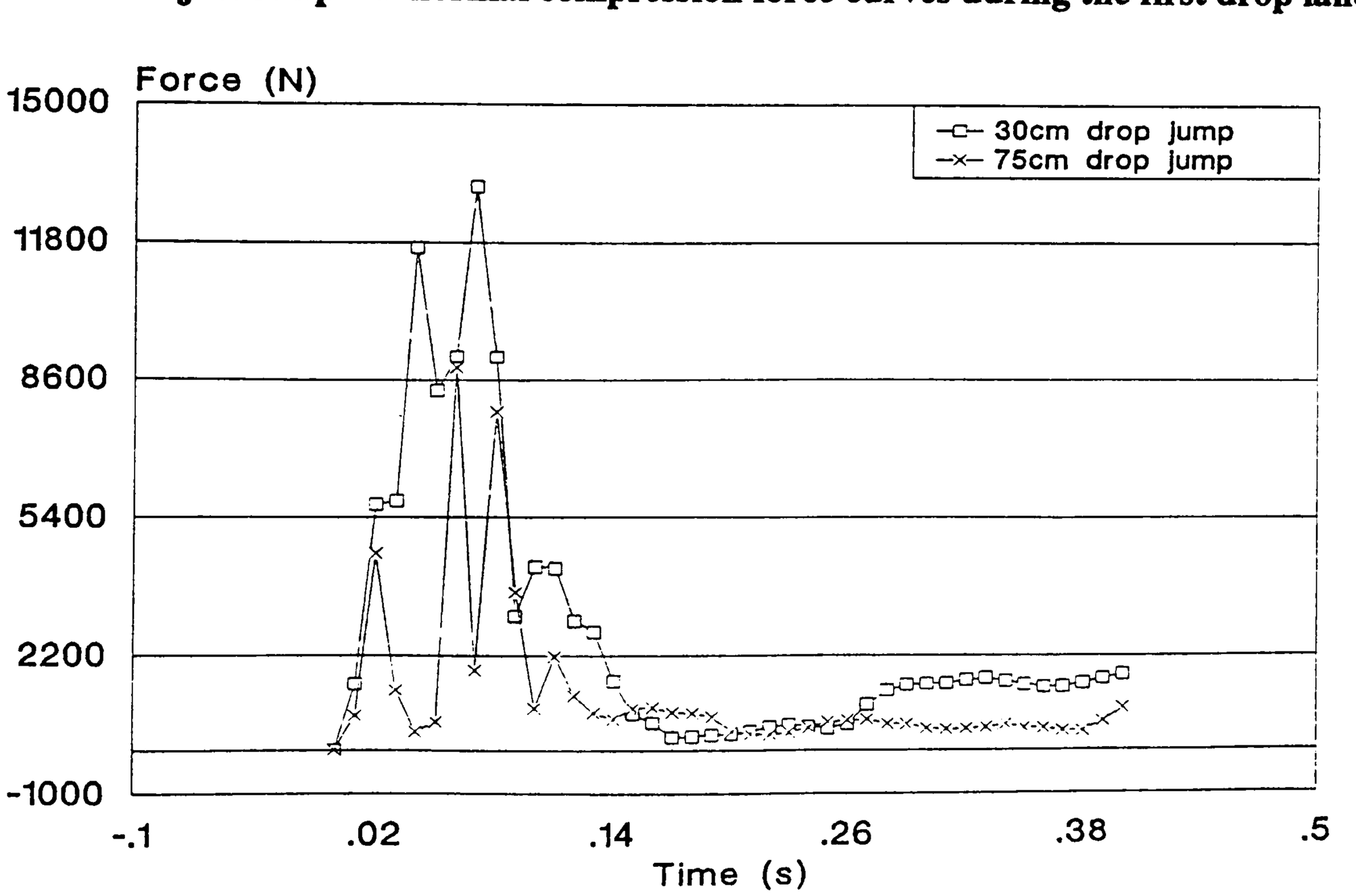


Figure 58. Subject 3: Spinal normal compression force curves during the second drop land.

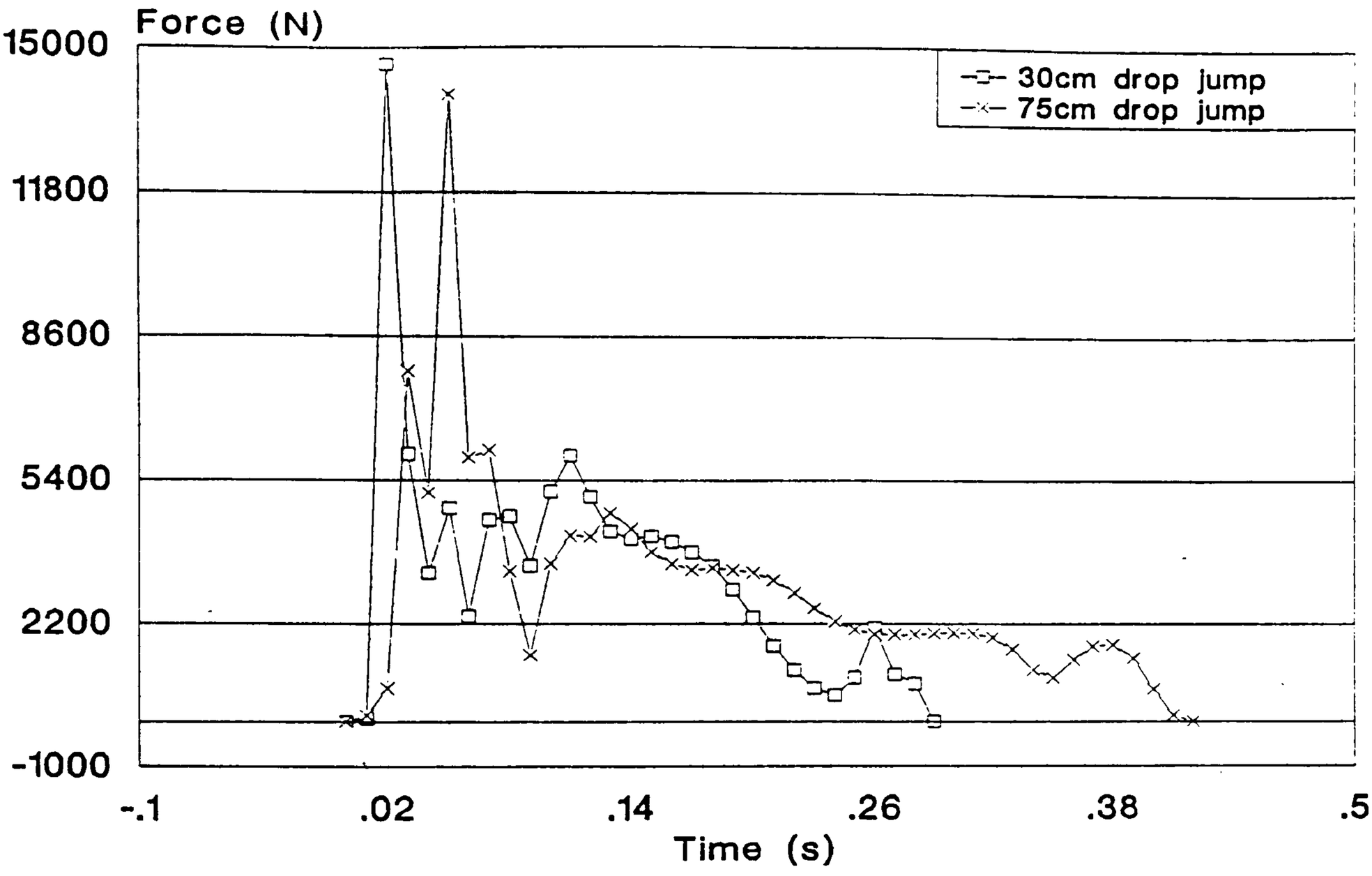


Figure 59. Subject 4. Spinal normal compression force curve during the first drop land.

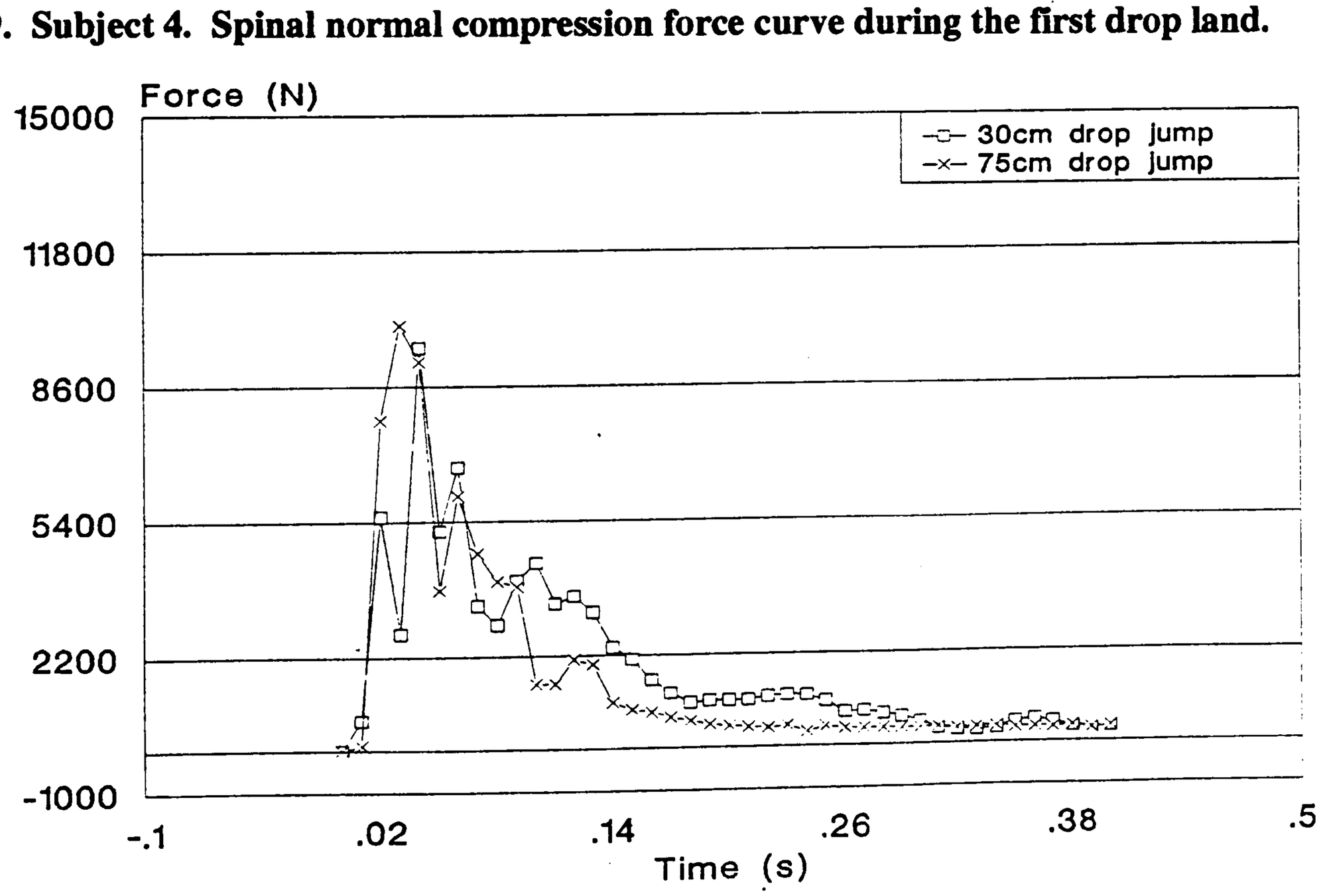


Figure 60. Subject 4. Spinal normal compression force curve during the second drop land.

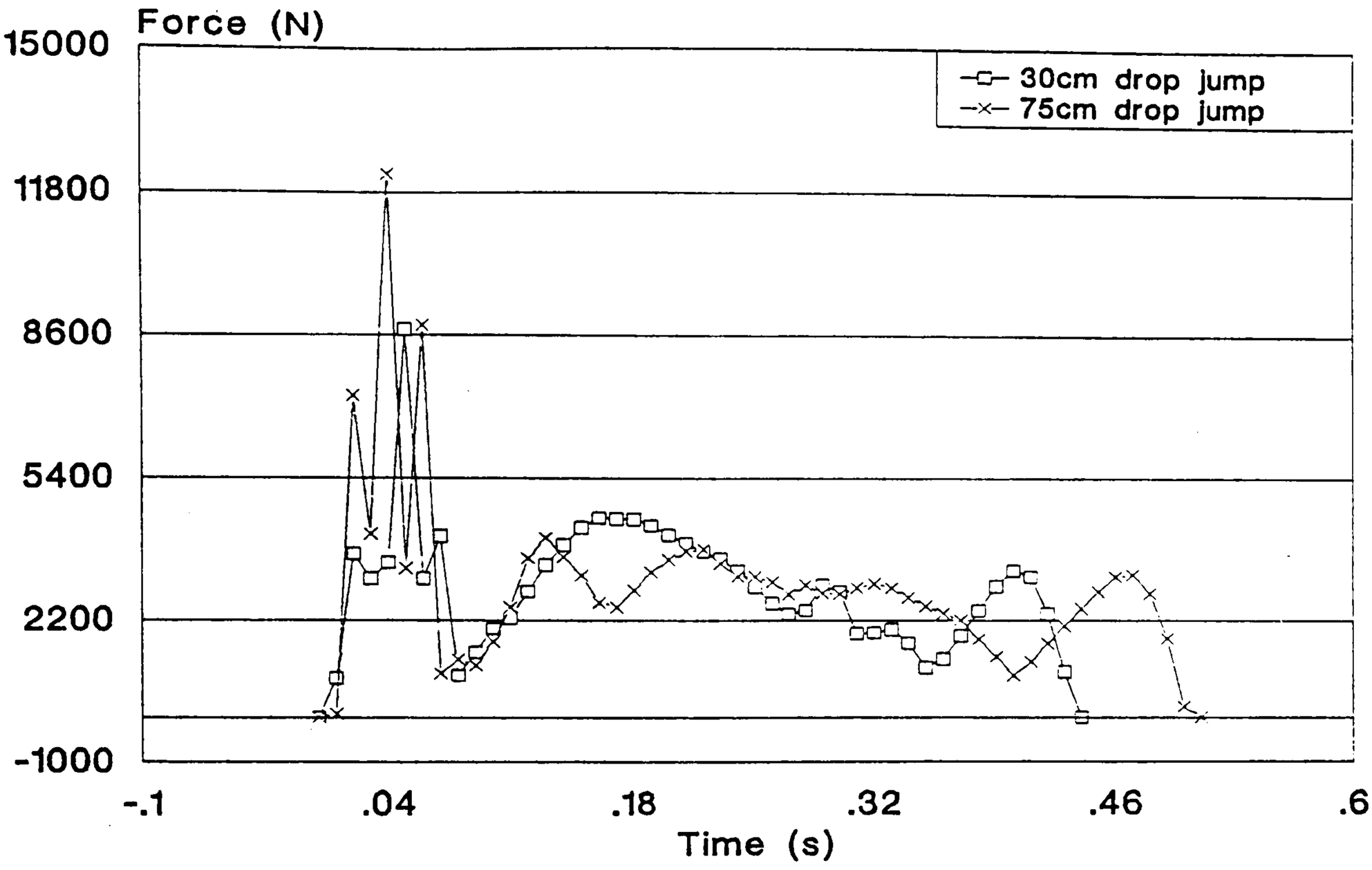


Figure 61. Subject 5: Spinal normal compression force curves during the first drop land.

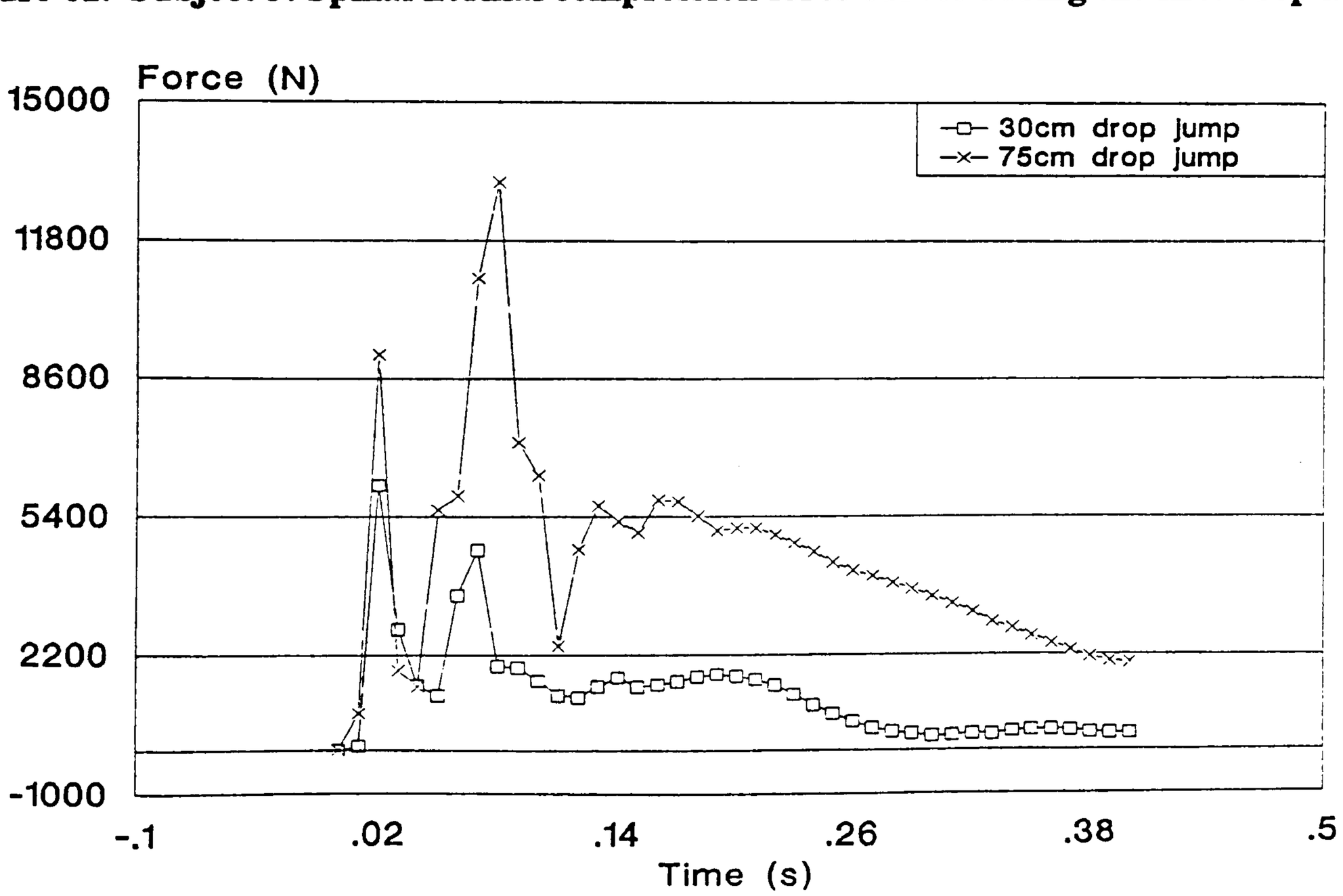


Figure 62. Subject 5: Spinal normal compression force curves during the second drop land.

