CONTRIBUTIONS OF POSTURE AND DEFORMITY TO THE BODY-SEAT INTERFACE CONDITIONS OF A PERSON WITH SPINAL CORD INJURIES

BY

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ABSTRACT

Over the last two decades research emphasis has been placed on studying the effects of pressure on the buttock tissues of individuals with spinal cord injury. It is evident that the prolonged application of pressures above certain threshold levels will initiate a pathological process in the tissues that can lead to necrosis and ulceration. The exact mechanics of the process is not clearly understood. There is also evidence which suggests that a number of additional factors could be involved in the formation of pressure sores. For example, shear stresses and tissue distortion, repeated loadings, impact stress, temperature and humidity, metabolic stress, nutritional status, age, body stature, and psychological factors have all been implicated as possible contributing factors. In spite of this extensive research effort very few clear quantitative measures and guideline have been developed that can be used for clinical decision-making. That is, there are still missing pieces to the pressure sore problem.

This study investigates the possible contributions of pelvic and spinal deformity and body posture to the variables occurring at body-seat interface. Four variables were investigated involving two study groups; a normal control group and a group of individuals with spinal cord injury. The variables are: spinal/pelvic alignment, pressures across the buttock support area, tangential shear at the support surface, and locations of centre of gravity. The latter three variables where measured for both groups in nine standardized postures commonly assumed while sitting in a wheelchair. Spinal/pelvic alignment involved a radiographic series taken in three of the nine sitting postures.

The results indicate that pressure distribution and tangentially-induced shear forces are highly influenced by body posture. The results also indicate differences between the study groups in pelvic alignment and movement of the ischial tuberosities during changes of body posture. It is proposed that these findings have important implications relative to the design of future seating devices and in the clinical practice of pressure sore management.

ACKNOWLEDGEMENTS

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Many individuals and fellow staff either directly or indirectly contributed to this research endeavour. The technical contributions of Stan Cronk, Glen Ellis, Sam Hanks, B.J. Seaton, D.W Davis and Greg Shaw and others are gratefully acknowledged. The tremendous secretarial contribution by Beverly Wilson and the financial management by Linda Mills are greatly appreciated. The support and encouragement given by Drs. Calandruccio, Tooms and Perry are acknowledged with appreciation. The guidance provided my academic advisers, Profs. Paul and Barbenel of the Bioengineering Unit, University of Strathclyde, and the kindness and support given by Joan Wilson of the Unit office have been important contributions to this effort.

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And last but definitely not least, to my immediate family, Elaine and Sean, for their unwavering companionship and encouragement, without which this work would not have been undertaken and completed. I am especially grateful for their understanding of the many hours I had to be remote from family life over the four year period of this work.

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PART I - CURRENT KNOWLEDGE AND PRACTICE

CHAPTER 1. INTRODUCTION

1.1. THE PROBLEM

1.1.1. Background

The development of pressure sores continues to be a major clinical problem, especially for people with spinal cord injuries (SCI). In spite of the significant research effort that has been carried out over the past several decades, relatively limited clinical impact has been made in reducing the debilitating effect of pressure sores. Earlier studies conducted in the Greater Glasgow Health Board area found an incidence of pressure sores to be 8.8 percent among all hospital patients and individuals receiving a home visit from a district nurse (Clark et al, 1978). More recent data from the United States Regional Spinal Cord Injury System for the years 1978-1980 reported that 40 percent of the 7,000 spinal cord injured individuals followed developed pressure sores during the time period from initial injury to initial rehabilitation discharge. Approximately 6 percent of the individuals had severe pressure sores involving subcutaneous adipose tissue and muscle (Young and Burns, 1981a). The incidence dropped to approximately 30 percent in each of the follow-up years beyond initial rehabilitation discharge. The study also revealed a number of predictors that correlate with development of pressure sores. The completeness of the injury was associated with pressure sores, but the level of injury was not. Surprisingly, it was found that the incidence among quadriplegics was a little lower than among paraplegics. Women had significantly fewer sores than men. People who lived in a nursing home had by far the higher number of sores. The young and elderly had decreased incidence of sores compared with the middle aged, as did those who were married (Young and Burns, 1981b).

Richards et al report that attempts to study the psychological correlates of pressure sore formation have been few and inconclusive. Anderson and Andberg (1979) suggest that individuals with pressure sores may have lower responsibility for their own skin care and lower levels of satisfaction with activities of daily living. Some individuals tend to be more impulsive and have lower ego strength (Richards et al 1981).

Pressure sores are a severe and potentially life threatening complication for many individuals. A USA-Veterans Administration (VA) report (1968) estimated that 50 percent of all quadriplegic veterans will require hospitalization because of pressure-related problems during their lifetime and more than 30 percent of the paraplegic population will have a similar fate. It was also estimated by the VA that approximately 1/4 of these veterans will die as a direct consequence of pressure sores.

The magnitude of the problem is further emphasized in analyses reported by Robinson (1978) in Canada, Manley (1978) in South Africa, Motloch (1978) in California, Noble (1981) in Australia and Barbenel (1984) in Scotland. Manley (1978) reports an incidence of 4.5 percent of the total in-hospital population with an additional 5.2 percent showing signs of skin discolouration from excessive or prolonged pressure. Lawes (1984) reports on pressure sore re-admission of 260 spinal cord injured patients discharged between 1975 and 1978 from two spinal cord units in the United Kingdom. Twenty-five percent of the outpatients were re-admitted with pressure sores over the five year period. Thirty percent reported having suffered a sore in the spinal unit during the initial period of treatment, and 15 percent had been re-admitted with a pressure sore. Others have reported that over four percent of the deaths among the spinal cord injured can be attributed directly to complications arising from pressure sores (Geisler, 1977).

Young and Burns (1981b) reported an additional average cost of approximately $15,000 per year for hospital treatment of SCI individuals with pressure sores compared to SCI individuals without pressure sores. An earlier estimate also suggested an average cost of $15,000 to treat a single severe pressure sore, with an estimated total annual cost of over $2 billion in the United States alone (Motlock, 1978).

Professor Robert Roaf (1976), speaking at the Strathclyde Bioengineering Series on Biomechanics of Tissue Viability, said, "I am always appalled by the bad sitting posture into which our present design of wheelchair seems to force patients. I feel we have given inadequate attention to this as a contributory factor in the aetiology of pressure sores". Zacharkow (1984) also focused attention on the potential contribution of the abnormal seated posture imposed by the conventional wheelchair towards the onset of pressure sores. After an extensive review of the literature combined with clinical observations Zacharkow concludes that, "poor sitting posture is one of the major factors, and definitely the most ignored factor, in the aetiology of pressure sores". In a 1984 presentation at a national symposium on the care, treatment, and prevention of pressure sores, the author outlined several biomechanical factors of abnormal seated posture that could predispose a person with a SCI to increased pelvic loading, when compared to individuals seated in a biomechanically stable posture. The concluding recommendations called for research studies that would address the relationship between postural factors, pressure sore management, and functional performance (Hobson, 1984).

The research study upon which this dissertation is based addresses the problem of abnormal seated posture and skeletal deformity as observed in the disabled population with a spinal cord injury. Specifically, the study addresses the relationship between skeletal deformity, seated posture and its effect on body/seat pressure distribution, body/seat interface shear and centre of gravity location. Comparisons are made between a sample of able-bodied subjects and a sample of individuals that have had a SCI for at least five years.

The research question(s) and the objectives of this investigation were formulated in 1984. Since that time a number of research advances have been made that have had a direct bearing on the outcome and to some degree originality of this study. To the extent possible these advances have been acknowledged, incorporated, and built upon throughout the latter course of the study. As a result, it is felt that the study makes a unique, current, and significant contribution to the body of knowledge related to management of non-ambulatory individuals with problems related to postural deformity and tissue breakdown.

1.1.2. Clinical Manifestations

Clinically, it has been generally concluded that the formation of decubitus ulcers is mainly a result of excessive or prolonged application of pressure to the weight bearing tissues; therefore the term "pressure sores" has been adopted universally as the clinical term for the problem. In the more severe cases pressure sores are categorized by the type of open wound where necrosis has penetrated the deeper tissues in response to the cessation of blood flow to the tissue region.

There are many other groups of disabled people who are predisposed to the debilitating effects of pressure sores besides those with spinal cord injury. These groups are:

1. the geriatric population, particularly those confined to nursing homes;
2. trauma patients who are confined to intensive care units for extended periods;
3. people with demyelinating neurologic diseases, such as multiple sclerosis;
4. people who suffer from limited physical and personal mobility, such as stroke, post-polio, and muscular dystrophy patients;
5. those with spinal birth defects, such as spina bifida.

The most commonly cited causes of pressure sores include:

1. prolonged sitting during daily activities, particularly during work, travel and recreational activities, such as card playing or video games;
2. use of old and ineffective wheelchair cushions;
3. activities that involve sitting on uncushioned areas such as a bathtub, or a floor to play with young members of the family;
4. falls while transferring from the wheelchair or bed;
5. sitting too soon after surgical procedures to correct a vertebral defect, or even during the comprehensive rehabilitation process;
6. excessive sweating or irregular attention to skin conditions;
7. wearing clothing that has exaggerated seam lines such as 'jeans', which can cause pressure to concentrate in areas that normally would not carry significant loads.

All of the above, except 6, can be related to increased levels or prolonged periods of pressure, tissue shear or direct trauma to the supporting tissue.

In summary, it is apparent that the incidence and treatment of pressure sores remains a major health problem. It affects a significant percentage of the total population resulting in major social and medical costs. The incidence is probably in excess of six percent of the population when other groups at risk in addition to the spinal cord injury population are taken into account.

The clinical and scientific evidence, although not conclusive, is biased towards excessive and/or prolonged application of pressure as being the primary causative factor. The other contributing factors such as; interface temperature and humidity, nutrition, circulatory, psychological, and general medical status, etc., will be briefly discussed in the appropriate following sections. And of course, the contributing factors associated with posturally induced deformity, shear and pressure stresses are the focus of this investigation and therefore have been given the primary emphasis throughout the literature review and following study.

1.2. HYPOTHESIS AND SCOPE OF STUDY

1.2.1. Development of the Hypothesis

It will become evident from the discussion of the relevant literature that the majority of the research work to date has focused on the relationship between buttock/seat interface pressure and the formation of pressure sores. Much less attention has been given to other factors such as the interface effects that can potentially result from pelvic and spinal deformity. For example, it has been observed by the author that

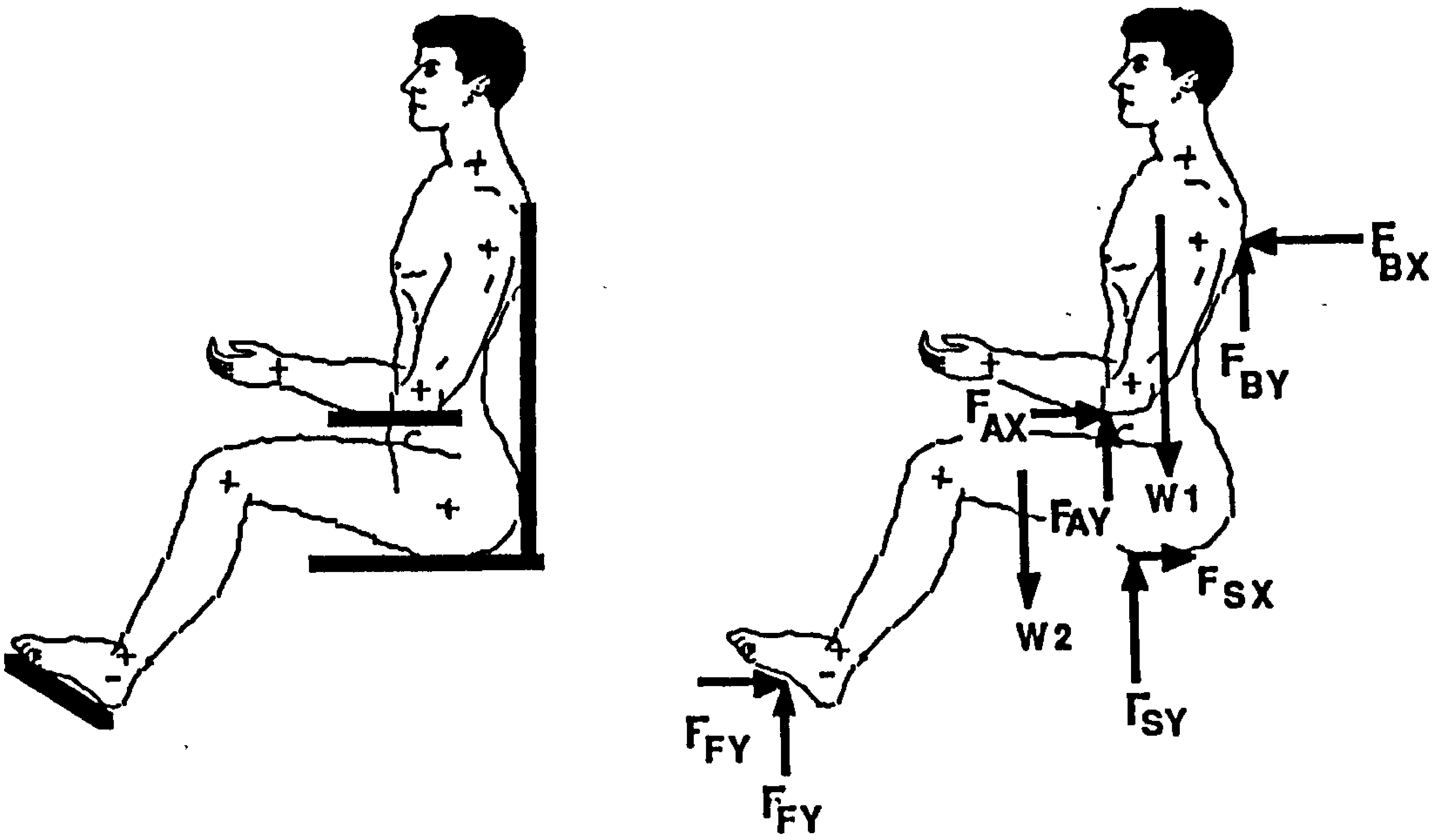


Fig 1.1 'Normal' supported seated posture illustrating the extrinsic forces acting on the body.

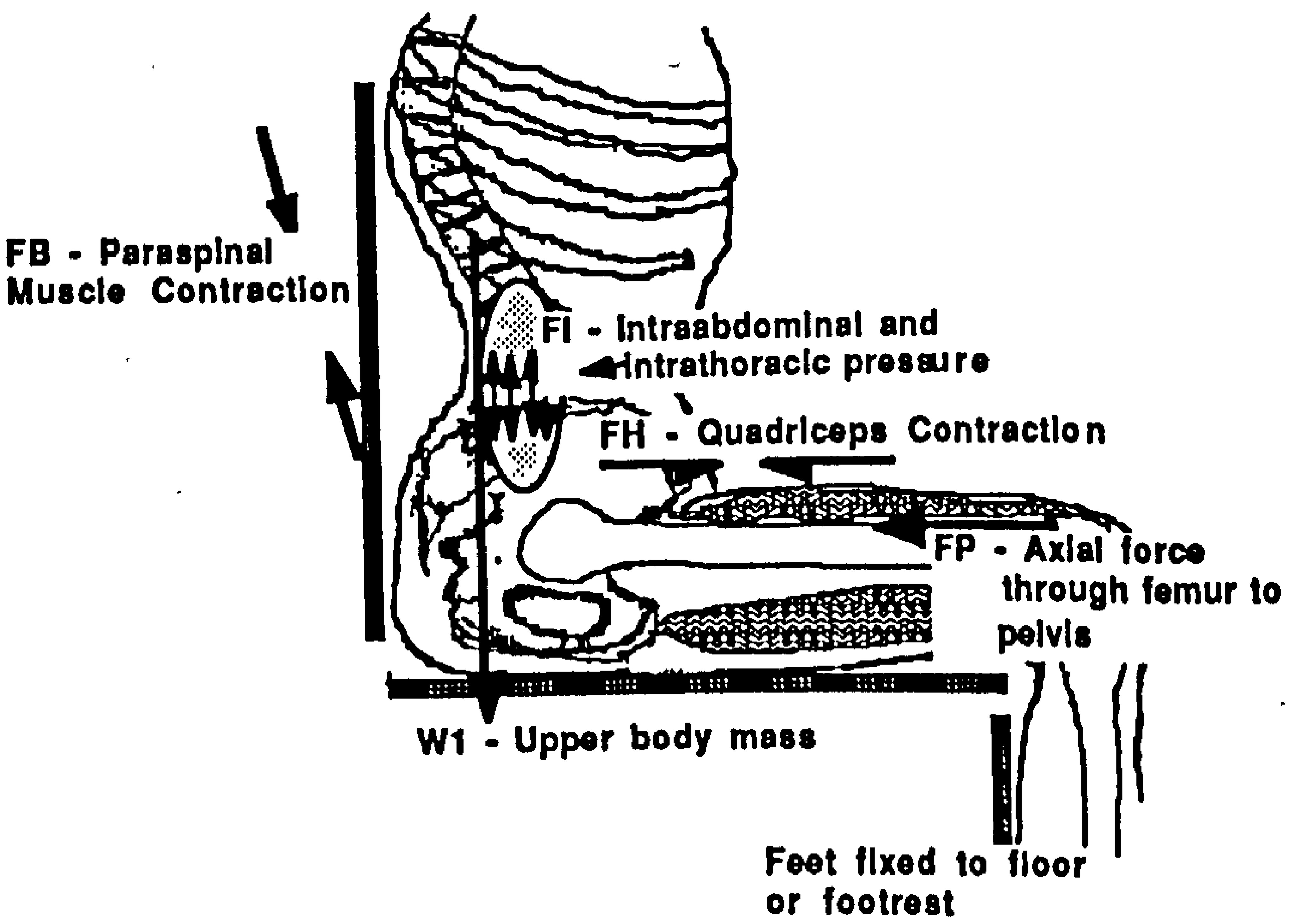


Fig 1.2 'Normal' seated posture illustrating the intrinsic forces that interact to maintain upright stability.

many individuals with spinal cord injury, who have used a conventional wheelchair for at least five years, have a typical abnormal seated posture. Their posture is characterized by a forward position of the pelvis with unsupported space between the sacrum and the backrest, and a 'C' shape thoracal spine with an extended neck and head position. It is postulated that a rigorous comparison of the seated postures between the able-bodied population and the spinal cord injured should reveal significant differences. The following cursory biomechanical analysis suggests some of the reasons why these differences may exist.

First, it can be stated that there really is no "normal" seated posture. This can be confirmed subjectively by viewing a large number of people seated in identical seats receiving the same visual and auditory stimulation. The observation will reveal a wide variance in seated posture (Wotzaka et al, 1969). However, if able-bodied individuals are asked to assume an upright seated mid-line posture then many similarities of seated posture between individuals begin to emerge. This 'simulated' normal upright posture can then be used as a reference posture to which comparisons of abnormal posture can be made.

Figure 1.1 depicts the extrinsic forces that act on the body in the sagittal plane during upright normal supported sitting, such as that provided by a wheelchair. The extrinsic forces are those of gravity (W1 and W2) and the counterbalancing reaction forces of the supporting surfaces (F_B, F_S, F_A, and F_P).

Of particular interest are the intrinsic forces. That is, the forces resulting from muscular contractions, skeletal reaction forces, ligaments, etc. that counter-balance the effect of gravity in order to maintain the trunk and pelvis in the upright position. Figure 1.2, simplistically illustrates the primary intrinsic forces that interact in a most complex manner in order to maintain an upright trunk and pelvis. The centre of gravity is positioned over the pelvis in such a manner as to minimize the energy expenditure necessary to maintain the upright posture. It can be seen that the posterior trunk musculature (the paraspinal muscles, F_B) contract to maintain the normal curves of the spine and thereby counter-balance the spinal flexion moment of gravity (W1). It has been shown that contractures of the abdomen, diaphragm, and intercostal muscles (F_I) create a supporting hydraulic reaction to the spinal column through compression of the viscera and stabilization of the thoracic cage (Morris et al, 1961). It can also be demonstrated that contraction of the quadriceps with the feet firmly planted on the floor or wheelchair footrests can create a reaction force along the axis of the femur (F_P), which in turn creates a righting moment on the pelvis via its reaction through the hip joint (acetabulum). The intricate interplay of the muscle contractions, combined with

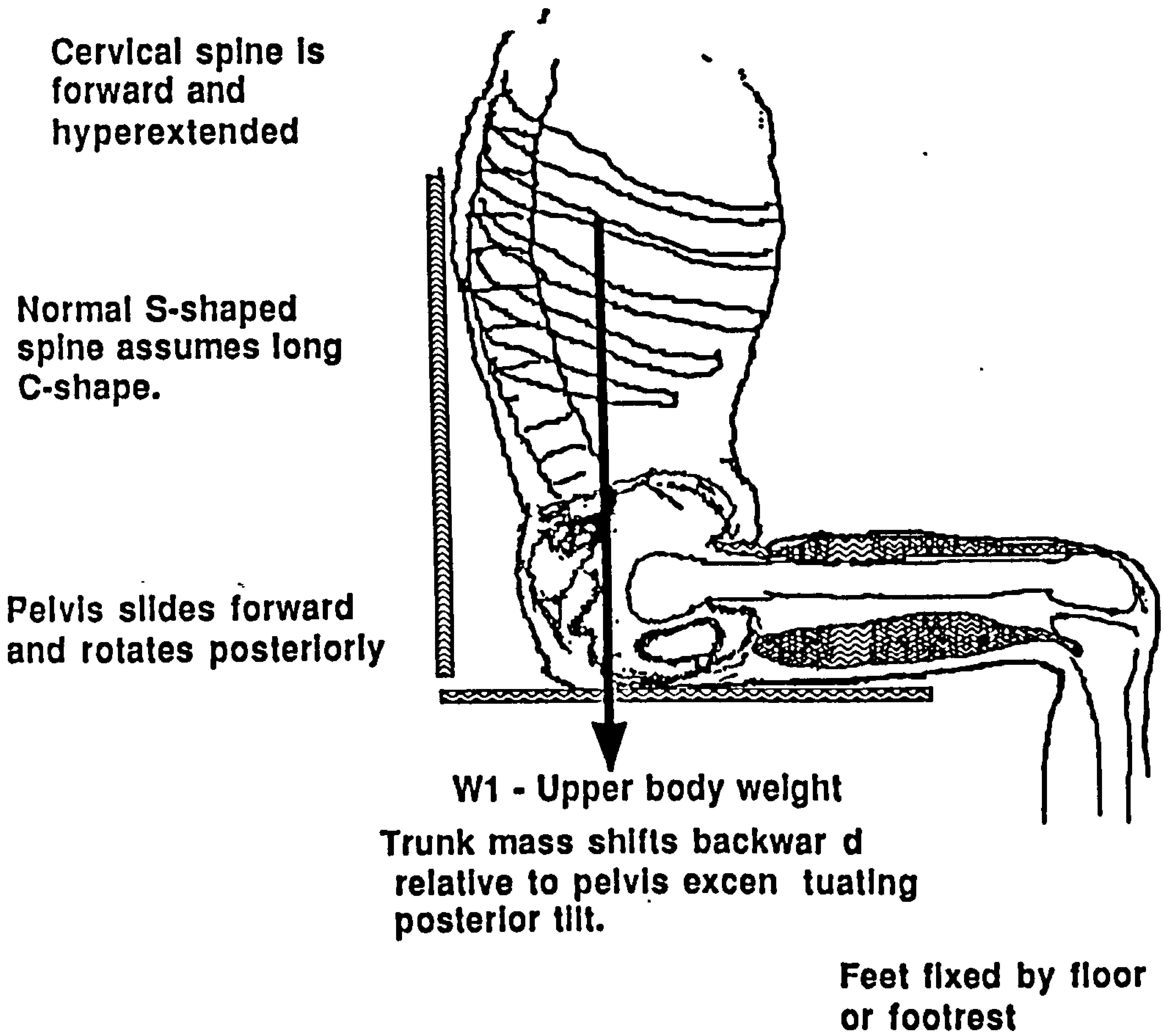


FIG 1.3 An observed abnormal seated posture that results with loss of voluntary intrinsic function distal to the mid thoracic level or higher.

their ligamentous structures and the forces transferred through the skeletal system all combine to maintain an upright seated posture with the trunk neutrally balanced over the pelvis.

A biomechanically normal erect supported sitting posture exemplifies the following characteristics:

* An "S" shaped spine as viewed in the mid-sagittal plane;
* Lordotic lumbar spine;
* Anterior rotation or neutral position of pelvis;
* Weight-line of the trunk over or anterior to the ischial tuberosities;
* Head located over the mid-line of the spine;
* Low horizontal forces on the upper backrest thereby minimizing the counteractive horizontal friction force acting at the seat surface;
* Maintenance of normal hamstring length;
* Controlled transfer of the sitting loads to the femoral shafts and posterior aspect of the thighs via the hip joint; and
* Minimized trunk muscle activity to maintain an upright trunk and head posture.

Examination of these same intrinsic factors in the extreme situation of a paralyzed person that has lost the use of the muscle function below the level of their spinal lesion presents a vastly different biomechanical picture. The spinal extensors (F_B), the supportive effect of the abdominal contractions (F_I) and the pelvic righting effect of the quadriceps and femur (F_H) are diminished or no longer present. Given the inability to internally counter-balance the trunk mass and given the posture imposed by the configuration of the wheelchair seat, a paralyzed person has little biomechanical alternative but to assume the posture shown in figure 1.3. The normal thoracic and possibly lumbar curves are lost and the spine assumes a long 'C' (kyphotic) shape. The pelvis slides forward and rotates (tilts) posteriorly due to the relentless effect of gravity being unopposed by any significant righting forces. The centre of gravity now shifts posteriorly relative to the ischial tuberosities. The pelvis becomes orientated in a fixed and relatively poor orientation for weight transmission via the supporting tissues. Furthermore, forward sliding of the pelvis can only be resisted by increased shear forces (F_{SX}) acting at the tissue/support interface. Movement of the upper trunk, such as in forward reaching, will be restricted by loss of the normal spinal mobility. It is further postulated that forward trunk motion beyond that which can occur by flexion of the thoracic spine will cause rotation of the complete trunk mass over the posteriorlydisplacing ischial tuberosities, causing elevation of both normal and shear stresses at the ischial/tissue/seat interfaces.

In summary, an abnormal sitting posture induced by the absence of muscle function and the postural constraints of the wheelchair can exhibit the following characteristics:

* Posteriorly tilted pelvis with impingement of the coccyx in severe situations;
* Kyphotic lumbar spine with a long 'C' shaped curve of the complete spine;
* Trunk CG-line falling posterior to the ischial tuberosities;
* Cervical hyperextension in order to maintain upright gaze;
* Shortening and permanent contractures of the hamstring muscles;
* Development of a fixed kyphoscoliosis of the spine;
* High forces against the backrest which will need to be counterbalanced by high shear forces at the seat surface.
* Actual unloading of the posterior aspect of the thighs causing higher ischial tuberosity pressures;
* Increased neck muscle activity necessary to maintain upright head position; and
* Poor ergonomic position for optimum wheelchair propulsion and cardiorespiratory function.