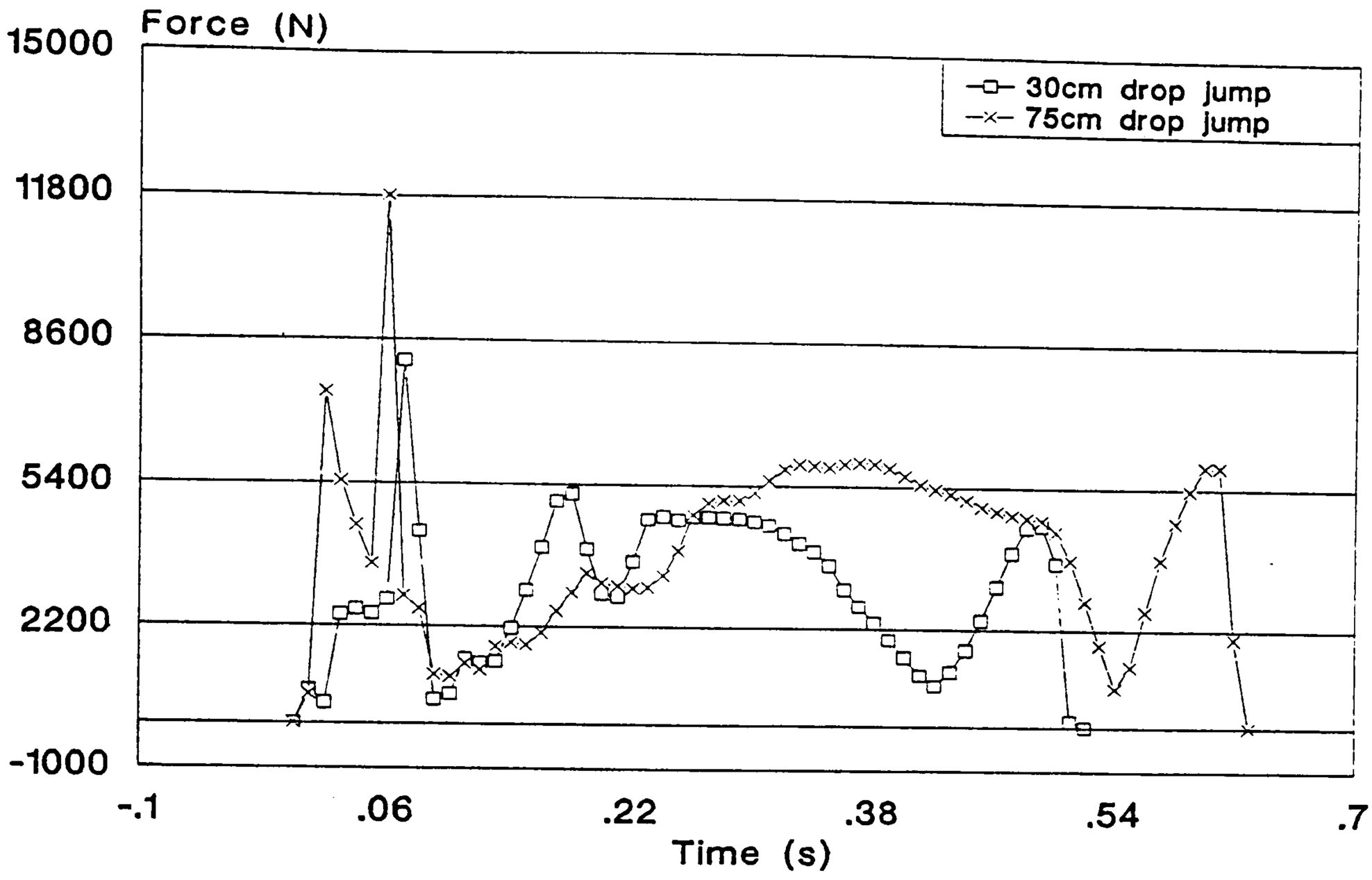


Figure 63. Subject 6: Spinal normal compression force curves during the first drop land.

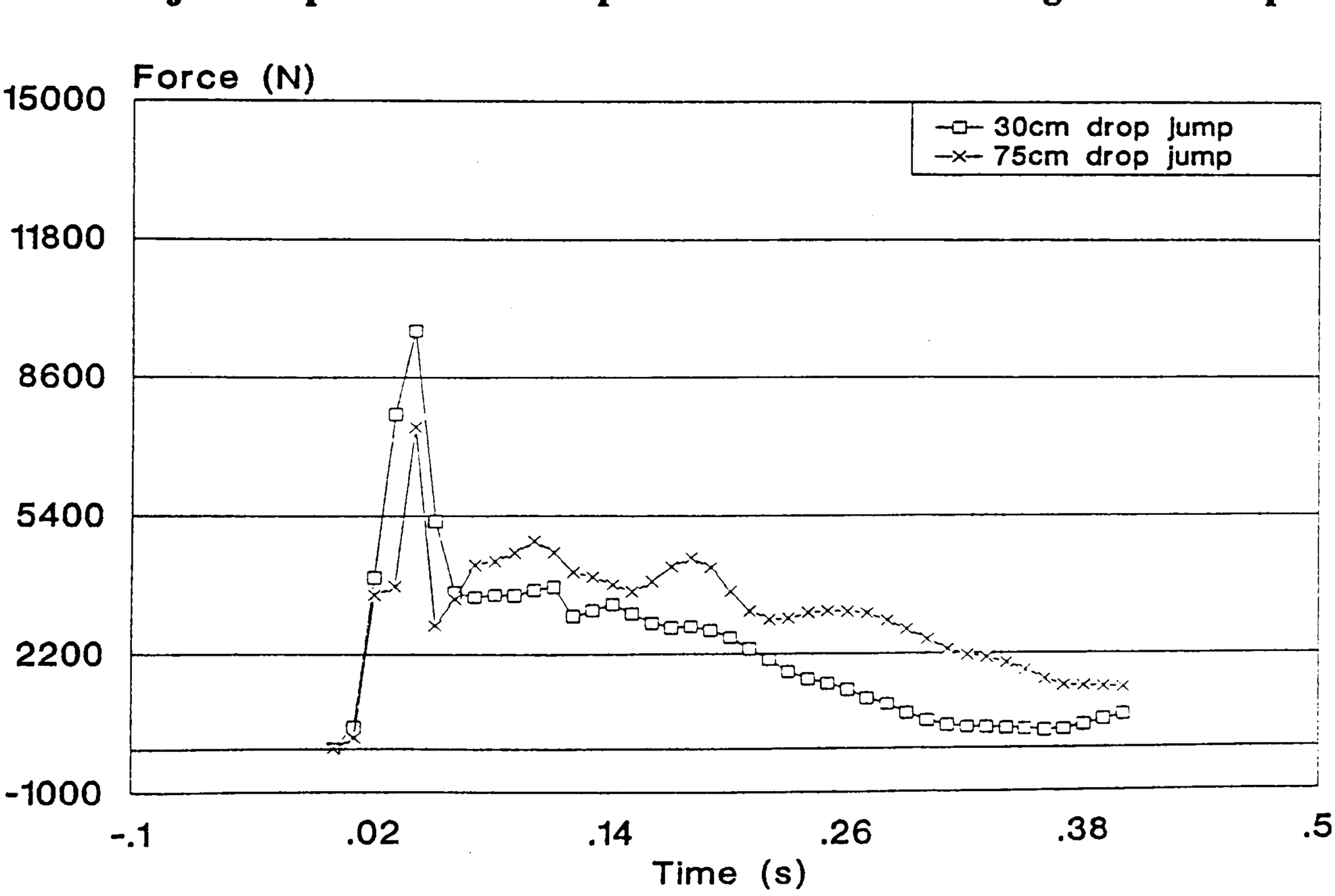


Figure 64. Subject 6: Spinal normal compression force curves during the second drop land.

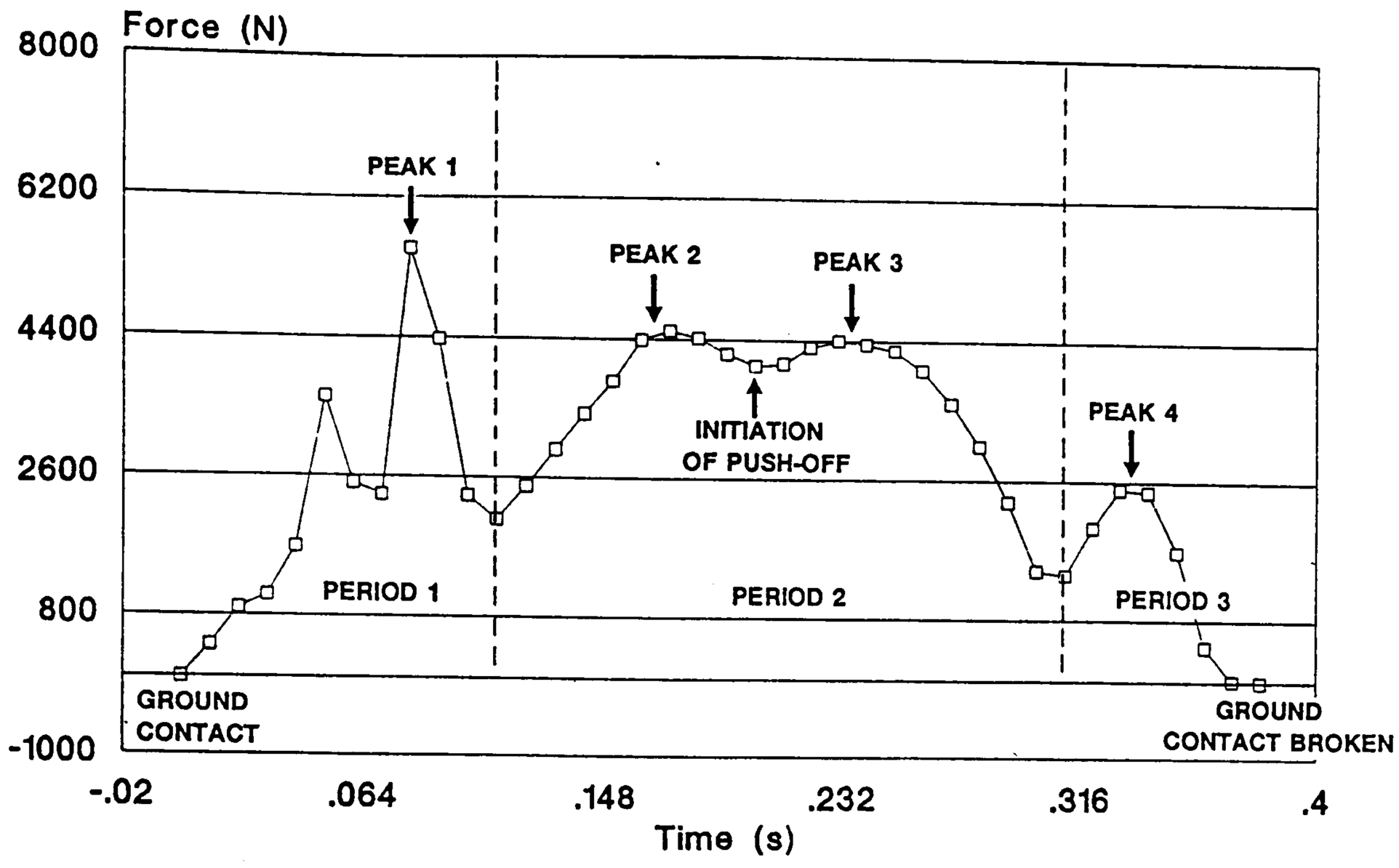


Figure 65. Salient features of the normal spinal compression force curve identified during the first land of each drop jump.

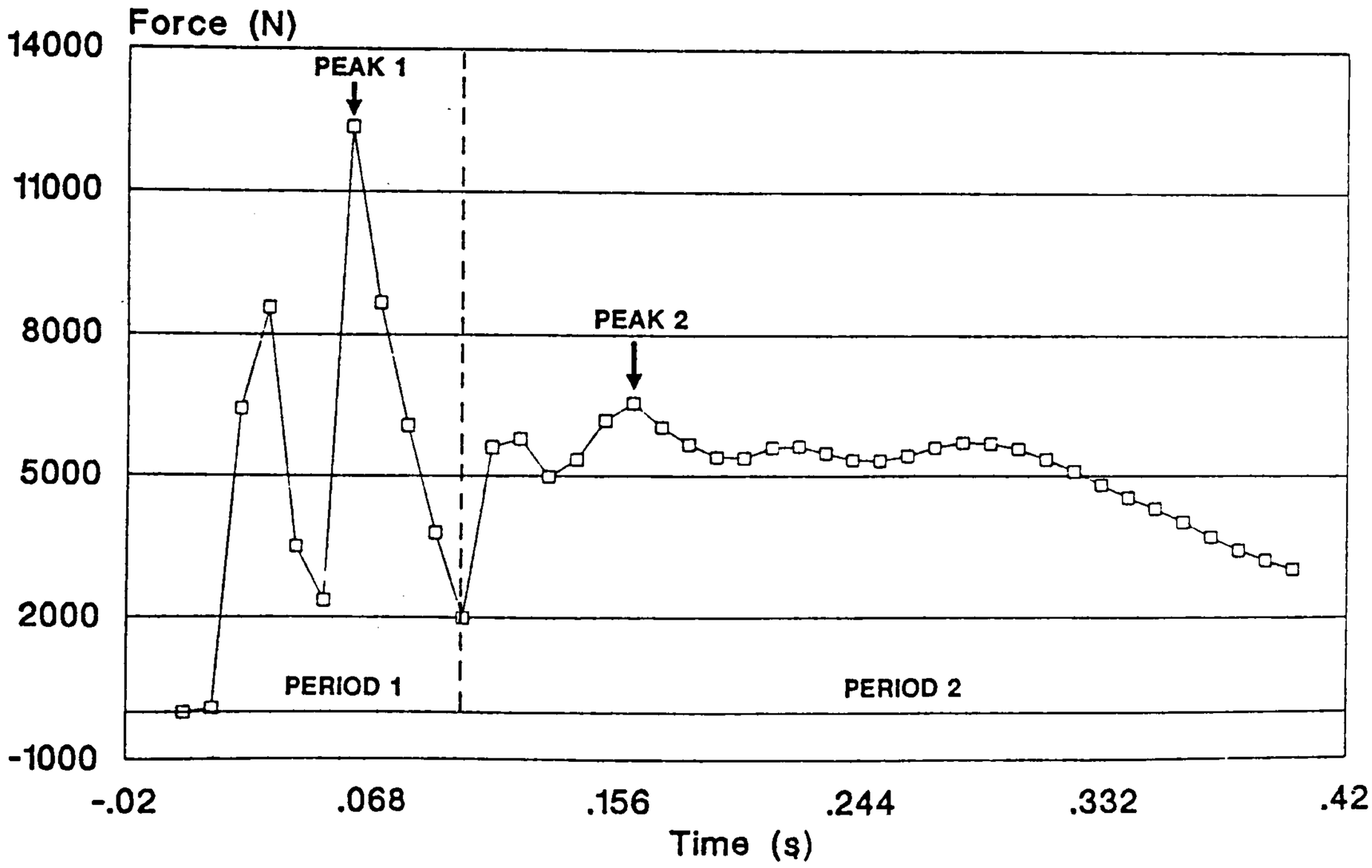


Figure 66. Salient features of the normal spinal compression force curve identified during the second land of each drop jump.

which this shift in the direction of the net muscle moment occurred to the instance at which subjects broke ground contact (Period 3). Analysis considered four peaks, peak 1 occurring within period 1, peaks 2 and 3 occurring within period 2, and peak 4 within the third period. Peak 1 was the maximum force for either forefoot or heel strike. Peaks 2 and 3 represented the maximum forces within period 2 during knee flexion and knee extension, respectively. Peak 4 was the maximum force during period 3. The compression curves were also integrated to find the areas encompassed by the total and individual periods identified, thereby providing an indication of the cumulative load acting on the lower spine.

For the second land only two periods during the compression curves were identified. The first again extended from ground contact through to completion of heel strike, the second being the time thereafter. Individual peaks during each period were analysed along with the area encompassed by the total and individual periods of each force trace. The peaks and periods identified during each land are depicted in Figures 65 and 66.

Tables 32 to 35 show the mean spinal compression peak forces and integrals (±SD) for drop jumps performed from 0.30 and 0.75 m. These forces are for a biomechanical model with an ESMMA of 50 mm and 70 mm. Bar charts to compare mean peak spinal compression and shear forces, and spinal compression integrals for both landings during the execution of the 0.30 and 0.75 m drop jump are shown in Figures 67 to 72. The biomechanical model used to predict these forces utilised an ESMMA of 70 mm.

8.9.5.1 Peak Spinal Compression and Shear Forces

When consideration is given to the initial landing of the 0.30 and 0.75 m drop jumps, the spinal compression force curves show rapid fluctuations during period 1 of the land (Figures 53, 55, 57, 59, 61 and 63). On average two distinct peaks occurred within approximately 0.1 s of making ground contact and could be related to initial forefoot contact followed by heel impact. The model estimated that the maximum of these forces, usually the result of heel strike, attained extremely large values following both the 0.30 and 0.75 m drop jumps, 13537 ± 2189 N (18.7 ± 4 BW) and 7595 ± 2693 N (10.6 ± 4.4

Table 32. Mean peak spinal normal compression forces (±SD) calculated at various phases of the initial drop jump land for the two heights and for two erector spinae muscle moment arms.

Erector Spinae muscle moment arm	Peak 1				Peak 2			
	Force (N)		Unit of Body Weight		Force (N)		Unit of Body Weight	
	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump
5 cm	10019 (±3698)	17218 (±3349)	14.0 (±6)	23.8 (±6)	6791 (±974)	6405 (±1271)	9.4 (±1.9)	8.8 (±1.6)
7 cm	7595 (±2693)	13537 (±2189)	10.6 (±4.4)	18.7 (±4)	5258 (±685)	4927 (±922)	7.3 (±1.4)	6.7 (±1.2)

Erector Spinae muscle moment arm	Peak 3				Peak 4			
	Force (N)		Unit of Body Weight		Force (N)		Unit of Body Weight	
	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump
5 cm	6486 (±1076)	6078 (±1026)	8.9 (±1.5)	8.3 (±1.6)	2827 (±568)	3030 (±1514)	3.9 (±0.9)	4.2 (±2.1)
7 cm	5044 (±800)	4652 (±710)	6.9 (±1.1)	6.4 (±1.1)	2692 (±713)	3030 (±1514)	3.7 (±1.1)	4.2 (±2.1)

NB : Refer to text (section 8.9.5) for explanation of each peak

Table 33. Mean peak spinal normal compression forces (±SD) calculated at various phases of the second drop jump land for the two heights and for two erector spinae muscle moment arms.

Erector Spinae	Peak 1 Force (N)		Peak 1 Unit of body weight		Peak 2 Force (N)		Peak 2 Unit of body weight	
	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump
5 cm	13660 (±2214)	14529 (±2697)	18.5 (±2.1)	19.8 (±3.2)	4601 (±1481)	5369 (±2111)	6.2 (±1.8)	7.3 (±2.6)
7 cm	11257 (±1765)	11057 (±2219)	15.3 (±1.9)	15.0 (±2.6)	3562 (±1014)	4216 (±1444)	4.8 (±1.2)	5.7 (±1.7)

NB: Refer to text (section 8.9.5) for explanation of each peak

Table 34. Mean spinal normal compression force integrals (±SD) at various phases of the initial drop jump land for the two heights and for two erector spinae muscle moment arms.

Erector Spinae muscle moment arm	Period 1 Compression force integral (Ns) 30 cm drop jump	75 cm drop jump	Period 2 Compression force integral (Ns) 30 cm drop jump	75 cm drop jump
5 cm	312 (±77)	536 (±76)	1164 (±371)	1337 (±366)
7 cm	256 (±59)	456 (±59)	912 (±263)	1036 (±252)

Erector Spinae muscle moment arm	Period 3 Compression force integral (Ns) 30 cm drop jump	75 cm drop jump	Total Compression force integral (Ns) 30 cm drop jump	75 cm drop jump
5 cm	117 (±53)	168 (±137)	1594 (±329)	2042 (±436)
7 cm	116 (±56)	167 (±137)	1283 (±242)	1659 (±329)

NB: Refer to text (section 8.9.5) for explanation of each period.

Table 35. Mean spinal normal compression force integrals (±SD) at various phases of the second drop jump land for the two heights and for two erector spinae muscle moment arms.

Erector Spinae muscle moment arm	Period 1 Compression force integral (Ns)		Period 2 Compression force integral (Ns)		Total Compression force integral (Ns)	
	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump	30 cm drop jump	75 cm drop jump
5 cm	541 (±137)	546 (±73)	814 (±545)	644 (±385)	1355 (±505)	1518 (±531)
7 cm	471 (±118)	475 (±68)	971 (±563)	757 (±407)	1115 (±341)	1232 (±377)

NB: Refer to text (section 8.9.5) for explanation of each period.

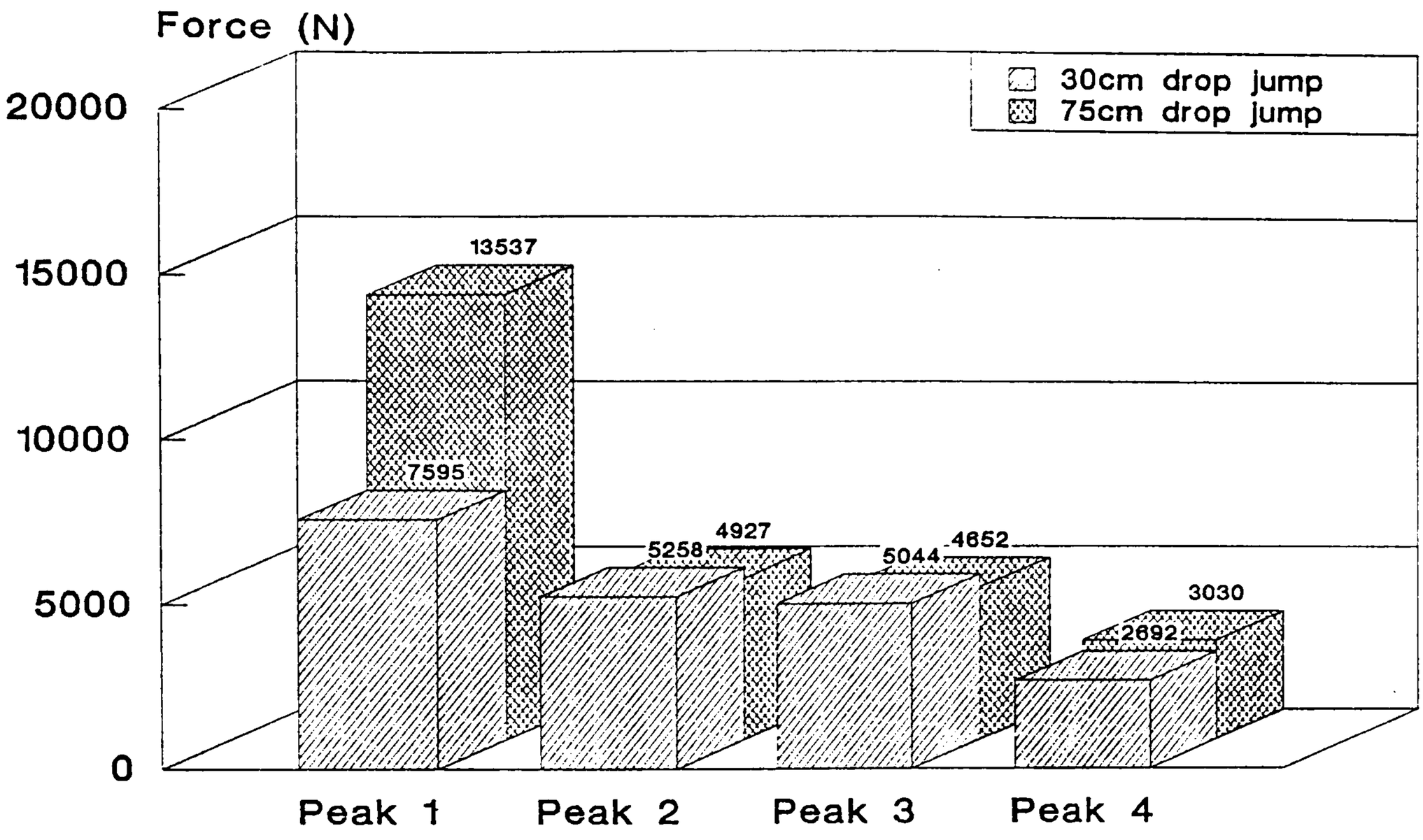


Figure 67. Mean peak spinal normal compression forces at various phases of the first land whilst performing 0.30 and 0.75 m drop jumps.

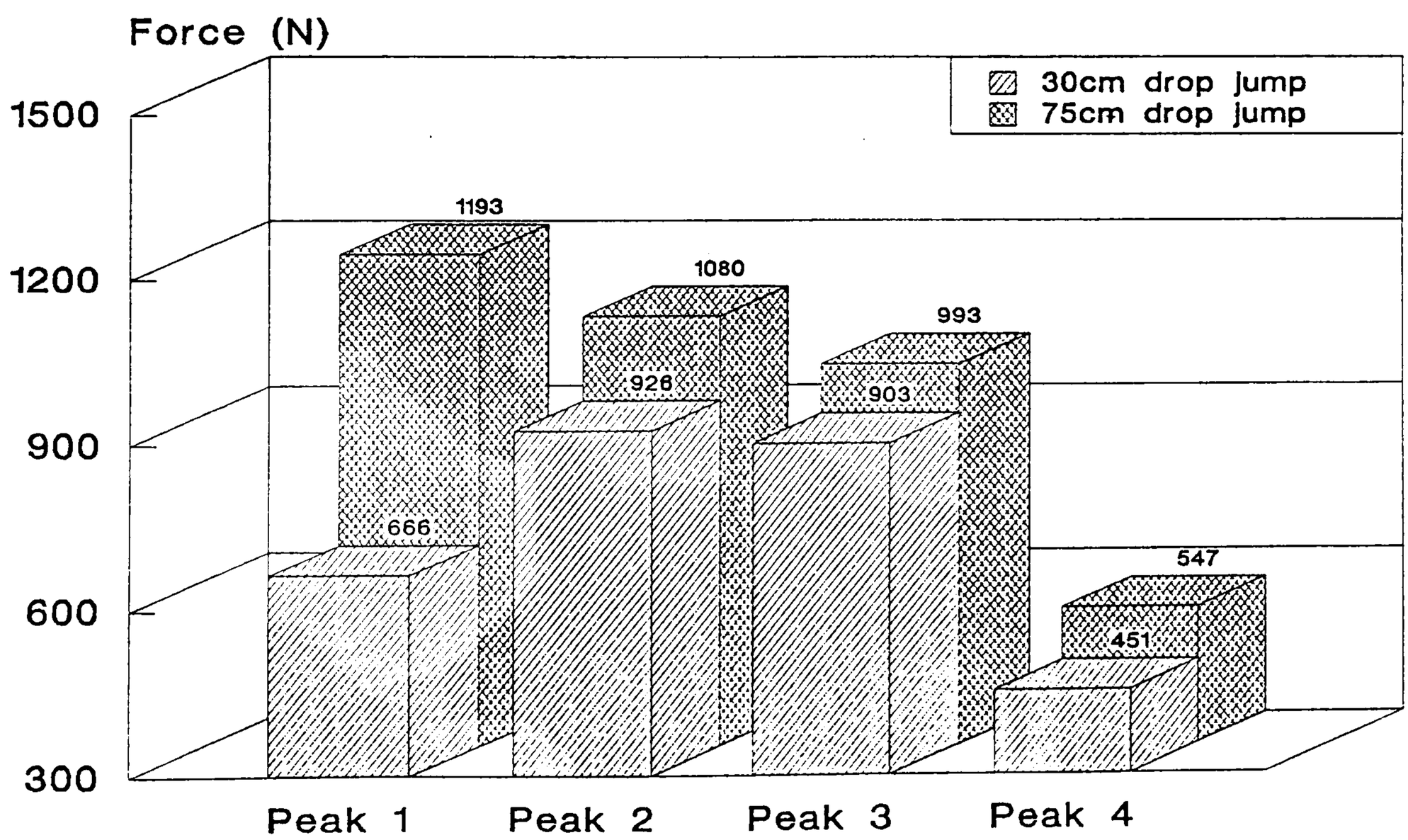


Figure 68. Mean peak spinal shear forces at various phases of the first land whilst performing 0.30 and 0.75 m drop jumps.

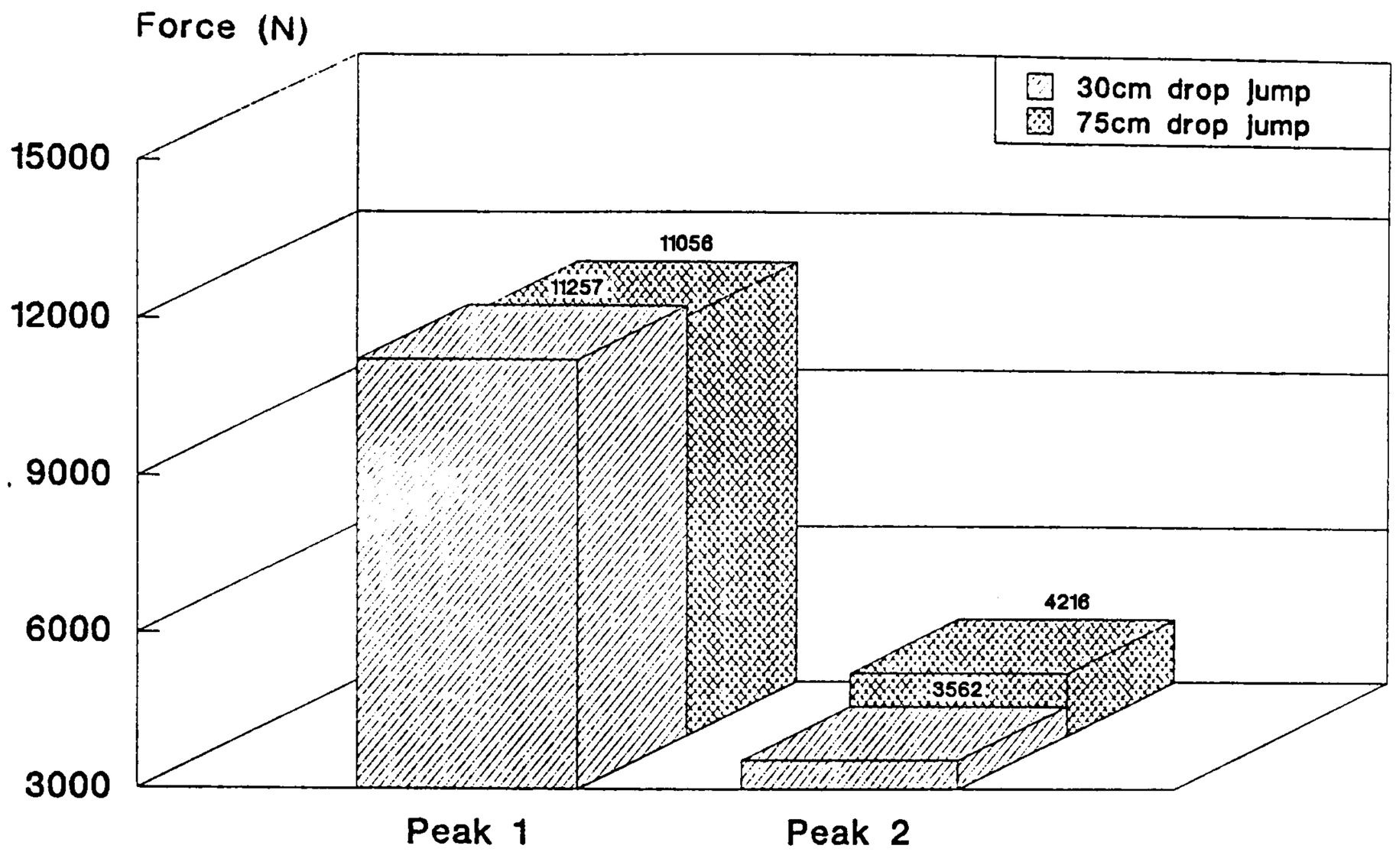


Figure 69. Mean peak spinal normal compression forces at various phases of the second land whilst performing 0.30 and 0.75 m drop jumps.

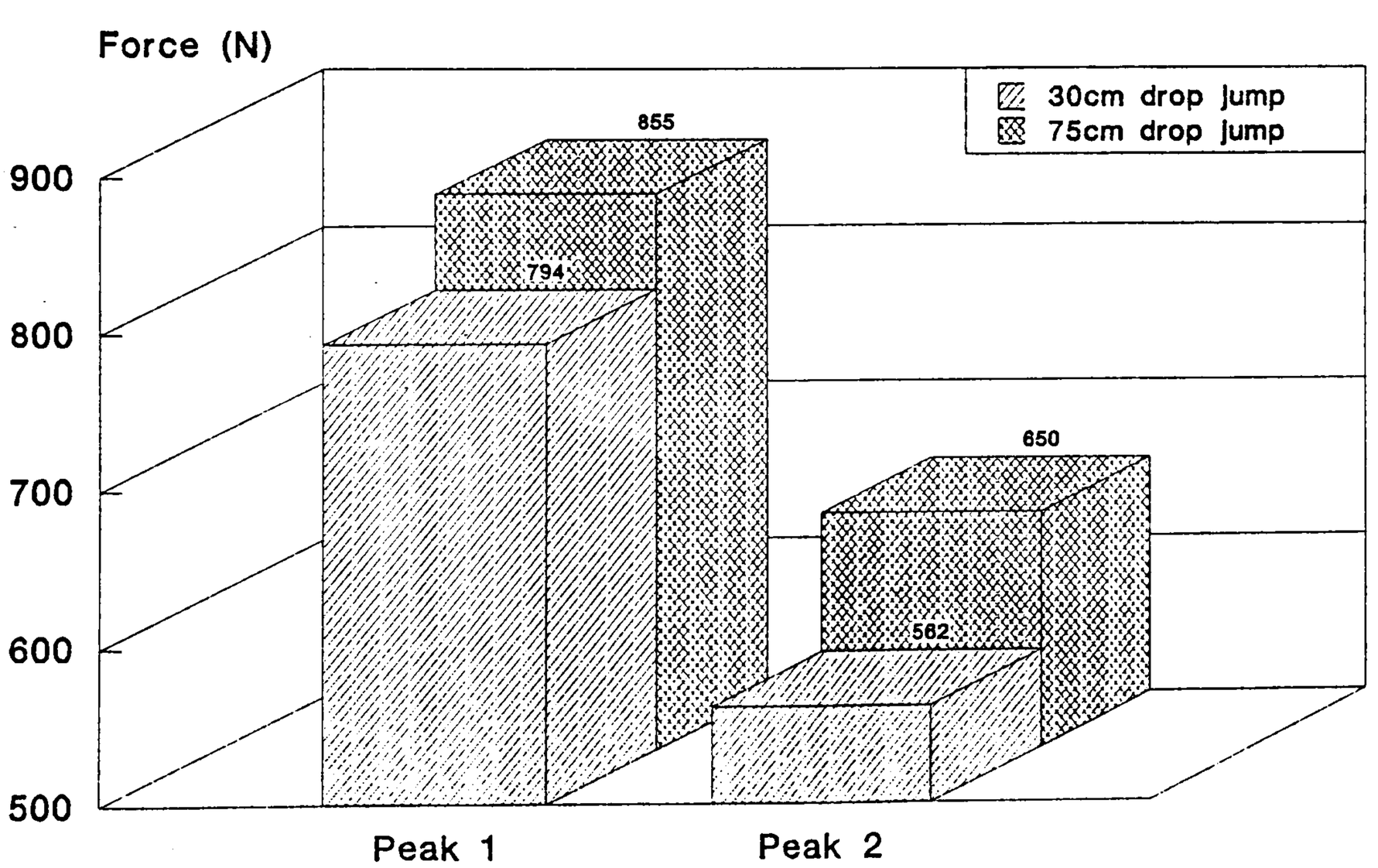


Figure 70. Mean peak spinal shear forces at various phases of the second land whilst performing 0.30 and 0.75 m drop jumps.