The detrimental effect of several of the above factors related to ischial loading may well be increased significantly during functional activities, such as; propulsion of the wheelchair, reaching, transferring, etc.

A 'correct sitting posture' in the frontal (side to side) plane is fairly obvious. Mid-line trunk orientation over a near level pelvis supported by a firm horizontal surface seems optimal. In contrast, problems are encountered by disabled individuals that are seated on the sling-type seats common to conventional wheelchairs. These problems may occur as:

* A pelvic 'drift' to one side or the other since very few wheelchair cushions provide mediolateral (M/L) stability to the pelvis, especially if a scoliosis is present;
* Persistent lateral pelvic drift and rotation (obliquity) which can drastically increase ischial pressures and probably shear forces as well, on one side; and
* The presence of other factors that can lead to asymmetrical postures and higher pressures such as; unilateral wasting of the gluteal muscle mass; unilateral hip abduction or extension contractures, and unilateral removal of pelvic bone (ischiaectomy).

Most importantly, it is apparent that the potentially detrimental effects of asymmetrical posture and deformity are not fully recognized by researchers and clinicians. In turn, these abnormalities can contribute significantly to elevated tissue pressure, increased shear stresses, sitting instability, tissue breakdown, and suboptimal functional performance.

Therefore, it is hypothesized that the abnormal sitting posture exhibited by individuals with spinal cord injuries can have a significant effect on the body-seat variables, that in turn can increase the probability of tissue breakdown and formation of pressure sores.

The above statement can be made in terms of several null hypothesis (Ho);

1) That sitting posture has no effect on the seat/body interface variables, and

2) That individuals with spinal cord injuries have body/seat variables that are no different from the able-bodied population.

In order to test the above hypotheses, four interrelated studies were carried out. The studies compared four parameters between an able-bodied control group and a sample of subjects with varying levels of spinal cord injury. The study variables are: a) spinal/pelvic alignment, b) body centre of gravity location, c) tangentially-induced shear forces, and d) interface pressure.

Pelvic alignment comparisons between subject populations is based on a radiographic series. The radiographs were taken in the same seating device with the subjects positioned in specified postures. The centre of gravity (CG) measurements were made using specially designed instrumentation, which permitted determination of the C G of the body while seated in nine reproducible postures. Gross shear forces acting tangentially to the seat surface were measured for both the normal and disabled subjects positioned in the same seated posture. Buttock/seat interface pressures were measured simultaneously using a commercially available pressure monitor. Analyses and comparison of results between the study samples have permitted conclusions relevant to the stated hypotheses.

1.2.2. Scope of the Investigation

In the above development of the hypothesis many statements were made that require substantiation. It was implied that elevated or prolonged seat/body interface pressure is a primary factor that leads to the onset of pressure sores. Is this in fact true? What is the current knowledge regarding the relationship between pressure and the onset of pressure sores? What are the other factors that can potentially contribute to the onset of pressure sores? For example, what role does interface shear force play and to what extent have other investigators substantiated its contributing role? What are the biomechanics of the normal seated posture and how does the body maintain an upright seated position? How, in fact, is the seated posture altered when certain muscles become weakened, unbalanced, or are no longer functional to maintain the upright position? How is seated posture defined, measured, and recorded? How have other investigators measured spinal and pelvic position and deformity? What is the state of the art in pressure relief technology and upper body positioning in order to prevent tissue breakdown and skeletal deformity, respectively? What are the technologies that are currently used in order to measure shear force and pressure at the interface surfaces, and what are the limitations of these technologies as they apply to both research and clinical environments?

A systematic review of previous work was carried out in an attempt to find answers to most of the above questions. In those cases in which answers were not forthcoming or complete, the current consensus (or lack of consensus) has been developed and stated. In those cases in which the measurement tools or conventional descriptive methods were not available or established, work was done to establish those necessary to conduct the study. An example being the absence of a suitable method to define the seated posture and the orientation of the seated body in space; further examples being a classification system to categorize the generic technologies used clinically in specialized seating and pressure sore management; and the instrumentation necessary to measure and record tangentially-induced shear force acting at the body/seat interface.

1.3. ORGANIZATION OF DISSERTATION

The dissertation is organized into nineteen chapters subdivided into three major parts, followed by appendices and two lists of references.

PART I (Chapters 1-5) contains the introductory section and the major review of the scientific and clinical literature. The literature review attempts to relate the relevant findings of previous work to the focus of this study. It formulates the scope of the clinical problem and reviews the current knowledge of both normal and abnormal seated posture. The biomechanics of body support is reviewed and the response of skin to external loading, particularly related to pressure and shear stress is emphasized. Current clinical practices are summarized, complete with a classification and description of current seating technologies in common use. In conclusion, Section I summarizes the current research and clinical knowledge and practice and develops the rationale for the following research study.

PART II (Chapters 6 -15) contains the essence of the investigation and the unique contribution of the work to the field. It begins with defining terminology and a method for defining the seated posture. The four inter-related studies involving; deformity, pressure, surface shear and CG measurement are detailed individually. The study populations, methods, instrumentation and initial findings are described. The methods used to analyze the study data, as well as estimations of error and statistical implications are presented. To facilitate presentation most details related to the design and construction of the laboratory instrumentation have been placed in the Appendix.

Part III (Chapters 16-19) contains the limitations on the findings, implications for changes to clinical practice and the recommendations for continuing research efforts.

Diagrams have been used liberally throughout the dissertation, especially in Part II and in the appendices in an attempt to enhance clarity of presentation and reduce the volume of text.

By way of a note to reviewers, chapters 2-4, inclusive, are a review of the current research knowledge and clinical practice. Readers wishing to more quickly review these chapters may proceed directly to the summary statements at the ends of most subsections or chapters. Chapter 5 is a summary of chapters 2-4, thereby making it possible to bypass many of the details contained in the literature review. This latter route is suggested for those familiar with current literature and clinical practice.

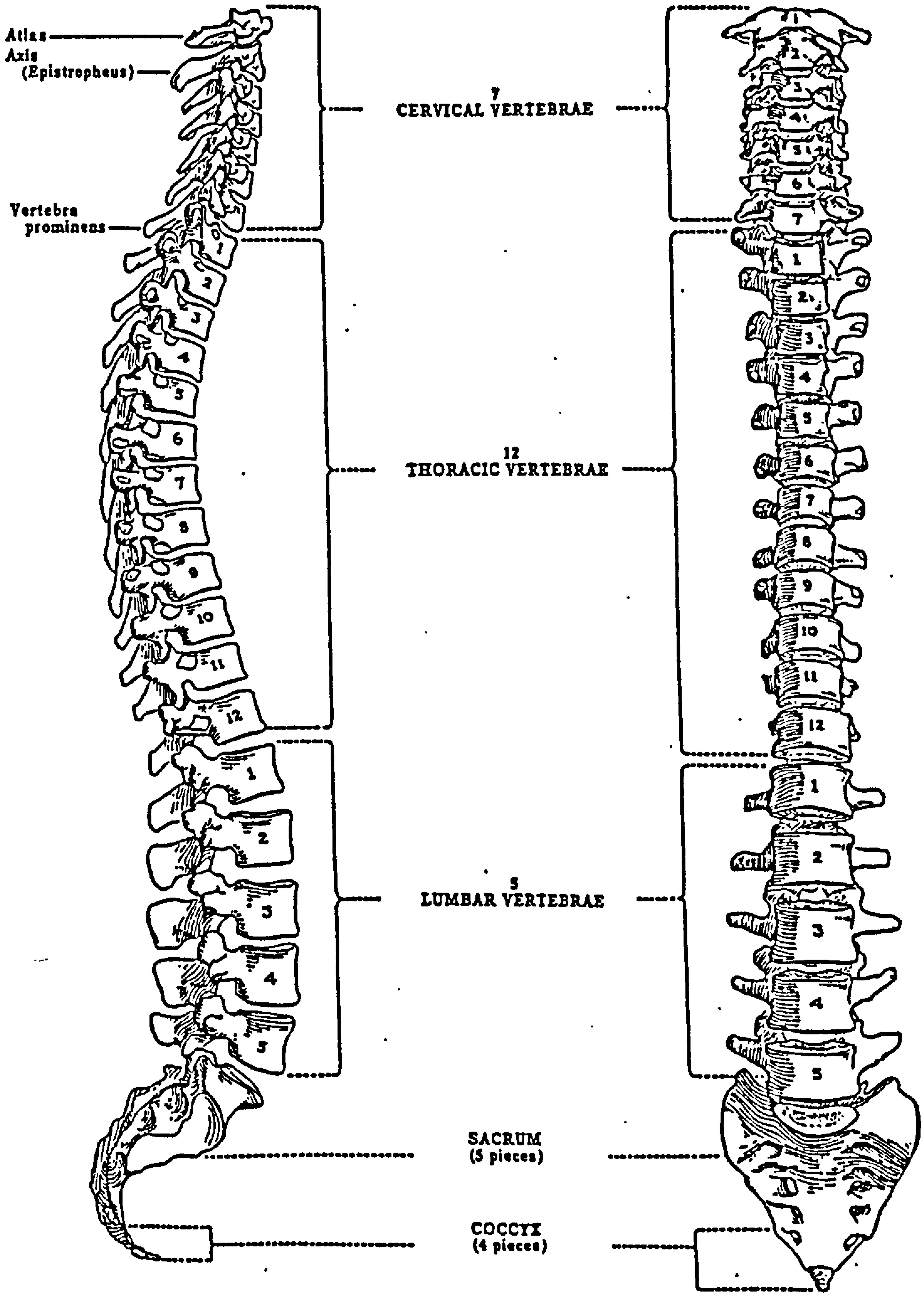


Fig 2.1 Diagram of erect spine showing normal curves and four segmental areas of the spine. (modified from Grants Atlas of Anatomy, Williams and Wilkins, Baltimore.)

CHAPTER 2. THE SEATED POSTURE

2.1. ABLE-BODIED SEATED POSTURE

Before analyzing the inherent problems of abnormal seated posture it is useful to identify the characteristics of able-bodied sitting posture. A useful beginning point of reference is the posture of the spine and pelvis in the erect anatomical standing position, sometimes referred to as the 'normal' posture of the spine.

2.1.1. The Spine

The classical anatomical or "military" erect spine consists of three normal spinal curves (fig 2.1). The vertebral column is straight in the frontal plane and curved in the sagittal plane; there is a cervical lordosis, a thoracic kyphosis, and a lumbar lordosis. When one sits down, the knees and hips flex, the pelvis tilts backward and the lumbar lordosis flattens (flexes). The normal spinal curves act as shock absorbers by reducing its longitudinal stiffness (Adams and Hutton, 1985). Of particular interest to this study are the normal ranges of motion of the lumbar spine because of its direct linkage with the pelvis.

In the living adult, the lumbar curvature, as defined in fig 2.2, varies with posture; from 80° when leaning backward to 0° when bending fully forward. In the erect standing posture the lumbar curvature is about 60°. When changing from a standing to a supported sitting posture, the pelvis rotates posteriorly and flattens or flexes the lumbar curvature to 45-50°. The curvature will normally decrease to 20-25° in unsupported sitting (Andersson et al, 1979; Adams and Hutton, 1983).

Currently, two divergent theories of proper sitting posture are being pursued by ergonomists and related researchers. One theory postulates that slight lumbar flexion (flattening) provides the optimal sitting position. Supporters of the flexed sitting posture commonly recommend elevating the knees above the hips. Those who support the flexion theory claim that a reduction in lumbar lordosis decreases pressure on the posterior elements of the lumbar spine (Adams and Hutton, 1985) .

The second theory of proper sitting posture promotes "normal" lumbar lordosis, which has been defined as the degree of lumbar extension established in the erect standing position, as being the ideal lumbar orientation for a correct sitting posture. Andersson et al (1979), Keegan (1953), and Mandal (1976) propose the use of a specially designed chair to achieve the correct sitting posture and possibly, to reduce the incidence of low back pain. This position is further supported by the classical work of Nachemson (1964, 1972) who found that, by reducing the normal lumbar lordosis in unsupported erect

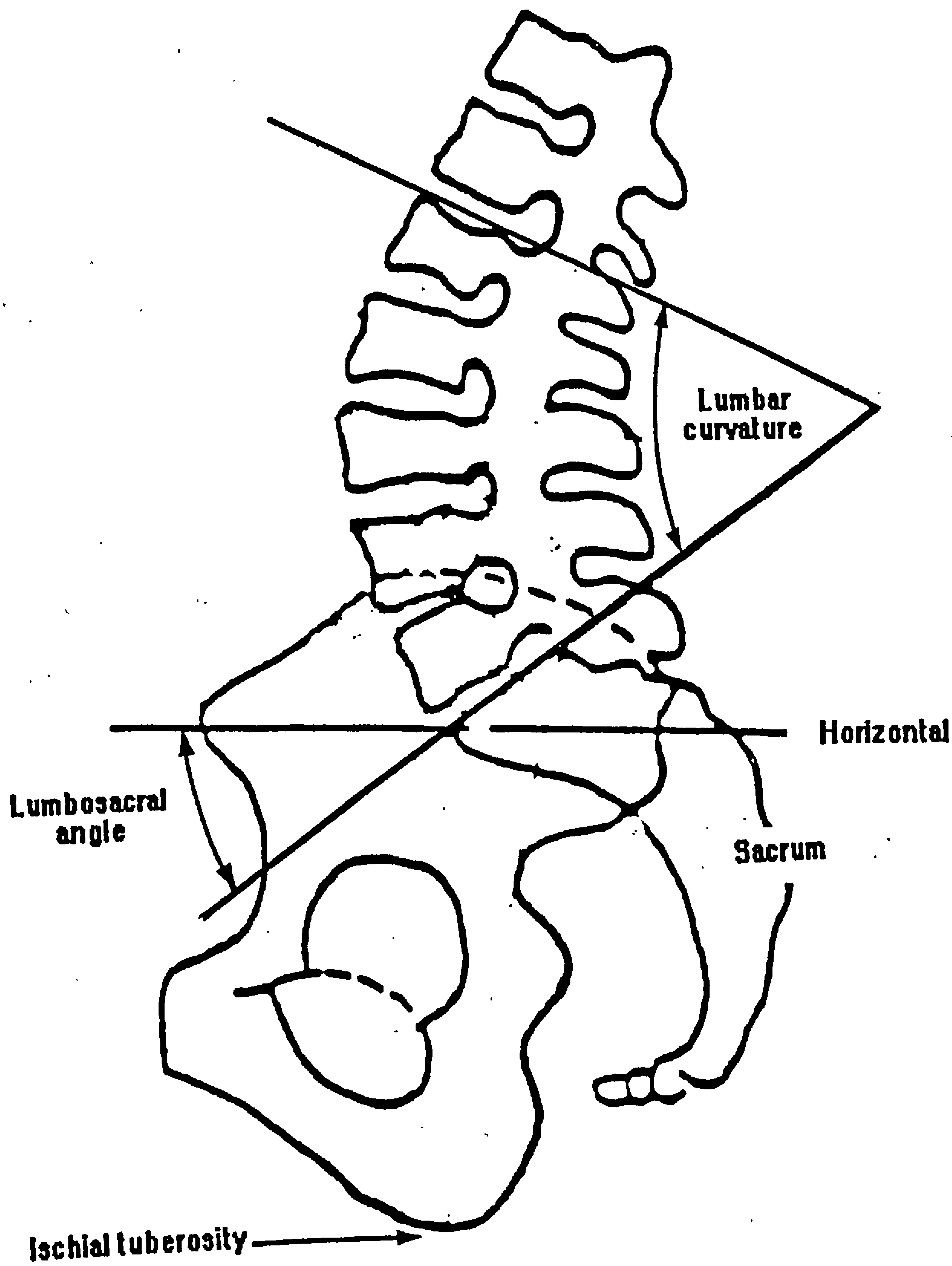


Fig 2.2 The lumbosacral angle is commonly used to measure the angle of the pelvis relative to the horizontal. The lumbar curvature measures the flexion or lordosis of the lumbar spine.

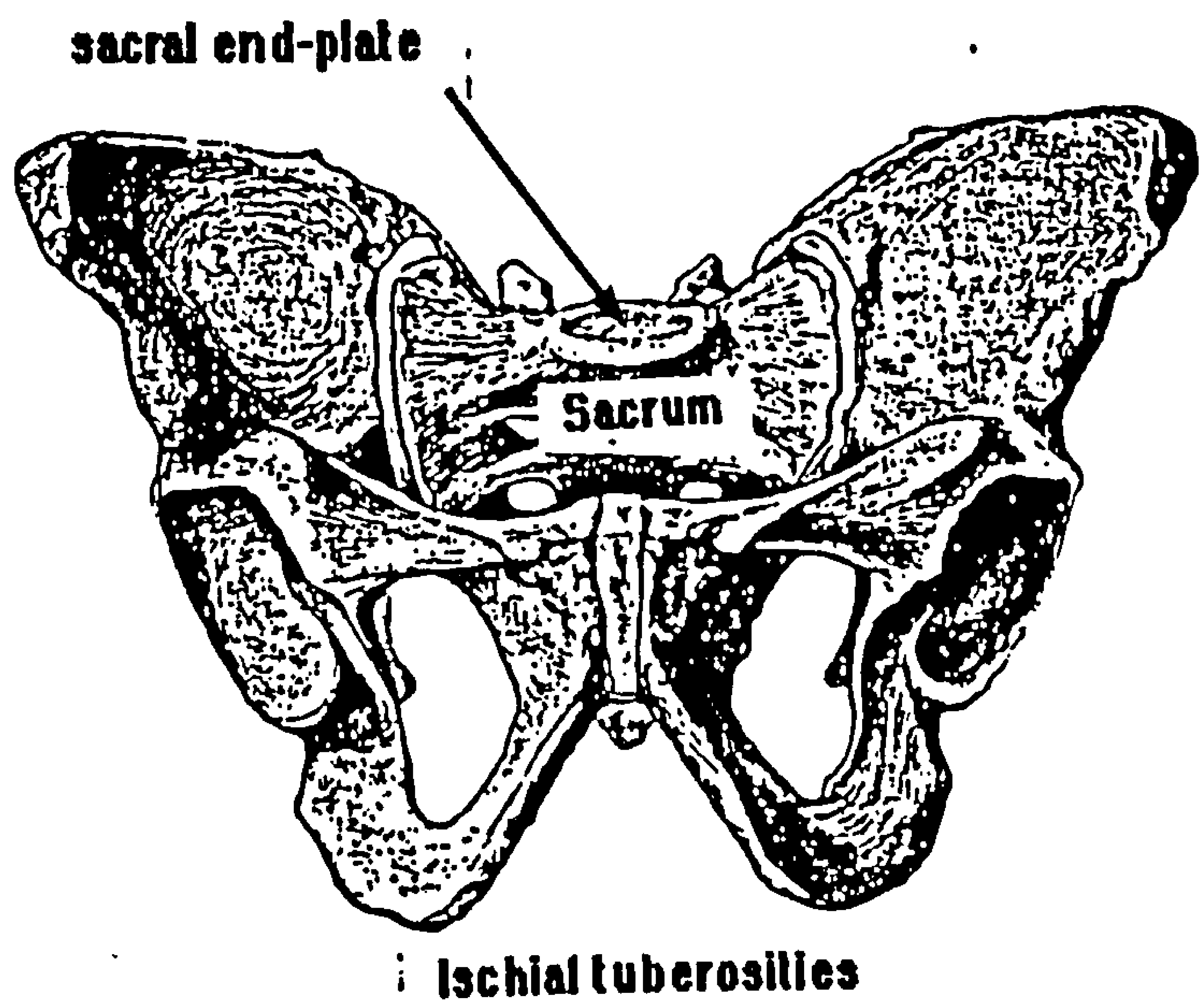
sitting, greater lumbar flexion occurred and the *in vivo* pressure on the third lumbar disk increased. Conversely, when sitting in a lumbar posture that more closely approximates the standing lordosis, intradiscal pressure decreases. It is this theory that supports the use of the forward reclining seats (15-20° of recline) in order to maintain the lumbar spine in its normal (standing) lordotic curvature, i.e., the Westnofa Balans chair. Keegan (1953) concluded that a thigh-trunk angle of about 135° in sitting preserved the desired normal lumbar lordosis. This is further supported by the observation that people often sit on the edge of a chair and lean forward. This is viewed as an attempt to improve comfort by restoring the normal lumbar lordosis and thereby reducing posterior tension on the lumbar spine (Mandal, 1976). In contrast, population studies have shown that lumbar disc degeneration is rare among people who habitually sit or squat in postures which flatten or reduce the curvature of the lumbar spine (Fahrni and Trueman, 1965).

These opposing theories have definite implications on the design of workstation seating for the able-bodied population, especially when muscle fatigue, postural fixation and low back pain are the primary considerations (Greieco, 1986). For the disabled the conventional wheelchair sling-type backrest provides virtually no support to the lumbar spine, and the seat is traditionally designed to maintain the knees at the same level as the hip joint. The backrest is usually reclined about 100° from the horizontal but this varies considerably with the age of the backrest upholstery. The footrests maintain the knees extended to approximately 35° of flexion.

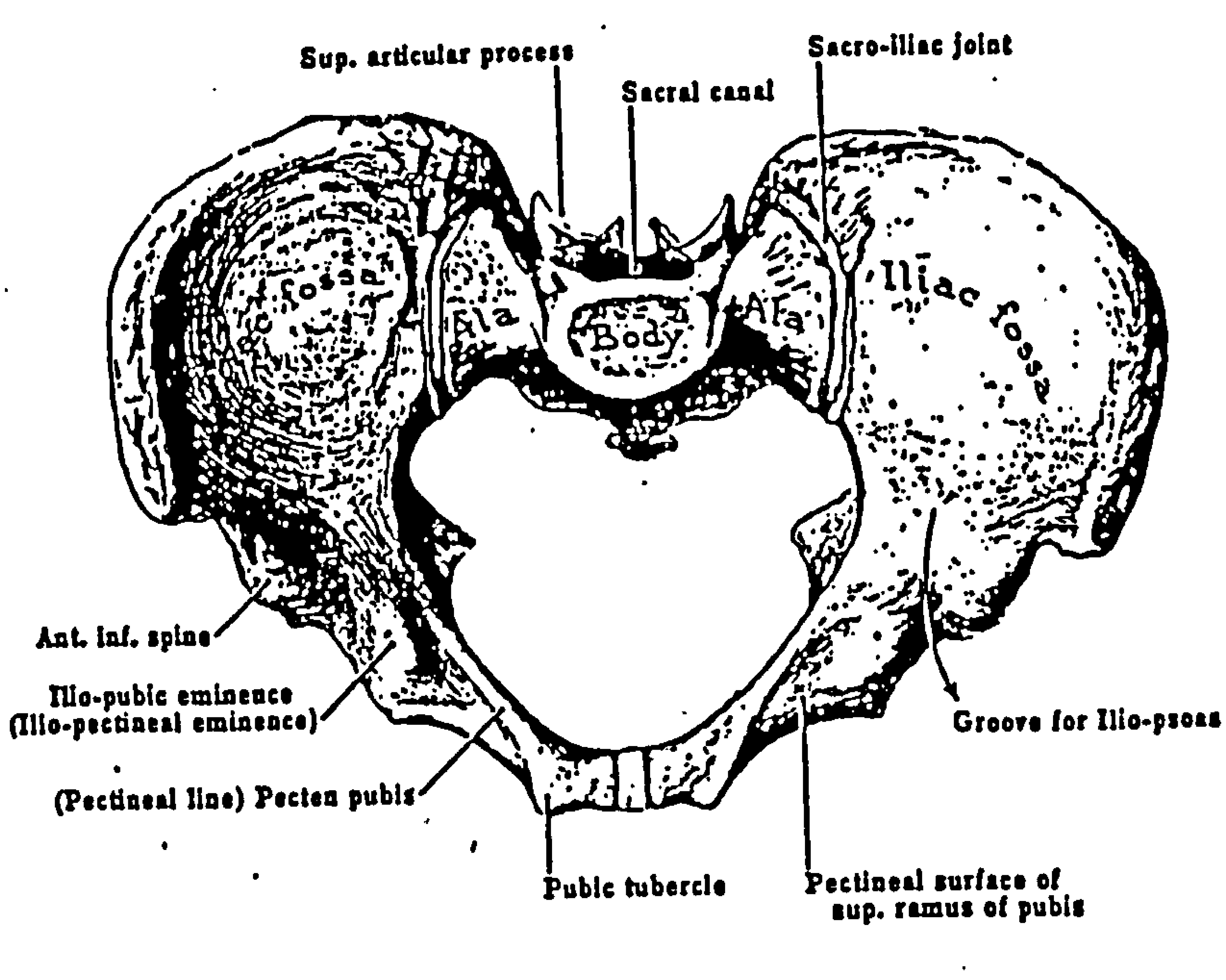
When an able-bodied person sits in a wheelchair they also assume a posture essentially dictated by the seat configuration. The lumbar curve tends to flatten to a value of 20-25°, which is similar to the unsupported sitting posture described by Adams and Hutton, (1985). This posture is further characterized by a backward tilt of the pelvis with the line of the centre of gravity of the torso falling either through or behind the ischial tuberosities (Kroemer and Robinette, 1969, Andersson et al, 1979). Schoberth (1962) described this posture as a posterior sitting posture. The observed posture of a person with weak or paralyzed trunk muscles will be described in a following section.

2.1.2. The Pelvis

Of particular interest to this study is the curvature of the lumbar bodies immediately superior to the sacrum (L3-L5) and the comparative alignment of the pelvis in both able-bodied and paralytic seating postures. By virtue of their inter-relationship, a discussion of pelvic movement must include reference to the sacrum. The sacrum is firmly



Male Pelvis, from the front



Male Pelvis, from above

Fig 2.3 Diagrams of male pelvis identifying bony landmarks referred to throuhout the thesis (modified from Grants Atlas of Anatomy, Williams and Wilfins, Baltimore).

bound to the two iliac bones of the pelvis by means of the anterior, posterior, and interosseous sacroiliac ligaments. It is further reinforced by the iliolumbar, sacrotuberous and the sacrospinous ligaments and by the lower portion of the erector spinae muscle. Because of this firm attachment, the sacrum is considered a part of the pelvic girdle. Some consider the sacrum to be the last vertebrae of the spine (fig 2.1). Figure 2.3 shows the relationship of the sacrum to the iliac bones of the pelvis.