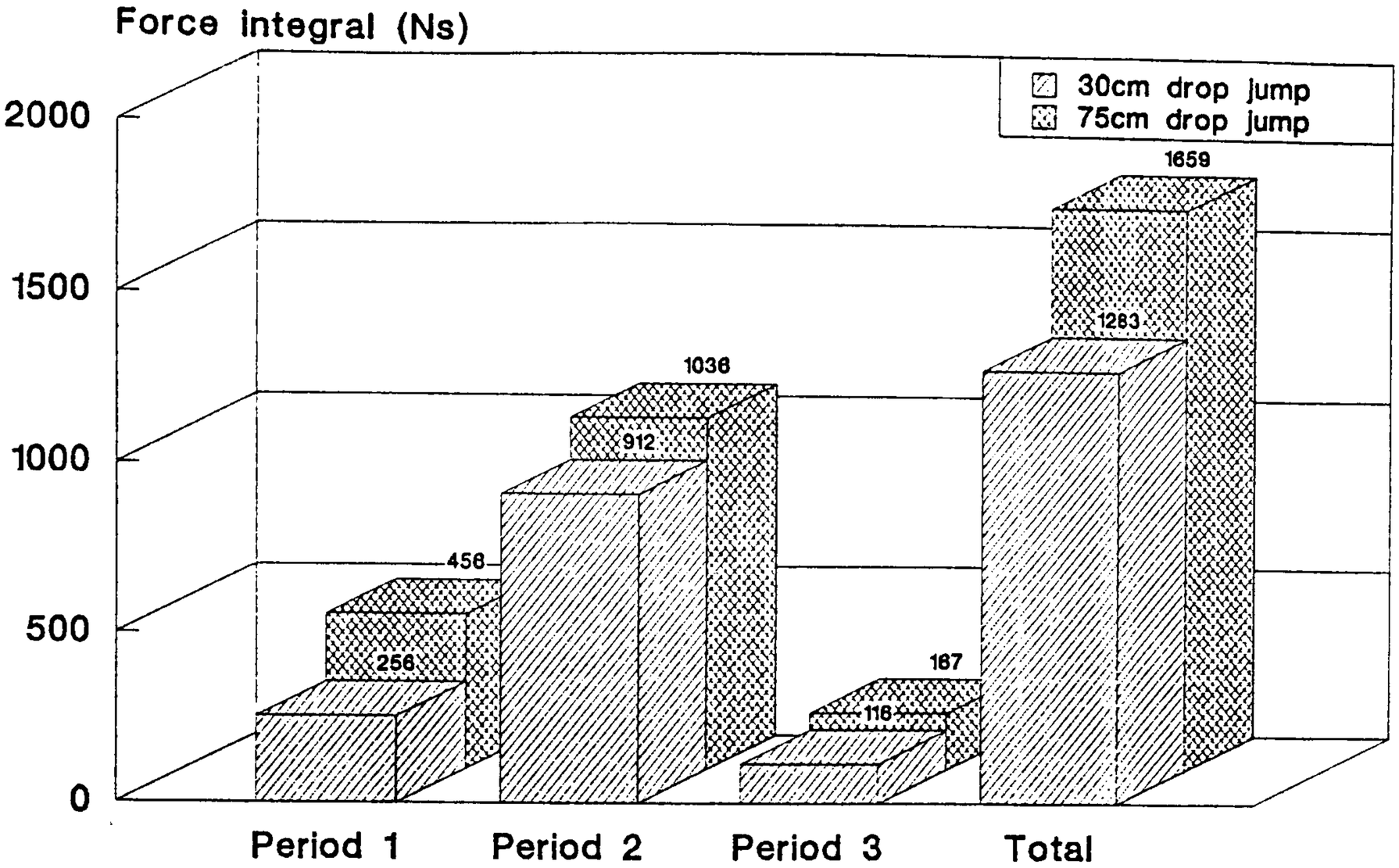


Figure 71. Mean spinal normal compression force integrals during various phases of the first land whilst performing 0.30 and 0.75 m drop jumps.

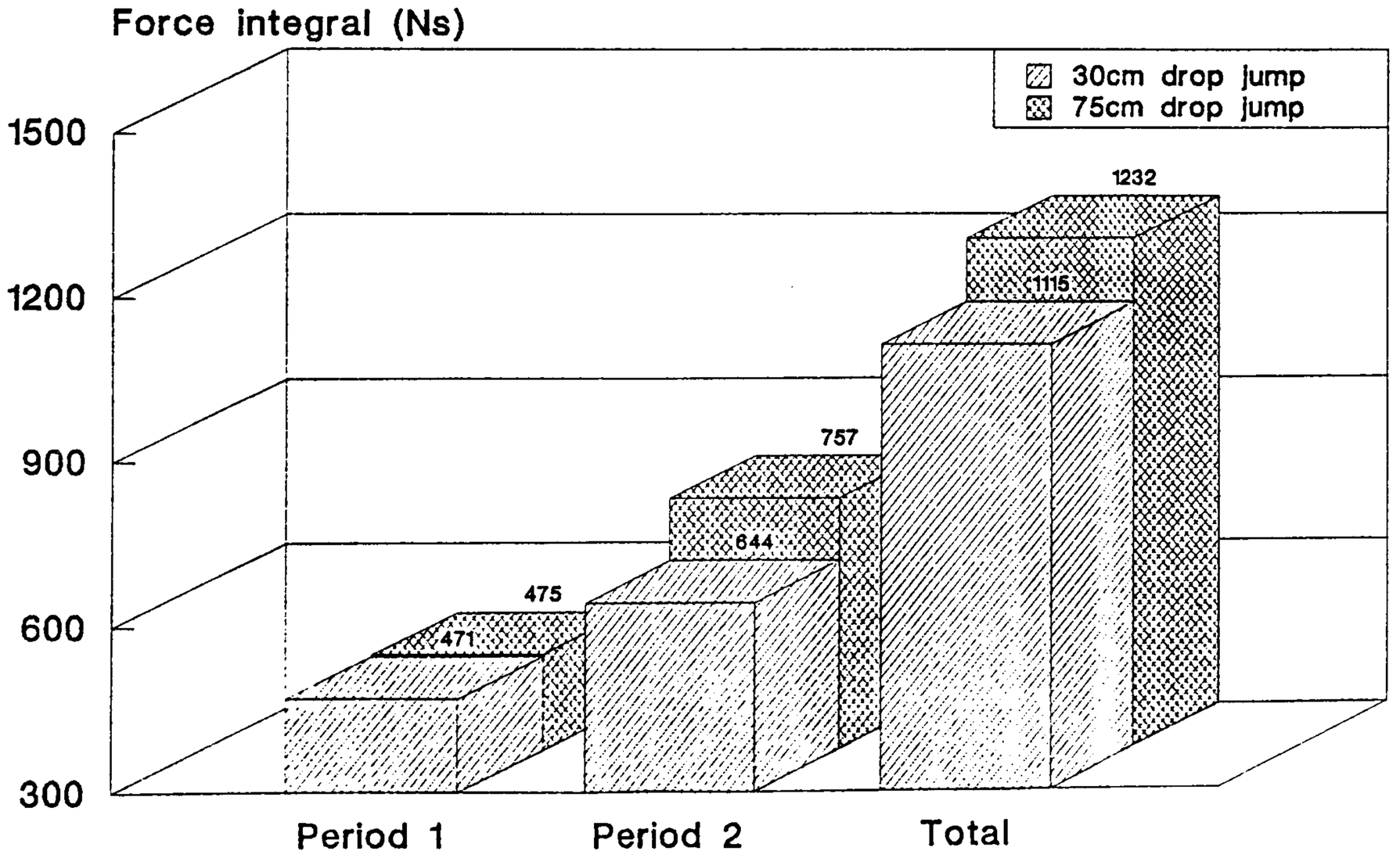


Figure 72. Mean spinal normal compression force integrals during two periods of the second land whilst performing 0.30 and 0.75 m drop jumps.

BW) for a model with a 70 mm ESMMA (Figure 64). A t-test revealed the difference to be significant ($p<0.01$). When a model with a ESMMA of 50 mm was used in the analysis this initial peak force increased in magnitude by 32 and 27% for the 0.30 and 0.75 m drop jumps, respectively ($p<0.01$). The quasi-static model resulted in an estimated increase in this peak force by approximately 12 and 8% for the 0.30 and 0.75 m drop jumps, respectively ($p<0.05$).

The two mean peaks in the compression curve recorded during the second period of the initial drop land were found to be greater during the 0.30 m drop jump, although the differences were not found to be significant ($p>0.05$). For both drop jumps, the first of these peaks (peak 2) exceeded the latter (peak 3) and coincided with maximum trunk flexion which appeared to precede maximum knee flexion (Tables 22 and 26). The model, implementing the 70 mm ESMMA, estimated the second peak to attain mean values of 5258 ± 685 N (7.3 ± 1.4 BW) and 4927 ± 922 N (6.7 ± 1.2 BW) for the 0.30 and 0.75 m drop jumps, respectively. Similarly, peak 3 mean values were 5044 ± 800 N (6.9 ± 1.1 BW) and 4652 ± 710 N (6.4 ± 1.1 BW) for the 0.30 and 0.75 m drop jumps, respectively. A model utilising a 50 mm ESMMA arm increased mean peak 2 and peak 3 forces by 29% for both drop jump heights ($p<0.01$). The quasi-static modelling approach caused a slight decrease (1%) in peak 2 ($p>0.05$) and a slight increase in peak 3 (6%), these differences being non-significant ($p>0.05$).

The final peak (peak 4) attained mean values less than any of those that had preceded it. The shift, in the majority of instances, to a net negative lumbar muscle moment and consequently, the model's implementation of the external obliques's muscle force produced peak compression forces of 2692 ± 713 N and 3030 ± 1514 N (4.2 ± 2.1 BW) for the 0.30 and 0.75 m drop jumps, respectively ($p>0.05$). At this phase of the land the exclusion of segmental accelerations appeared to have the biggest influence on the peak forces, a mean increase of 16% for both the 0.30 and 0.75 m drop jump ($p>0.05$).

As with the initial land, the estimated compression forces fluctuated rapidly during the initial 0.09 s of the second land, reaching sharp, high peaks. Following the forefoot and heel impact these curves were found to vary for each subject, mirroring the observations noted for

the ground reaction forces. Those who adopted a pronounced knee and trunk flexion displayed a more prolonged increase in compression loading.

During the second land, the biomechanical model, comprising of an ESMMA of 70 mm estimated mean compression forces for peak 1 to be 11257 ± 1765 N (15.3 ± 1.9 BW) and 11056 ± 2219 N (15 ± 2.6 BW) following the 0.30 and 0.75 m drop jumps, respectively ($p>0.05$). Similarly, the second peak (peak 2) showed no significant differences following drop jumps from each height ($p>0.05$); 3562 ± 1014 N (4.8 ± 1.2 BW) and 4216 ± 1444 N (5.7 ± 1.7 BW) for the 0.30 and 0.75 m drop jumps, respectively. Again, it would appear that on each occasion the second landings are alike irrespective of drop height.

As noted for the first land, altering the ESMMA from 70 mm to 50 mm had a pronounced effect on the magnitude of the peak forces. Mean increases of 26 and 28% for the first and second peaks, respectively, were found to occur. The quasi-static model increased mean peak 1 forces by 11%, whereas mean peak 2 forces decreased by 5% during the 0.30 m drop jump and increased by 9% during the 0.75 m drop jump.

The predicted mean shear forces (Figure 68) were found to be higher during the first land from the 0.75 m drop height when compared with the 0.30 m drop height. However, only the initial impact peak (peak 1) was found to differ significantly ($p<0.01$); -666 ± 114 N (-0.9 ± 0.2 BW) and -1198 ± 198 N (-1.6 ± 0.2 BW) for the 0.30 and 0.75 m drop jumps, respectively. For the 0.75 m drop jump this represented the largest shear force experienced throughout the land. During the 0.30 m drop jump the second peak was found to achieve a maximum value, -926 ± 125 N (-1.3 ± 0.2 BW). All peak shear forces acted in a posterior direction.

8.9.5.2 Integrated Spinal Compression Forces

Integration of each compression force curve was thought to provide a better indication of the overall spinal loading imposed by each drop jump. Figures 71 and 72, along with Tables 34 and 35, provide information on the force integrals (Ns) for each land and each drop height. During the first land the total force integral was found to be significantly higher ($p<0.01$) when landings occurred from a height

of 0.75 m; mean values of 1283 ± 242 and 1659 ± 329 Ns occurred for the 0.30 and 0.75 m drop jumps, respectively.

Periods 1 and 2 were found to be stages of the land where significant differences were evident ($p<0.01$). Mean force integrals of 256 ± 59 and 456 ± 59 Ns were calculated during period 1 for the 0.30 and 0.75 m drop jumps, respectively. Similarly, period 2 produced integral values of 912 ± 263 and 1036 ± 252 Ns, respectively. The third period did not show any significant differences with respects to drop height.

A 50 mm muscle moment arm, compared to 70 mm, increased the force integrals for period 1 and 2 of the initial land, a mean increase of 19 and 27%, respectively. Consequently, the overall mean increase in the total force integral was found to be 23%. As period 3 represented a shift in the direction of the net muscle moment from involvement of the erector spinae to external obliques, no difference occurred during this period.

As one might expect, exclusion of segmental accelerations had the most noticeable effects on the force integrals during periods 1 and 3 of the initial land. On average, the quasi-static model increased values by 23 and 39%, respectively. Only a slight increase (2%) occurred during period 2.

Although, during the second land, compression force integrals were higher following the drop jumps from 0.75 m, neither individual periods or the overall integral showed any significant differences ($p>0.05$). Mean overall integrals were 1115 ± 341 and 1232 ± 377 Ns for the 0.30 and 0.75 m drop jumps, respectively. A 50 mm ESMMA increased this estimated overall force integral by an average of 22%, whereas exclusion of segmental accelerations increased it by only 12.5%.

8.9.6 EMG Analysis

Figures 73 and 74 provide typical examples of the traces obtained from the chart recorder of the Polygraph during two drop jumps performed from a height of 0.30 and 0.75 m, respectively. The three upper most traces are EMG signals recorded from the three major muscle groups,

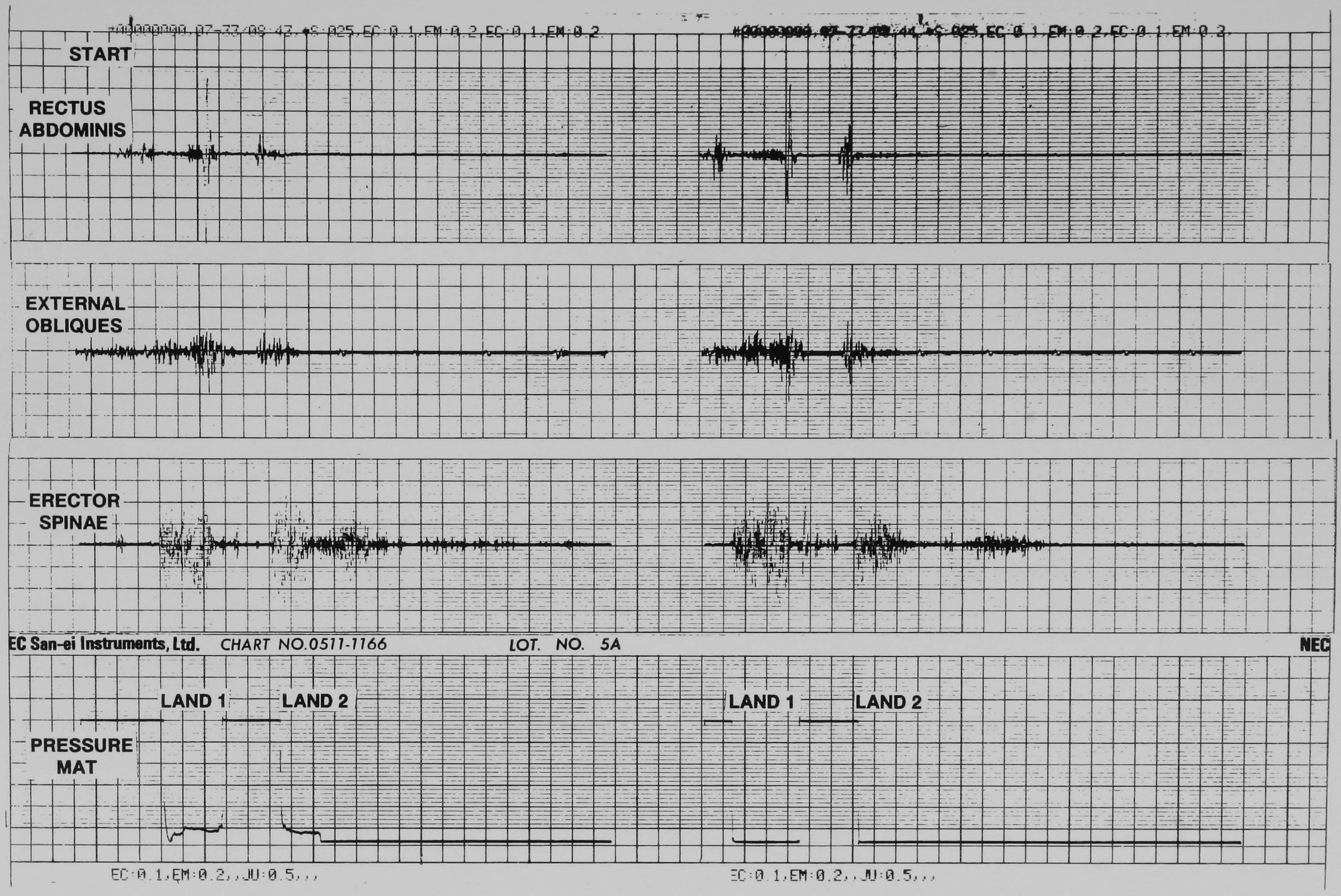


Figure 73. EMG and pressure mat signals during two 0.30 m drop jumps.

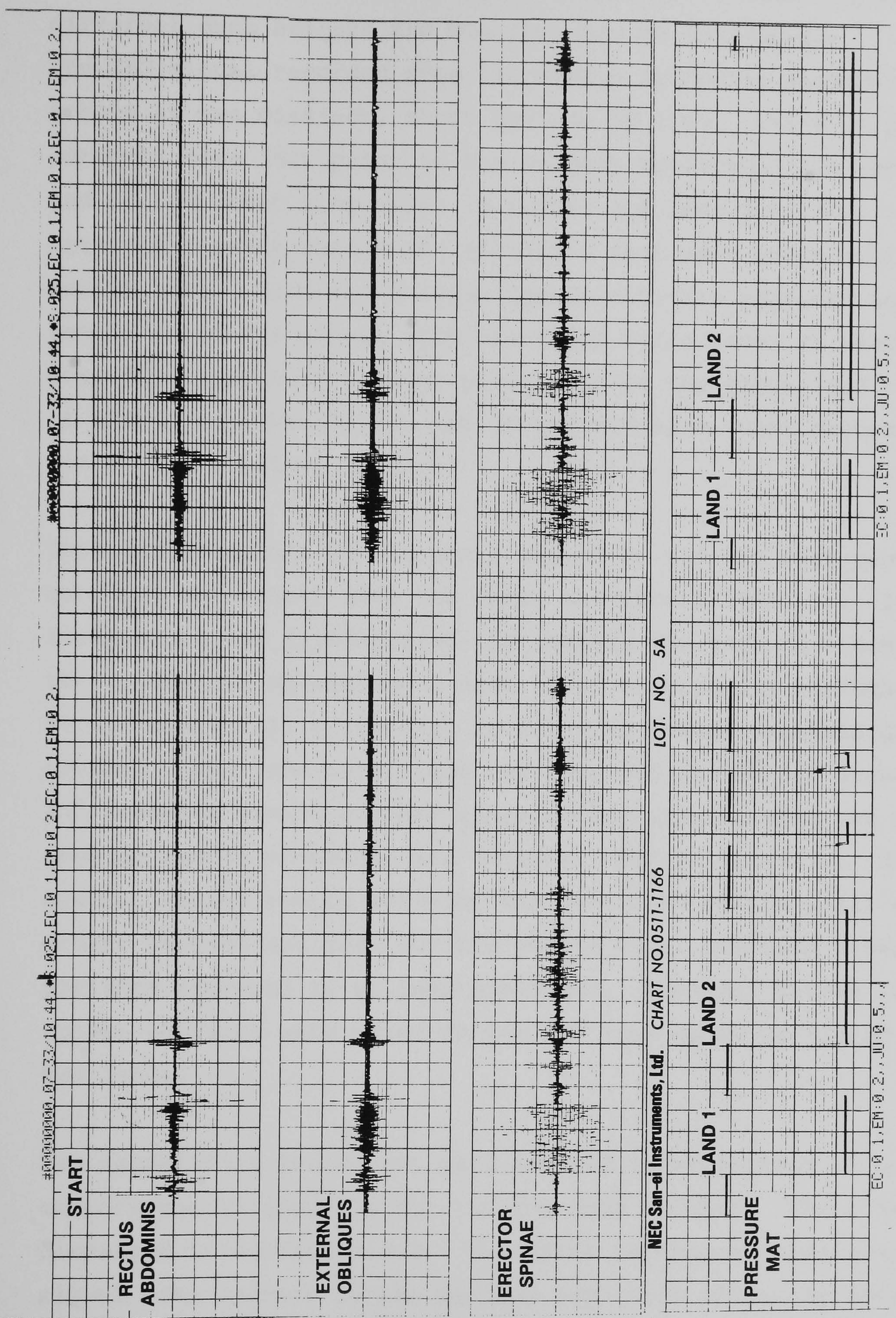


Figure 74. EMG and pressure mat signals during two 0.75 m drop jumps.

whereas the bottom trace is that obtained from the pressure mat, showing the various ground contact phases. Clearly there are common characteristics inherent to each trace showing the involvement of each muscle group throughout the drop jump. For reasons outlined in the review (section 3.4.4.2) comparisons of the raw EMG signals for different drop heights are not possible at this stage.
The EMG signal recorded from the rectus abdominis showed it to be active at two distinct, short periods of time. The most pronounced activity came just prior to the subject breaking ground contact at the end of the first drop land. A second and less prominent signal occurred moments before the subject's feet made ground contact for the second time. External oblique muscle activity appeared to mirror the rectus abdominis signal with one major difference; it showed more activity throughout a large proportion of the first land. The level of muscle involvement was not as pronounced as that occurring moments before the vertical jumps.

The erector spinae muscle group showed by far the most prominent activity throughout both the first and second landings. Its involvement was initiated at the instant the feet made contact with the ground and maintained a high level of activity throughout the first land, until a fraction of a second before breaking ground contact. The time separating the drop off in erector spinae activity and breaking ground contact appeared to coincide with the momentary peak in rectus abdominis and external oblique activity. Similarly, erector spinae muscle activity was initiated at the point at which ground contact was made for the second time. On this occasion its presence appeared to be less prolonged. Periodic bursts in erector spinae activity were noted during the time spent in flight between the two lands.

Two normalising procedures were adopted in attempting to quantify and compare muscle activity for the two drop heights. These have been discussed in section 8.8.2. The first technique normalised each signal with respect to the maximum amplitude occurring during the activity (dynamic amplitude) and the time duration of each land. A second procedure normalised maximum amplitudes within each signal against maximum amplitude during a controlled isometric contraction.

The former of these two procedures involved a hand drawn and digitised envelope around the positive amplitudes within the signal and was, therefore, subject to error. However, to assess the reliability of this technique in predicting the enclosed area, the whole procedure was repeated on two separate days. The RMS error associated with predicting erector spinae, external oblique and rectus abdominis muscle activity was found to be 2.3, 1.8 and 2.2%, respectively.

Figures 75 and 76 show the results of mean values calculated for each muscle group and for each drop height, using the first of these techniques. During the first land percentage values were found to be higher for each muscle group when jumps had been performed from the lower height, although only the external oblique muscle group showed a significant difference ($p<0.01$); 39.5 ± 7.8 and 29.1 ± 9.9% for the 0.30 and 0.75 m drop jumps, respectively. Muscle activity during the second land again showed the erector spinae muscle group to dominate, no significant differences being observed following the drop jumps from different heights ($p>0.05$).

When the second normalising procedure was adopted and mean peak amplitudes during the activity were compared (Figure 77), a significant difference between drop heights was found for the erector spinae muscle group ($p<0.05$), 110.2 ± 34.1% and 140.7 ± 40.6% for the 0.30 and 0.75 m drop heights, respectively. The fact that these percentage values were greater than 100% indicates that the maximum EMG amplitudes found during the activity were higher than those recorded during the isometric contractions. This was not the case for the other two muscle groups which produced similar peaks of intensity for each drop height. A similar analysis was not performed on the second land as peaks were not always evident and tended to be considerably less than those during the first land.

8.9.7 Changes in Stature

In accordance with previous experiments, measurements of stature were recorded at four periods in time prior to and after the two drop jumping regimens. The alterations in stature recorded for each subject are given in Table 36; overall mean stature changes throughout the experiment are depicted in Figure 78.

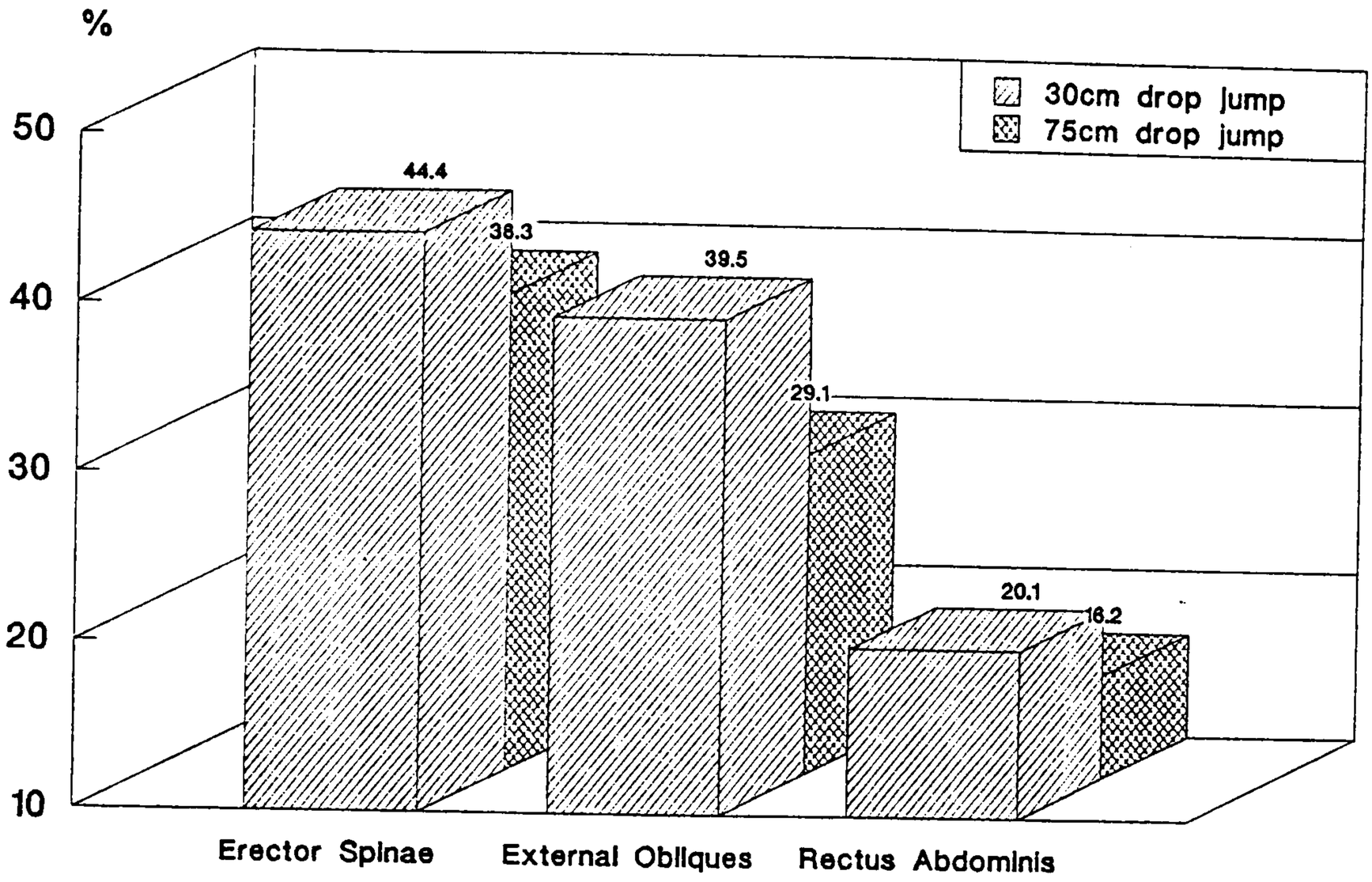


Figure 75. Normalised EMG activity with respects to the maximum dynamic amplitude and time duration for the three major muscle during the first land.

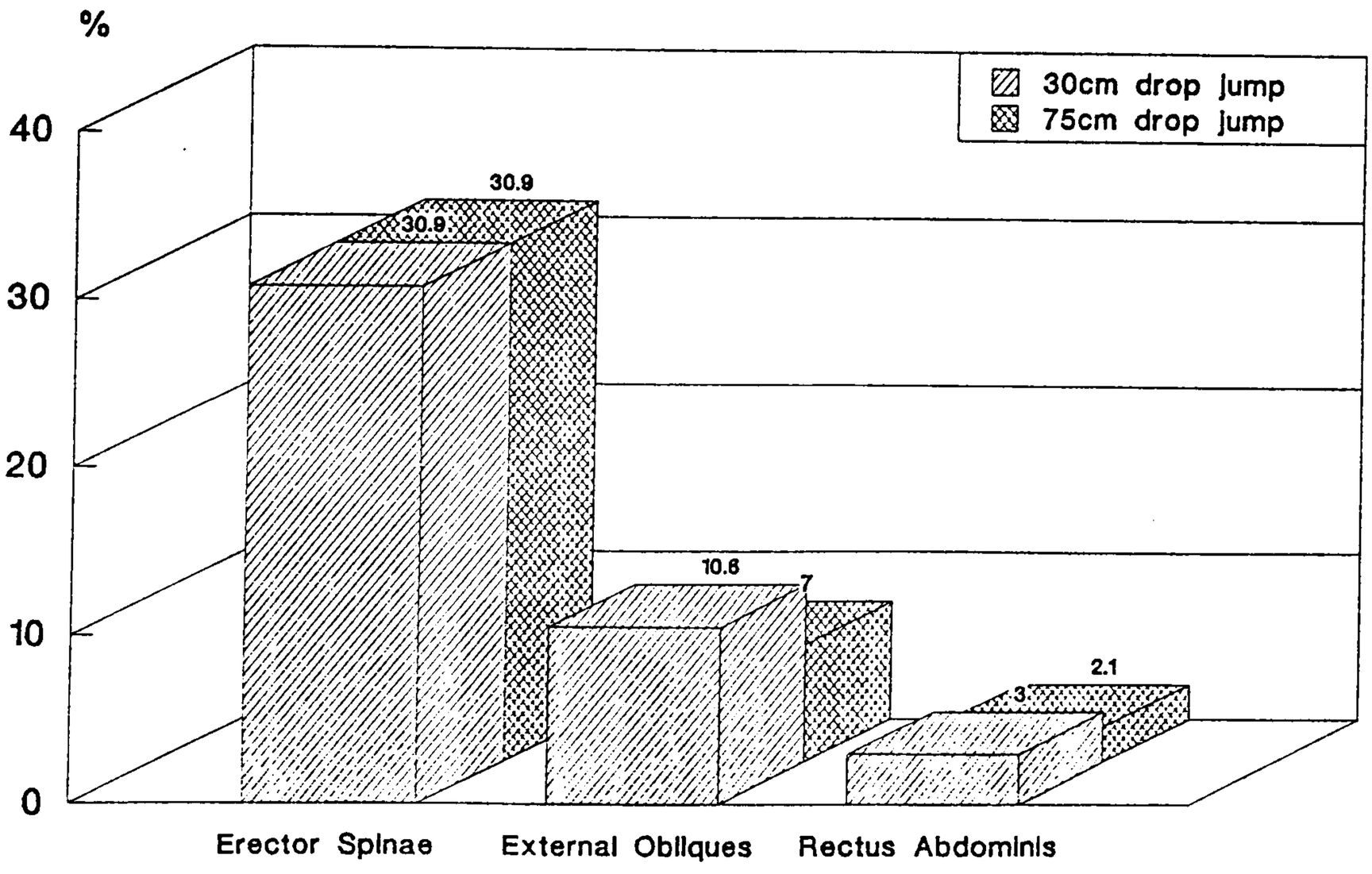


Figure 76. Normalised EMG activity with respects to the maximum dynamic amplitude and time duration for the three major muscle during the second land.

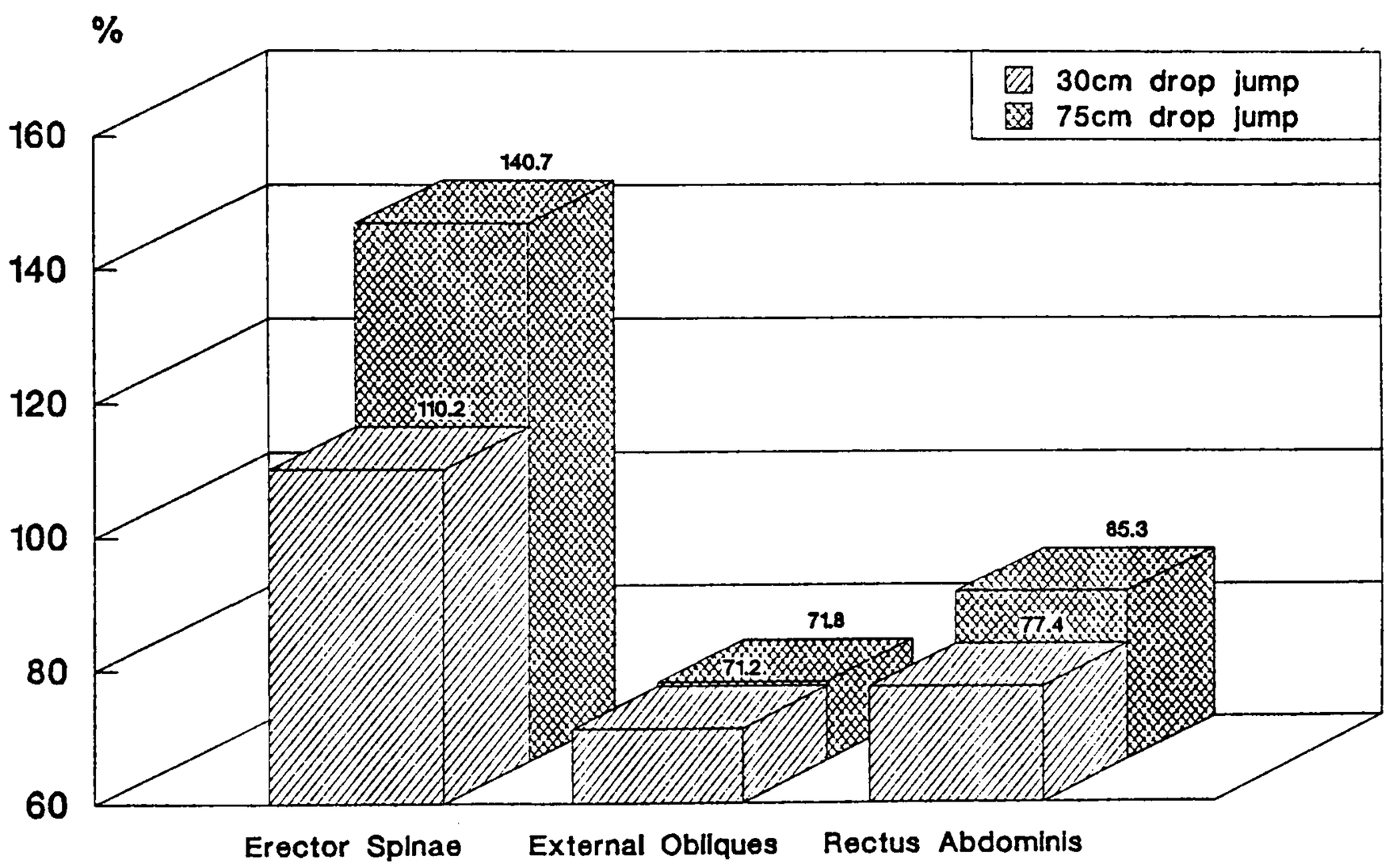


Figure 77. Maximum EMG amplitude normalised against the maximum isometric amplitude for the three major muscle groups during the first land.