The joints by which the movement of the pelvis occurs are the two hip joints and the joints of the lumbar spine, particularly the lumbosacral articulation. Since the pelvis depends on the joints of the lower spine and those of the hip joints for its movements, it is not surprising that its motion is sometimes associated with the motion of the trunk, the lumbar spine, and sometimes with the thighs. Clinically, the orientation of the pelvis in the sagittal plane is determined radiographically by measuring the angulation of the lumbosacral joint from the horizontal. The lumbosacral angle is the angle formed by two lines, one horizontal and the other drawn in line with the superior plateau (end plate) of the first sacral vertebrae (fig 2.2). The lumbosacral angle is approximately 30° in the erect standing position and reduces to approximately 15-20° in the erect seated position. Anterior tilting of the sacrum increases the lumbosacral angle.

In the erect normal position, the line of the centre of gravity passes slightly anterior to the sacroiliac joints. The gravitational moment that is created at the sacroiliac joint tends to cause the superior portion of the sacrum to rotate anteriorly and inferiorly. This tends to force the inferior portion in a posterior direction. The sacrospinous and the sacrotuberous ligaments counter-balance the gravitational torque and prevents the inferior portion of the sacrum from moving posteriorly. The superior portion of the sacrum is kept from being thrust anteriorly by the sacroiliac ligament. Pelvic balance is achieved when the line of gravity passes through the body of the 5th lumbar vertebra and the axis of rotation of the lumbosacral joint. In this position the gravitational movement relative to lumbosacral joint is zero and therefore the pelvis has a minimal tendency to tilt in either the posterior or anterior direction. However, during sitting voluntary control of trunk and pelvic musculature is required to achieve this ideal alignment of the trunk mass over the pelvis.

The pelvic structure is an integral part of the biomechanical structure of the body. In the upright standing position it provides the articular sockets for lower limb moment and upright mobility. When standing most of the torso loads are transmitted via the lumbosacral joint to the pelvis, down through the hip joints to the femurs, tibiae and ankle joints and eventually the standing surface. In sitting the pelvis has two inferior bony prominences (ischial tuberosities) that provide the main load transmission via the underlying tissues to the supporting seat surface (fig 2.3). The remaining load is transmitted via the femoral shafts and their underlying tissues depending on the intactness of the hip joint complex. Spinous processes on the pelvis serve as the attachment points for spinal, abdominal, hip joint, and lower limb muscles and ligaments which all interact in a complex manner to stabilize the body posture and permit voluntary movement.

Examination of the topography and normal alignment of the ischial tuberosities is of special importance to this study. The ischial tuberosities form a prominent bony projection on the posterior-inferior aspect of the pelvis when viewed in the sagittal plane (fig 2.2). The tuberosities are wedge shaped, larger in their superior than inferior dimension. Given the variances throughout the population the inferior aspect presents a bony projection to the supporting tissues of approximately 2 cm^2. When the pelvis tilts posteriorly with a reduction of the lumbosacral angle to approximately 15-20°, as in the case of moving from standing to sitting, the area of support offered by the wedge shaped tuberosities increases slightly. Also, because of the angulation of the ischial tuberosities as viewed in the transverse plane, posterior pelvic rotation tends to broaden the base of support on the seated surface.

The anterior spinous processes of the ilium provide origins for the powerful hip flexors, namely, Sartorius and Rectus Femoris. The ischial tuberosities serve as the origins for the powerful hamstring muscles that cross both the hip and the knee joints. The hamstring muscles, Biceps Femoris, Semitendinosus, Semimembranosus, insert into the tibia and fibular heads and therefore serve as extensors of the hip and flexors of the knee. In the seated position with the the feet fixed they can have strong posterior tilting effects on the pelvis.

The sacral bodies (S1-S5) are rough bony projections which lie in the posterior midline of the pelvis (fig 2.1 and 2.3). The coccyx (tailbone) is the most inferior projection of the sacrum and therefore can readily receive impingement with excessive posterior tilt of the pelvis in the seated position.

The pelvis also serves as the attachment (insertion) point for the primary anterior and posterior muscles of the torso, i.e., Rectus Abdominous and Erector Spinae. These muscles, combined with the intrinsic muscles of the thorax and spine provide the stability and voluntary movement of the torso.

In normal sitting the femurs are usually abducted at the hip joints providing a broadened base of support to both the pelvis and indirectly the trunk. Most of the upper body weight is transmitted via the pelvis and femurs to the underlying support surface.

The weight of the lower limbs and part of the trunk weight is transmitted through the posterior thigh tissues to the seat to the floor, or through the lower limb skeletal structure to the floor via the feet. This normal transmission of lower limb weight may be significantly altered when seated in a wheelchair.

Variations in normal seated posture constantly shift the loads from one body area to another; which includes the arms (via chair armrests), through the back (via the backrest in slouched sitting) and the feet (via controlled anterior trunk flexion). When a person is confined to a much smaller body movement envelope, either due to loss of voluntary movement or through design of the seating device, the ability to shift loads to alternate body tissues can be dramatically restricted.

2.1.3. Spinal Stability

Biomechanically, the spinal column may be viewed as a sustaining rod that maintains the upright position of the body and, as such, is subjected to a complex system of forces and stresses of different types. The spinal column itself may be considered to have both intrinsic and extrinsic stability. Intrinsic stability is provided by alternating rigid and elastic components of the spine which are bound together by the ligamentous system; whereas extrinsic stability is provided by the paraspinal and other trunk muscles (Morris et al, 1961). When the isolated ligamentous spine is loaded as a modified elastic rod fixed at the base, the critical load value is approximately 2 kg (4.5 lbs), much less than the weight of the torso it supports. If the load is increased further buckling occurs (Lucus and Bresler, 1961). The stability of the spine is therefore dependent largely on the action of extrinsic forces provided by the trunk muscles. Morris et al, (1961) studied the magnitude of the forces that are withstood by the lumbar vertebrae and discs during activities of heavy lifting. They showed that the structures would only withstand the high forces if the contribution of the hydraulic effect of the abdominal and thoracic cavities contributed to the extrinsic supporting forces of the paraspinal muscles. The action of the trunk muscles converts the thoracic and abdominal cavities into nearly rigid-wall cylinders of air, liquid and semi-solid material (viscera). Both cylinders are capable of providing added stability to the spine (Morris et al, 1961). Although the study emphasized the supportive contribution of the trunk cavities during heavy lifting activities, it also demonstrated that a reflex mechanism caused involuntary contractions of the trunk muscles, thereby fixing the rib cage and compressing the abdominal contents. In summary, it seems reasonable to assume that, to varying degrees, intracavitary pressures aid in the support and stability of a normally seated trunk.

Other investigators have modeled the biomechanics of the human spine. Most investigators have analyzed only segments of the spine such as the deformation of the rib cage during simulated car crashes (Roberts and Chen, 1970). Belytschko (1973) used a three dimensional model of the vertebral column, but did not include the action of the muscles. Yettram and Jackman (1980) make the point that muscles of the trunk are essential to the stability of the vertebral column and they act as one with the structure, not as an externally applied load. They describe an equilibrium model for the spine that includes the reaction forces of many of the spinal muscles. However, they concluded that it was difficult to assess the accuracy of the results obtained, especially due to the difficulty in accurately determining the muscle forces from EMG measurements, and the forces generated in the passive elastic tissues of the intravertebral joints. Scholten and Veldhuizen (1988) studied the influences of curvatures on the mechanical stability of the spine. They used a three dimensional, geometrical non-linear biomechanical model to study the critical buckling loads with increasing lordotic and kyphotic curvatures. They found that when the body weight is taken into account as a load distributed along the whole spine, the calculated initial buckling load is twice the value that it is in the case of a single concentrated load acting at the top of the spine (Lucas and Bresler, 1961). According to their model they found that initial buckling decreases with increased lordotic and kyphotic curvatures.

For purposes of this study it is suffice to say that a comprehensive biomechanical model of the 'normal' human spine that has wide spread acceptance remains to be developed. The evidence to date suggests a complex interaction of intrinsic (ligamentous) forces and extrinsic (muscle) forces acting in conjunction with the skeletal structures and the involuntary hydraulic effect of the mainly abdominal cavity. They interact in an integrated manner in order to provide a stable biomechanical structure that supports both the upright standing and sitting postures, while still permitting voluntary movement of the torso in a wide variety of functional activities. Most importantly, this conceptual model of 'normal' provides a biomechanical point of reference when considering a person with 'abnormal' spinal mechanics in the seated posture.

2.2. ABNORMAL SEATED POSTURE

Although the emphasis of this work is on problems specific to the spinal cord injured (SCI) population other groups of nonambulatory people, to varying degrees, experience similar difficulties with postural instability, deformity, discomfort, tissue breakdown and less than optimal upper body function. The important differences unique to the SCI group are the total or partial absence of gravity-opposing muscle function distal to the lesion level, and the total or partial absence of tissue sensation over the weight bearing surfaces. Furthermore, it has been postulated that these two latter characteristics place the spinal cord injured group at increased risk due to progressive deformity, which can lead to asymmetrical sitting postures, which in turn can precipitate elevated body/seat interface stresses and therefore an increased predisposition to tissue breakdown. In order to gain further insight into these potentially adverse characteristics of abnormal sitting postures, involvement of the spine, pelvis, and problems inherent to the wheelchair are reviewed and presented separately.

2.2.1. Involvement of the Spine

Compared to 'normal' seated posture, much less research work has been done on abnormal seated posture resulting from physical disability, especially spinal instability. The majority of the literature on spinal instability is related to surgical management of the scoliotic spine. Scoliosis secondary to neuromuscular disease is probably the oldest form of recognizable spinal deformity. Until the virtual elimination of poliomyelitis all spinal curves without metabolic or structural deformities were considered to be due to muscle paralysis (Fisk and Bunch, 1979). More recently, research and treatment of neuromuscular and idiopathic scoliosis has occupied the attention of many spinal surgeons. Deformities in children with cerebral palsy, especially in the spine and hips, has been ascribed to muscle imbalance and increased tone, or to the type of cerebral palsy (Fulford and Brown, 1976).

Mayfield et al (1981) reported on a retrospective study of 40 children who had incurred spinal cord injuries between birth and the age of 18 years. The population was divided into pre-adolescents and post-adolescent groups. Paraplegia and quadriplegia in the preadolescent patient resulted in spinal deformity in all cases in the study, and 96 percent had a progressive deformity. These results were similar to other studies in the orthopaedic field (Kilfoyle R. M., 1965; Campbell and Bonnett, 1975). Although all of the post-adolescent patients had either a scoliosis or kyphosis secondary to instability at the fracture sight only one case had paralytic spinal deformity, in contrast to all of the pre-adolescent patients (Mayfield et al, 1981). This work suggests that individuals receiving a spinal injury prior to puberty will develop progressive paralytic spinal curves, whereas those acquiring the injury later in life will mainly have spinal deformities associated with the injury site. Mayfield et al (1981) also reported on a variety of spinal deformities including scoliosis, kyphosis, and lordosis. They reported paralytic scoliosis with either pelvic obliquity or spinal decompensation, or both, in 23 subjects. Paralytic kyphosis was present in 16 subjects of whom 13 had progression of the kyphosis. The five in which lordosis developed, all had secondary hip flexion contractures. There was no apparent reason why scoliosis developed in some and either kyphosis or lordosis developed in others. Furthermore, there was no correlation between the level of injury or the type of injury and the predominant deformity that developed. The authors did not report on malalignment of the pelvis in the sagittal plane.

Lancourt et al (1981) report on fifty pre-adolescent cases of spinal column loss of muscular support following spinal injury. Thirty-one had scoliotic curves of more than 20°; twenty-one of these were long paralytic curves of 40° or more. Age of injury and spasticity were the most important risk factors. Patients with lesions at all neural levels were at risk. On lateral roentgenograms the predominant finding was a reversal of the lumbar lordosis (curvature) into a kyphosis, with the development of a long thoracolumbar kyphosis. In five patients the opposite deformity, thoracolumbar lordosis, developed. Again, the study group were under seventeen years at time of injury, therefore extrapolation to those with post-adolescent injuries may not be appropriate.

Pope (1985) reports on a postural study conducted in Putney, England with a group of 266 very disabled adult subjects; ninety-two percent had a primary diagnosis of central nervous system (CNS) deficits resulting from the following disorders; multiple sclerosis, cerebral vascular disease, cerebral palsy, trauma, and other CNS disorders. The remaining eight percent had diagnosis of other less common disorders. Two typical abnormal postures were reported which involved deformities of the spine. The first group had a forward curvature of the upper trunk and head, a flattened lumbar spine, and a pelvis tilted backward. The mid-thorax was supported against the backrest and the upper body assumed the characteristic "C" shape. The second group reflecting instability of sitting posture presented with the body arched backwards from coccyx to vertex. The upper trunk was supported by the top of the backrest, with legs flexed under the seat (Pope, 1985). Unfortunately, the study did not delineate what diagnoses constituted each group. It can be assumed that the spinal cord population fell into Group I, ie. the group that demonstrated a "C" shaped unstable trunk posture with the flattened lumbar spine. More importantly, the study does clearly show that there are other groups of disabled people, in addition to the spinal cord injured, who have spinal instability which dramatically effects their sitting posture and functional performance.

Gibson et al (1978) reported on a study of progressive spinal deformity in 62 boys with Duchenne Muscular Dystrophy (DMD). Through review of clinical records they noted two pathways of spinal deformities. One route taken by only a few patients was gradual stiffening of the spine in a somewhat hyperextended position over a period of years. This group were usually good sitters for a long time, but if they lived long enough even the stiff spine would develop severe lateral deformities. The more usual and less preferable pathway involves moving gradually from a straight spine, which is subject to small lateral curves but still correctable by muscular effort, to one that has a constant and fairly marked kyphosis, which the patient cannot correct. Thereafter, there is a fairly steady progression into severe kyphoscoliotic deformity that can be predicted with certainty. Unchecked, the spinal deformity progresses relentlessly to severe scoliosis without kyphosis that makes sitting very difficult and even recumbency a problem (Gibson et al, 1978). They make the case for prophylactic surgical stabilization of the spine from T4 or T5 to the sacrum, which can lead to improved sitting posture.

In a previous report by the same group, they suggested that it may be possible to have more patients with DMD follow the first and more acceptable pathway thereby postponing or offering an alternative to surgery. To accomplish this they proposed to maintain intrinsic spinal stability by keeping the spine extended or hyperextended with the pelvis level from the beginning of wheelchair confinement (Gibson and Wilkins, 1975). This concept lead to the design of the Spinal Support System, which is a seating insert designed to maintain the desired midline orientation of the hyperextended spine. Although the authors reported success in delaying the onset of lateral curvature in a significant percentage of cases (Gibson and Wilkins, 1978), this concept has not received widespread acceptance in the non-surgical management of DMD.

Other than the above clinical studies, there is limited biomechanical analysis in the literature arising from laboratory studies on spinal instability related to specific disabilities. It is reasonable to conclude that the nature and progression of a disorder will have a direct bearing on the type and degree of muscle or skeletal involvement. For example, the results of a detailed biomechanical analysis of trunk instability for individuals with DMD will differ from the results of analysis done with individuals with spinal cord injuries. And even within a specific population biomechanical analysis will need to include variables related to the level of lesion, or degree of residual spasticity, or, if progressive, the stage in the progression of the disease. These are all extremely complex biomechanical variables, the analysis and determination of which may apply only to a particular individual at a particular time in the natural history of their disease.

To summarize, it can be stated that there are populations of disabled people in addition to those with spinal cord injury that have widely varying degrees and types of spinal instability and postural deformity. The nature of the spinal instability and abnormal posture is usually diagnoses or disease specific and varies significantly between individuals depending on the severity of the disease; and if it is progressive, the stage of its progression, and possibly the age of onset of injury or disease. Information on the biomechanical analysis of the unstable spine is even less complete than the available work on the normal spine. Clinical studies have indicated that abnormal spinal postures create discomfort, impair respiratory function, increase the risk of tissue damage, and reduce upper body function (Pope, 1985). Orthopaedic studies involving the traumatic paralytic spine commonly recommend surgical stabilization of the spine by bony fusion extending from the trauma site extending down to the sacrum (Mayfield et al, 1981). In the case of neuromuscular disorders like DMD, early surgical stabilization has become the treatment of choice in many centres.

Given the dearth of scientific and clinical knowledge on spinal biomechanics, especially as it relates to the paralytic spine, the simplistic analysis presented in Section 2.1 above seems reasonable in its assumptions. In essence, the model proposes that 'normal' spinal stability involves a complex interaction between spinal and abdominal muscles, supported by involuntary intra-abdominal and inter-thoracic 'hydraulic' pressures, all interacting to support a flexible spinal column that is maintained in the normal 'S' configuration during sitting and standing activities. Diminution of muscle function either due to partial or total paralysis renders the spine inherently unstable and subject to the extrinsic gravitational forces. In addition, the extrinsic forces imposed by the postural alignment of the wheelchair seat support surfaces does little to compensate for losses of intrinsic trunk function. Loss of voluntary trunk stability, combined with the imposed configuration of the wheelchair seat, biomechanically necessitates that a person with diminished trunk control assume an abnormal sitting posture. This posture is characterized by a long 'C' shaped kyphotic spine, with an extended cervical spine, a flattened lumbar spine and a posteriorly tilted pelvis. If lateral trunk deformities are present trunk imbalance and pelvic deformities in the frontal plane can exist. Finally, it is postulated that this combination of trunk and pelvic abnormalities can be contributing factors to tissue problems, the effect of which is not fully appreciated by many researchers and clinicians.

2.2.2. Involvement of the Pelvis

It is difficult to separate the spine and pelvis because they are so intimately associated, especially when deformities are present. For example, in the management of cerebral palsy it is debated as to the temporal sequence of pelvic and spinal deformities.

Does a spinal deformity precipitate a pelvic obliquity and hip dislocation or does the hip dislocation occur first and then the other deformities follow? Letts et al (1984) in an indepth clinical and radiological review of 22 teenage subjects found that the most common temporal sequence was the dislocation of the hips, followed by pelvic obliquity and finally scoliosis. Although this finding serves to illustrate the intimate structural relationship between the spine and pelvis, the results do not apply directly to most other populations since the deforming mechanisms are likely quite different. As indicated previously, asymmetrical muscle contractions due to spasticity in the cerebral palsied group are reported as dominant factors (Fulford and Brown, 1976); whereas diminished muscle activity is the influencing factor in the spinal cord injured and DMD groups (Gibson et al, 1978).

Drumond et al compared sixteen paralytic subjects with spinal cord involvement, fourteen with spina bifida cystica and two with traumatic paraplegia. All had a neurological level of L2 or higher with no sensation in either the buttock or sacrococcygeum. Fifteen able-bodied individuals were used for controls. Nine of the disabled subjects had previous pressure sore ulcers and the remaining six were ulcer free. All nine subjects with ischial and sacral decubiti showed redistribution of the sitting pressures posteriorly, asymmetrical loading of the ischiae, and higher than normal pressures under the sacrococcygeum. The conclusions were that the abnormal pressures were a result of unbalanced scoliosis, pelvic obliquity, and loss of physiological lordosis following spinal fusion. They also found that the neurosegmental level of paralysis did not correlate with the development of ulcers. They concluded from the study that in excess of 30 percent of body weight on an ischium is the critical amount that should not be exceeded. And that greater than 55 percent posterior weight distribution can be associated with increased risk, particularly when spinal fusion does not restore or retain an ample amount of lumbar lordosis (Drumond et al, 1985).

The above study involved mainly a population of congenitally (pre-adolescent) spinal cord injured subjects. As indicated by Mayfield et al (1981) caution must be exercised in extrapolating findings to the traumatic post-adolescent spinal cord injured. Although the possible detrimental effects of surface shear were indicated in the Drumond study, no attempt was made to measure shear forces. Also, interface pressure was measured only in one posture, therefore the possible effects of alteration of seated posture on pressure distribution were not considered in the study. And finally, no attempt was made to compare centre of gravity location between the disabled and normal control populations.

2.2.3. Contributions of the Wheelchair to Abnormal Posture

The early history of wheelchairs remains somewhat obscure. Patent records in the United States Patent Office indicate that designs similar to the present configuration of user propelled wheelchairs date back to 1894 (Wilson, 1986). In 1937, Hubert A. Everest and Harry C. Jennings patented the present day cross-frame folding wheelchair (Everest and Jennings, 1937). This development rapidly replaced the pre-existing wood seat non-folding design, especially for nonambulatory people wanting a lighter weight wheelchair that would collapse small enough to be transported in an automobile. Probably no other single development has influenced personal mobility as much as the Everest & Jennings design, since it became the world-wide standard for the industry that remains to this day.

In order to achieve light weight and ease of folding, the Everest & Jennings design used vinyl sling-type upholstery to replace the firm seat and back components of its predecessor. Unfortunately sling-type surfaces do not provide a stable sitting surface, especially when pelvic and/or spinal instability or deformity is present. Gibson et al (1975b) feel that the hammock seat invites pelvic obliquity. Zacharkow (1984b) makes the point that the hip on the high side of the pelvis may become subluxated and dislocated from being kept in a chronically adducted and flexed position.

The hammock type backrest is designed to give approximately 10° trunk recline from the vertical. The backrest provides no lumbar support thereby allowing the superior aspect of the pelvis to rotate backwards into a posterior sitting position. This places the centre of gravity of the trunk either over or behind the ischial tuberosities with less weight being distributed over the posterior thighs (Zacharkow, 1984). With increased time the backrest upholstery stretches allowing further posterior pelvic sitting and an increase of the kyphotic spinal deformity. Continued deformity may allow the coccyx to become weight bearing (Howarth, 1978). Koreska et al (1976) and Gibson and Wilkins (1975a) felt that a scoliosis can be initiated by kyphotic sitting through unlocking the facet joints of the lumbar spine. Although one needs to use caution in extrapolating results from one group of patients to another, it is reasonable to assume that a biomechanically unstable spine in the lateral direction could yield similar lateral deformities in both children with progressive muscle weakness and those with pre-adolescent traumatic spinal cord injuries. And finally, Bunch and Keagy (1976) suggested that sitting in a wheelchair with marked kyphotic posture makes diaphragmatic breathing more difficult.

Most wheelchairs are made with non-adjustable armrests. Armrests that are too low encourage a slumped kyphotic posture. Or, they can result in leaning to one side in order to obtain trunk support. This will result in uneven pressure distribution across the buttocks, increase spinal imbalance, and possibly induce a subluxated shoulder joint on the weight bearing side (Zacharkow,1984a). Armrests that are too high will constantly maintain the shoulders in an elevated position (Diffrient et al, 1974) and provide a less than optimum reaction point for weight relief through routine decompression.

The wheelchair frame of most standard wheelchairs reclines the new seat upholstery 3-5° from front to back, with the back reclined 7-10° from the vertical beginning just above the armrests (Zacharkow, 1984a). When sitting on this near horizontal surface with a reclined backrest, the angle of which usually increases with age, there is a constant tendency for the legs and pelvis to slide forward into the posteriorly-rotated sitting position. This results in an increased upper rotation of the pelvis (posterior tilt), an increased kyphosis of the lumbar spine and more weight bearing concentrated over the ischial tuberosities, coccyx, and lower sacrum (Zacharkow, 1984b; Drumond et al, 1982), with potentially dangerous shearing (friction) forces occurring at the seat surface (Zacharkow, 1984b).

In reclining backrest wheelchairs with large seat to backrest angles, there is even a greater tendency for forward sliding which must be resisted by interface friction and shear stresses forces at the seat interface. In this position the weight of the abdomen also tends to flatten the lumbar curve when sitting in a semi-reclining position (Keegan, 1953).

The height and anteroposterior position of the feet relative to the seat surface are also variables dictated by the design of the wheelchair. In order to allow propulsion and steering with large diameter rear wheels, castered front wheels are used on most wheelchairs. In order to maintain optimum seat to floor height and allow use of the largest possible diameter of front caster wheels, the footrests are positioned forward of the caster wheels. This configuration, although allowing free rotation of the large casters, means that the knee joint must be extended to approximately 30-35°. Stokes and Aberey (1980) showed correlation supporting the hypothesis that straightening the knees produces flexion of the spine to an extent dependent upon hamstring tightness. It was found that compared to the standing curvature of the lumbar spine, individuals with tight hamstrings have a greater loss of lumbar lordosis when the knees are straightened from 90° to 45° of flexion. It is common for nonambulatory individuals to have hamstring tightness or spasticity that prevents extending of the knee joints in the seated position. Therefore, being forced to sit with the knees in 30-35° of flexion can exacerbate the posterior pelvic sitting posture due to the pull of the hamstrings on the ischial tuberosities.

Regarding the height of the footrests Lindan et al (1965) found that pressure under the ischial tuberosities markedly increased with the feet supported. They therefore concluded that the feet of a person in a wheelchair should bear little or no weight. Zacharkow (1984b) contests this view indicating that unsupported feet for most SCI individuals will create an unstable posture. He maintains that the thighs will be slanted downwards on the seat increasing the tendency for the buttock to slide forward. He further cautions that with no weight bearing through the feet there will be very high pressure along the distal posterior thigh. This may in turn obstruct the venous blood flow from the lower legs. He therefore concludes that the feet should definitely bear some weight when sitting, but only when combined with 10° of seat inclination, 15° of back inclination and a lumbar pad (Zacharkow, 1984b).

2.2.4. Anthropometric Factors

Differences in body build, an atrophied lower body, and decalcification of bone mass can significantly alter the biomechanics of tissue loading and the orientation of the centre of gravity. Garber and Krouskop studied the relationship between body build and pressure distribution in 67 subjects with severe spinal cord injury; 60% had quadriplegia and 40% had paraplegia. It was found that thin subjects had a predisposition for a maximum pressure in a critical area, and as a group showed significantly higher pressures over bony prominences than did average weight or obese subjects (Garber and Krouskop, 1982a).

Another study involved anthropometric measurements on 131 nonambulatory subjects with cerebral palsy and compared the results to anthropometric data derived from normal subjects. It was found that the differences in most critical seating measurements gradually increases with age yielding an approximate 20 percent difference at maturity. Body weight being the exception in which a 30-35 percent increase of normal individuals over disabled exists at maturity (Hobson and Shaw, 1987).

Atrophy of paralytic muscles can affect the distribution of body mass. Therefore, data from able-bodied studies must be used with caution when analyzing the centre of gravity, for example, of the seated person with a spinal cord injury. Shifts in the centre of gravity resulting from reduction in the weight of lower body segments may have to be taken into account when analyzing alterations to pressure distributions and stability of seated people with weight-altering disabilities.

2.2.5. Other factors

There are a number of additional factors that can directly or indirectly precipitate an abnormal sitting posture and therefore supposedly an increased predisposition for added tissue stress over susceptible areas of the pelvis and buttock.

Perpetual sitting in the cross-legged position causes adduction on the elevated side and transfer of additional weight bearing to the contralateral side. Pelvic obliquity with lumbar spinal involvement can lead to fixed structural deformities of the spine and pelvis.