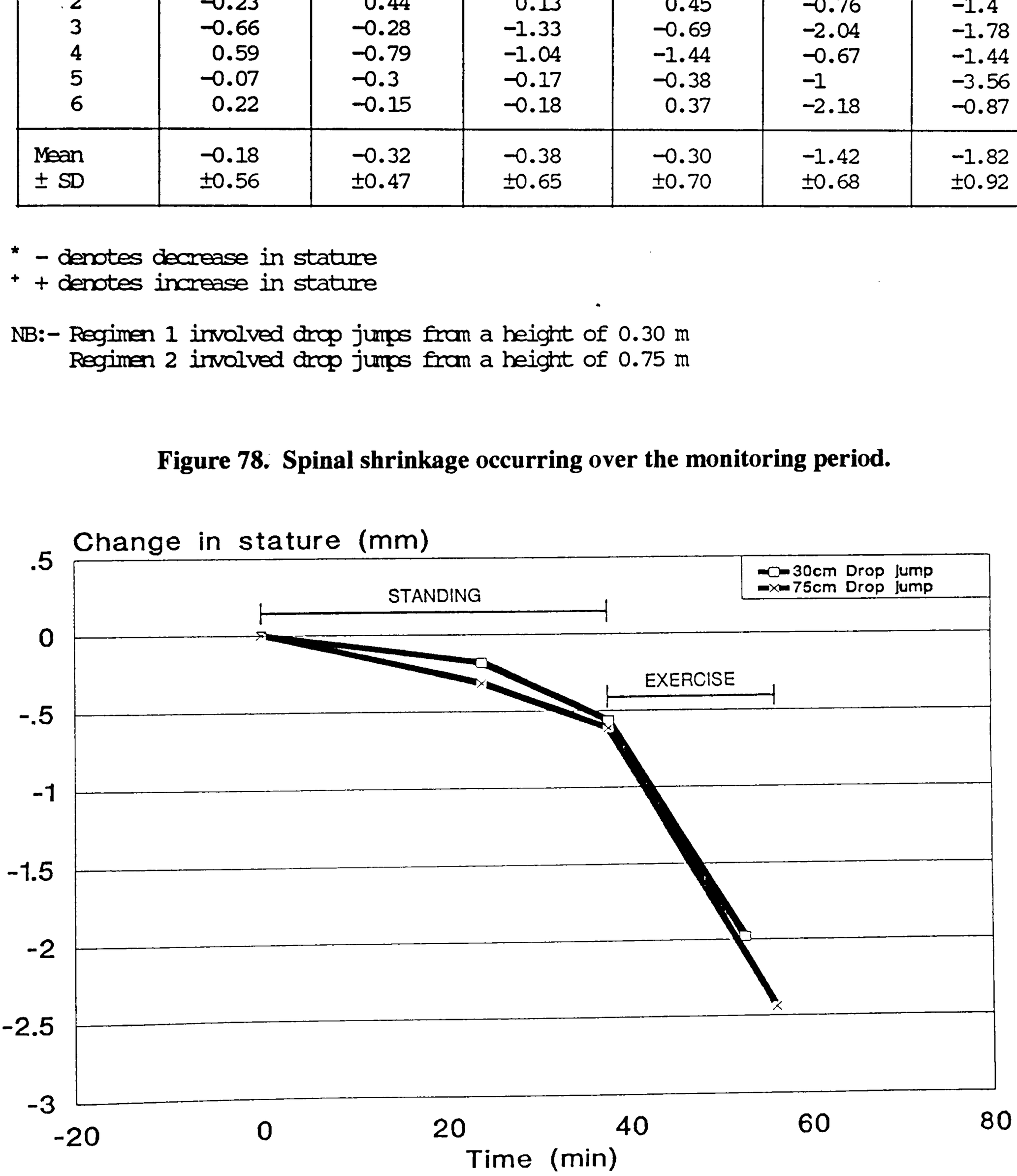
Table 36. Changes in stature (mm) recorded during the execution of two regimens composing of 25 drop jumps from a height of 0.30 and 0.75 m.

Subjects	20 min standing		10 min standing		Drop jump regimen	
	Regimen 1	Regimen 2	Regimen 1	Regimen 2	Regimen 1	Regimen 2
1	-0.95*	-0.83	0.3+	-0.09	-1.85	-1.9
2	-0.23	0.44	0.13	0.45	-0.76	-1.4
3	-0.66	-0.28	-1.33	-0.69	-2.04	-1.78
4	0.59	-0.79	-1.04	-1.44	-0.67	-1.44
5	-0.07	-0.3	-0.17	-0.38	-1	-3.56
6	0.22	-0.15	-0.18	0.37	-2.18	-0.87
Mean ± SD	-0.18 ±0.56	-0.32 ±0.47	-0.38 ±0.65	-0.30 ±0.70	-1.42 ±0.68	-1.82 ±0.92

* - denotes decrease in stature
\+ + denotes increase in stature

NB:- Regimen 1 involved drop jumps from a height of 0.30 m
Regimen 2 involved drop jumps from a height of 0.75 m

Figure 78. Spinal shrinkage occurring over the monitoring period.



Each activity period during the experiment was found to cause a mean decrease in stature. The initial 20 min standing produced shrinkage of 0.18 ± 0.56 and 0.32 ± 0.47 mm on the occasions prior to the 0.30 and 0.75 m drop jump regimens, respectively ($p>0.05$). Similarly, a further 10 min standing caused shrinkage of 0.38 ± 0.65 and 0.30 ± 0.70 mm, respectively ($p>0.05$).

The 0.75 m drop jump regimen produced a greater overall loss of stature when compared with the 0.30 m regimen; however, this difference of 0.4 mm was not found to be significant ($p>0.05$). It was also noted that the time taken to complete the exercise and stature measurements following the 0.75 m drop jump regimen was on average 3.47 min longer. This probably reflected the slightly further travelling distance required between drop jumps and the slightly longer time required to attain accurate stature measures post-exercise, a reflection of the more intense physical nature of the exercise.

8.9.8 Comparisons Between Approaches

The results obtained from the three analytical approaches, - the biomechanical modelling, EMG and study of stature changes - were all subjected to a cross correlation using Pearson Product Moment Correlation. Probability values of less than 0.05 were deemed significant.

During the first land spinal shrinkage was found to correlate with the third peak of the spinal compression curve ($r=0.94$, $p<0.01$) and the area enclosed under the second period of the compression force curve ($r=0.89$, $p<0.01$), for drop jumps from 0.75 m. Shrinkage showed no correlation with any of the measures of muscle activity at this height. Similarly, measures obtained for the EMG and biomechanical analysis showed no significant correlations. For the 0.30 m drop jumps, no significant correlations were found for any of the measures derived from the three separate approaches. No trends were present when each approach was compared for the second land.

Included in the correlations were the body segment angles measured at various phases throughout the lands. Minimum trunk angle featured strongly, being inversely correlated with the total and second period of the integrated area under the compression force-time curves for

both drop heights and each land ($r = 0.73 - 0.91$, $p<0.05$). In addition, this same measure was significantly correlated with both contact and flight times, the relationships being inverse for both drop heights and each land ($r = 0.84 - 0.95$, $p<0.01$).

8.10 DISCUSSION

8.10.1 Kinematic Parameters and Drop Jump Height

The nature of the experimental set-up dictated that subjects repeated drop jumps from the same height under slight differing laboratory conditions. This provided the opportunity to assess not only a number of kinematic parameters as a function of drop height but also the repeatability of the kinematic parameters for jumps performed at the same height, but on different days.

Some 22 kinematic parameters during the landing phase of each drop jump, plus the initial contact and flight times for each drop jump were compared. Repeatability of these measures for each drop height was good with only a significant difference occurring for final and initial trunk angle during the first and second lands of the 0.30 m drop jumps, respectively. It was felt such differences were more a reflection of the method of trunk angle detection than of individual variations.

When these same parameters were compared for each drop height during the first drop land, several differences were apparent. Not unexpectedly the higher drop height significantly increased ground contact time by 0.064 s. As a consequence of the greater drop height subjects had been found to decrease both knee and trunk angles significantly in an attempt to cushion the higher ground reaction forces. On both occasions the time to reach maximum knee and trunk flexion was greater but only the latter showed a significant difference ($p<0.01$). For both lands, minimum trunk flexion preceded minimum knee flexion.

It would seem that the 0.75 m drop height did increase the angular velocity of both ankle dorsi-flexion and knee flexion, although only the former showed a significant difference. Thus, increases in these angular velocities should imply that higher eccentric velocities of knee extensors and plantar flexors are attained and the resulting

force output enhanced. The similarity in flight times for each drop height would imply that vertical jump performance was not affected by the higher drop height. In addition, mean maximum angular velocities for ankle and knee extension were alike as well as the mean maximum ankle and knee moments during the push-off phase. Thus, the mechanical output of muscle appeared to be unaffected by drop height within the range of heights in this experiment, findings in line with those proposed by Bobbert et al. (1987b) but contrary to findings elsewhere (Asmussen and Bonde-Petersen, 1974).

It is accepted that the latter assumption should be viewed with some caution as this study did not take a direct measure of net power output at either the ankle or knee joints. The reasoning is based solely on the non-significant variation in mean peak angular velocities and net joint moments about the ankle and knee joints prior to take-off, for the two drop heights.

Although, Fahmi (1986) has questioned the use of flight time as an accurate predictor of actual vertical height attained during drop jumping, this should not have influenced the assumption that jumping performance had been comparable for each drop height. Flight time was only used to compare drop jump performance and consequently, the systematic error will have been constant across subjects.

The lack of a measurable increase in jumping performance from the higher of the two drop heights may raise questions about the validity of this investigation, from the point of view of subject motivation. Every attempt was made to verbally motivate subjects throughout the course of the exercise period. Fahmi (1986), who observed similar trends in performance, provided a physical form of motivation in the shape of a rubber ball above the subject's head which they attempted to contact.

The fact that drop height did not have an effect on jumping performance may be explained by the enhanced knee flexion occurring on the initial land. The literature suggests that the shortness of the latency period within the stretch-shortening muscle action is a critical factor in the muscle's ability to utilise stored elastic

energy (Bosco et al., 1981). Boase (1983) proposed that an over-deeping of the knee joint on landing will lead to an over-stretching of the muscle fibres and result in muscle spindles being stimulated asynchronously, leading to reduced power efficiency. It would also seem logical to assume that the more knee flexion there is the less biomechanically efficient the movement becomes.

8.10.2 Forces Encountered During the Impact Landing Phases of the Drop Jump

The three approaches adopted - a biomechanical modelling approach, EMG and measurements of stature change - all sought to examine the nature and magnitude of forces transmitted to the spine during drop jumping. Consideration was given to the initial and second landing phases of the drop jump and to the influence of a 0.30 and 0.75 m drop height.

The biomechanical modelling approach had necessitated measurements of ground reaction forces. Both the vertical and horizontal forces displayed common characteristics with the exception of one subject, during the 0.30 m drop jump. The first ground reaction force curves possessed sharp peaks within approximately the first 0.102 s of landing. In the majority of instances, the largest of these peaks coincided with the heels striking the force platform. Bobbert et al. (1987b) had recorded a similar effect for 0.60 m drop jumps but not for 0.20 or 0.40 m drop jumps. Subjects in the present study were relatively inexperienced in drop jumping, unlike those in the study by Bobbert et al. (1987b), all of which had previous experience with drop jumps as a training mode. This may well explain why such peaks were still associated with the 0.30 m drop jumps found in this study, although one subject did appear to control the heel impact.

The largest peak vertical and anterior-posterior force recorded by the force platform occurred during the 0.75 m drop jumps, the difference being significant. Mean peak values for 0.75 m drop jumps were found to be 2336 N. Bobbert et al. (1987b) measured values of 4496 N for the 0.60 m drop jumps but subjects had been required to land with both feet on the force plate. Thus, it would seem that peak forces reported here attained comparable magnitudes. They did report that this peak was reached 52 ms after landing, approximately 50 ms earlier than in this study. This may again reflect the experience of the

subjects and the posture adopted at landing. Once heel impact had occurred, peak vertical forces during the push-off phase were of a similar value for both drop heights.

During the second landing, ground reaction force curves were again characterised by sharp peaks, representative of forefoot and heel impact. Interestingly, for both drop jump heights the maximum amplitudes attained similar values to those recorded during the first land of the 0.75 m drop jump. Thus, it would appear that the second land may be as important in terms of its potential for incurring injury and should be given the necessary consideration.

A potential source of error in ground reaction force measures could have emerged as a consequence of the experimental requirements. Subjects were required to land with only one foot on the force platform, the biomechanical model assuming that the land was symmetrical and forces distributed equally through both legs. In practice, jumpers may favour one particular leg as it is stronger than the other and therefore, offset weight over the preferred leg. The jumps selected for analysis did not, on visual inspection, show marked trends towards this, nor was there any evidence of large lateral forces.

Analysis of the spinal compression-force curves sought to examine two basic features, instantaneous peak loads and cumulative loading by integrating the area under each compression curve. These were carried out at selected phases of both lands as force-curves appeared to display similar characteristics. The first land identified three distinct phases - initial forefoot and heel impact, knee flexion and extension, and a final phase where there was a shift in the net muscle moment acting about the trunk. Four peaks were identified, one in the initial and latter phases and two in the middle phase, prior to and following maximum knee flexion. The second landing considered two periods and two corresponding peaks, initial forefoot and heel impact, and the period thereafter.

Biomechanical modelling does possess the advantage of being able to alter certain parameters within the model to investigate their

influence on the ultimate predicted loading. Consideration was given to both the length of the ESMMA and inertial forces. An ESMMA of 70 mm was considered to be the more realistic; however, the more commonly used 50 mm moment arm was included for comparison. It would seem from more recent work using both CT scans and MRI (Table 3) that the moment arm may lie somewhere between the two. It would of course also be dependent upon the trunk angle.

Initial impact peaks for both landings appear to be extremely high and raise questions about the predictive power of the model. Peak forces during the first land attained values of 13537 and 7595 N for the 0.75 and 0.30 m drop jumps, respectively. These become more disturbing given that the literature reports *in-vitro* failure loads for motion segments of between 3 to 12 kN (Brinckmann et al. 1988). Guidelines provided by NIOSH for sagittal plane lifting consider a spinal compression force of 3425 N to be potentially hazardous to some workers, whereas forces approaching 6361 N are regarded as high risk to most workers. The maximum spinal compression force found in the literature using biomechanical models (Table 2) was 10000 N by Jäger and co-workers (1989). They investigated 50 kg lifts involving a 'humpback' technique.

This apparent paradox may well be explained in two ways. The first takes into consideration the time duration of these initial impact peaks. These peaks were only found to be present for an extremely short time and it is conceivable that under such circumstances this magnitude of force could be tolerated by the tissue. Perey (1957) has shown that an inverse relationship exists between the strength of the spine and the duration of the applied load. Further, Hodgson et al. (1963) identified that failure of spinal segments is related to the peak acceleration and the rise-time or jerk of the applied load. Clearly, peak forces alone provide an insufficient measure of the likelihood of structural failure during highly dynamic activities. These forces should be considered in conjunction with their corresponding rise-times. Unfortunately, the present investigation could not ascertain with certainty the rise-time of spinal compression forces due to restrictions emanating from the hardware. This limited the sampling of force data to a frequency of 100 Hz. Future

investigations must consider rise-times more closely, employing much higher sampling rates.

The second explanation and more plausible lies with the potential shortcomings of the modelling approach. The biomechanical model considered body segments to be composed of rigid-link segments. Thus, forces which were imposed at one end of the segment were transmitted directly and unaltered to the other. While body segments may be thought of as rigid, they are in fact composed of a number of deformable soft tissues with varying viscoelastic properties (Nordin and Frankel, 1989). Consequently, the forces that are transmitted through the lower limbs are likely to been changed by the shock absorbing properties of the soft tissues. Studies by both Smeathers (1989) and McGill et al. (1989) would seem to support this notion. McGill et al. (1989) concluded from their studies on the transmission of force through the spine that the rise-time of the force is important, the longer the rise-time the closer the body conforms to a rigid system. It may be that rigid-link and cantilever modelling approaches are inappropriate methods by which to mechanically represent the spine. Aspden (1988) has proposed an arch model which he stated gives rise to potentially lower compression forces within the vertebral column. Further work is needed to assess the validity of such a model.

Once the initial forefoot and heel strike had taken place the peak spinal compression forces occurring pre- and post-minimum knee flexion attained values comparable with those found for other activities using biomechanical modelling techniques. In this study these forces ranged between 6078 and 6791 N, forces found during certain lifting techniques of 15 kg and upwards (Table 2). This provides partial validation of the model's ability to predict post-impact forces and also implies that this form of activity induces loads that can be considered dangerous, given present guidelines (NIOSH, 1981). The 0.30 m drop jumps produced the higher peak forces at this stage of the land, although not significantly different, suggesting that both drop jump heights impose a similar risk. The second peak exceeded the third peak and was related to trunk flexion. A significant correlation between compression forces and trunk angle ($p<0.05$)

suggests that the amount of trunk flexion during the activity is an important factor governing spinal loading.

A shift in the direction of the net muscle moment towards the latter stages of the first land was consistent across subjects. It would seem to reflect the attainment of an upright posture and a shift in the net ground reaction force vector behind the L5/S1 joint complex.

Peak forces during the second land appeared to mirror the first land during the initial impact. Again, the model predicted extremely high forces of over 11 kN. Drop height did not seem to influence the magnitude of these forces, re-emphasising the similarity of drop jump performance.

Peak L5/S1 shear forces were found to be higher throughout the 0.75 m drop jump and reached a magnitude of 926 and 1193 N for the 0.30 and 0.75 m initial drop lands, respectively. These values are within ranges predicted elsewhere in the literature (Table 2). The model's prediction of shear forces depends to a large extent on the trunk angle and this would seem to have greatly influenced these findings. The model's assumption that muscle actions are parallel to the angle of the trunk eliminates any influence they might have on the prediction of shear forces. In reality this would appear unlikely and serves to emphasise a major limitation of the model in calculating such forces. The model also serves to ignore the potential action of any ligaments which would induce a restraining action against the applied load.

The area under the spinal compression force curve provided a measure of the cumulative load acting on the spine, a technique which has been used elsewhere (Leskinen et al., 1983b). Again, during the first land the model predicted that this cumulative stress was significantly greater following the 0.75 m drop jump, 1283 and 1659 Ns for the 0.30 and 0.75 m drop jumps, respectively. The second land again showed no significant difference with regard to this measure. These findings, in conjunction with those found for the peak forces, lead to an acceptance of hypothesis 10; the biomechanical model would predict higher L5/S1 spinal compression and shear forces for 0.75 m drop jumps than for 0.30 m drop jumps.

The influence of both inertial forces and the ESMMA on compression forces was studied. The latter was found to have the most profound effect on spinal force prediction. On average, total integrated compression force was 23% higher as a result of the shorter 50 mm ESMMA. For the same measure, a static model over-estimated forces by approximately 11%. Clearly the ESMMA is a critical variable when biomechanical modelling and needs to be given careful consideration. The findings of the static analysis are opposite to those reported for lifting, McGill and Norman (1985) reporting an increase of 19% on peak L4/L5 as a result of dynamic forces. Considering the way in which these forces would act during drop jumping, this finding would not seem surprising. It would also suggest that dynamic forces have a significant influence on the compressive load and should be included during any modelling approach.

Monitoring of muscle activity from the three major muscle groups served not only to assess their contribution towards spinal loading but also helped to validate the biomechanical model. The reliability of EMG signals, particular surface recorded signals, is subject to debate. Quantitative analysis during the present study was further hampered by the nature of the stored signal.

Overall, the raw EMG signals would on the one hand seem to support findings from the biomechanical model. However, they would also seem to identify potential shortcomings. A pronounced increase in rectus abdominis and external oblique muscle activity, just prior to the vertical jump, would support the shift in the direction of the net lumbar muscle moment, predicted by the model, at a corresponding phase of the jump. A limitation in the model is clearly its inability to cater for antagonist muscle activity. Such muscle activity was present throughout the majority of the first land, evidence coming from prolonged activity of the external obliques. This would suggest that the model may have under-estimated the overall compression loading. However, the model also failed to take account of any beneficial effects changes in intra-abdominal pressure might have had.

No consideration was given to spinal loading during the periods prior to the two landing phases of the drop jump. It is unlikely that a great deal took place during the initial drop, but given the kinetic energy within the system between the first and second land, the flight stage of the vertical jump may have contributed markedly to loading. Short bursts of erector spinae muscle activity were common features within the EMG traces during this phase of the jump. Presumably this reflected subjects' attempts to counteract rotations in flight and prevent over-balancing on landing.

Normalising and quantifying the EMG signals proved difficult. Two approaches were adopted, the first seeking to normalise the signal with respect to both a peak 'dynamic' amplitude and time, the second normalising with respect to a maximum isometric peak amplitude. The former approach implied that less muscle activity had been present for each of the muscle groups following the higher drop land. This apparent contradiction to the increased loading predicted by the model was felt to emanate from limitations inherent in the normalising procedure. The second normalisation procedure identified that peaks about which normalisation had been conducted for the first technique were higher during the 0.75 m drop jumps. Consequently, as these amplitudes had not been sustained throughout the land it resulted in a much lower overall estimation of muscle activity. It would seem that these two analysing techniques would only permit the assertions that during the first land peak erector muscle activity for the 0.75 m drop jump attained higher values, overall external oblique activity was greater during the 0.30 m drop jump, rectus abdominis activity was similar for both drop heights. During the second land there was no evidence to suggest that drop height had an effect on muscle activity recorded from each muscle group.

When maximum peak amplitudes during the first land were normalised against maximum isometric peak amplitudes, a significant difference between drop heights only occurred for the erector spinae, the 0.75 m drop jumps producing higher percentage values. These findings prevent an acceptance of hypothesis 11. They also imply that the major role of the abdominal muscles, particularly the external obliques, is one of trunk stabilisation.

The drop jumping regimens had been found to increase the rates of stature loss; however, there was not found to be any significant difference in the magnitudes of shrinkage for the two drop heights. Thus, the null hypothesis (hypothesis 12) was accepted. The pattern of shrinkage found here was similar to that reported in a previous experiment (section 7) involving an identical number of drop jumps from a height of 1 m. There was a slight difference in jumping requirements; a maximum vertical jump was required in this experiment as opposed to a jump over a 0.5 m hurdle in the previous experiment. The increased physical exertion of a maximum jump may have had a more pronounced effect on spinal loading than drop height and may explain the slightly increased shrinkage values - 1.82 and 1.74 mm for the 0.75 and 1 m drop jumps, respectively.

The biomechanical modelling approach and study of EMG data were taken from a small selected sample of drop jumps from within the regimen, whereas shrinkage was the cumulative effects of the entire jumping regimen. In addition, subjects were not regarded as proficient in these types of activities. It is felt that these factors contributed significantly towards the lack of difference in shrinkage for drop height and the failure for losses in stature to correlate with the other measuring procedures. The biomechanical modelling approach has distinct advantages when assessing discrete periods of time within any activity, whereas measurement of stature change provides a much longer term assessment of cumulative loading. For example, the biomechanical model identified that initial heel impact and maximum trunk flexion during drop landing lead to forces that could be regarded as hazardous to the spine. Ensuring proper cushioning of these forces, through good shock absorbers, and encouraging a more upright trunk posture throughout the land would be recommendations stemming from this study. The rapid loss in stature over a short period of time, as recorded by shrinkage, suggests that these training sessions should be interspersed with regular periods of spinal unloading to reduce the risk of injury.

8.11 SUMMARY OF FINDINGS

1) Kinematic parameters showed good repeatability for drop jumps performed from the same height.

2) Drop height increased the angular velocity of ankle dorsi-flexion and knee flexion but this was not found to increase net ankle or knee muscle moments, or ankle and knee angular velocities prior to take-off. Flight times, used as a measure of vertical jump performance, were not increased by drop jump height, further evidence to suggest that drop height did not increase performance.

3) The larger drop height caused greater knee and trunk flexion, which in turn increased the ground contact time during the first land. It is likely that, as a result, the efficiency of the stretch-shortening cycle was decreased.

4) Characteristic sharp peaks in ground reaction forces were evident during heel impact. During the second lands, impact peaks were found to reach values close to those which had been recorded during the first land of the 0.75 m drop jumps. Adequate footwear and floor matting should be used to cushion these forces.

5) A biomechanical model was developed which recorded normal compression and shear forces about the L5/S1 intervertebral joint complex. The model predicted extremely high peak spinal compression forces during heel impact which appeared to reflect shortcomings within the model. Peak forces at maximum trunk flexion during both drop jumps were found to reach values that were considered to be theoretically hazardous to the athlete, though dependent on the strain rate. Trunk angle was an important factor governing this peak force.

6) The modelling approach identified a significant increase in the total spinal stress when performing drop jumps from the higher drop height. Mean peak normal compression and shear forces were found to attain higher values when drop jumps were performed from 0.75 m as opposed to 0.30 m.

7) Use of EMG did show that the erector spinae muscle and the external obliques were active throughout the majority of the landing phases. A significant increase in the maximum amplitude of the erector spinae EMG signal was found during drop jumps from the increased height. Muscle activity for the external obliques and rectus abdominis were not found to increase with the increase in drop height.

8) The rate of stature loss was increased by an increase in drop height; however, this was not found to be significant

9) Spinal shrinkage did not consistently correlate with either biomechanically derived measures of loading or measurements of muscle activity. The lack of correlation may reflect the limited sampling restrictions imposed by the latter two techniques. For example, shrinkage measurement considered the effects of the entire regimen, whereas, the latter two approaches were restricted to two drop jumps.

9. SUMMARY OF THE WORK

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The primary purposes of this work were to:-

i) Develop and enhance the present system by which to record small changes in stature as a measure of load imposed on the human spine.

ii) Examine the effects of short term plyometric exercises on spinal loading.

iii) Assess spinal responses to pre- and post-exercise activities.

iv) Compare three *in-vivo* spinal load measuring techniques - biomechanical modelling, EMG and spinal shrinkage.

The specific aims and objectives were set out in section 2.

9.1 UPGRADING OF STADIOMETER

Although a pilot study was undertaken to examine the influence of a bounding regimen on spinal loading (section 4), its primary aim was to investigate the apparatus and procedure used at the time to record small changes in stature.

The work identified shortcomings with the stadiometer's calibration procedure and the method by which to store and analyse data. Consequently, a fully automated computer system was developed to enhance data capture and processing. The software package was designed to be 'user-friendly' with several multi-option menus, e.g. main menu, training and test data capture, training and test data display, stadiometer settings storage and display, calibration routine.

To ensure reliability between head disc displacements and digital recorded signals, a series of repeated trials were conducted. These compared the digital displacement signal recorded by the computer against the Mercer dial gauge readings on the head gantry (sensitive

to 0.01 mm). For ten trials the standard deviation of the recorded signal was found to be 0.1 mm or less in the middle 30 mm of the head disc displacement range.

9.2 THE EFFECTS OF SPINAL UNLOADING PRIOR TO EXERCISE

Section 5 details a study which examined the effectiveness of a spinal unloading procedure prior to exercise as a means to reduce the likelihood of incurring spinal injury during subsequent exercise. Based on the physical nature of bounding exercises it was hypothesised that:

Hypothesis 1: A short duration bounding regimen places a mechanical load on the spine and leads to an incurrence of spinal shrinkage.

The second hypothesis considered the physiological and mechanical conditions known to be present within more compressed discs.

Hypothesis 2: During a period of recovery, regains in stature are greater for those subjects who have undergone a greater initial shrinkage.

Based on the time duration of the mechanism's controlling disc height changes, it was hypothesized that:

Hypothesis 3: Spinal shrinkage resulting from a short duration bounding regimen is greater after a period of unloading.

Results from this experiment supported both hypotheses 1 and 3. Although a trend in the pattern of stature recovery was consistent with hypothesis 2, shortcomings within the experiment prevented its acceptance. The study implied that the benefits of short duration unloading procedures, prior to loading, have a limited value as a preventative method against back injury.

The work also examined the influence of unaccustomed bounding regimens on post-exercise muscle symptoms. Two hypotheses were put forward:

Hypothesis 4: Following a bounding regimen, reported muscle soreness is most prominent between 24 and 72 hours post-exercise.

Hypothesis 5: A bounding regimen repeated within a week of the first execution shows a reduced severity of muscle soreness ratings after the second performance.

Post-exercise muscle soreness rating questionnaires produced data to support both of these hypotheses.

9.3 LONGITUDINAL REPEATABILITY OF STATURE MEASUREMENTS AND SPINAL CHANGES FOLLOWING A WARM-UP ROUTINE

The purpose of this experiment (section 6) was two-fold:- to examine the reproducibility of spinal responses to various activities and determine whether a 'typical' 10 minute warm-up routine could significantly alter spinal loading or unloading. The former has implications regarding the reliability of the measuring procedure and the latter towards the implementations of such routines in an exercise programme. Two hypotheses were proposed:

Hypothesis 6: On separate days, mean fluctuations in stature for a specified regimen are repeatable.

Hypothesis 7: A short duration warm-up routine increases the normal rate of stature loss and consequently, spinal loading.

The changes in stature recorded during the two identical activity regimens (one week apart) were found not to differ.

In addition, shrinkage occurring during an activity regimen identical to that conducted during a previous experiment (section 5) provide further support to the reproducibility of the measuring technique. On the basis of these findings hypothesis 6 was accepted.

Although, a short duration (10 min) warm-up routine had increased the rate of stature loss when compared directly to a period of standing,

the difference was not found to be significant. Longer duration warm-up routines may present a greater problem to the athlete than an equivalent shorter routine, by causing losses in disc height that leave the disc susceptible to injury in the subsequent exercise period. It may be prudent to structure the warm-up routine in such a way that spinal recovery promoting exercises are interspersed between exercises that increase load on the spine.

9.4 DROP JUMPING AND POST-EXERCISE SPINAL UNLOADING

Section 7 details an experiment which explored whether a short duration drop jumping regimen would impose a similar loading effect to that which had been observed for bounding exercises. The hypothesis was:

Hypothesis 8: A short duration drop jumping regimen increases the rate of spinal shrinkage when compared to an equivalent period of standing.

Shrinkage did result from the drop jumping regimen and exceeded stature loss over a similar period of standing. Thus, hypothesis 8 was accepted. It appears that short duration drop jumping routines, like bounding exercises entailing a similar number of ground impacts, induce loading effects on the spine that lead to rapid losses in intervertebral disc height. Consequently, one type of exercise routine cannot be advocated in favour of the other for lessening the risks of incurring back injuries.

The work also sought to investigate the influence of gravity inversion as a means of quickly negating the effects of the loading post-exercise. The hypothesis stated:

Hypothesis 9: Gravity inversion quickly promotes spinal recovery following a period of exercise-induced loading.

The shrinkage resulting from the short duration drop jumping regimen was negated within approximately 7 min and therefore, led to hypothesis 9 being accepted. The study went on to consider spinal responses post-inversion and it was observed that stature appeared to

return quickly to its 'normal' diurnal curve, supporting findings reported elsewhere (Klingerstierna and Pope, 1987). Recommendations from this study were directed towards the regular use of unloading postures between brief bouts of exercise, within the single training session.

9.5 BIOMECHANICAL MODELLING OF SPINAL LOADS AND ITS COMPATIBILITY WITH SHRINKAGE DURING A DROP JUMPING REGIMEN

A simple five-link segment, dynamic, symmetrical, two-dimensional, sagittal plane biomechanical model was developed to investigate the load transmitted to the L5/S1 intervertebral disc during the landing phases of the drop jump. A suite of computer programs was developed to collect, combine and process relevant data. The model required three data sources, kinematic, anthropometric and ground reaction forces. The computer programs allowed each data set to be combined to provide information on kinematic parameters, joint reaction forces, muscle moments, L5/S1 compression and shear forces. Analysis routines within the final suite of programs enabled salient features within a specified time curve to be derived, as well as permit parameters inherent to the model to be varied or eliminated from the analysis, primarily the ESMMA and inertia forces.

Measurements of muscle activity from three major trunk muscle groups and the changes in stature recorded over the drop jumping regimen were compared with data derived from the model. Repeatability in performance was assessed by comparing salient kinematic parameters. Hypotheses were based upon the measuring procedures adopted and the relative spinal loading imposed by a drop height of 0.30 and 0.75 m.

Hypothesis 10: Drop jumping activities from a height of 0.75 m induce higher spinal compression and shear forces on the L5/S1 intervertebral disc, as measured using a biomechanical model, than those performed from a height of 0.30 m.

Hypothesis 11: When undertaking 0.75 m drop jump there is an increased muscle activity of the erector spinae, external obliques and rectus abdominis, above that recorded during 0.30 m drop jumps.

Hypothesis 12: Dropping from a height of 0.75 m during the execution of drop jumps causes greater spinal shrinkage than when performing the same jumps from a height of 0.30 m.

Comparisons between kinematic parameters for drop jumps performed from the same drop heights showed good daily repeatability. During the first drop land, drop height significantly increased ground contact time, maximum knee and trunk flexion. Although the eccentric muscle phase of the activity was enhanced by drop height, there was no evidence to suggest a net increase in the mechanical output. It seems likely that the former kinematic variables may have served to decrease performance by increasing the latency period between the eccentric and concentric muscle actions.

The biomechanical model predicted sharp peaks in the spinal compression force curves at initial forefoot and heel impact. For both drop heights and both drop lands, these forces attain values which were considered to be excessive given present methods of calculation. An over-prediction of such forces was felt to stem from limitations inherent to the rigid-link modelling approach adopted by the model. However, given these limitations there appeared to be good reason to recommend that athletes should take every effort to cushion and minimise these initial impact forces. In addition, there was sufficient evidence to suggest that the second landing may impose a similar risk as the first landing.

During the first drop land, peak forces following heel strike attained values that were comparable between drop heights. The rise-time for this peak was longer and therefore more likely to represent true loading. Maximum forces were representative of those reported in the literature and were correlated with the degree of trunk flexion. Maintaining a more upright trunk would appear to lessen the magnitude of the load and consequently, the risk of injury.

Integration of the spinal compression force curve was used to assess the total stress throughout the landing phase. The 0.75 m drop jumps resulted in significantly higher loading than the 0.30 m drop jumps

and this, coupled with higher shear forces, led to an acceptance of hypothesis 10.

The EMG findings showed the erector spinae and external oblique muscles to be active throughout a majority of the landing phases. A drop off in erector spinae activity and increases in external oblique and rectus abdominis activity, towards the later stages of the first land, support the model's reversal of the net lumbar muscle moment. Only the normalised peak erector spinae EMG amplitudes were found to be significantly higher following the 0.75 m drop jumps; thus hypothesis 11 was not accepted.

Spinal shrinkage resulting from the exercise regimen involving 0.75 m drop jumps was greater than that found for the regimen composed of 0.30 m drop jumps, although differences were not found to be significant. Therefore, hypothesis 12 was not accepted. No consistent correlation was identified when comparing measures derived from each analytic approach.

10. RECOMMENDATIONS AND FUTURE WORK

10. RECOMMENDATIONS AND FUTURE WORKS

Increased loading and subsequent losses in intervertebral disc height, brought about during bounding and drop jumping regimens, are theoretically disadvantageous in view of the discs' increasing vulnerability over the period of exposure. As identified in the biomechanical modelling approach, there is a need to achieve maximal performance with the minimal spinal jerk. Establishing alternative methods of invoking the same training stimulus but reducing the likelihood of incurring spinal injury must be a prime future objective.

Prescribing spinal unloading postures prior to exercise was found to have little benefit, given the accelerated rates of shrinkage during the exercise period. A post-exercise unloading posture was found to negate exercise-induced loading quickly and it would therefore, seem prudent to recommend the intermittent use of such procedures throughout the training period. Establishing the acceptable time intervals for which an athlete can be exposed to loading before recovery postures are adopted, needs to be an area for future investigation. Clearly, there would first need to be more information identifying the amount of shrinkage that can be regarded as detrimental.

Warm-up routines are regularly advised to all forms of athletes as a means to prepare prior to more strenuous exercise. Although, results failed to show a significant variation in shrinkage for a 10 min warm-up routine and a period of standing, an increased loss in stature had occurred due to the warm-up routine. Longer warm-up routines would seem to pose a greater threat towards injury and therefore, attention should be focused towards developing routines which employ a series of unloading exercises, interspersed between those known to induce shrinkage.

Attempts were made to assess the influence of physical and functional measures on rates of stature loss. Clear patterns did not emerge, mainly due to the relatively small number of subjects studied and the length of period over which athletes were monitored. Future work needs to examine in greater detail the role these physical and functional parameters play in protecting the spine. For example, such studies could examine how long-term back and abdominal training programmes

influence the spines' response to loading exercises. Preventative measures may then be advocated prior to embarking upon this form of training.

Improvements were made to the method by which changes in stature could be accurately and reliably recorded. However, there are a number of areas in which the technique should be improved. At present the system requires a number of training sessions during which familiarisation with the apparatus is necessary. This obviously creates restrictions, particularly when implementing the procedure in an occupational setting. Reducing the hardware would certainly make it more amenable for use in the field.

An in-built time recording system within the computer set-up identified that on average 4 min was required to undertake a measure of stature. This may well have failed to detect the precise change in stature that had occurred, particularly directly after exercise or a period of unloading. The majority of this time was necessary to ensure that heel compression had normalised. Consequently, developing a system which eliminates the possible effects of heel compression, or ideally restricts measurements to the length of the spine, would be a direction in which researchers should be striving. More recent developments in stadiometer design (Burton et al., 1992) have seen an elimination of the head measuring gantry and the potential errors it incurs. Measurements have been restricted to, and below C7 using a lens and marker attached to the spinous process of C7; the assumption is that minimum height loss occurs within the cervical region of the spine. Alterations have also been made to the procedure by which stature loss is measured, measurements being calculated with respects to diurnal changes.