Unilateral wasting or atrophy of the buttock tissues may occur when incomplete spinal cord injuries are present. If left uncompensated, pelvic obliquity can result in increased tissue stresses on the atrophied side (Zacharkow, 1984b).

Asymmetrical upper body activities such as writing or similar unilateral manual tasks can foster asymmetrical postures that result in unbalanced pelvic and femoral loadings.

Unilateral removal or reshaping of an ischial tuberosity (ischiectomy) can create asymmetrical pressure distributions, with the load being shifted to the intact ischial tuberosity (Guttman, 1976). Zacharkow (1984b) expresses concern regarding the person having a recurrent sore over the resected area, or developing a new pressure sore over the posterior aspect of the greater trochanter on the same side. He claims that following a ischiectomy a gluteal pad will be necessary under the cushion on the resected side in order to level the pelvis.

Bilateral ischiectomies effectively lower the remaining bony structures of the pelvis, such as the coccyx and posterior aspects of the greater trochanters. Although asymmetrical loading may not result, malalignment of the pelvis such as may occur in slumped sitting may place the inferior aspect of the sacrum at high risk of impingement on the seat surface.

Unilateral hip disarticulation amputations can create unbalanced sitting. Unless trunk support and compensation is made to the seating surface the individual will attempt to shift the body mass over the remaining hip in order to achieve sitting balance, resulting in increased pressures over the nonamputated side (Zacharkow, 1984b).

A unilateral hip flexion contracture which limits hip flexion unilaterally to less than 90°, will often cause a person to sit with the trunk shifted over the nonrestricted side. This position can cause increased asymmetrical pelvic loading on the unrestricted side which could lead to elevated pressures and eventually tissue breakdown.

2.3. SUMMARY AND RELEVANCE TO STUDY

This subsection has highlighted a variety of factors that can induce elevated normal and possibly shear stresses within the critical pelvic tissues. Many of the factors are not well understood by clinicians involved in pressure sore prevention and treatment. The contribution to abnormal posture induced by the design of the conventional wheelchair is rather evident. Additional factors, such as unilateral bone resections or amputations can create imbalances to sitting postures that result in increased unilateral loading to either side of the pelvis and its supporting tissues. Contractures, residual spasticity, perpetual sitting postures and unilateral atrophy can be present in varying degrees thereby contributing to altered and asymmetrical force distributions.

This study is primarily concerned with alterations in pressure distribution and tangentially-induced shear force patterns resulting from deformity and/or changes in body posture, and in quantifying the differences that may exist between normal and disabled populations. It is therefore important to be aware of the possible contributions of the numerous other factors highlighted above when attempting to draw any conclusions from the results of this study. In addition to the findings of this study it will be equally important to underscore the relevant and consistent findings that exist throughout the literature, especially when making recommendations for improvement to clinical practice.

CHAPTER 3. CONCEPTS ON THE BIOMECHANICS OF BODY SUPPORT

3.1. INTRODUCTION

In contrast to the biomechanics of abnormal sitting posture, a considerable volume of literature exists on the mechanics of body/interface support, especially as it relates to tissue viability of the spinal cord injured (SCI) population. Identification of the stresses occurring at the body-support interface and the response of tissues to these stresses provides important background for this study. Also, techniques for measuring interface stresses and centre of gravity locations are reviewed in an effort to identify methodological approaches that can be adopted in total or in part for this study.

3.1.1. Identification of the Primary Factors

Everyone, disabled and able-bodied alike, is subjected to gravitational forces acting vertically downwards. We remain stationary because equal and opposite forces are exerted by the surfaces which support us. For individuals sitting in a wheelchair, the support surfaces of the wheelchair exert forces on the body in identifiable locations. In particular, forces exerted by the backrest, seat surface, footrests, and armrests maintain the body in the stationary position (fig 1.1). In general, the body weight resulting from the gravitational effect is supported and transmitted by the skeletal structure. Transmission of the body weight to the external support surfaces, i.e., the wheelchair, occurs across or through soft tissues. These tissues are skin, underlying fat, fascia and muscle. The transmitted forces have a potential for damaging the soft tissues.

The critical factor is not simply the force, but the ratio of the force to the surface area over which it acts. This concept is known as stress and defines the 'force intensity'. Another important factor is the direction in which the force acts on the tissue. It is convenient to divide the forces into two types; direct and shear components. Direct or normal forces act perpendicular to the skin surface and when defined in terms of 'force intensity' are known clinically as normal stress or interface pressure. Shear forces act tangential to the skin surface and when defined in terms of 'force intensity' are termed shear stresses. This combination of normal and shear forces are thought to cause complex distortions and compressions within the underlying tissue structures that interrupt the perfusion of oxygen and nutrients essential for cell survival.

Direct or normal supporting forces are the most common in erect upright sitting. However, when slumping or reclining of the trunk against the backrest occurs, reaction frictional forces at the support surfaces must occur in order to maintain equilibrium and prevent motion of the body (fig 1.1). The frictional forces are in fact shear forces that ultimately must react between the support surface, the clothing, the skin and its underlying structures.

The above biomechanical picture applies only to a body remaining at rest, i.e., under static equilibrium. Movement of the body generates forces that are proportional to the mass of the body segment being moved times the acceleration of the movement (i.e., F = ma). Active people in wheelchairs are constantly moving and therefore generating dynamic interface forces, of both types, that can have peak values higher than the resting or static forces. These forces are sometimes termed impact forces and may cause quite different responses in the supporting tissues (Brand, 1976).

3.1.2. Contributions of Posture and Deformity

Of particular concern to this study is the potential contribution of body posture and deformity to the normal and shear stresses. As indicated previously (vide 2.2) many people sitting in wheelchairs acquire asymmetries of sitting postures which can drastically alter the force distribution patterns across the support surfaces and therefore the supporting tissues. Alterations to locations of centre of gravity of the trunk relative to the pelvis in the anteroposterior (sagittal) plane is a common observation. Lateral deformities of the spinal column with involvement of the pelvis generates similar asymmetrical stresses when viewed in the mediolateral (frontal) plane. Dynamic loads created during movement or functional activities can further alter the magnitude of the stresses throughout the supporting tissues. These stresses, as well as applying compressive loads to the tissue, can also cause distortional effects to both the skin and the underlying tissue structures.

3.1.3. Other Contributing Factors

In addition to the normal and shear stresses other extrinsic factors have been identified that can have detrimental effects on the supporting tissues. Build up of moisture between the skin and the body support surface is thought to effect the mechanical properties of the skin (Trandel and Lewis, 1975b). Temperature increases have been measured at the interface surface(s). There is evidence to suggest that temperature has a direct bearing on the metabolic requirements of the cells of the skin and underlying structures (Hyman and Artique, 1976).

When disease or paralysis are present systemic factors become issues of concern. For example, protein deficiencies, neurogenic disorders, biochemical imbalances, general nutritional status, oxygen transfer rate, venous congestion, infection all have been identified by researchers as potential contributing factors to tissue necrosis and breakdown. These latter factors, although not directly related to the scope of this study, are important to recognize since they can alter the clinical perspective of the mechanical variables central to this study.

3.2. A REVIEW OF THE STRUCTURAL PROPERTIES OF SKIN

Forces acting on the body surface produces stresses and strains (distortion) to the underlying body tissues. The relationship between stress and strain depends upon the mechanical properties of the tissue. Kennedi et al (1975) applied a variable force to excised tissue and measured the strain. The resulting stress-strain curve shows skin to be non-linear with a large initial deformation occurring upon application of the force. They also showed the existence of variability of skin extensibility depending on its orientation to loading and the body site from which it was excised. Other investigators have studied the mechanical properties of skin. Findlay (1971) studied the torsional properties, Hickman et al (1966) the compression properties; Evans (1973); Kenedi et al (1975) examined the time required by skin to recover from a period of maintained stress. They found that skin will not return to its original condition unless it experiences the stress and strains to which the skin is normally subjected. This may have important implications for the period of time required for pressure relief of supporting tissues.

Other studies investigated the response of skin and superficial tissues to repetitive loadings. It was concluded that application of cyclic loading or deformation of tissue produces an apparent change in the mechanical characteristics which is related to the loading frequency and time the load is applied (Evans, 1973). This finding is supported by the work of Brand that showed correlation between repetitive mechanical stress with specific threshold levels and the ulceration of tissue in the foot pads of rats. It was concluded that so long as the repetition of stress stays below certain threshold levels of magnitude and frequency, the inflammatory process seems to be associated with a protective hypertrophy (Brand, 1976; Brand and Mooney, 1977). Again, this has important implications relative to the etiology of pressure sore formation in the seated person, especially if they are active and at risk of subjecting their supporting tissues to high repetitive loads, without giving the tissues sufficient time for cellular recovery.

When viewed as a whole, the body tissues are composed of different structures that exhibit inhomogeneity as a mechanical structure. On the macroscale there are clear divisions between the skin, the underlying fat, fascia, muscle and bone. At the various interfaces there are inhomogeneities in the distribution of stress and strain (Gibson et al, 1976). Many of the arteries that supply the skin are at right angles to the boundary between the skin and subcutaneous tissues. Distorting shear forces acting in the plane of the skin could produce sufficient deformation to cause reduction in the nutritive supply to the skin.

At the microscopic level skin consists of collagen, elastin, and reticulin - embedded in a viscous ground substance (Gibson et al, 1976). Survival of the structure depends on the living cells within the skin. Mechanical stresses which exceed some critical threshold may cause micro-circulatory failure and if protracted may lead to tissue damage and necrosis. Less than optimal systematic conditions may contribute to this process by lowering the threshold at which irreversible damage occurs.

And finally, models are being developed in an effort to describe and predict the quantitative relationships between stress, strain, time, and the physiological factors at play within the tissue structures (Hickman et al, 1965; Brown, 1971; Pawan, 1971; Gibson et al, 1976; and Bennett, 1975, 1979). At this time, there is still not a quantitative method to make accurate predictions regarding the viability of supporting tissues at the body support interface of the seated or lying person. However, many of the parameters that have been studied have significantly increased our knowledge as to the aetiology of decubitus ulcers.

3.3. DEVELOPMENT OF CONCEPTS ON THE AETIOLOGY OF DECUBITUS ULCERS

Many investigators have published on the pathogenesis, prevention and treatment of pressure sores (Rudd, 1962; Exton-Smith, 1961; Lindan, 1961; Merlino, 1969; and Guthrie and Goulian, 1973). An accepted theory is that pressure ulcers are primarily the result of tissue death (necrosis) which has occurred in response to externally applied mechanical loads. The structure of normal skin and physiological processes involved in maintaining healthy tissue are fairly well understood (Ferguson-Pell, 1977). In contrast, the exact causes and mechanisms of soft tissue breakdown resulting in pressure sores are not as well known. During normal activities such as sitting or lying, relatively small volumes of flesh are compressed between the internal bony skeleton and the external surface. Since most of the body weight is carried by the skeleton, extremely high tissue stresses can be generated. Therefore, decubitus ulcers (pressure sores) are assumed to be caused by pressure-induced vascular ischemia resulting in tissues being deprived of oxygen and nutrients. The non-rigid walls of blood and lymph vessels collapse under pressures that are higher than that of the fluids inside. Also mechanical deformations of the flesh due to high levels of sustained loads or more moderate repetitive forces are of importance in producing tissue damage (Brand, 1979)

Studies have focused on the biomechanics of pressure sore formation. Investigators have studied blood flow and tissue mechanics in attempts to quantify the relationship between externally applied loads and the internal stresses that result in cessation of blood flow. The classical work done by Kosiak (1959, 1961) extended the studies of the variables governing soft tissue breakdown to include time and pressure. While Kosiak's important inverse relationship between time, pressure and breakdown has considerable variances associated with it, the clinical significance was shown by Rogers et al, (1974) and Reswick and Rogers, (1976).

Many studies have been designed with animals in order to determine the amount of externally applied pressure that will create tissue breakdown. As a result, substantial data has been collected which indicates that pressure ranges from 30 mmHg upwards are necessary to produce tissue breakdown (Kosiak, 1959; Dinsdale, 1974; Cochran and Palmeri, 1980).

These early studies and subsequent clinical experiences tended to evolve the common belief that continuous application of pressure above a certain level will impair tissue perfusion and result in ulcer development. The pressure/time relationship proposed by Kosiak (1961) and later by Reswick and Rodgers (1976) has spurred the development of a number of seat cushions that have the ability to minimize the maximum pressure under the ischial tuberosities to 60-80 mmHg (Souther et al, 1974; Ferguson-Pell et al, 1980b; Cockran and Palmari, 1980). These pressure levels have become generally accepted as desirable safe levels for people sitting with insensitive buttock skin. It further assumes that relief of sitting pressure will be obtained at least several times an hour by decompression or shifting of the torso.

Reichel (1958); Bennett (1971, 1972); Dinsdale (1974); Chow and Odell (1978) and Bennett et al (1979, 1984) have researched the role of shear stress. Their work suggests that shear forces can add significantly to the effect of externally applied normal forces towards occlusion of blood flow in soft tissues. However, it remains unclear as to exactly how these findings can be related to buttock tissues under stress.

As these early studies became publicized subsequent investigators began to appreciate that tissue breakdown is probably a multi-faceted process. The additional variables identified included such factors as; temperature, humidity, general medical condition, local tissue integrity and viability, age, activity level, nature of the tissue loading, altered sensation, as well as psychological and psychiatric factors. However, work in these areas has not resulted in clear qualitative relationships between the variables and ulcer formation. Therefore, clinical indicators or guidelines for acceptable values for these parameters do not exist.

As indicated, rat studies done by Brand (1976) has contributed to the knowledge about the effects of repetitive stress on soft tissue breakdown. In Brand's experiments, repetitive sub-critical loads were applied to the rats foot pad over a period of three weeks in different patterns. These loadings created necrotic areas in the soft tissue that had characteristics similar to those of pressure sores. Of particular importance was the finding that during the stressing of the soft tissue, an introduction of rest intervals followed by restressing caused the soft tissue region to hypertrophy and become capable of bearing much greater external loads than before. Manley (1980) also conducted similar studies involving repetitive mechanical stress. The extent to which this 'tissue conditioning' phenomenon may apply to the buttock tissues of an active spinal cord injured person is unclear. One possibility is that an individual may inadvertently subject their tissues to repetitive sub-critical stresses that results in 'work toughening' whereas another person may exceed the critical threshold and progress to soft tissue breakdown.

The role of skin temperature and perspiration in the process of soft tissue breakdown has also become an area of research interest. The soft tissue has been modelled as a chemical engine in which slight increases in the operating temperatures in the environment produce dramatic effects by increasing the metabolic demands of the cells in the local region (Hyman and Artique, 1976). The use of temperature change as a predictive tool has been explored in several laboratories. Infrared thermography has been employed to study thermal effects in the skin in an effort to monitor its responses to external stresses (Brand, 1976; Hahn and Black, 1977 and Manley and Darby, 1980). It has also been used to monitor the rate of healing of established pressure sores (Barton and Barton, 1973) .

Paterson questioned the dominant importance of pressure as the causative factor in formation of ulcers. He argues that several investigators (Dinsdale, 1973 and Danial et al, 1981) have shown that the pressure that develops ulcers under controlled conditions is higher than the acceptable pressure level (60-80 mmHg) that will result in an ulcer-free condition in patients (Paterson, 1984). The growing number of reports that show no correlation between level of spinal lesion and formation of pressure sores casts further question on the static pressure theory. Paterson concludes that a significant factor is shear and oscillatory changes in pressure which may result in peaks much greater then average. He makes the point that paraplegics are more active in body movements which results in more shearing and repeated dynamic pressure loading, which explains the unexpected higher incidence in ulcer formation when compared to tetraplegics (Paterson, 1984). Paterson's argument that the tissues of paraplegics experience higher shearing and repeated dynamic loading appears a logical conclusion from his review of the relevant finding; however, several questions remain. As discussed above, Brand's experiments indicate that 'work toughening' occurs as a result of repetitive dynamic loading. Paterson's argument does not appear to take this finding into account. Also, dynamic shear stresses that occur at the surface of the skin may have only a coincidental relationship to the deeper shear stresses that are thought to cause occlusion of the nutrient supply to the deeper tissues.

Neumark (1981) makes a strong claim that tissue deformation and not pressure are the prime cause of pressure sores, and therefore pressure should not be cited as being the major aetiological factor.

In summary, of all of the variables that can have a detrimental effect the supporting tissues, the concern for *excessive and prolonged interface pressure* has had the greatest impact on present day clinical practise. Unfortunately, most research studies isolated for examination only one or two variables, leaving the other variables uncontrolled or assumed constant. It is now difficult and often impossible to compare the results obtained in one laboratory directly with the results from another. As a result, now twenty years later, the general *trends* that have resulted from the pressure studies are clinically valuable, but agreement on specific threshold values is often found to be lacking. Futhermore, the relative contributions of temperature, humidity, shear stress, posture, deformity, dynamic stresses, psychological factors, and activity level are even less well understood. As a result clinical measurement tools and guidelines for their application are not generally available. And finally, of even greater concern is the apparent lack of general awareness as to the possible deleterious effects of these latter variables, especially if their effects can be cumulative.

3.4. TECHNIQUES FOR QUANTIFICATION OF THE BODY-SUPPORT INTERFACE VARIABLES

3.4.1. Identification of the Variables

Support of the body weight in lying, in sitting, and in standing necessitates the transmission of stabilizing forces via the supporting tissues to the support surfaces. As indicated previously, under certain conditions which are not fully understood, the stabilizing forces can damage the tissues to the point of ulceration. There is convincing evidence that mechanical stresses acting at the body support interface can cause damage to the underlying soft tissues. Subcutaneously this process is associated with the vascular occlusions, ischemia and eventual necrosis (Kosiak, 1961; Fernie, 1973; Brand, 1976; and Ferguson-Pell, 1977). Numerous techniques have been developed to study the mechanics of the interface stresses and the detrimental effects caused to underlying tissues.

Early efforts focused on developing techniques to quantify normal stresses and temperature/humidity effects. More recent work has been directed towards the measurement of tissue distortion or changes in tissue shape. In parallel with efforts to study interface parameters, work has been done on studying the underlying interstitial and arterial systems. For example, studies on alterations to blood supply, lymphatic drainage, biochemical changes and transcutaneous oxygen tension (tc PO_2) have been undertaken.

Since this study investigates the contribution of posture and deformity to interface stresses, an appreciation of the attributes and limitations of the relevant laboratory techniques used in previous studies is essential. The following review is intended to be selective rather than exhaustive, since many of the earlier methods used are now more of historical interest than of practical value to current work. Therefore, emphasis will be placed on the more recently reported techniques that appear to have particular relevance to the objectives of this study .

3.4.2. Normal Stress (Pressure) at the Body/Support Interface

Ferguson-Pell (1980) outlined the analytical criteria that should be considered when designing or assessing devices for use in the measurement interface pressure. Although a number of assumptions were made that may or may not apply to a particular interface situation, the criteria serve as useful guidelines. His specifications for a general purpose transducer for use in resolving pressure differences across non-uniform pressure surfaces without causing undue perturbating artifacts are as follows.

> A suitable transducer should have an aspect ratio (diameter: thickness) of 1:30, yielding a maximum diameter of 14 mm and 0.5 mm thickness in order to adequately differentiate peak pressures. It must have a low response to non-axial stress and to hysteresis effects, particularly when curved surfaces are present. The device should respond independently of environmental temperature and humidity conditions. Calibrations should be possible in a manner that is comparable to the actual interface usage conditions. Finally, the device should be reasonably priced, relatively easy to use, and be sufficiently durable to withstand the rigors of clinical usage.

Lindan et al (1965) pioneered the pressure mapping field using a device termed the "bed of nails". The device consisted of spring-loaded nails mounted on a 2 cm matrix to form the contours. Measurement of the nail movement (1000 for supine subjects, 150-300 for sitting subjects) against calibrated springs gave the contour information. However, the linear movement devices of this kind do not respond well to tangential forces, which causes the nails or sliding elements to jam and thereby limit their vertical movement.

Fernie (1973), in his Ph.D. dissertation, prepared an exhaustive analysis of the transducer technology used prior to that time. Of particular note was the early work of Swearingen et al (1962) who used absorbent paper over inked corduroy cloth to obtain the pressure distribution of the buttock and thighs of a seated person. Frisina and Lehneis (1970) used an acid indicator. Brand (1969) used microcapsules to determine pressure distributions under insensitive feet. All of these systems, although ingenious in their time, were sensitive to environmental conditions, awkward and time consuming to use, and most responded to the time rate of loading and non-axial forces.

Aronovitz et al (1963) reported on the early design of a pneumatic cellular matrix system that was used to measure isobars of pressure. This early technique which has since been refined forms the basis for a number of commercially available pressure monitoring systems.

During the late 1960's and 70's a host of transducers were developed that relied on either resistive, inductive, or capacitive electrical principles. The basic approach is to deform an elastic material between conductive plates and measure the response electronically by resistive, inductive, or capacitive techniques (Ferguson-Pell et al, 1976). Problems associated with the early designs were hysteresis (resistive), electronic noise (inductive) and nonlinearity (capacitive). Bush and Brooks (1969) reported on the use of a multilayer capacitive device. The development and use of strain-guaged transducers was reported by Fernie (1973). Maslin (1969) detailed the sources of error in the "Katie" beam design as did Fernie (1973) on the modified version. Fernie demonstrated the problems of in-plane forces corrupting the normal forces.

None of the miniature electronic devices have been refined to the point of a self-scanning matrix system that is commercially available for routine clinical use. A partial exception to this statement is the 64 cell (8 x 8) matrix reported by Drummond et al, (1982b). They constructed a pressure scanner which used 64 strain guage resistor transducers mounted on an aluminum plate. The limitation of the approach is its inability to monitor curved surfaces and the cost and complexity of the electronics. The device was never refined to the point of commercial availability, therefore no confirming studies have been possible by other investigators.

Patterson and Fisher (1979) studied the accuracy of five different miniature transducer designs. They placed the transducers under a pressure cuff located over various bony and soft tissue areas of the leg. The average error between the transducers before placement varied from 20.4 to 54.9 mmHg depending on hardness of the underlying surface.

A number of attempts have been made over the years to use barographs for monitoring pressure patterns. Minns (1986) reports on the design of an 'Ischiobarograph' which used reflected light, a television camera, and a colour monitor to display the results. The limitations of this approach is that it requires the patient to sit on an artificially flat surface. Therefore, monitoring of pressure directly under a person sitting on a cushion in their wheelchair would be very difficult, if not impossible.

Holley et al (1976) reports on the development of a water filled, air tight transducer connected to a physiological pressure transducer. A 10 cell version was constructed using a ten-way valve. The system was used successfully to compare the pressure distributing qualities of three different cushions. This approach was never defined for use by other researchers or clinicians.

Mooney et al (1971) described the use of a copper grid on either side of the inner surfaces of the flexible PVC air-tight pneumatic cell. When the external pressure exerted on the sensor was equivalent to the internal pressure in the cell the copper grid would make or break an electrical contact. This concept has received widespread use in several commercial versions (Scimedics Pressure Evaluation Pads in the USA and Talley Skin Pressure Evaluation in the UK). The device uses only one large sensing area (7 cm^2). This necessitates moving the patient if multiple areas under the patient need to be monitored. The need to move and reposition patients is probably one of the main sources of errors in pressure monitoring (Bader, 1982). Mayo-Smith and Cochran (1981) offered a partial solution to the problem by developing a manifold system that allowed connection of five single-cell pneumatic transducers of the Scimedics design.

Garber et al (1978) describe a 12 x 12 air cell pressure pad with pneumatically controlled switches. The 144 sensing cells in the matrix pad have a 22 mm diameter spaced on 29 mm centres. Each cell contains a pair of silver contacts connected by flexible silver leads to a data output module. Its make or break state is monitored by 144 lights on the readout display. The cells are inflated until all the contacts are broken and the display lights are out. The air is then slowly bled from the cells and the pressure noted when the lights come on, i.e., contact is made within the cell. It is claimed that the switches remain in contact with the cushion and patient for moderate changes in body contour. For any particular supply pressure a map of cells is produced for which the applied interface pressure exceeded the supply pressure. A maximum error of less than 5 percent in clinical situations was determined (Garber et al, 1978).

The commercial version of the system is called the Texas Interface Pressure Evaluator (TIPE).

Lincoln et al, (1988) report on the development of an electronic interface for the TIPE, designed to be used with an IBM PC or PC compatible. The interface electronically monitors the pressure profile of the 144 cells of the TIPE and displays the results in coloured graphics on the computer screen. They emphasize the potential for error in using absolute values and indicate that more accurate information results when relative interface pressures are used.

Although conceptually advanced, several problems have been encountered with the TIPE system. First, the silver contacts rapidly deteriorate and the pad gives spurious results. Secondly, the inflated cells throughout the matrix can act as an air cushion, which is likely to reduce the applied pressure in adjacent areas leading to inaccurate recordings (Bader, 1982). Calibration of the TIPE system is also difficult since a calibration instrument does not come with the system. Measurements are limited to a maximum value of 150 mmHg.

Bader (1982), Bader et al (1984) and Bader and Hawken (1986) report on the development and use of a pressure monitoring device termed the Oxford Pressure Monitor (OPM). The OPM has been designed to eliminate the problems inherent in the TIPE device; namely, short pad life due to breakage of electrical connections and the air cushion effect. Briefly, the OPM uses a 3 x 4 cell pneumatic matrix transducer pad without internal electrical contacts. The control system continuously monitors the pressure gradient within the cell and can detect the point at which internal and external pressures equalize. It then records and displays the pressure at that point. Since the OPM has been chosen for use in this study, it will be described in more detail in Part 2.

Several investigators have done comparative studies between various types of transducers, interface materials, and surface configurations. Reger and Chung (1985) compared three commercial pneumatic pressure monitoring devices; the Scimedics pressure guage, the TIPE, and the Oxford Pressure Monitor. They calculated, measured and compared pressure values under three different laboratory conditions, each designed to simulate different loading configurations at the body-seat interface. The tests simulated point, planar, and curvilinear loading conditions. The calculated values of the interface pressures were based on measurement of the contact area and the normal load. The point loading was done against a rigid surface, the planar loading between pneumatic bladders and the curvilinear test was done on polyurethane foam. In no case did the calculated values equal the measured values, and in some cases the difference was ten fold. Good agreement between transducer types was found in the planar loading condition in which two opposing pneumatic bladders applied the planar loads. The results clearly suggest that interface configuration and the compliance of the underlying surface materials are important considerations when selecting and using pressure measurement devices. Also, caution must be exercised when interpreting the absolute values of any of the existing pneumatic devices.