The combined analytical approach to studying spinal loading during two drop jumping regimens, from two drop heights (0.30 and 0.75 m), did identify areas of concern for those athletes taking part in these activities. A biomechanical modelling approach showed the cumulative stress placed on the spine during the higher drop height was significantly greater. Shrinkage over the entire regimen was also found to increase, although differences between regimens were not significant. Consequently, it is recommended that athletes should not embark upon regimens involving a higher drop height without shortening the duration

of the exercise period, or introduce more frequent periods of recovery throughout a single training session.

Given limitations inherent to the biomechanical model, the normal compression force curves identified distinct periods throughout the landing phases where forces may attain values which could place the health of the performer at risk, if not properly controlled. These occur during heel impact and towards maximum trunk flexion, increases in trunk flexion reflecting increases in compressive loading. Therefore, athletes are advised to cushion ground impact forces, via the use of suitable footwear or floor matting, and maintain a more upright technique. Future work should be directed towards examining the shock absorbing properties of cushioning materials. In addition, studies should also consider how movement of the upper limbs might influence softer landings.

Validation and improvements to the biomechanical model must be a prime objective of any future work. The model was found to possess deficiencies both in terms of the simplified muscle model and the rigid-link segment modelling approach adopted. Applying the model to other forms of activity, particularly lifting, and comparing data against standard reference material would provide one form of validation. Improvements in the model should be in the form of a mathematical function to take account of soft tissue absorption of forces. A muscle model driven directly by EMG would significantly enhance model predictions.

Enhancing the biomechanical model and EMG data collection, processing and analysis time would bring major advantages in terms of providing instantaneous feed back of information and the ability to study a much larger sample population. At present, a considerable amount of time is required to process the cinematographic film and raw EMG trace. Improvements should be directed towards on-line data processing, in the form of image analysis systems which directly process the kinematic data and computer systems to convert and analyse the electrical EMG signals.

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APPENDICES

APPENDIX 1

EFFECTS OF BOUNDING ACTIVITIES ON SPINAL SHRINKAGE

Abstract of the poster presentation for the one day Symposium on the Effects of Work Activity on the Spine, 24th October 1986, University of Nottingham.

BOOCOCK, M.G., GARBUTT, G., and REILLY, T., Dept of Sport and Recreation Studies, Liverpool Polytechnic. LINGE, K., and TROUP, J.D.G., Dept of Orthopaedic & Accident Surgery, University of Liverpool.

Static and dynamic loading of the upper torso with weights has been shown to cause a loss in stature which has been attributed to a decrease in intervertebral disc height. A pilot study investigated changes in stature following a bounding regimen, the type often undertaken by jumpers and sprinters as a training technique. Four male subjects performed twenty repetitions of five double footed bounds from a stationary position. Prior to exercise, subjects maintained the Fowler position for 10 min. Post-exercise recovery involved 40 min sitting in a relaxed posture. Measurements of stature were taken at two specific times of day (09:00 and 18:30 h) using a purpose built stadiometer.

The Fowler position, prior to exercise, caused all subjects to increase in stature by an average of 1.33 and 1.25 mm at 09:00 and 18:30 h, respectively. Following the bounding exercise, all subjects displayed spinal shrinkage. A loss in stature of 2.19 and 2.32 mm were found at 09:00 and 18:30 h, respectively.

The results suggest that further investigations should be carried out to determine the importance of pre- and post-exercise behaviour on spinal shrinkage. Assessing the extent to which the spine is loaded as a result of shock forces will allow the athlete to structure his bounding regimen so that spinal stress is minimised.

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APPENDIX 3

BORG'S (1970) PERCEIVED EXERTION RATING SCALE

6.

7. VERY, VERY LIGHT

8.

9. VERY LIGHT

10.

11. FAIRLY LIGHT

12.

13. SOMEWHAT HARD

14.

15. HARD

16.

17. VERY HARD

18.

19. VERY, VERY HARD

20.

APPENDIX 4

ELEVEN POINT LOWER BACK PAIN RATING SCALE

0 NO PAIN

1 EXTREMELY MILD PAIN

2 VERY MILD PAIN

3 MILD PAIN

4 FAIRLY MILD PAIN

5 MEDIUM PAIN

6 FAIRLY SEVERE PAIN

7 SEVERE PAIN

8 VERY SEVERE PAIN

9 EXTREMELY SEVERE PAIN

10 WORST PAIN IMAGINABLE

APPENDIX 5

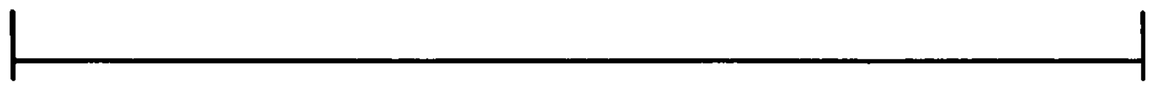
POST-EXERCISE STATE QUESTIONNAIRE

Please complete the following questions at the times stated immediately following your exercise session. Where necessary use the rating scale provided.

NB. LOW BACK PAIN refers to pain experienced between the mid-back and buttocks.

9.00PM ON THE DAY OF EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.



no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

10.00AM ON THE DAY AFTER EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.

no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

9.00PM ON THE DAY AFTER EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.

no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

10.00AM TWO DAYS AFTER EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.

no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

9.00PM TWO DAYS AFTER EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.

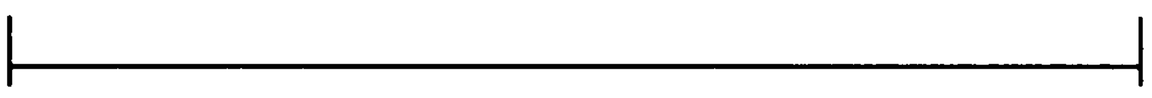
no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

10.00AM THREE DAYS AFTER EXERCISE

1. Indicate the degree of MUSCLE SORENESS on the line below.



no soreness severe soreness

2. Indicate the degree of any LOW BACK PAIN (insert a number from the scale provided)

.....

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APPENDIX 7

A. METHOD FOR CALCULATING SEGMENTAL PARAMETERS

Description of anthropometric measures

1	Body mass: (BM)	Subjects were weighed with a set of medical weighing scales and values taken to the nearest gram.
2	Stature: (STAT)	Subjects stood erect with heels together and looked directly forward. An anthropometer was used to measure the vertical distance from the floor to the top of the head.
3	Trochanter height: (TROCH)	Subjects stood erect with heels together. An anthropometer measured the distance between the floor and greater trochanter on the subjects' right side.
4	Epicondyle height: (EPH)	Subjects stood erect with heels together. An anthropometer measured the vertical distance from the floor to the lateral femoral epicondyle on the subjects' right side.
5	Tibiale height: (TIBH)	Subjects stood erect with heels together. An anthropometer measured the vertical distance between the floor and the right tibiale.
6	Lateral malleolus height: (LATMH)	Subjects stood erect with heels together. A tape measure was used to record the vertical distance between the floor and the subjects' right lateral malleolus.
7	Sphyrion height: (SPHY)	Subjects stood erect with heels together. A tape measure was used to record the vertical distance between the floor and the subjects' right sphyrion.
8	Foot length: (FOOTL)	Subjects stood with weight evenly distributed. Using an anthropometer a measure of the maximum distance between the most posterior point on the heel (acropodion) to the tip of the most anteriorly projecting toe (pternion) was taken.
9	Knee breadth: (KNEEB)	Subjects stood with weight evenly distributed. An anthropometer was used to measure the maximum breadth of the right knee across the femoral epicondyles.

10	Calf depth: (CALFD)	Subjects stood with weight evenly distributed. An anthropometer measured the horizontal distance from the anterior to the posterior surface of the right calf at the level of the maximum calf circumference.
11	Upper thigh circumference (UTHIGC):	Subjects stood with weight evenly distributed. A tape measure, perpendicular to the long axis the leg and positioned just below the lowest point of the gluteal furrow, was used to meast the circumference around the right thigh.
12	Mid thigh circumference: (MTHIGC)	Subjects stood with weight evenly distributed. A tape measure, perpendicular and at a mid point along the long axis of the thigh, was used to measure the circumference around the right thigh.
13	Calf circumference: (CALFC)	Subjects stood with weight evenly distributed. A tape measure, perpendicular to the long axis of the shank, was used to measure the maximum calf circumference.
14	Ankle circumference: (ANKC)	Subjects stood erect with weight evenly distributed. A tape measure, perpendicular to the long axis of the shank, was used to record the minimum circumference of the right ankle.
15	Arch circumference: (ARCHC)	Subjects stood erect with weight evenly distributed. A tape measure, perpendicular to the long axis of the foot and passing over the highest point in the arch, recorded the circumference of the right arch.
16	Iliac crest fat: (IILC)	Subjects stood erect and using skinfold calipers, a measure of skinfold thickness, one inch from the iliac crest at an angle of 45° towards the midline of the body was taken. The skin fold was obtained using a grasp of the thumb and index finger, enough to include two thicknesses of skin and subcutaneous fat but not muscle or fascia.

17	Omphalion height: (OMPH)	Subjects stood erect with heels together. An anthropometer measured the vertical distance from the floor to the omphalion. Subjects were instructed to relax the stomach.
18	Subischial height: (SUBH)	Subjects stood erect with heels together. An anthropometer measured the vertical distance from the floor to the inferior aspect of the isctual tuberosity and any compressed soft tissues.
19	Buttock circumference: (BUTC)	Subjects stood erect with weight evenly distributed. A tape measure, perpendicular to the long axis of the trunk and passing over the greatest protrusion of the buttocks, was used to measure the buttock circumference.
20	Buttock depth: (BUTD)	Subjects stood erect with weight evenly distributed. An anthropometer measured the horizontal distance from the anterior to the posterior surface of the buttocks at the level of maximum protrusion.

B. FORMULAS USED FOR CALCULATING SEGMENTAL PARAMETERS BASED ON THE WORK OF SEVERAL AUTHORS

(NB: R - RIGHT LIMB, L - LEFT LIMB)

1 **Segmental Masses (kg)**

Thigh

DEMPSTER (1955)	BM x 0.1 (R*) BM x 0.101 (L^+)
CLAUSER et al. (1969)	(0.074 x BM) + (0.123 x UTHIGC) + (0.027 x ILCF) - 4.216
CHANDLER et al. (1975)	((0.126 x BM x 1000) - (1688))/1000 (R) ((0.127 x BM x 1000) - (1511))/1000 (L)
ZATSIORSKY & SELUYANOV (1983)	(0.1463 x BM) + (0.0137 x STAT) - 2.649

Shank

DEMPSTER (1955)	BM x 0.046
Clauser et al. (1969)	(0.111 x CALFC) + (0.047 x TlBH) + (0.074 x ANKC) - 4.208
CHANDLER et al. (1975)	((0.038 x BM x 1000) - (179))/1000 (R) ((0.044 x BM x 1000) - (178))/1000 (L)
ZATSIORSKY & SELUYANOV (1983)	(0.0362 x BM) + (0.0121 x STAT) - 1.592

Foot

DEMPSTER (1955)	BM x 0.014 (R) BM x 0.0146 (L)
CLAUSER et al. (1969)	(0.003 x BM) + (0.048 x ANKC) + (0.027 x FOOTL) - 0.869
CHANDLER et al. (1975)	((0.008 x BM x 1000) + (343))/1000 (R) ((0.009 x BM x 1000) + (253))/1000 (L)
ZATSIORSKY & SELUYANOV (1983)	(0.0077 x BM) + (0.0073 x STAT) - 0.829

Pelvis

ZATSIORSKY & SELUYANOV (1983)	(0.0976 x BM) + (0.04896 x STAT) - 7.498

*R - Right limb
*L - Left limb

2. Segmental Centres of Gravity (% Distance from Proximal Joint)

Thigh

DEMPSTER (1955)	1- (((TROCH + 3) - EPH) - ((TROCH + 3) -EPH) x 0.433))/(TROCH - EPH)
CLAUSER et al. (1969)	(0.227 x TROCH) + (0.989 x KNEEB) - (0.033 x ILCF) - 13.362/(TROCH - EPH)
CHANDLER et al. (1975)	0.3657 (R) 0.3681 (L)
ZATSIORSKY & SELUYANOV (1983)	0.4549

Shank

DEMPSTER (1955)	0.433
CLAUSER et al. (1969)	(((EPH - LATMH) - (TIBH - LATMH)) + (0.309 x TIBH) - (0.558 x KNEEB) + 5.786))/(EPH - LATMH)
CHANDLER et al. (1975)	0.3769 (R) 0.3739 (L)
ZATSIORSKY & SELUYANOV (1983)	0.4047

Foot (NB. % DISTANCE FROM HEEL)

DEMPSTER (1955)	0.429
CLAUSER et al. (1969)	((0.153 x FOOTL) + (0.137 x ANKC) + (0.444 x LATMH) + 1.403)/FOOTL
CHANDLER et al. (1975)	0.4232 (R) 0.4291 (L)

Pelvis

ZATSIORSKY & SELUYANOV (1983)	1 - (((OMPH - TROCH) - (OMPH - SUBH) x 0.3541))/(OMPH - TROCH))

3 Segmental Moments of Inertia ($Kg.m^2$)

Thigh

DEMPSTER (1955)	0.1226 (R) 0.1195 (L)
WIDULE (1976)	(0.1131) x (BM/60.78) x (STAT/170) (R) (0.1096) x (BM/60.78 (STAT/170) (L)

CHANDLER et al. (1975)	((24.102 x (BM x 1000)) - 433522) x (0.113 x 0.884 x 10^6) (R) ((20.31 x (BM x 1000)) - 172235) x (0.113 x 0.884 x 10^6) (L)
McCONVlLLE et al. (1980)	((16713 x STAT) + (61238 x MTHIGC) + (46410 x (TROCH - TIBH 6813237) x (0.113 x 0.884 x 10^6) (R) ((1465 x BM x 2.20462) + (23443 x STAT) + (63128 x MTH1GC) -6263299) x (0.113 x 0.884 x 10^6) (L)
ZATSIORSKY & SELUYANOV (1983)	((32.02 x BM) + (19.24 x STAT) - 3670) x 0.0001

Shank

DEMPSTER (1955)	0.0446 (R) 0.0429 (L)
WIDULE (1976)	0.0437 x (BM/60.78) x (STAT/170)2 (R) 0.04194 x (BM/60 .78) x (STAT/170)2 (L)
CHANDLER et al. (1975)	((5.434 x (BM x 1000)) + 37127) x (0.113 x 0.884 x 10^6) (R) ((6.434 x (BM x 1000)) - 24410) x (0.113 x 0.884 x 10^6) (L)
McCONVILLE et al. (1980)	((10851 x STAT) + (65309 x CALFD) + (26079 x ANKC) - 2723962) x (0.113 x 0.884 x 10^6) (R) ((10683 x STAT) + (83844 x CALFD) + ((TIBH - SPHY) x 10801) -276144) x (0.113 x 0.884 x 10^6) (L)
ZATSIORSKY & SELUYANOV (1983)	((4.594 x BM) + (6.815 x STAT) - 1152) x 0.0001

Foot

DEMPSTER (1955)	0.00299 (R) 0.003196 (L)
WIDULE (1976)	0.00252 x (BM/60.78) x (STAT/170) (R) 0.00289 x (BM/60.78) x (STAT/170) (L)
CHANDLER et al. (1975)	((0.433 x (BM x 1000)) + 5371) x (0.113 x 0.884 x 10^6) (R) ((0.371 x (BM x 1000)) + 8974) x (0.113 x 0.884 x 10^6) (L)
McCONVILLE et al. (1980)	((691 x ANKC) + (1018 x SPHY) + (77 x STAT) - 28110) x (0.113 x 0.884 x 10^6) (R) ((1000 x ANKC) + (63 x STAT) + (578 x SPHY - 29972) x (0.113 x 0.884 x 10^6) (L)
ZATSIORSKY & SELUYANOV (1983)	((0.414 x BM) + (0.614 x STAT) - 97.09) x 0.0001

Pelvis

McCONVILLE et al. (1980)	((30171 x BUTC) + (13754 x ILCF) + (46623 x BUTD) - 3214802) x (0.113 x 0.884 x 10^6)
ZATSIORSKY & SELUYANOV (1983)	((11.8 x BM) + (3.44 x STAT) - 934) x 0.0001

C. Examples of computer output of segmental parameters based on the work of several authors

TABLE OF SEGMENT MASSES (KG)

	THIGH		CALF		FOOT		PELVIS
	RIGHT	LEFT	RIGHT	LEFT	RIGHT	LEFT	
	MEAN		MEAN		MEAN		MEAN
DEMPSTER 1955	8.6000	8.6860	3.9560	3.9560	1.2040	1.2556	0.0000
	8.6430		3.9560		1.2298		
WIDULE 1976	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.0000		0.0000		0.0000		
CLAUSER 1969	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	9.8751		4.4848		1.3105		
CHANDLER 1975	9.1480	9.4110	3.0890	3.6060	1.0310	1.0270	0.0000
	9.2795		3.3475		1.0290		
McCONVILLE 1980	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.0000		0.0000		0.0000		
ZATSIORSKY 1983	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	10.0022
	12.4818		3.7718		1.1910		

TABLE OF SEGMENT C OF G FROM PROXIMAL JOINT (%)

	THIGH		CALF		FOOT		PELVIS
	RIGHT	LEFT	RIGHT	LEFT	RIGHT	LEFT	
	MEAN		MEAN		MEAN		MEAN
DEMPSTER 1955	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.3926		0.4330		0.4290		
WIDULE 1976	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.0000		0.0000		0.0000		
CLAUSER 1969	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.4854		0.3961		0.4438		
CHANDLER 1975	0.3657	0.3681	0.3769	0.3739	0.4232	0.4291	0.0000
	0.3669		0.3754		0.4261		
McCONVILLE 1980	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
	0.0000		0.0000		0.0000		
ZATSIORSKY 1983	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.5466
	0.4549		0.4047		0.0000		

TABLE OF SEGMENT MOMENTS OF INERTIA (Kg.m.m)

	THIGH			CALF			FOOT			PELVIS
	RIGHT	LEFT	MEAN	RIGHT	LEFT	MEAN	RIGHT	LEFT	MEAN	MEAN
DEMPSTER 1955	0.1226	0.1195	0.1210	0.0446	0.0429	0.0437	0.0030	0.0032	0.0031	0.0000
WIDULE 1976	0.1916	0.1856	0.1884	0.0740	0.0710	0.0725	0.0043	0.0049	0.0046	0.0000
CLAUSER 1969	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
CHANDLER 1975	0.1637	0.1573	0.1605	0.0504	0.0528	0.0516	0.0043	0.0041	0.0042	0.0000
McCONVILLE 1980	0.1873	0.1933	0.1903	0.0762	0.0749	0.0756	0.0012	0.0011	0.0012	0.1161
ZATSIORSKY 1983	0.0000	0.0000	0.2642	0.0000	0.0000	0.0511	0.0000	0.0000	0.0053	0.0721