Reddy et al (1984) compared the performance characteristics of air cells and a cluster of minature semiconductors transducers, sandwiched between materials with differing mechanical properties. They found that in all but one exception the transducers gave significantly higher measured readings when compared to the actual calculated or normal applied stress. The transducer performance was highly dependent on the properties of the interface materials and on the ratio of transducer surface area to the contact area of the interfacing materials. They concluded that the perturbation effect of the transducer caused local concentrations of stress depending on the compliance and thickness of the transducer. The problem tends to be minimized if the transducer surface area is either very small, or is equal to the area of the interfacing surfaces. The 'enveloping' properties of the interface materials seem to be an important factor effecting transducer performance. They suggest the use of a correction factor when making clinical pressure measurements - unless only comparative measures using the same seating materials are required (Reddy et al, 1984).

In a second study, the performance of a skin surface mounted air cell transducer (Scimedics) was compared to a subcutaneous wick catheter device. Wick catheters are widely used for measuring *invivo* subcutaneous interstitial pressure (Snashall et al, 1971). They found good correlation between the external air cell pressures and the indwelling catheter pressures. They proposed that the favorable results are due to the excellent compliance properties of human tissues under low strains in which the tissue acts to distribute the load evenly over the transducer. They caution that air cell transducers, in general, reflect average rather than peak pressures within there measurement area, and any relatively stiff interface cushion material, clothing, prominent bony areas may tend to drastically distort the results (Reddy, 1984).

In summary, considerable research and clinical experience have been gained in the measurement of interface pressures. Unfortunately, a pressure transducer/monitor device for general purpose use that meets the criteria proposed by Ferguson-Pell, (1980) remains to be developed. Unless one is prepared to undertake the design of a device to meet these guidelines, it then becomes a matter of choosing an existing approach that presents the least number of compromises for the task at hand. Selection decisions should be based on a knowledge of the limitations of existing monitoring techniques, particularly their shortcomings in repeatability, durability, linearity and accuracy. The above design criteria and laboratory comparisons suggest that air cell devices are the most attractive from the cost, durability, availability, and performance viewpoints. Although not offering all the desirable features set out by Ferguson-Pell, pneumatic transducers appear to offer several practical advantages with the least number of compromises. However, caution must be exercised when interpreting pressure values, especially when used in non-planar environments consisting of materials with compliance properties significantly different from human soft tissue. Use of absolute values for research and clinical decision making should be done with extreme caution. Findings to date suggest that the most useful (accurate) general purpose application of the air cell transducer is obtained when comparisons are made using the same interface materials. That is, comparative values rather than absolute values are likely to be more meaningful (Reddy, 1984). Accuracy of comparative values is enhanced when the interface environment (cushion material, clothes, etc.) is kept constant. Interface materials that have mechanical properties that most closely approximate the properties of human skin will yield transducer values that are most closely representative of the subcutaneous interstitial pressures (Reddy, 1984).

3.4.3. Shear Stress at the Interface Surface

Soft body tissues are readily deformable but practically incompressible, i.e., the Young's Modulus, E, is very low and the bulk modulus, K, is very high. It has been shown that localized normal pressure of sufficient magnitude and duration can cause mechanical damage through the occlusion of blood vessels supplying the supporting tissues. Interface friction or shear stress acting tangential to the support surface has also been identified as an important factor in the formation of ulcers (Riechel, 1958; Chow, 1974; Dinsdale, 1974).

Husain makes the point that hydrostatic pressures (with equal components in all directions) cause little or no deformation. Body tissues can tolerate 1,655 kPa (240 psi, 12.4 mHg) of hydrostatic pressure with no difficulty; whereas a uniaxial pressure of less than 6.7 kPa (1 psi, 50 mmHg) will induce pathological changes (Husain, 1953). Based on this concept Chow and Odell (1978) argue that since the stress observed in the buttocks can be considered to be a combination of shear and hydrostatic; and since the hydrostatic is relatively harmless to biological tissues, then the harmful stress must be shear. "Shear stress is involved in uniaxial pressure, localized pressure, and any non-uniform pressure distribution or any pressure that causes distortion. It does not matter which quantity one chooses to quantify this harmful stress, because uniaxial stress and shear stress can be converted one to another" (Chow and Odell, 1978). They proceed to develop a shear stress model using finite element analysis and Mises stress equations. Based on the results of the model study they state the following conclusions. "Hydrostatic stress causes no distortion to shape and very small changes in volume. The distortion of the tissues is often more severe at internal locations than at the surface of the buttocks." They further indicate that the absolute value of one pressure measurement on the surface is not as important as an indication of the uniformity of the pressure distribution. This suggests that measurements at multiple locations are necessary so that gradients in pressures can be measured (Chow and Odell, 1978). The pressure gradients may give an indication of the shear stress being induced in the underlying tissue. Chow and Odell conclude by indicating that interface shear is very difficult to study because it is a result of differential radial expansion between the buttocks and the supporting cush-ion, or cushion cover; which can be affected or limited by the coefficient of friction of the interface material (Chow and Odell, 1978).

The basic assumptions made in the above hydrostatic model is that the supporting buttock tissues are homogeneous and exhibit isotropic properties. Gibson et al, (1976) clearly demonstrated that body tissues, when viewed as a whole, exhibit inhomogeneity as a mechanical structure. This results from the varying properties inherent to adipose tissue, fascia, muscle, and bone. At the various interfaces there are also inhomogeneities in the distribution of stress and strain which are more typical of anisotropic structures than those that exhibit equal mechanical properties along all axes. It is postulated that the effects of structural inhomogeniety and anisotropic properties are further exacerbated by the reality that the buttock tissues are most often subjected to non-uniform interface pressures. In summary, conclusions drawn from a comparison of stressed buttock tissues to ideal hydrostatic models can result in misleading over simplifications; thereby masking the true complexity of the stresses acting upon the tissues and their nutrient supply.

Other investigators have indicated a concern about the effects of shear acting at the interface. Guttman emphasized the importance of distinguishing between purely vertical pressure and shear stress. "Shear stress is much more disastrous for it cuts off larger areas from their vascular supply" (Guttman, 1976). Reichel expressed an opinion that raising the head of a hospital bed by even a few inches was capable of producing sufficient shear force over the sacral area to deprive large areas of tissue of blood supply (Reichel, 1958). Roaf (1976) also believes that "avoidance of shear force is as important as avoidance of direct pressure". Bennett reported on the analysis of stresses that exist within prosthetic sockets. He analyzed compressive forces (Bennett, 1971), shear stress (Bennett, 1973) and the flesh reaction to contact curvature (Bennett, 1972). In the late 1970's Bennett et al developed a device that could simultaneously quantify external pressure, shear and pulsatile arteriolar blood flow. This development was important since it allowed the quantification of the shear effect and clarification of its relative contribution to ulcer formation. The measurement transducer had a shear-upon-pressure interaction error of less than 6% on pressure magnitude, which was significantly less than previous electronic transducers. The pressure-upon-shear interaction error was measured to be negligible. A specially designed photoplethysograph measured the change in blood flow in precapillary vessels or arterioles. The thenar eminence of 4 healthy subjects was chosen as the anatomical study site. Acknowledging the limita-tions of size and compliance mismatch with body tissue, and the differences that may exist between the thenar eminence and buttock tissues, Bennett makes the following concluding statements. "Only the existence of large pressure values permits the stable development of large shear values. It is the combination of pressure plus shear that is particularly effective in promoting occlusion. The value of pressure necessary to produce occlusion can be nearly halved when accompanied by sufficient shear. Pressure remains the primary force, not only because of the greater hindrance given blood supply by a unit of pressure as compared to a unit of shear, but also owing to the necessity of the preexist-ence of sufficient pressure to permit the stable development of shear" (Bennett et al 1979).

In a subsequent study, Bennett and Kavner used the same instrumentation to compare shear, pressure, and changes in blood flow between normals, geriatrics and ill patients and paraplegic subjects. The instrumentation was installed into a flat "plexiglass" seat which resulted in a "hard-surface" instrument. The subjects were freely placed on the test seat so that the three parameters could be measured in the vicinity of the ischial tuberosities. Again, given the limitations of the experimental setup, he reports the following conclusions. "Pressure and shear act in an additive fashion in terms of blood flow inhibition. In the case of roughly equal median sitting pressures (52-60 mmHg), median shear results for the paraplegic and quadriplegic groups are nearly three times corresponding normal values; and median blood flows are roughly one-third the normal values" (Bennett and Kavner, 1981).

Even though Bennett et al (1979) were unable to offer any specific causal relationship between the increased shear and lessened blood flow characteristics of paraplegic and geriatric groups, their findings are important. First it gives quantitative evidence that shear is a primary causitive factor and its related to the interface pressure. Secondly, the evidence presented indicates a marked difference in the shear stress values between normal and disabled subjects when the interface pressures are approximately equal. However they did not offer any definitive explanation as to why these differences may exist between the disabled and normal populations.

3.4.4. Temperature and Humidity Parameters

Protection of soft tissues over the buttock area is vital to people that spend many hours daily in a wheelchair or in lying positions. Changes, primarily increases, in temperature and humidity at the buttock-seat interface have been implicated as secondary causative factors in the generation of pressure ulcers (Trandel and Lewis, !975a). Rèsearch evidence has shown that the application of pressure reduces blood flow and affects skin temperature. Removal of the pressure leads to increased blood flow or reactive hyperemia. Goller et al have shown in healthy subjects that this results in a transient increase in the skin temperature or a 'thermal flare', (Goller et al, 1971). Also, the chemical engine model suggests that slight increases in operating temperature produces dramatic effects by increasing the metabolic demands of the cells in the local region. Incipient impairment of nutrient supply by the occluding pressures or for other reasons may place the cells at risk of survival.

Studies have shown that the tensile strength of skin is decreased by 75 percent with an increase in relative humidity of from 10 to 98 percent (Wildnauer et al, 1971). Clinically, skin with reduced strength due to a moist environment may be more prone to damage due to shear stress or abrasions. In addition, dry skin offers less risk to infection. Traditional nursing practice employs the principle, 'keep the patient dry'. Standard nursing procedures for detecting incipient pressure sores is a skin examination that especially searches for areas of redness (erythema). It is known that concomitant with such redness, and often persisting subsequent to its disappearance, is elevated skin temperature (Rogers et al, 1974).

Many investigators have attempted to use temperature as a quantitative index for detecting incipient sores, as well as studying the aetiology and healing of existing decubitus ulcers. Barton and Barton (1973) used infrared thermography to investigate the rate of healing of existing ulcers. Harris and Brand (1966) and Bergtholdt and Brand (1979) investigated the thermal response of insensitive feet and amputee stumps to repetitive stresses. Hahn and Black (1977) used thermography to detect thermal flare areas to assist in the design of seating systems for young paraplegic patients. The thermal flare was interpreted as a danger sign of incipient pressure sore formation.

Infra-red thermography is a technique of obtaining a picture (thermogram) of the distribution of temperature over the skin. Thermograms can be obtained by either using liquid crystals, a heat sensitive camera, or a thermograph. However, it is usually difficult to obtain absolute skin temperatures from it (Davis and Neuman, 1981). Traditional thermographic machines are bulky and expensive, and in practice a thermographic examination requires that the patient go to a special area which precludes widespread decentralized assessments. Davis and Neuman report on the development of a portable thermal vidicon camera that was used for the bedside scanning of new admissions to a geriatric facility (Davis and Neuman, 1981). In general, the thermographic approach requires the compilation of thermographic norms through screening a relatively large number of normal subjects. Subsequent thermograms that differ from the norm can then be interpreted as a danger sign. The precise quantification and correlation to actual tissue breakdown has not been clearly demonstrated using the thermography technique.

Localized skin temperature measurement using direct skin probes, thermistors or thermocouples have been demonstrated. Patterson and Fisher (1980) reported the use of a thermistor (Yellow Springs Model 427) to study the daily temperature patterns of a group of paraplegic subjects sitting on a 100 mm thick foam cushion. Their results indicated a slow rise in interface temperatures to near body temperature after approximately 2 hrs. They did not observe any rise in temperature following pressure relief as reported by several other investigators (Brand, 1976; Mahanty and Roemer, 1979a). In the majority of the studies that monitored skin temperature after applying a known pressure for a specified time, the skin temperature typically, but not always, rose to some higher peak value and then slowly declined toward its original value. The implicit assumption is that temperature change, which is relatively easy to measure, is a reliable indicator of the physiological responses of the underlying tissue, in particular the metabolic and circulatory responses (Mahanty and Roemer, 1980, 1981). Since these latter variables are more difficult to measure it is useful to be able to infer their responses from the easier skin temperature measurements.

Mahanty and Roemer report on the successful development of a mathematical model that relates measured thermal responses to physiological responses. To obtain the experimental thermal responses they applied pressures of 100, 200, and 300 mmHg for specified time periods to the trochanter area of normal subjects. The resulting model predicts the skin temperature after pressure release and evaluates the reactive hyperemia blood flow that is caused by the pressure application. From the results of the model they inferred the following upon the release of pressure; 1) the main factor causing the observed skin temperature increase was an increase in blood flow; 2) that even in those areas in which skin temperatures decrease, there may still be an increase in blood flow rate; 3) that the model could be used to predict individual subject results in a clinical environment; 4) extension of the model could eventually result in a quantitative clinical measure of the susceptibility of an individual to tissue trauma and the extent and cause of such trauma when and if it occurs (Mahanty and Roemer, 1979a).

In a subsequent study by the same group , they applied the model and compared the results of the previous study with normals to a small group of paraplegic subjects (Mahanty and Roemer, 1981). Eight thermocouples were used to measure skin temperature changes around the load application site (greater trochanter). They observed a similar trend as in the non-paralyzed (normal) subjects. They again concluded that peak temperature rise is dependent on applied pressure magnitude and duration, with higher peak values associated with the higher values of pressure and duration. In the previous study with normal subjects they observed increases in temperature in only 12 of the 18 subjects (Mahanty and Roemer, 1979a). Whereas the mathematical model predicted increasing blood flows even in the cases where temperature was decreasing. Clinically, this means that the direction in which temperature changes (increase or decrease) following pressure release may not always be a sufficient indicator of the direction in which blood flow rate is changing. Therefore, if temperature change is used clinically as a measure of hyperemia following pressure release, it may need to be used in conjunction with a mathematical model to estimate the direction of blood flow rate when decreasing temperatures are observed (Mahanty and Roemer, 1979b,). Another complicating factor is that Goller et al (1976) reported on the fact that skin redness and increased temperatures are not necessarily related.

The above studies suggest that there is a causative relationship between pressure and skin temperature values, but that the relationship is not a simple one and may not correlate directly with the physiological processes that take place within the underlying tissues.

Studies have also been done to compare the relative temperature and humidity effects of a variety of different seat interface configurations and materials. Steward et al (1980) reports on the comparison of 24 commercial wheelchair cushions. Brattgrad and Severinsson (1976) did a similar study, as did Fisher et al (1978) with five cushions. Stewart makes the point that neither study evaluated typical types of commercial cushions for sufficient periods of time to reach a steady state condition. Stewart also studied heat flux as an additional correlation for temperature. The Stewart study used commercially available transducers to measure temperature, humidity and heat flux. Two humidity sensors were used, one electronic and one chemical device. Atmospheric conditions in the free standing position were used as the base line to which subsequent values were compared (Stewart et al, 1980).

The results suggest that interface temperature, humidity, and heat flux vary with the type and specific design of each cushion. Basically, for non-critical applications, the relatively high temperature produced by foam-based cushions can be accepted because of the relatively low humidity build-up, convenience, cost and acceptable mechanical performance. Gel flotation cushions are good for controlling temperature but have a tendency to increase humidity. Water flotation cushions cause significant drops in skin temperature associated with high heat flux. And finally, the behavior of each cushion, especially in terms of humidity build up, is influenced strongly by the covering as well as the primary structural material for the cushion (Stewart et al, 1980).

Other investigators have also used quantitative methods to measure changes in interface humidity. Trandel outlines the development of a reliable humidity transducer which was used to study relative humidity (RH) changes with 28 subjects. The active part of the sensor was a 2.5 cm disc of blotting paper containing RH sensors for 20% to 90% in 10% increments. Each spot sensor is impregnated with a spot of cobaltous chloride solution in a concentration that changes from blue to pink at the specified humidity. The active element was sandwiched between two latex, open-cell foam rings with a filter disc used to protect the sensor from skin contamination. Accuracy of approximately 5% is possible. A special reader is used to read the RH values manually against a permanent colour guide using natural light. The author claims that the sensor is extremely pliable and convenient for determining RH in a range of 20-90 percent (Trandel and Lewis, 1975b).

Results showed RH readings varying from below room values to over 70 percent in the two areas measured; sacral and scapular regions. They reported that RH recordings were often higher for inactive patients than for those who were active. Unfortunately, no direct readings under the buttock area were made, possibly because of the rather large size of the transducer and the discomfort that would result.

To summarize, successful efforts have been made to quantify temperature and humidity parameters at the seat interface. Temperature measurement devices have been developed, ranging from thermography to thermisters and thermocouples. Most study results have shown an increase in surface temperature upon pressure relief, which appears to be related to an increase in the blood flow rate, i.e., reactive hyperthemia. Correlations between physiological parameters of blood flow and metabolic rate have been shown in a mathematical model but not without a number of unanswered questions. Extensions of the concepts to provide clinical useful measures and guidelines regarding the susceptibility of tissue trauma seem promising (Mahanty and Roemer, 1980). However, this development remains to be accomplished.

Relative humidity measures have been demonstrated which show increases over ambient values of 30-40%. The clinical significance of this information remains to be demonstrated. For example, none of the studies compared RH values under the buttock area of able-bodied people sitting for prolonged periods of time. Although high humidity has been shown to reduce tensile strength of tissue, it may add other properties, such as increased compliance, which could be advantageous under certain conditions of loading.

Parallel research efforts have shown the temperature and humidity effects due to various cushion support mediums and their covering materials. Again, temperature and humidity increases were observed, the amounts dependent on the cushion design and the porosity of the covering material. Guidelines have been developed for clinical decision making that can help in resolving clinical problems in which temperature and humidity extremes may impose additional detrimental factors (Stewart et al, 1980; Brattgrad and Severinsson, 1976).

None of the investigators debated the primacy of pressure and shear as contributing factors to the formation of pressure sores.

3.4.5. Tissue Distortion and Buttock Shape

Quantification of the relationship between pressure, shear, and blood flow has stimulated more recent studies in the area of tissue distortion or buttock tissue shape. Since the intrinsic effects on the tissue due to shear stress are difficult to measure, it is assumed that inferences can be made from the easier measurement of changes in tissue shape. The inferences are that tissue distortion can be directly associated with pressure and shear acting at the interface surface. Investigators have developed a number of techniques for quantifying the changes in the buttock shape while sitting on various cushion arrangements. Pressure distributions are also measured simultaneously using techniques previously outlined.

Reger and his colleagues reported on the development of a modular contour gauge consisting of 64 linear potentiometers mounted on 50.8 mm centres, arranged in an 8 x 8 array. The probes on the ends of the potentiometers extend through the cushions, drilled out on the same 50.8 mm centres. The transducer array was multiplexed to 4 channels of a 12 bit A/D converter for computer scanning. Discrepancies of 21% between applied and measured loads were determined using the modified wheelchair cushion and contour gauge. Tissue deformation studies were conducted on three types of typical wheelchair cushions with three types of subjects. The cushion types were viscoelastic foam (7.5 cm thick), polyurethane foam (7.5 cm thick) and a bladder cushion (10 cm Roho). The subjects were normal, male and female and paraplegic males. Curve fitting techniques were used to extrapolate the data to give a 33x33 array from the 8x8 measurement array. Pressure recordings were done using the Texas Interface Pressure Evaluator (TIPE).

The data was plotted to give a three dimensional model of the entire cushion surface. Comparisons were made between nonweight bearing and weight bearing shapes and between normal and paralyzed subjects. Characteristically a single peaked surface was seen for the paraplegic, whereas the normal subjects showed double peak surface contours with peaks following the contour of the ischial tuberosities. Furthermore, the results showed inversely proportional relationships between pressure and contour for four cushions, and a direct relationship for the bladder cushion (Roho). The researchers indicated that the instrumentation was useful in showing the extent and rate of tissue deformation (Reger and Chung, 1985). It is suspected that the linear potentiometers exhibit similar deficiencies to the "bed of nails' design when shear loads of any magnitude are applied.

A subsequent study by the same group investigated the contour-pressure relationship on four foam cushions of different densities, with six normal subjects. Pressure measurements were done using the Oxford Pressure Monitor. The measured contour of the unloaded buttock was used to contour the foam cushions using a CAD/CAM process. An interactive technique was used to obtain the unloaded shape of the buttock in foam under weight bearing conditions thereby minimizing tissue distortion.

The results showed reduced peak pressures as a result of using the interactive contouring approach. Variances between foam types and individual subjects were observed. They also observed that peak pressure areas occurred in the pubic and perineal areas rather than under the ischial tuberosities. The investigators concluded that improvement of the resolution of the contour gauge to allow more precise body measurements was necessary to resolve the problem (Chung et al, 1988).

Laliberte and Masiello (1985) studied the use of an ultrasonic dimension gauge, a standard gel pad, and a pressure monitor to obtain the simultaneous measurement of deformation and pressure contour at the buttock-cushion interface. The system produced two-dimensional contours for the buttock shape. The technique is based on an ultrasonic pulse traveling in a multi-layered medium which is reflected at each interface that has dissimilar acoustic properties. Using the reflected echos it is possible to measure the deformation contour of the gel surface under weight bearing conditions. The pressure contour was measured using the TIPE monitor. Three normal subjects were used to demonstrate the capabilities of the system.

The investigators concluded that the technique will be useful for assessing the degree of asymmetric loading and in identifying the highly localized pressure distributions of patients with flaccid tissue covering bony prominences. The measurements can be used to identify those individuals who produce substantial indentation of their seating surface and therefore may be considered at risk for developing ulcerations (Laliberte and Masiello, 1985).

Although these developments have made progress toward measuring and recording changes to buttock shape under weight bearing, the correlation to pending tissue damage remains unclear. The instruments used to measure pressure are known to be unreliable especially when used on curved surfaces. The ultrasonic approach appears to be limited to gel cushions, since ultrasound does not propagate readily through foam materials. This will greatly limit the clinical application of the device. It remains uncertain that the CAD/CAM contouring technique can be simplified to the point of clinical practicality. The time and associated cost must compete against already available commercial cushions that yield pressure-relieving capabilities similar to those obtained through CAD/CAM custom contouring. The use of free-hanging shapes as the shape ideal for weight bearing is an assumption that needs further validation.

In conclusion, although these early efforts at tissue distortion mapping appear scientifically stimulating, their long term clinical relevance and direct relationship to shear stress effects remains to be demonstrated.

3.4.6. Physiological Factors

Although not occurring directly at the buttock-seat interface, changes in the underlying tissue physiology as a result of the overlying surface stresses has been studied in relation to pressure ulcers. Measurements involving blood flow, clinical hyperemia, necrosis, infection, local anatomy, metabolic state, and transcutaneous oxygen tension have been undertaken. The details associated with these research studies extend beyond the scope of this work. However, since some of these measurements form the basis of our current clinical practice, a brief review of the salient quantitative techniques and the results derived seems appropriate.

It is known that pressure causes an ischemic event of some nature to the skin in the region of localized pressure points. However, the nature by which the surface stress invokes responses in the deeper tissues, specifically subcutaneous tissue and muscle, is less well known. More importantly very little information is available on the three dimensional full thickness changes which occur since most of the studies have been done on individual tissues. The initial problem seems to be that of disorders of the microcirculation system with partial or complete occlusion of the vascular lumens involved. Variations in the intensity of ulceration appear to be due to the extent of the vascular occlusion. In experimental settings ulceration usually occurs about four days after pressure application, although microscopic changes occur in the muscle after 24 hours, or sometimes less. Thus the pathological changes give some indication that a vascular process is occurring.

It remains unclear whether muscle and skin respond to stress in different ways. There is evidence that suggests that some pressure sores may actually begin in the muscle and then progress upward towards the surface (Davis and Newman, 1981). These studies have been done using various models and test species with varying end points, including necrosis, ulcer formation, skin hyperemia, blood flow change, and transcutaneous PO_2 (Holloway, 1984a). The lack of standardization has made it difficult to compare results of physiological studies and thereby draw specific causal relationships.

Quantification of blood flow has been done using digital photoplethsmography (Bennett et al, 1981), 133 Xenon radioisotope clearance (Holloway et al, 1976) and a lazer doppler system (Holloway, 1984a). Bennett's work using the blood flow plethsmography combined with pressure and shear recordings suggests that metabolic changes, nutritional changes and the age of the patient may indeed be factors contributing to ulcer formation (Bennett et al, 1981).

The 133 Xenon clearance technique, first introduced by Sjersen (1961), has been used to study pressure loadings in normals and paraplegics. The technique uses small doses of 133Xe introduced into the skin. The blood flow rate is determined from the ^{133}Xe isotope disappearance using a variety of diffusion models. The method is eventually limited by the local retention of the isotope in the subcutaneous fatty tissues. Studies on the forearms of normal subjects showed an initial marked fall in blood flow with pressures in the range of only 10 mmHg, but then a plateau in flow with pressures up to 40 mmHg. Studies in the parasacral area of both normal and paraplegic individuals showed a similar pattern with basically no difference between the two groups, suggesting that control of microcirculatory flow in denervated areas is not markedly different from that of normal areas (Holloway et al, 1976). This finding seems to be in conflict with Bennett and his colleagues whose studies showed a marked difference between normals and paraplegic subjects (Bennett et al, 1981).

The Doppler method utilizes the Doppler shift of lazer light backscattered from moving red blood cells in the tissues being examined. It provides a continuous and non-invasive measure of blood flow of any exposed tissue surface. The technique was used to examine blood flow and its changes in susceptible areas in paraplegics before, during, and after sitting on several different wheelchair cushions. Preliminary results show that one can not generalize in terms of the causal relationship between specific cushions and changes in blood flow. Flow will decrease to varying low levels in some patients sitting on some cushions, with significant hyperemic flow following. In others, flow may remain high or even increase and not show any hyperemic following. These findings suggest that pressure readings alone may not be sufficient clinical indicators for cushion selection, and that non-invasive evaluation of blood flow may also be required (Holloway and Tolentino, 1984a).

Barbenel et al, (1976) described the development of an optical technique to study blood flow in abdominal skin folds. Compressive loads and displacements were applied to the fold via a circular anvil of one cm^2 area. Both the load applied and the resulting displacement could be continuously monitored. Light sampled from the transilluminated skin fold by a photodiode pick up was used to detect the change in blood flow. The study failed to disclose any level at which there as a marked and sudden decrease in blood content. A study with paraplegic patients showed similar results to normals. These findings appear to differ from those of Bennett et al (1984), and Daley and Chimoskey (1976) who showed a marked decrease in blood flow in geriatrics and paraplegics compared to normal values. However, the differences in the anatomical study sites and the methods of loading the skin may account for the different outcomes.

Another approach to studying physiological changes within supporting tissues is to measure the changes in the transcutaneous oxygen tension ($tcPO_2$) and partial pressure of oxygen (PO_2). Newsom and Rolfe (1982) describe experiments in which a 'Helligie Servomed Oxymonitor SM261' was used to incrementally load and measure PO_2 at the skin surface of three healthy subjects at three anatomical sites; the sacrum, the greater trochanters and the lateral aspect of the thigh. The applied pressure was derived through calculations over a range of 0-200 mm Hg. The results showed that $tcPO_2$ values reduce gradually at first and then at an increasing rate as the pressure is increased until eventual cut off. They did not disclose a plateau effect as found by Holloway (1984c). Although variations were found between the test sites and test subjects, all showed cut-off pressures in excess of 100 mm Hg (Newsom and Rolfe, 1982).

Bader and Grant measured $tcPO_2$ using a radiometer TCMI TC Oxygen Monitor in twenty debilitated subjects prone to tissue breakdown. The tissue load was applied incrementally to the sacral area until a maximum threshold of 2.7 kPa (20 mmHg) was reached and the load was removed. Pressure was measured using a single cell from the Oxford Pressure Monitor. The unloaded resting $tcPO_2$ at the sacrum ranged from 6.7 to 11.7 kPa (50 - 88 mmHg) The values of the interface pressures which reduce the $tcPO_2$ value to below 2.7 kPa (20 mmHg) ranged from 3.6 to 14.4 kPa (27 - 108 mmHg).

Similar to other investigators, they observed that the $tcPO_2$ decreased gradually upon initial application of pressure and then at an increasing rate as the applied pressure was increased. The $tcPO_2$ recovered to the unloaded resting levels following removal of the load. They also found considerable variation between individuals with no obvious trends with respect to age, sex, or clinical diagnosis. They did observe relatively low pressure threshold values in three subjects who had a history of pressure sores. Generally, recovery of $tcPO_2$ to resting levels occurred within several minutes. The investigators acknowledged the potential effects of repeated loading and recovery and its possible role in cumulative tissue breakdown (Bader and Gant, 1988).

In summary, it is evident that a wide variety of research studies have been undertaken, some conflicting, regarding the physiological responses of tissue to external loading. The reported safe threshold pressure levels range from average capillary pressure of 4.3 kPa (32 mmHg) to in excess of 100 mmHg provided a relief regime is adopted. Clinically, the author has measured pressures of 100-125 mmHg with no consistent pressure relief regime and no reported pressure sore problems. This rather wide range of often conflicting research and clinical experiences has caused confusion for clinicians and people prone to pressure problems. At best it suggests a wide variation of 'norm' exists, or at worst that the reporting investigators have not been measuring and comparing identical variables. It also gives rise to the question as to whether normal pressure loadings can be compared to those that have significant shear components, such as in the case of buttock tissues of a sitting person. The studies of Bennett and Chow strongly suggests that shear is a highly important factor, the amount of which increases the effect of the pressure (Bennett et al, 1984; Chow and Odell, 1978). Finally, are skin surface measures of blood flow, by whatever technique, representative of the deleterious effects that are occurring in deeper tissues, such as muscle and bone? Can the measured results from one anatomical site be readily used to infer events which will occur at another site? Or, can the physiological studies done with animals be readily equated to events that will occur in humans, especially when dynamic loading conditions are considered? Fernie (1973) urges caution in making these types of assumptions.

Comparison and integration of the results from physiological studies to yield concise clinical guidelines has not readily occurred. It appears that this is largely due to the lack of standardized physiological models and repeatable procedures. The field seems to beg for this standardization so that consensus can be established and limited research resources directed to the remaining host of unanswered questions.

3.4.7. Mechanical Properties of Supporting Materials

The specific details on the properties of materials used at the body-support surface interface are beyond the scope of this study. However, for reasons of completeness the salient references will be briefly summarized.

The pressure distribution, heat and moisture dissipation and longevity properties of materials used to fabricate wheelchair cushions are important. Cochran and Palmieri (1980) identified the important mechanical loading properties required by materials used to fabricate wheelchair cushions, and the laboratory methods for testing the various properties. These studies concluded that a cushion design having properties offered by multiple materials would most likely yield an improvement to body supporting properties. Ferguson-Pell et al (1986), building on the previous work of Cochran and Palmieri (1980), provide guidelines for designing a multilayered seat cushion that possesses many of the desired mechanical properties.

The influence of environmental aging and time factors on cushion materials are also important. This is particularly vital from the viewpoint of follow-up planning in order to identify degradation of pressure distribution properties due to changes in mechanical properties prior to the onset of pressure problems. Noble and Goode (1985) conducted studies over a six month period on polyurethane foam samples. They showed changes that led to a 68% gradual decrease from the initial properties within the study period. In general, most polyfoam cushions today are considered to have a useful life of between 6 and 12 months.

As detailed previously, Stewart et al (1980), Brattgrad and Severinsson (1976) and Fisher et al (1978) studied the temperature and humidity characteristics of various cushion materials and their covering fabrics. They demonstrated that temperature buildup was directly related to the insulating properties of the material, whereas humidity build up was related to the permeability of the covering material. One surprising finding of the Stewart study was the high humidity buildup of the tufted balloon design. The ballon-type cushion was designed to allow air circulation between the balloons. However, it seems that the tufts squeeze together so that the effect is the same as sitting on a cushion of an impermeable material (Stewart et al, 1980).

Various studies have been done to evaluate and compare the pressure distribution, pressure redistribution, and pressure relief properties of various commercial cushions and related materials used for prevention of decubitus ulcers (Mooney et al, 1971; Souther, 1974; Peterson and Adkins,1982; Garber and Krouskop, 1984; Nelham, 1984; Lim et al, 1988). Trials have been done including normals, paraplegics, and geriatric subjects. Several of these studies have involved trials with modified foam cushions (Mooney et al, 1971; Ferguson-Pell et al, 1980a; Garber and Krouskop, 1984; Lim et al, 1988;). These studies have served to demonstrate the potential advantages in customizing support materials to yield improved pressure relief characteristics to meet individual needs. Unfortunately, most studies have not looked beyond maximum pressure in a neutral static posture for quantitative indicators of clinical acceptance. For example, shear stresses, dynamic loading and alternate postures have been virtually ignored. The point being that a particular cushion configuration may exhibit excellent pressure relief characteristics under static conditions, but the presence of elevated shear stress or sitting instability may well obviate the pressure relief advantages.

It has been generally accepted that shear combined with pressure results in distortion of the tissues, both internally and externally, which can lead to vascular occlusion and ultimate tissue breakdown. The Roho balloon design and the new Talley Medical bellows design offer low shear support surfaces due to the ease of lateral displacement of the air sacs. However, clinical problems have been reported by users of the Roho cushion regarding the lack of sitting stability. Also, minimal control was provided to prevent forward sliding of the pelvis into posterior pelvic sitting. More recent designs have attempted to resolve this problem through compartmentalizing the air cells in the cushion. Individually inflated segments can give higher resistance to movement without sacrifice of the pressure relief characteristics in critical areas. Also, accommodation of pelvic and hip deformities have been identified as desirable properties of cushion materials. However, most commercial designs assume a symmetrical pelvic alignment and therefore do not accommodate pelvic and hip deformities without inducing unilateral elevated pressure and shear stresses on the involved side.

The more sophisticated measurement technology and related mechanized means for customizing cushioning materials are not widely used in clinical environments. For example, application of the CAD/CAM approach is only in its infancy and may eventually provide the quantitative means of altering the shape of supporting materials to precisely meet individual needs. However, other more direct approaches are also emerging for rapidly shaping supporting materials to meet custom needs. The author has reported on the development of a cushion design termed the Foam-In-Place (FIP) system (Hobson et al, 1978). The FIP features the custom fabrication of a flexible foam cushion by allowing the foam to polymerize and rise directly against the person, suspended in an appropriate posture. Clinical evaluation of this development over the past 10 years with spinal cord injured individuals has shown pressure relief characteristics equivalent to the best commercial devices (Hobson and Nwaobi, 1985). However, since the support surface is contoured directly to the shape of the individual, it is postulated that the shear stresses are greatly reduced, while still providing good lateral stability and pelvic control. Since it is now routinely possible to fabricate seat and back cushions like the FIP in half a day, the more sophisticated approaches like CAD/CAM may be difficult to justify outside the research arena. These latter concepts remain to be scientifically and clinically determined.

3.5. MEASUREMENT OF CENTRE OF GRAVITY LOCATION AND DISPLACEMENT

It is convenient to conceptualize the effect of gravity on the seated body as a point though which the composite effect of gravity acts. This point in three dimensional space is termed the centre of gravity (CG). The CG may fall either inside or outside the body, depending on the position of the body segments in space. A line projected vertically downward through the CG is termed the line of gravity. The location of the line of gravity relative to the major joints of the body is immensely important from a biomechanical perspective. Generally, the magnitude of the moments due to gravity relative to the axis of joint rotation determines the magnitude of the muscle activity necessary to provide static equilibrium. Under most conditions the body strives to maintain alignment of the mass of the various body segments as close as possible to the axes of joint rotation in order to minimize muscle activity and thus energy expenditure. This condition biomechanically defines the "resting position" in most static body postures. Examples are the alignment of the centre of gravity of the trunk relative to the hip joint in upright standing, or the alignment of the trunk relative to the first sacral vertebrae (S1) during sitting, or the alignment of the gravity line of the trunk relative to the alignment of the pelvis (ischial tuberosities). Shifts of the gravity line away from the resting position necessitates intrinsic muscle activity in order to maintain biomechanical equilibrium. This biomechanical reality becomes vitally important if muscle function is impaired such as in the case of people with spinal cord injuries.

Analysis of the position and movement of the centre of gravity in sitting and lying postures has been used by several investigators concerned with pressure distribution. For example, displacements or shifts in centre of gravity can provide indications of the frequency and nature of pressure redistribution. This has been accomplished in most cases by projecting the two dimensional movement of the CG onto the standing or sitting surfaces.

In the sitting posture the location and movement of the centre of gravity relative to the anatomical areas at risk may provide useful information, especially when compared to normal data. The magnitude of the CG shifts may provide indications of the magnitude of the pressure relief to critical anatomical areas. Determination of asymmetrical CG locations may also indicate the presence and extent of skeletal deformities or other pathologies leading to asymmetrical sitting postures and elevated pressures. Analysis of movement of the CG may also provide insights into abnormalities of balance that can be associated with disease, muscle weakness or injury to the central nervous system.

Bardsley (1977) used CG displacement to study the relationship between trunk movements and pressure sore history in a group of spina bifida children. He compared the results of six normal children to those of 24 sitting disabled children, many of whom had a history of pressure sores. The instrumentation consisted of a force platform mounted on three load cells. Pen traces of the projected CG movement while the subjects underwent an external stimuli (watching TV) produced the measurement data. Unfortunately, there was limited information on the extent of the deformities and no real way of relating its influence on the CG movement pattern observed. Based on his results, which showed large variations within and between subjects, there was no significant difference observed in the movement patterns between normal and disabled subjects. It was further concluded that M/L (medio-lateral) movements are preferable to A/P (anteroposterior) movements in terms of relieving pressures over the critical buttock areas. The results also suggested that the extent of movement and incidence of pressure sores were related to levels of spinal lesion. Bardsley also used the same instrument to quantify the movement of people during sleep (Bardsley, 1977; Bardsley et al, 1983).

More directly associated with the goals of this study is the work at the Helen Hayes Hospital in New York State. Sequential student projects have resulted in the development of load cells and interfacing instrumentation that can be connected directly to an IBM PC computer. Building upon the previous work of Wheatley et al (1980), Wheatley (1982), and Grewe (1984), Snow (1985) describes the design of a load cell transducer device, specifically designed to compute and display the position of the centre of gravity of a person sitting in any wheelchair. This development will be discussed more fully in the following section since it has been adopted as part of the instrumentation for this study.

Ferguson-Pell and his colleagues report on the most recent study using the latest version of the same equipment. They monitored the trunk control and head position during reaching tasks of a small group of persons with head trauma. The four load cells were mounted under the wheelchair wheels. A commercial data acquisition system combined with locally developed software permitted the on-line computation, display and recording of CG or "centre of pressure" movement. Early results indicated significant differences in sitting stability between able bodied and disabled subjects; with large standard deviations and both intra and inter subject variability (Ferguson-Pell et al, 1988).

And finally, Reinecke et al reported results on studies relating centre of gravity shifts to tolerance of seated postures in four static sitting positions. They reported low back pain as the most prevalent form of discomfort followed by muscle fatigue. The CG was monitored by a force plate onto which the six commercial test chairs were bolted. Muscle activity was monitored by EMG activity of the erector spinae muscle. Among the several observations they noted that the seat pan pressure on the ischial tuberosities increased as the trunk posture changed from 70° to 90°; and decreased as the subject reclined to 120°. Unfortunately, the investigators did not report on the method used to measure the pressure distribution. Since the test subjects were all able-bodied subjects it can be assumed that deformity and other factors related to disability were not considered in the study (Reinecke et al, 1987).

In summary, monitoring of the centre of gravity (centre of pressure) location and pressure appears to be a viable method of quantitatively detecting postural differences between various study populations. Differences in the centre of pressure related to buttock areas of high risk or relative to the sitting balance point of the trunk and pelvis may have important implications relative to deformity and possibly the onset of pressure sores. The magnitude and directions of the trunk centre of gravity shifts can effect comfort, trunk alignment and most importantly critical relief of ulcer - inducing pressures. None of the reported studies have investigated the effects of deformity and centre of gravity shifts with respect to pressure distribution. These latter aspects are central to the focus of this study.

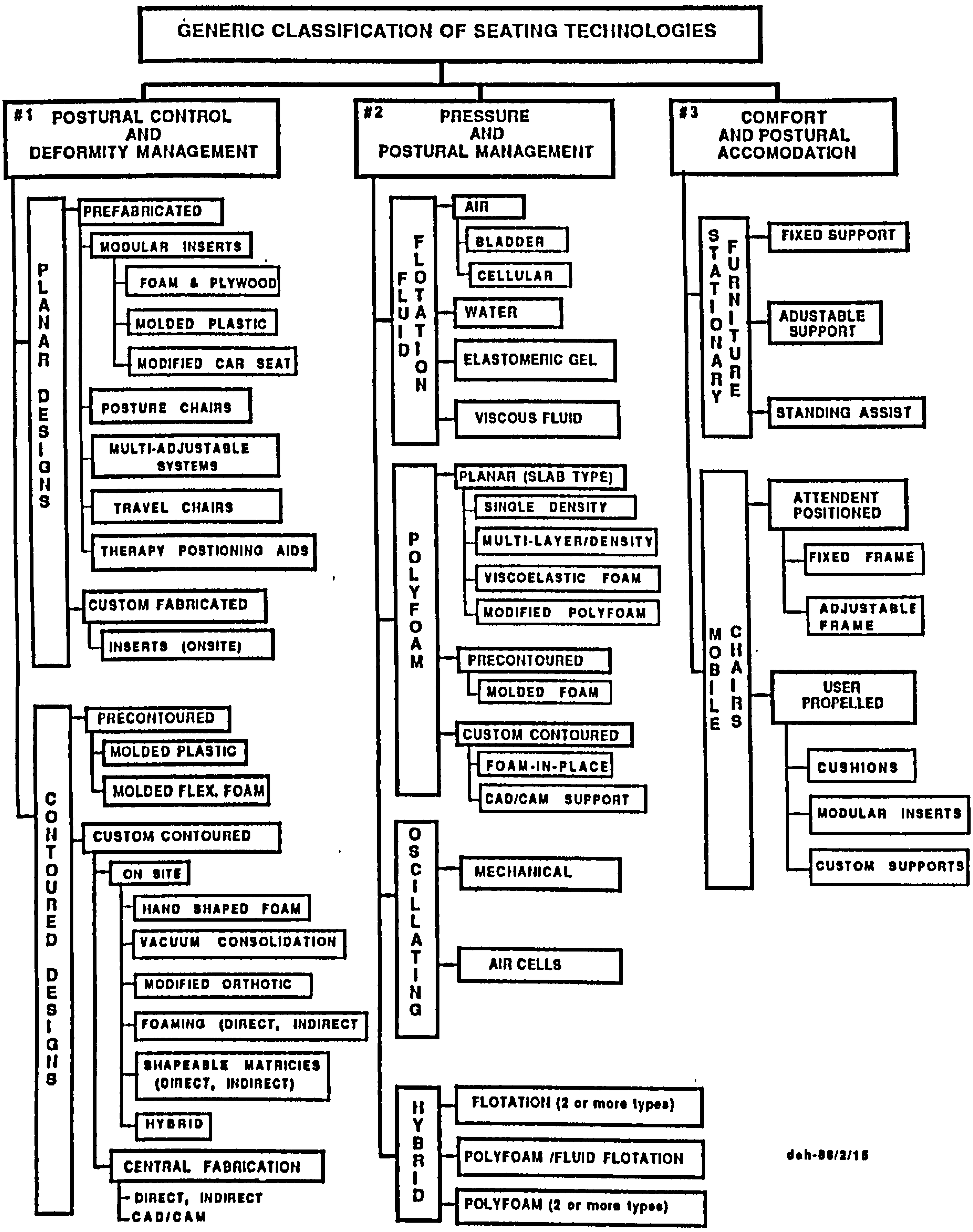


Fig 4.1(a) A generic classification of technologies used in the field of specialized seating and mobility.

CHAPTER 4. CLINICAL PRACTICE - A REVIEW OF TECHNOLOGY AND CONCEPTS

4.1. CLASSIFICATION AND DESCRIPTION OF SEATING DEVICES

4.1.1 Seating Technologies - Classification and Terminology

The rapidly increasing number of commercial options in seating and mobility present many new selection challenges for clinicians and consumers, especially for those first entering the field. No standard system of classification or terminology has been established in order to facilitate communication between researchers, clinicians, students, suppliers, and consumers. The classification scheme that follows is an initial attempt and therefore should not be regarded as definitive. The author has contributed to a comprehensive text on rehabilitation engineering. A chapter entitled ' Seating and Mobility for the Severely Disabled' has been accepted for publication (Hobson, 1988). The following section is based on relevant extracts from this more broadly based work.

The field of specialized seating is evolving along three distinct tracts, each of which have been guided largely by the needs of distinct user populations. These tracts are: a) seating for postural control and deformity management (cerebral palsy); b) seating for pressure and postural management (spinal cord injured); and c) seating for comfort and postural accommodation; (multiple handicapped and elderly) (fig 4.1a). For reasons of relevance and space only tract b) 'seating for pressure and postural management' will be presented. The classification scheme for tract b) builds on the scheme proposed by Garber, (1979, 84) and has been extracted from fig 4.1a as fig 4.1b.

Individuals with spinal cord injury are at risk of developing pressure sores in the tissues overlying the pelvic area. Many types of wheelchair cushions have been developed in an attempt to control pressure over the bony prominences of the pelvis. Dissipation of body heat to prevent moisture buildup and maintenance of postural stability are also important factors that have influenced cushion design. A wide range of materials, geometric configurations and mechanical technologies have been tested and used in cushion designs. Clinical experience clearly indicates that the requirements of individuals vary widely and that no single design meets every need (Garber and Krouskop, 1984).

Development of seating technologies designed for pressure and postural management has followed an evolutionary path somewhat different from the cerebral palsy group (postural control and deformity management). In general, cushion designs for

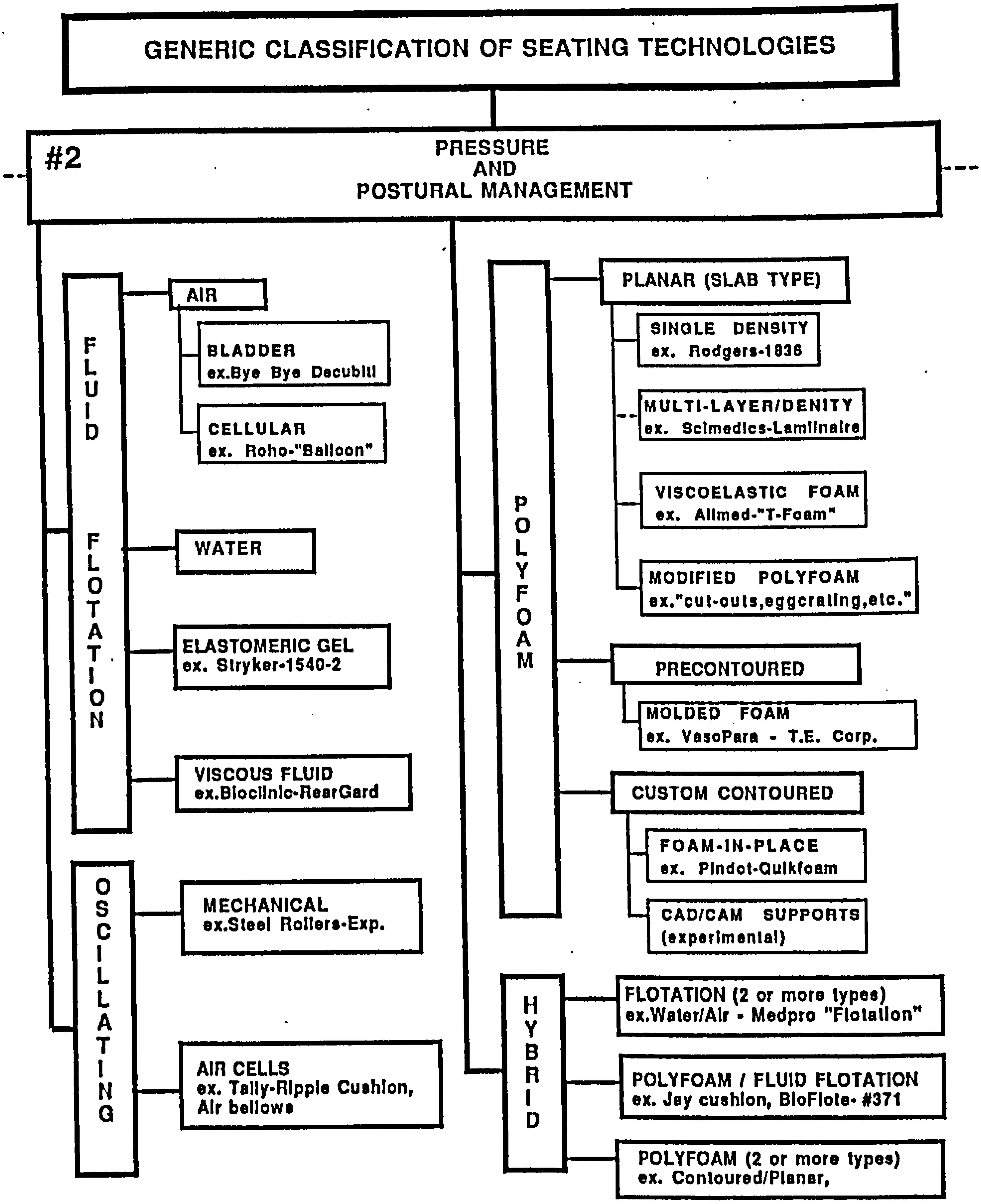


Fig 4.1(b) Generic classification of technologies used in the sub-field of pressure and posture management.

the spinal cord injury population are first categorized by the support medium or mechanical technique used to obtain the pressure relief, rather than on whether the surface profile is planar or contoured. As indicated in figure 4.1(b), the basic design approaches are: a) fluid flotation, b) polyfoams, and c) oscillating devices. A fourth subcategory (hybrid) provides a means of classifying designs that use two or more of the first three approaches.

a. Fluid Flotation

Four fluid or semi-fluid mediums have been used in cushion designs; air, water, viscous fluid, and elastomeric gels. Air filled designs have taken several directions. Two of the most common approaches are: a) configured bladders or rubber membrane designs and b) a cellular approach. The most common cellular cushion (Roho) has a large number of individual balloons (cells). The developer claims this minimizes any 'hammocking' effect as well as allows some air movement through the support structure. It is important in any air filled design that the appropriate air pressure be maintained. Over inflation or inadvertent loss of support pressure is possibly the most serious drawback to the air type flotation devices. Newer designs use compartmentalized peripheral sections at higher pressures then the ischial support area in an attempt to provide increased lateral and anterior stability.

Water filled cushions are less common. Although they provide good dissipation of body heat, leaks and excessive weight are noted problem areas. Again the hammocking effect created by the covering membrane can negate the pressure equalization attributes of the fluid suspension.

Elastomeric gels can be considered a very high viscous fluid since the gel does exhibit flow and pressure distribution properties similar to a fluid. Gel cushions are produced from materials such as silicone elastomers, chosen to have a viscosity or consistency similar to body fat. Gel cushions also provide good initial heat dissipation, but are usually heavy and gel leakage has been a problem with some designs. Pressure distribution characteristics can be effected by the design of the confining membrane and its outer covering material(s). More recently, high viscosity fluids contained in oversize flexible membranes are being used. The high viscous fluid exhibits pressure equalization properties of a fluid, but is free to flow within the confines of the membrane from areas of highest loading. This in effect creates a custom contouring effect.

b. Polyfoam Designs

Polyurethane foams (and some latex based varieties) have been the mainstay of cushion technology over the years, particularly for individuals at low to moderate pressure sore risk. The pressure relief characteristics of foam cushions depends on the inherent mechanical properties of the foam; whether it is used in a single or multi-layered 'slab' configuration or contoured to the pelvic area, either through precontoured molding or custom contouring to the precise shape of a person's pelvis.

i. Planar (Slab type) - In general, planar foam cushions are the most versatile since bilayering with different densities, altering geometry through cutouts, etc., and finishing with suitable slip-on covers can be accomplished with minimal technical support and resulting cost. Cushion thickness can be easily altered and their inherent lightweight and simplicity of wheelchair insertion facilitates transfers. Disadvantages are that their life is relatively short (6 - 12 months) and most cushions can not be readily washed without reducing their effectiveness(Noble and Goode, 1983). Also, the pressure relieving characteristics may not be adequate without significant modifications, especially for those that are in the moderate to high risk group.

Planar polyfoam cushions are designed with many foam types and configurations. Slabs of foam up to 100 cm (4") thick may be used in a single density design. Foams of varying thicknesses, types, and densities may be combined to gain the advantages of each type in a multilayer/density design (Ferguson-Pell et al, 1986).

A variation of the conventional polyurethane foam technology is to introduce a viscous fluid into the cellular foam structure. This combination creates viscoelastic properties which exhibit characteristics of both the standard polyfoam and viscous fluid flotation devices. The pressure distribution capabilities are usually improved, but the mechanical properties can be highly influenced by ambient temperatures causing a 'bottoming-out' possibility. Shock absorption and postural stability characteristics are often superior to traditional polyfoams.

And finally, it is important to realize that in clinical practice a great deal of creative design is necessary in terms of modifying planar (slab type) foams. The geometry of the slab is modified throughout the ischial area by 'cut outs', multidensity layering, or cutting serrations in one or both surfaces to enhance pressure distribution or other mechanical properties of the foam. The art and science of modifying planar polyfoams to meet individual needs is addressed by Garber and Krouskop (1978, 1984) and Ferguson-Pell (1980b, 1986).

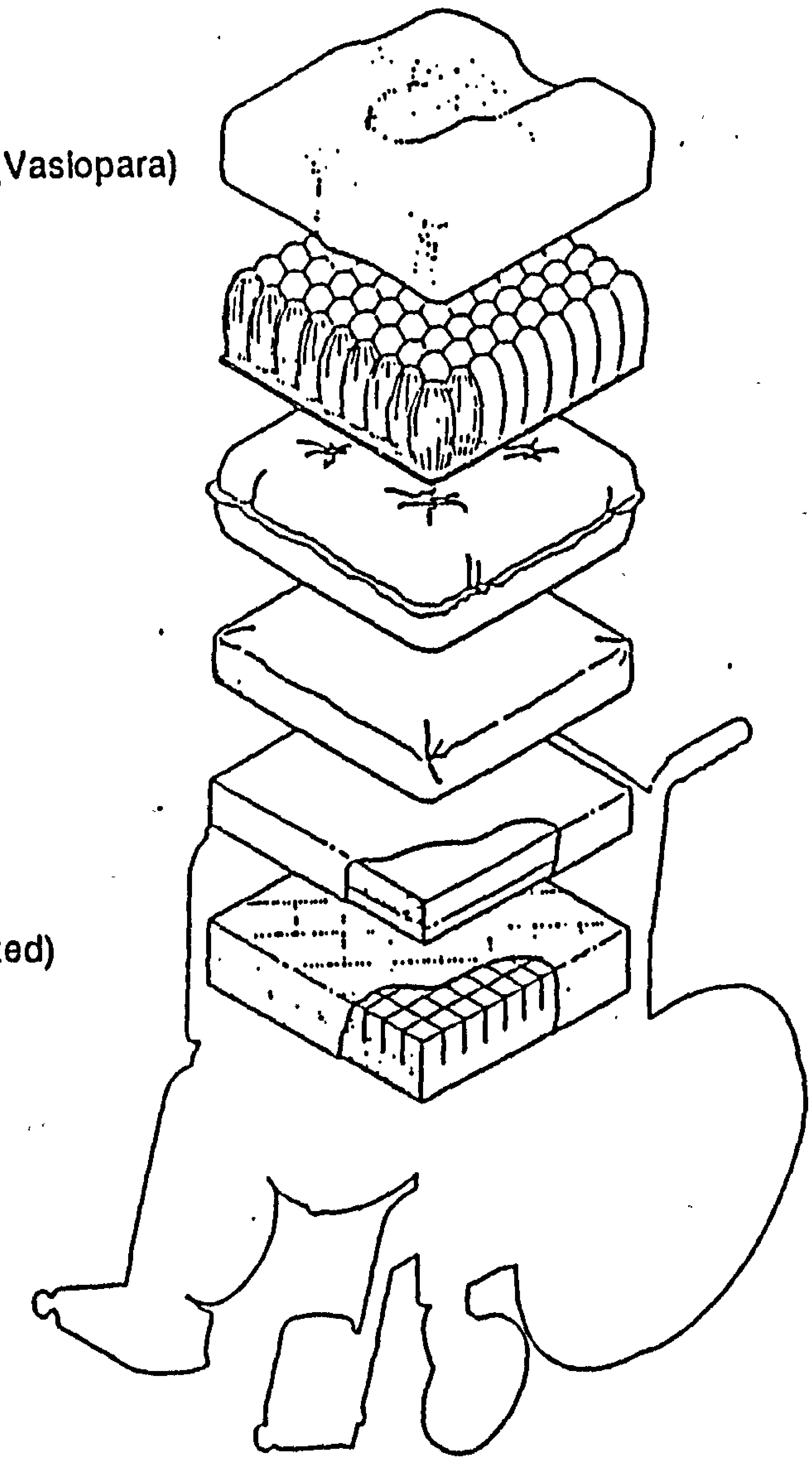


Fig 4.2 Generic types of wheelchair cushions used to provide pressure distribution (from Wilson, 1986).

ii. Precontoured Polyfoam - More recently, designers have attempted to distribute pressure away from bony prominences through the design of precontoured support surfaces. Precontoured designs provide an opportunity to reduce pressures in the genital area thus reducing the likelihood of constriction to the urethra. Also, increased air circulation is provided to reduce humidity buildup. Tissue loading is moved away from the ischial prominences to surrounding gluteal tissues. However, this tends to create a rim effect of high pressures. The prefabricated contoured cushions are made by injecting polyurethane foam into standard molds. A dip or spray-on covering material increases the durability of the cushion and reduces moisture penetration. The Vaso-Para cushion is an example of this technology. The developers of Vaso-Para have published information on the rationale for the shape used and the results of their clinical trials (Perkash et al, 1984).

Figure 4.2 illustrates a number of the more common commercial cushion technologies discussed above.

iii. Custom Contoured - Contouring of polyfoam to meet the shape of an individual has the promise of reducing the shear stresses inherent in slab foam designs. Design of the support surfaces can be readily individualized to redistribute pressure similar to the precontoured approach. An example of this technical approach is the Foam-in Place (FIP) system developed by the author (Hobson et al, 1978, Hobson and Tooms, 1981a,b). The FIP technique allows the formation of the polyfoam cushion by polymerizing the flexible polyurethane foams against the seated patient. Only a thin stretchy membrane separates the patient from the polymerising foam. In this manner the FIP approach allows custom contouring of the support surface to the buttock shape of the individual user. It is postulated that this process can significantly reduce interface tissue shear that normally occurs when one sits on planar foam cushions. Experience with over 300 (FIP) fittings has indicated that pressure distribution levels within safe clinical guidelines are routinely possible. The durability of the FIP has been exceptional (1 to 3 years). The FIP is advised for use only by those that have had training in the technique and in handling polyurethane polymers.

CAD/CAM technology has been used to produce seat cushions for spinal cord injured individuals on an experimental basis (Chung et al, 1985). The promise of designing and producing support surfaces to meet individualized pressure redistribution needs offers exciting possibilities for this computerized approach. The availability of clinical tools that can rapidly record body shapes will most likely dictate the feasibility and degree of acceptance of the CAD/CAM approach.

c. Oscillating Devices

The cushion designs discussed above are either static or quasi-dynamic devices in which periodic changes of body position are required to assure sufficient reduction in pressure levels through weight relief over supporting tissues. These static designs all attempt to equalize or redistribute pressure over the contact surface of the buttock tissues. That is, they attempt to minimize the pressure acting on the susceptible areas of the pelvis, ie. ischial tuberosities, sacrum and trochanters.

On the other hand, oscillating designs alter the pelvic pressure profile independent of the user. In general, they adhere to the principle that locally high pressures can be tolerated provided that the duration is kept within prescribed limits. Several approaches have been pursued; one basically a mechanical design, and the other undulating air cells or bellows. Kosiak (1976) described an experimental wheelchair seat that consisted of a series of rollers on a chain drive that moved under the person at a predetermined speed. Although, the time/pressure profiles were within tolerable limits described in his previous work (Kosiak, 1959, 1961) the complexity, cost, and bulk of the device precluded its commercialization and clinical use.

Talley Medical Ltd., England(a) has produced oscillating air cell devices. The first, termed the Ripple Seat, consists of 8 adjacently mounted air tubes each measuring 5 x 35.5 cm. Alternate air tubes are connected to a common manifold which is in turn connected to a small battery powered portable oscillating pump. Each series of 4 air tubes is inflated as the other series of 4 is being deflated. The effect is an oscillating system which produces intermittent high pressures followed by low pressure to alternate areas of the buttock. The more recent design, termed the Air Bellows Support, uses a matrix of cellular air bellows (7.5 cm high x 3.7 cm diameter) connected to a similar manifold/oscillating air pump. The pump is designed to allow adjustment of the pressure levels and the frequency of pressure changes across the bellows matrix. A warning device alerts users to system failures so corrective action can be taken. The approach is technically intriguing and may offer significant advances in pressure management in the years ahead.

d. Hybrid Devices

There are pressure management technologies in routine use that do not fit neatly into the above classification scheme. The Hybrid category permits the classification of cushion designs that employ two or more of the above approaches. For example, several of the fluid flotation mediums have been combined into a single cushion design. One design uses water within an air bladder. In another approach elastomeric gels and related high viscous fluids have been combined with polyfoam structures. For example, the Jay cushion uses a precontoured polyfoam substructure with a top surface membrane filled with a high viscosity fluid; whereas, the Akros-Zero Pressure uses a precontoured plastic substructure with a gel impregnated polyfoam top layer. Ferguson-Pell et al, (1986) succinctly describe the key factors involved in designing multi-media/layered modular cushions to address the specific needs of individuals with spinal cord injury. A review of this work is recommended to anyone contemplating design or development of hybrid type cushions for the management of pressure sores.

A glaring discrepancy from the generic classification scheme are technologies and devices that specifically address the *posturing* needs of individuals with spinal cord injury. Observation of individuals five years or more post-injury suggests that spinal and pelvic deformities due to inappropriate sitting posture is a common outcome among this population (Zacharkow, 1984b; Hobson, 1984; Hobson and Nwaobi, 1985). Slowly the posturing tools developed and the experiences gained with the cerebral palsy population are filtering over for use with the spinal cord injured population. Hopefully future research and clinical practice will take a more holistic approach to the seating needs of the spinal cord injured, and not remain overwhelmed by the immensity of the pressure management problem.

4.2. MANAGEMENT OF THE PRESSURE SORE PROBLEM

4.2.1 Prevention and Medical Management

Under normal conditions the body's sensory feedback system enables people in good health with voluntary mobility to consciously or unconsciously change posture and thereby avoid damage to the skin or deeper tissues. Paralysis, sensory deficits, mental apathy, heavy sedation and anaesthesia, coma, and immobility due to poor physical condition may lead to a lack of response to discomfort or painful stimuli arising from the skin or deeper tissues. It is known that when loss of sensation and mobility are coupled with circulatory insufficiency, incontinence, or other impairments most often found in elderly people, special precautions must be taken to protect the vulnerable areas of the body from undue external forces. Prevention through relief of pressure is then central to current medical management of the pressure sore problem. Skin hygiene, general nutrition, prevention of moisture and heat build-up are additional preventative measures that have been already discussed.

In spite of these recommended practices and the availability of predictive scoring systems like the Norton Scale (Norton, 1961), which quantify the risk factors for pressure sore formation so preventative measures can be taken, 20-30 percent of skin insensitive and/or immobile people will develop problems requiring hospitalized treatment (Clark et al,1978; Young and Burns, 1981). McDougal (1976) predicts that treatment of pressure sores occupies 60 percent of the time spent in hospitals for this population.

The first stage of conservative medical treatment involves determination of the general condition of the patient. Treatment of any anemia, correction of nutritional deficiency, and nitrogen imbalance (if present) are important factors in assisting the healing of sores (Guttmann, 1976). The second stage can involve localized treatment or a combination of local treatment and surgical repair.

Local treatment is largely dependent on the degree and type of pressure sore (Guttmann, 1976). Infected sores with necrotic underlying fascia are usually treated by surgical debridement, followed by application of a wide variety of possible medications and treatments. An adage that is often quoted is: 'you can put anything you like on a pressure sore except the patient'. The guiding principles are to keep the wound dry, clean and relieved of pressure to allow the natural healing process to proceed.

Surgical repair can involve a combination of excision of sloughs followed by conservative management, to epithelialisation by skin grafts, once the granulations are healthy and have reached the level of surrounding skin (Guttman, 1976). In the most severe cases in which osteomyelitis has occurred it is necessary to excise the infected bony area (Karaca et al, 1978). Similar criteria are followed by other clinicians (Campbell, 1959; Berry, 1980; Fowler, 1984).

In summary, most progressive medical treatment regimes, whether conservative or surgical, advocate the indoctrination of patients about the dangers of pressure and how to avoid it. Avoidance of pressure or inadvertent tissue injury through generous applications of common sense, coupled with the wise selection and use of appropriate technical devices, is the most important precondition in the prevention of tissue problems or their reoccurrence.

4.2.2. The Equipment Selection Process

This review section can not begin to summarize all the information necessary to successfully evaluate and prescribe (select) equipment for use by the target population. A complete treatise on therapeutic evaluation and related decision making involves a review of neuromotor dysfunction, pressure sore pathology, pathomechanics, and orthopaedic deformities, the details of which are beyond the scope of this chapter. Therefore the intent is to focus on the key factors associated with the decision-making process and thereby highlight the systematic steps that are followed within most successful multidisciplinary clinical settings. The following overview is intended to create an awareness of the importance of the steps rather than provide the details associated with each step.

The needs of individuals for seating and mobility equipment are tremendously variable. No one technical approach works for all people. Therefore, a process is necessary which systematically identifies and documents needs so that they can be interpreted and communicated in terms of specific technical requirements. This information can then be passed on to technicians, equipment vendors, and to funding agencies who must often pay for the services. This process is often termed '*evaluation and prescription*'.

It is important to note that it is not necessary for every disabled person to pass through a formal evaluation process in order to assess their needs prior to provision of technical services. For example, some people are seasoned users of equipment that already works well for them and they are simply seeking a replacement. Others have rather obvious needs and the technical solutions are generally straight forward. However, there are others that maybe first time users of devices, or they are more severely physically involved so that their equipment selection process is more complex; or the consequences of misapplication may be more serious, especially when funding is scarce. Also, integration of multiple technologies may be important such as; seating and mobility, augmentative communication, and computer access. In these later cases a carefully planned evaluation and prescription process is generally necessary and beneficial.

Another reason for observing the steps of a formal selection process is to guide prescribers and consumers towards joint selection decisions without omitting any important factors. In general, technology selection decisions should not be solely dictated by the local availability of specific types of seating or mobility devices. The process should be designed to generically define what technical approach is potentially the most appropriate to meet the needs of the individual. Compromises are then usually necessary as a result of many local influencing factors. However, more options are kept open if a 'conceptual ideal' can be kept in mind when the compromises are being made. Figure 4.3 is a schematic representation of a generalized selection process that encompasses both seating and mobility devices for all disabilities and age groups. The process can be conceptualized as consisting of three inter-related phases; 1) needs identification, 2) provision and 3) follow-up.

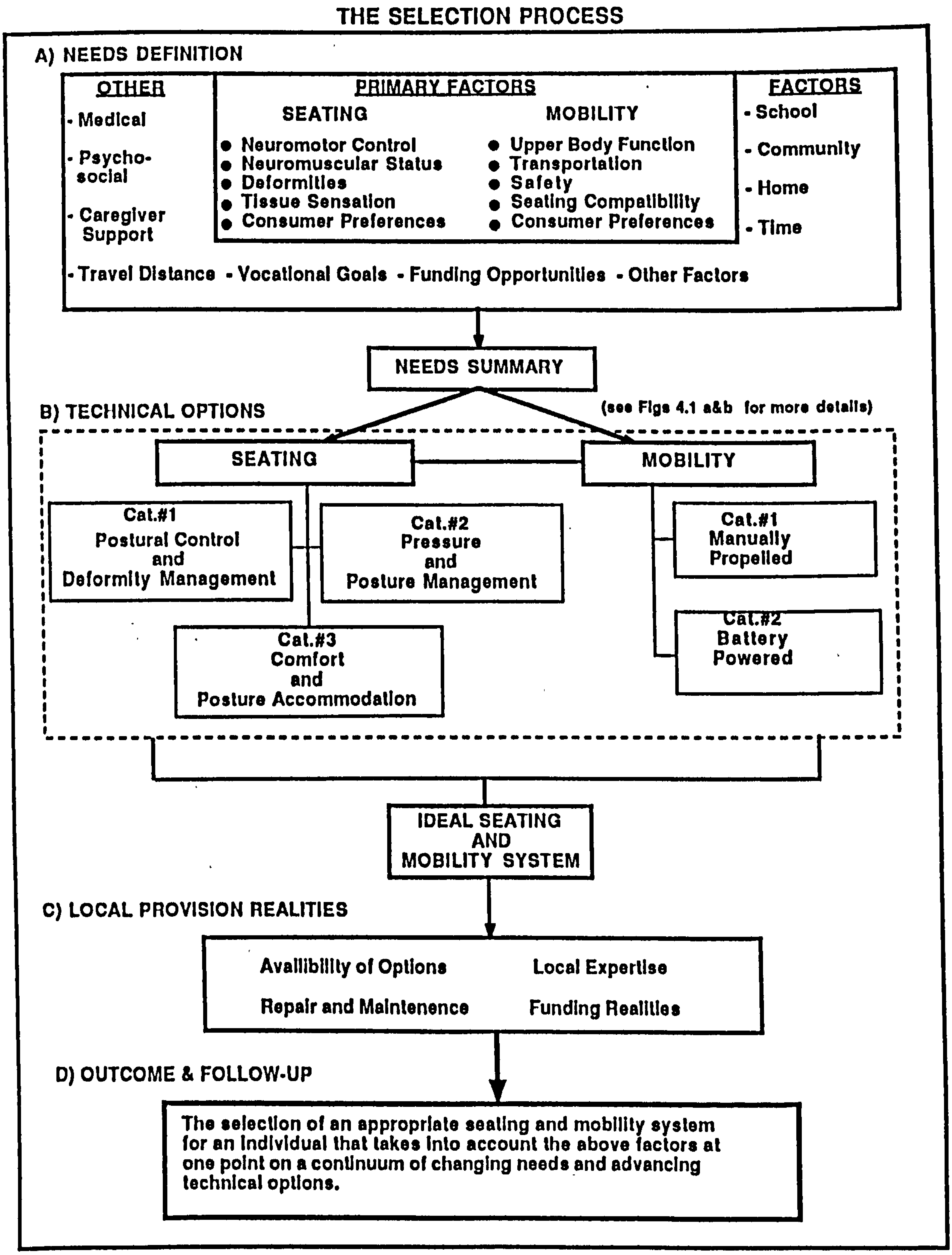


FIG 4.3 A schematic presentation of the equipment selection process used in the field of specialized seating and mobility.

1) The Needs Definition

As can be seen in figure 4.3 many factors can be involved in the needs definition phase. Since seating and mobility is the core focus of the process; factors that directly impact on seating and mobility decisions are denoted as the *primary factors.* The many additional factors that can influence the final needs definition (prescription) have been designated as *other factors.* In some cases the other factors may be more important than those designated as primary factors.

a. Primary Factors

i. Seating - Analysis of the functional status of an individual from the neuromotor and neuromuscular perspectives defines the nature of the body support and posturing that the seating system will need to provide. For example, severe damage to the neuromotor system of a child with cerebral palsy may result in strong hip extensor thrust patterns, minimal functional use of upper extremities and moderate head and trunk control with only minimum orthopaedic deformities. Others may have all of the above neuromotor involvement but also have severe deformities of the pelvis and spine. Needs for seating support will be different in each case.

Others may have loss of neuromuscular function that will vary throughout the body. For example, individuals with spinal cord injury will have loss of sensation and muscle function below the lesion level. Most will require pressure relief and some will require trunk support by the seating system depending upon the level of lesion. In contrast, individuals with Duchenne Muscular Dystrophy have progressive loss of muscle function with intact sensation, so that their needs for body support will vary with time and the progression rate of the disease; and also the extent of deformity present.

Lastly, and certainly not of lesser importance, is the expressed preferences of the individual or their primary care-giver. These preferences will often out-weigh the recommendations of clinicians carrying out the evaluation process.

When all of the above factors are combined they result in what may be termed as the *seating needs definition.* The next step is to determine the mobility needs of the individual.

ii. Mobility - Needs analysis for mobility can be equally complex especially when all the other factors are considered. A prime indicator, and usually the most obvious, is the level of physical function in the upper body that can be called upon to propel a manual wheelchair. If sufficient function is present this usually indicates the need for a manually propelled base. The total or partial absence of upper body function

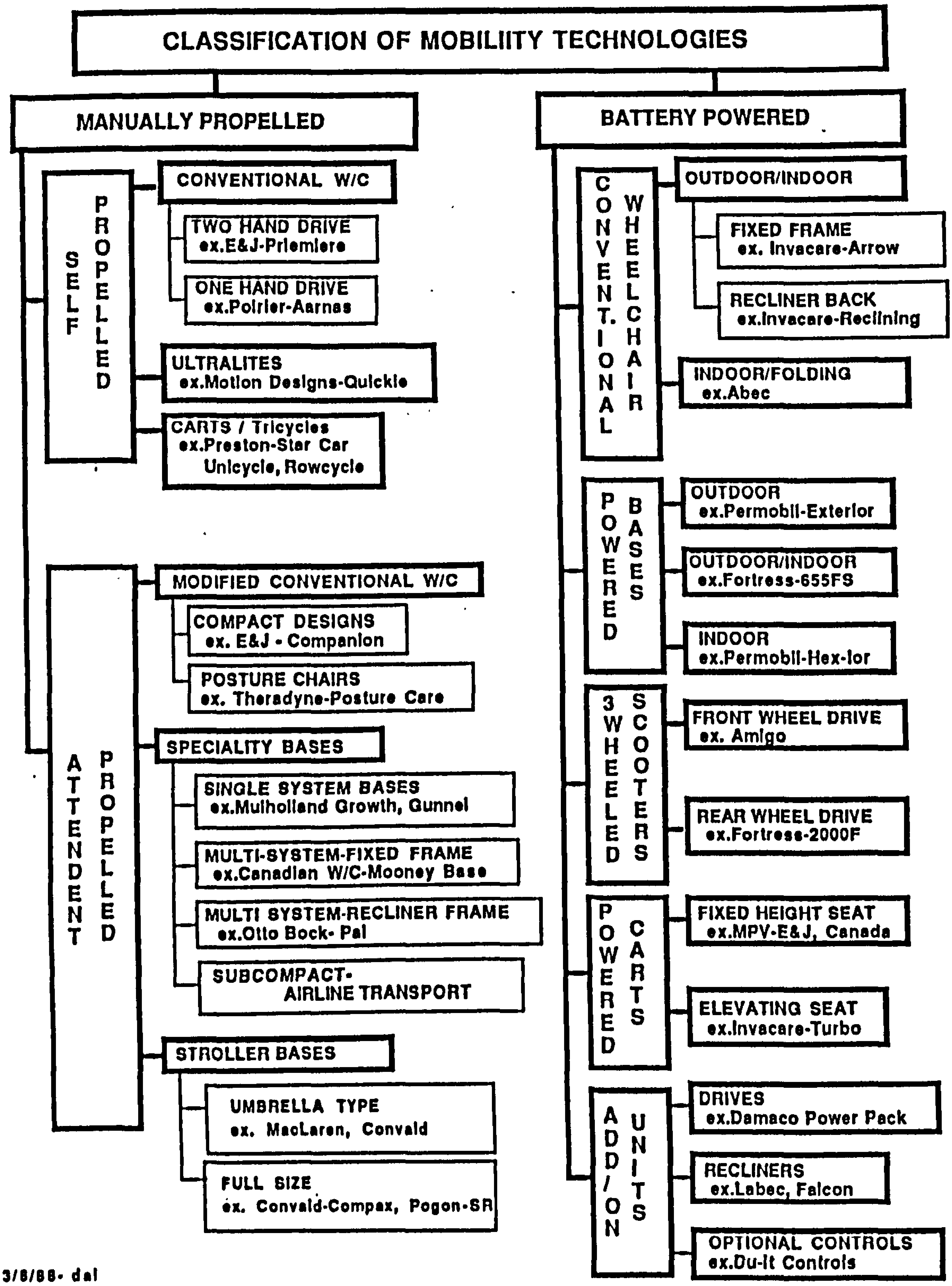


Fig 4.4 A generic classification of mobility technologies used in the field of specialized seating and mobility.

suggests the possibility of externally powered mobility. However, before powered options can be considered factors associated with transportation of the individual such as the family vehicle(s), school transportation, etc. must be identified. Also, the wheeled base needs are often dictated by the seating needs, so that compatibility can be assured. For example, some individuals require a seating system that will tilt the whole body in space. The mobility options considered must meet this requirement.

Safety factors such as; base stability, body mass distribution, tie down needs, cognitive control capabilities and securement of the seating system can also become dominant need considerations.

Finally, the expressed preference of the individual users in terms of gadget tolerance, lifestyles, esthetic concerns, etc. can all weigh heavily on the mobility needs definition. In some cases people will require both the manual and powered mobility bases in order to carry out the range of activities associated with their lifestyles. Figure 4.4 is a classification of generic mobility devices in common use today. Commercial options within these categories are growing at an amazing rate considering the relatively small size of the demand.

b. Other Factors

In addition to the above primary factors there are many other factors that can greatly influence the needs definition. For example, the general medical condition of the individual, their prognosis, and the overall rehabilitation goals established by their primary care team must be taken into consideration.

As indicated previously the individual may have need for other technologies such as; augmentative communication aids, computer access, ADL devices, etc. Although seating and mobility needs are often basic to the successful use of other devices, an awareness of these additional needs can often influence the selection of the seating and mobility systems. An example could be the selection of a powered wheeled base that offers compatibility with communication or environmental needs.

The psycho-social status of the individual and/or their care-giver support system can influence the durability requirements, esthetic expectations and tolerance for technical complexity of the devices recommended. Additional factors related to school, community and home activities and vocational pursuits can greatly influence the needs statement and ultimate technical selection. Also, the distance individuals must travel for follow-up problem solving can set limits on technical complexity or types of devices that are appropriate, as well as limit the time available for service provision and follow-up.

Prior knowledge of the basic funding categories in which people fall (private or public) can influence the initial needs statement. That is, although funding should not be the dominant factor, an early assessment of the funding possibilities may exclude or limit consideration of the more costly generic approaches.

In summary, there are many *other factors* that must be carefully identified and balanced in order to generate a *needs summary* that reflects the comprehensive seating and mobility requirements of an individual (fig 4.3). Once the needs are identified the next phase in the selection process is to consider the various technical options. These generic cushion options for the spinal cord injured group have been briefly outlined in the preceding section. In clinical practise this generic information must be further supplemented by a working knowledge of the actual cushions that exist within the commercial market place.

2) Local Provision Realities

Identified needs and awareness of the technical options leads to the final step in the process, the selection and provision of an actual cushion or seating system with a compatible model of mobility base. Additional factors that can drastically influence the outcome of the selection process are the realities associated with the local service environment. For example, although technical options may exist in the market place, they may not be available to consumers in a particular locale for a variety of reasons. One factor may be the absence of local stock or expertise to fabricate and fit a particular device. In some cases, an equipment vendor can obtain the seating and mobility device but there may be no mechanism for providing maintenance and repair. Finally, and usually the most dominant local factor is the unavailability of funds or health authority approval to purchase the required technology.

These factors, often in combination, constitute the realities that necessitate compromises towards selections that are less than ideal. The challenge then becomes a matter of weighing the needs and technical options against the local realities in order to arrive at a final selection that best meets the needs of an individual.

3) Outcomes and Follow-up

Any one selection/provision cycle is usually only one step in a continuum of changing needs and expanding technical opportunities for people with chronic long term disabilities. The brief history of seating and mobility field suggests that the next time the selection/provision cycle is repeated that improved opportunities will exist to more closely approach the ideal solution. It is this reality that makes the necessary compromises associated with each cycle a little more acceptable.

A delineation between mediocre and superior service facilities is the commitment to record keeping and follow-up practices. Organized and timely records are essential, not only for legal reasons, but also to assure continuity and improvement to future services. Aggressive follow-up is equally necessary in order to identify problems that if left unresolved could lead to serious consequences, such as onset of a pressure sore. A progressive follow-up program also creates the environment in which both clinicians and people with disabilities can more rapidly progress towards 'ideal' solutions in a field of advancing technical and therapeutic options.

4.2.3. Evaluation and Prescription Tools

In addition to developing seating and mobility devices scientists and engineers, working with clinicians, are developing new concepts and tools to aid in the selection and provision of specialized seating services.. As in most new fields clinical decisions are based largely on subjective observations and past experience. Tools are now being developed that allow clinicians to make improved assessment and provision decisions and quantify the results. These tools allow the measurement and recording of variables such as body posture, interface pressure, and deformity. They also permit clinicians to experiment with various body positions so that the accuracy and confidence of subjective assessments are improved. Once final decisions are made the tools facilitate measurement and recording of the final decision, so that communication with technical personnel can be more readily accomplished.

For example, simulators allow simulations of body posture for children with cerebral palsy. This is very helpful for therapists attempting to make decisions about the critical features of the definitive seating system. Another distinct advantage is that the definitive seating arrangement can be mocked up so that assessment of controls for communication aids or computer access can also be carried out during the assessment process. In the case of a powered mobility evaluation the seating simulator can be mounted on a mobility evaluator so that a variety of control options can be quickly tried. This latter tool assists the clinical team and the user, or their care-giver, in arriving at joint decisions in an efficient manner.

Measurement of interface pressure for those with spinal cord injury has vastly improved the clinician's ability to make decisions based on quantitative data. A review of these developments was presented in sec. 3.4.

The field of clinical tool development is just in its infancy. For example, with the advent of personal computers and expert systems its now possible to computerize clinical decision making strategies that are difficult to communicate via traditional educational programs. Tractman and Ferguson-Pell (1984) have demonstrated the use of an expert system, termed CushFit, for aiding clinicians in choosing the most appropriate cushion for pressure sore management. This and other similar tools, such as computer-aided shape definition, will surely be developed for routine use in specialized seating in the future.

CHAPTER 5. DISCUSSION AND SUMMARY OF CURRENT KNOWLEDGE AND PRACTICES

Studies conducted over the past decade have confirmed that incidence and related costs of pressure sore treatment remain a major health problem. Both clinical and research evidence, although inconcise and inconclusive, are biased towards excessive and/or prolonged application of pressure as being the primary causative factor. More recently, additional contributing factors have been implicated. Some of these factors are: shear stress, impact loading of tissue, elevated temperature and humidity, age, nutritional status, general health, activity level, deformity, body stature, and psychological deficits.

Studies have been conducted on the 'normal' seated posture, particularly with reference to alignment of the spine in various standing and sitting postures. Opposing theories exist as to the most appropriate alignment of the lumbar spine during sitting in order to minimize lower back pain and muscle fatigue. Biomechanical models have been developed in an attempt to better understand the complex mechanics associated with spinal stability and torsal movement. In spite of this excellent work, a comprehensive biomechanical model of the 'normal' human spine that has general acceptance remains to be developed.

Considerably less knowledge has been gained on the pathomechanics of the unstable spine. Most of the investigations have been retrospective studies associated with specific surgical techniques. Several investigators have reviewed the natural history of spinal deformity based on diagnosis and age of injury. It seems evident that individuals receiving a spinal injury prior to puberty will develop progressive paralytic spinal curves, whereas those acquiring the injury later in life will mainly have spinal deformities associated with the injury site. However, studies involving other populations, i.e., cerebral palsy, multiple sclerosis, muscular dystrophy, and stroke have shown that a wide variety of spinal deformities exists, especially in the older and more severely involved patients. Flattening of the lumbar spine and kyphosis of the thoracispinal was reported by at least two investigators.

Given the dearth of scientific and clinical knowledge on spinal pathomechanics, especially as it relates to the stability of the paralytic spine, a simplistic model has been proposed. In essence, the model proposes that 'normal' spinal stability involves a complex interaction between spinal and abdominal muscles, supported by involuntary intra-abdominal and inter-thoracic hydrostatic pressures, all interacting to support a flexible spinal column that is maintained in the normal 'S' configuration during sitting and standing. Diminution of the muscle forces either due to partial or total paralysis renders the spine inherently unstable and subject to the extrinsic gravitational forces. In addition, the extrinsic forces imposed by the postural alignment of the wheelchair seats does little to compensate for losses of intrinsic muscle function. Loss of voluntary trunk stability, combined with the imposed configuration of the wheelchair seat, biomechanically necessitates that a person with diminished trunk control assume an abnormal sitting posture. This posture is characterized by a long 'C' shaped kyphotic spine, with an extended cervical spine, a flattened lumbar spine, and a posteriorly tilted pelvis. If lateral trunk deformities are present, trunk imbalance and pelvic deformities in the frontal plane can exist. Finally, it is postulated that this combination of trunk and pelvic abnormalities can be contributing factors to tissue problems, the effect of which is not fully understood or appreciated by most researchers and clinicians.

It is difficult to separate the analysis of the spine and pelvis because they are so intimately associated, especially when deformities are present. Pressure studies done with paralytic subjects indicates a causal relationship between pelvic deformity and elevated buttock pressures as well as altered locations of maximum pressures. However, no attempt was found in the literature to analyze these relationships in alternate seated postures.

The conventional wheelchair has been both a blessing and a curse, especially for people with spinal cord injuries. The conventional X-frame folding design has afforded many people greater mobility, especially when independent transfers to personal motor vehicles are possible. The compromise has been that the flexible sling type seat does not provide adequate postural support for prolonged sitting. Also, the configuration of the conventional wheelchair seat is such that it fosters postural deformity of the spine and pelvis; rather than counteract deformity by facilitating the assumption of neutral postures commensurate with the user's functional abilities. Conventional cushion inserts, especially the commercial options, assume a symmetrical seated posture which is not the case for many people who use them.

Factors other than deformity and lack of independent alterations to body posture can also predispose a person to elevated interface pressure and eventual ulcer formation. Differences in body build, altered body mass through retarded growth and disuse tissue atrophy have also been implicated as factors that can drastically alter the pressure distribution characteristics across supporting tissues. Atrophy of muscle mass over the critical pelvic areas such as the ischia will mean elevated pressures and pressure gradients. Perpetual sitting in an asymmetrical postures or unilateral wasting or removal of pelvic structures (ischiectomy) can create asymmetries in posture that cause elevated pressures on the involved side.

When taking a holistic view of the problem systemic factors must also be considered. For example, protein deficiencies, neurogenic disorders, biochemical imbalances, venus congestion, infection, vascular insufficiency, and general nutritional status are all systemic factors that can have direct or indirect effects on tissue viability.

Mechanical models of tissues under stress have been developed in an effort to describe and predict the quantitative relationship between stress, strain, time, and the physiological factors at play within tissue structures. Hydrostatic models have been proposed that attepmt to explain the nature of the shear stresses acting upon the deep buttock tissues. These models are usually based on assumptions of structural homogeneity and isotropic properties which do not readily apply to the buttock supporting structures when viewed as a whole. It remains to develop a quantitative model that can make accurate predictions regarding the viability of supporting tissues at the body support interface of a seated or lying person.

Many early investigators have studied blood flow in attempts to quantify the relationship between externally applied loads and internal stresses that result in cessation of blood flow. As a result of this early work several clinical guidelines have emerged regarding safe thresholds of pressure and time. The inconsistencies in the investigative work suggested that the formation of a pressure sore is most probably multifaceted, and that many factors interact in a complex manner. As a result subsequent investigators have also studied temperature and humidity buildup, shear stress, centre of gravity displacement, tissue integrity, neurogenic factors and age among other factors. In spite of this extensive investigative effort, concise clinical guidelines for safe thresholds of pressure and time over the buttock tissues remains to be determined. Clinical practice regarding pressure management varies tremendously with a majority of the decisions still being made on mainly subjective information. This information is largely based on what has or has not worked in a particular locale for an individual patient or clinician.

Researchers have struggled for many years to design and develop tools that can facilitate the prevention and prescription process. The overwhelming majority of the effort has been placed on measurement of interface pressure at the sacrifice of many of the other implicated factors outlined above. At this time there are measurement tools that give only gross indications of the pressures experienced by the supporting buttock tissues. However, this is only one measurement, albeit probably the most important. Tools for quantifying the remaining factors are woefully lacking.

Interface shear stresses have been strongly implicated as a contributing factor. These implications are based on animal studies and work on the thenar emminence muscle of the hand. It is acknowledged that shear stress is difficult to measure since its effects are often realized deep in the supporting tissues. It has been recognized that shear or friction forces acting at the seat surface are only one aspect of the shear problem. High pressure gradients at critical locations on the buttock have been associated with elevated shear in the supporting tissues. It is postulated that differential strain properties between the tissues and the supporting materials can be a causative factor.

Studies using advanced blood flow monitoring techniques, partial pressure of oxygen (PO2) and biochemical changes offer promise for the future. Promise lies in the reality that if a practical method can be developed to actually monitor the tissue responses to external stress, rather than measuring the extrinsic factors that cause the response, a more precise and useful indicator of pending tissue problems may emerge.

Mechanical properties of the materials used as support tissues at risk are essential if the deleterious effects of the interface factors are to be minimized. Access to materials that are superior in: pressure distribution, resilience, heat dissipation, moisture dispersion, shear and impact attenuation, longevity, and other desirable mechanical properties are vitally important to both clinicians and disabled people. Laboratory tests have been developed that can quantify many of the above properties. Comparative clinical and laboratory evaluations between commercial cushions can produce additional guidelines for the selection and provision of appropriate interface support materials.

One method of quantifying posture is to determine the centre of gravity (CG) and then track the movement of the line of the CG relative to the area of support or some other similar reference. Frequency and magnitude of movement of the CG has been used to make inferences regarding the changes of pressure distribution across the support surface. Comparative studies between different populations conducting various functional tasks have yielded useful information, especially when combined with quantitative pressure data. Devices have been developed which provide a warning to the user when specified time/ pressure relief thresholds have been exceeded. In general these devices have not been accepted for wide scale use.

The clinical practice of specialized seating for non-ambulatory people is rapidly becoming a recognized sub-specialty in the rehabilitation field. Activities in the specialized seating field can be broadly categorized into three major areas; 1) postural control and deformity management (cerebral palsy), 2) pressure and posture management (spinal cord injured), and 3) comfort and postural accommodation (elderly). The clinical principles that are involved in the selection and provision of appropriate technologies for each of these groups are developing along converging paths, with some overlapping concepts and devices already evident. A generalized selection and prescription process has been proposed which involves the participation of a trans-disciplinary team.

The quality of the evaluation and provision process used for pressure management varies widely throughout clinical facilities. It is proposed that this may be due, in part, to the inconsistencies between the guidelines emerging from research and the emperical observations associated with actual clinical events. Also, relatively few tools are available to clinicians which can facilitate a more systematic approach to cushion selection, problem solving and follow-up. Changes in administrative attitudes combined with increased emphasis on staff education could make immediate and significant improvement to clinical outcomes.

It is evident that much remains to be accomplished in the area of pressure and postural management from both the research and clinical perspectives. A few investigators have questioned the primacy of pressure as the leading causal factor in pressure sore formation. The fact remains that there is a dearth of research evidence suggesting any other factor as being more dominant than interface pressure. However, it is also clear that many unanswered questions have persisted and it is now time to more closely examine some of the remaining implicated factors. For example, the role of shear stress, deformity, activity level and alterations of body posture are factors that require improved understanding. In particular, an understanding as to how these variables may interact with interface pressure to yield predictable physiological responses in the supporting tissues.

It is from the perspective summarized above that the following study has been formulated. This study has been designed to examine several factors which to date have received very limited attention. As indicated previously individuals with spinal cord injury currently often exhibit deformities of the spine and pelvis. In addition to defining the nature of the deformity, it is also important to examine how the deformity may alter the pressure distribution, particularly when alternate sitting postures are assumed. The study also attempts to evaluate tangentially-induced shear at the seat surface and determine how this alters with changes in body posture. Since changes in the CG of a seated person are relatively easy to quantify clinically, an attempt will be made to relate changes of CG to alterations in posture and pressure distribution.

The primary purpose of the work is to evolve additional guidelines that can complement and possibly clarify those guidelines already derived from studies related to pressure distribution. In this context it is hoped that additional tools will emerge that can be used for more systematic decision-making within clinical environments.

PART 2 - THE RESEARCH STUDY- SUBJECTS, METHODS, AND RESULTS

CHAPTER 6. ORGANIZATION OF THE STUDY

The study has been designed to investigate the possible contributions of deformity and body posture to the variables that occur at the body-seat interface; namely, pressure distribution and tangentially-induced shear forces. Sample groups of able-bodied ('normal') people and individuals with spinal cord injuries were selected to serve as the study subjects. Four distinct but interrelated variables have been measured and compared between study groups. These variables are; a) spinal and pelvic alignment, b) pressure distribution, c) tangential shear (friction) force and centre of gravity location. The same subjects (normal and disabled) participated in all four components of the study.

Each component of the study is presented separately. A summary section integrates the results from all four components in order that the interrelated aspects can be discussed and final conclusions drawn. The final Part 3 relates the findings to clinical practices and suggests topics requiring continuing research investigation. The limitations of the findings are also discussed in Part 3.

CHAPTER 7. THE SAMPLE GROUPS

It is evident from previous work that measurement of absolute values at the interface surface yields results that are questionable in terms of accuracy and repeatability. However, if relative values are obtained, many of the procedural uncertainties and inaccuracies associated with obtaining absolute values are minimized. This concept will be developed and discussed more fully in the procedural presentation of each component of the study. However, it is this fundamental notion that influenced the decision towards having a sample of normal subjects to serve as controls, or a reference baseline, to which the results from the disabled sample group could be compared. In this manner, importance is placed on the relative differences between the 'normal' and disabled population rather than the absolute values of either population. It then becomes increasingly important that the characteristics of the two study samples be clearly described, so that any inferences made regarding the larger population of disabled individuals can be kept in perspective.

7.1. THE ABLE-BODIED SAMPLE

Ten able-bodied adult subjects ('normals') were randomly selected from a university environment, which included both professional and support personnel. The only criteria imposed upon the selection was that the subjects be over 18 years of age (skeletally mature) and that they have no previous history of lower back or pelvic disorders. The population selected contained six males and four females with a mean age of 39.3 years and a range of 28 to 57 years. Mean body weight is 68.6 kg (range 50-95.5 kg) and a mean height of 174.1 cm (range 161.3 - 198.1 cm). All subjects reported no previous impairments of spinal or pelvic movements or abnormalities of the pelvic, buttock, or thigh musculature.

7.2. THE SAMPLE OF PEOPLE WITH SPINAL CORD INJURIES

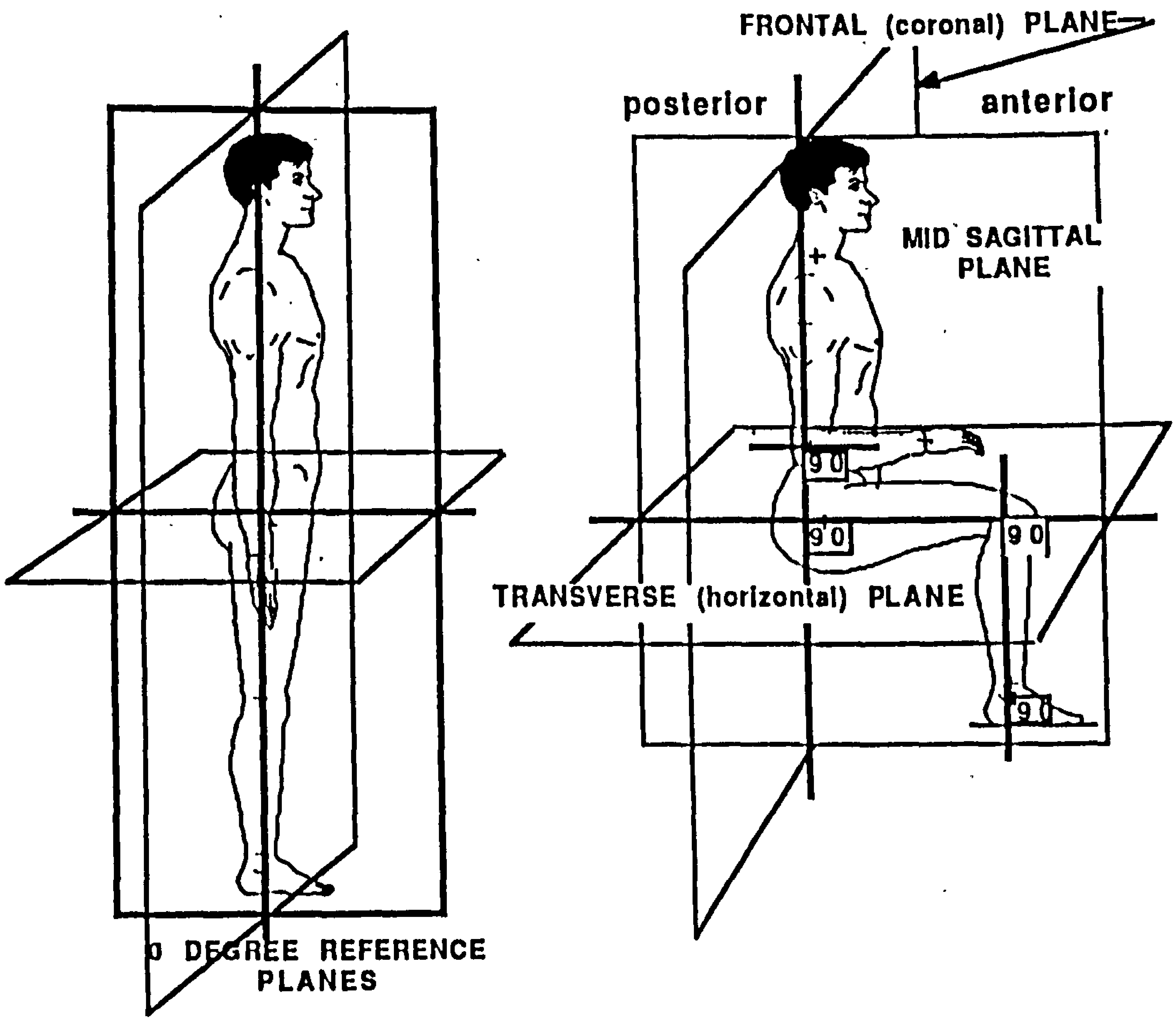
Since the effects of paralysis and related deformity are the primary focus, twelve individuals with complete lesions of their spinal cord were chosen. An advertisement for participation was placed with the local Centre for Independent Living. Respondents were interviewed by telephone as to their suitability to serve as test subjects. The selection criteria used was that each subject must have had their injury for at least five years, and had used a conventional wheelchair for a period of at least five years. They also should not have had any surgical procedures (fusions) that would directly impair motion of the lumbar spine and pelvis. On x-ray examination, one of the twelve subjects had to be eliminated from several aspects of the study since a fusion extending into the lumbar spine was evident. And finally, each subject had to be paralyzed and insensitive in the buttock and thigh areas.

The sample group of SCI people contains ten males and two females, a mix of paraplegics (7/12) and tetraplegics (5/12). Their level of injury extends from T11-L1 to C5-C6. Four subjects had a history of pressure sores (1 tetraplegic and 3 paraplegics). The mean age is 40.9 years (range 25-66 years) with a mean body weight of 59.8 kg (range 39-74.2 kg) and a mean height of 172.7 cm (range 162.6- 185.4 cm). The mean number of years since injury is 19.5 (range 6-54 years). None of the subjects had an active ulcer during (or after) the period that they participated in the research study.

In general, the subjects are all active individuals living in the community and regularly using their wheelchairs. One subject was injured at birth, but all others were injured during teenage or adult years. The sample of disabled people appears most closely representative of the SCI population that has had a spinal injury for the major part of their adult life, and have remained active users of wheelchairs in community or vocational activities. The sample population is most probably not representative of institutionalized or inactive individuals, children, elderly, or spinal cord injured people that have had a spinal injury for less than five years. Finally, the disabled sample appears to be mainly from a low to average socio-economic environment, although this was not formally assessed.

ANATOMICAL REFERENCE POSITIONS & TERMINOLOGY

CONVENTIONAL STANDING (CARP) **SEATED (SARP)**



Anterior:	Toward the front.
Posterior:	Toward the back.
Superior:	Toward the head.
Inferior:	Away from the head.
Medial:	Toward the midline of the trunk.
Lateral:	Away from the midline of the trunk.
Proximal:	Toward the trunk. Refers to the limbs.
Distal:	Away from the trunk. Refers to the limbs.
Palmar (Volar) surface:	Anterior portion of forearm and hand.
Dorsal (Dorsum) surface:	Posterior surface of the trunk and Upper limb.
Plantar suface:	Inferior surface of the foot.

Fig 8.1 Schematic showing both the conventional (CARP) and seated (SARP) reference positions. Reference planes and body directional terms within the planes are also defined.

CHAPTER 8. QUANTIFICATION OF THE SITTING POSTURE

A major segment of the following study is concerned with the biomechanics of people in the seated posture. The terminology for describing seated posture and movement of body segments is not well defined. The methods for recording and describing body measurements to meet the needs of the specialized seating field exhibit wide variability throughout research and clinical settings. In an effort to establish an international convention for communication the author has collaboratively published a definitions and terminology reference document that is currently being reviewed and modified by experts in the field for eventual adoption (Hobson, 1988). Relevant extracts from this work are presented in the following subsections as the basis for defining the terminology of body movements, measurement reference planes, and the linear and angular measurements used throughout the study. This terminology has been drawn largely form existing nomenclature used in medicine, engineering, human factors, and anthropometry.

8.1 THE ANATOMICAL REFERENCE PLANES

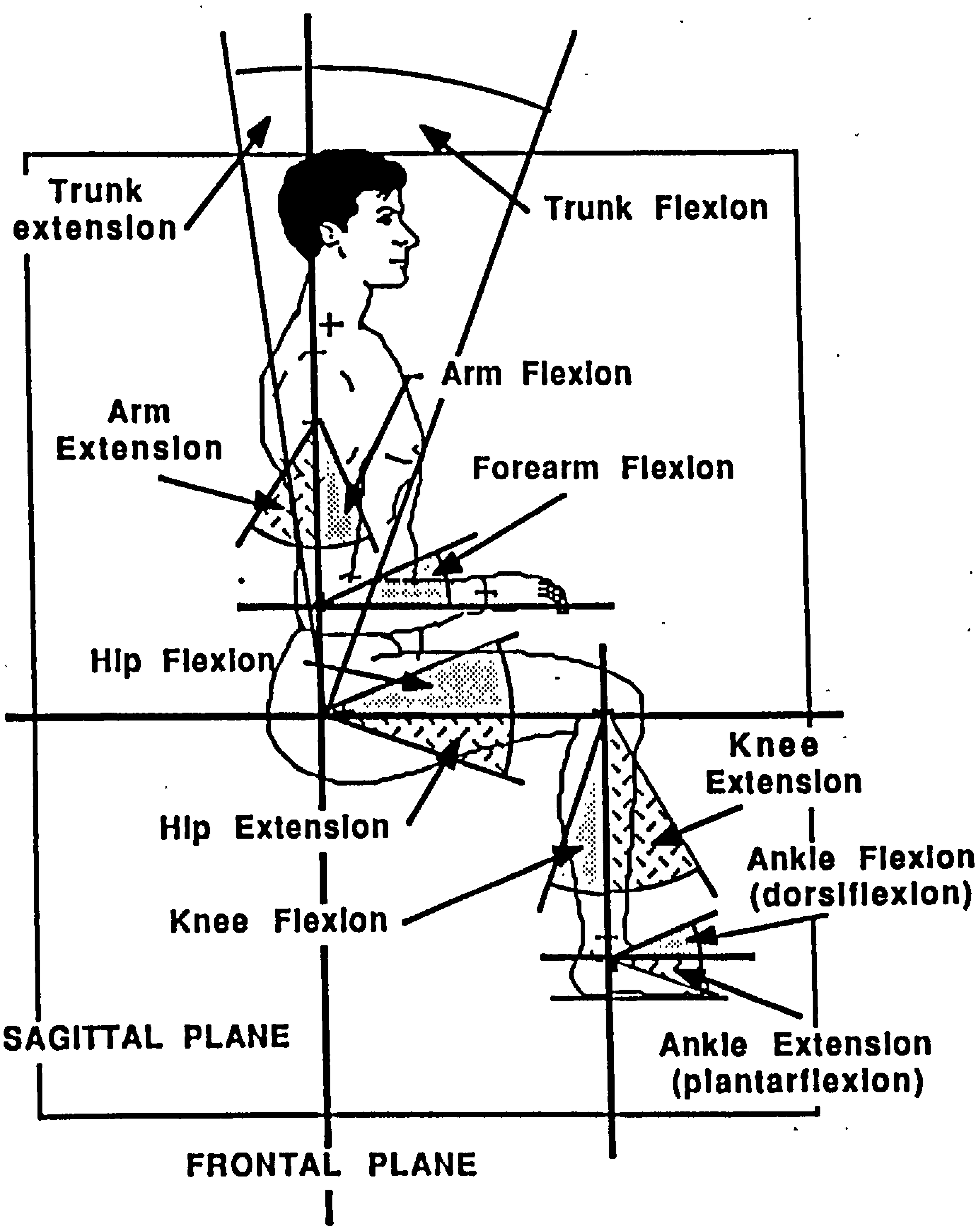
A convention is used throughout the medical field to describe the movement and therefore position of body segments relative to a neutral anatomical reference posture. The *conventional anatomical reference posture* (CARP) has been defined as a person standing erect with the head, hands, and feet facing forward and the arms parallel to the trunk and straight to the sides (Fig 8.1). Clinicians working in specialized seating are usually working with people in the seated position and therefore have informally developed a modified reference system which can be termed the *seated anatomical reference posture* (SARP). This posture is often referred to as the 90-90-90 posture. Since it is important to maintain consistency between the CARP and SARP, both systems are defined in figure 8.1.

If one defines the SARP with reference to the vertical frontal plane passing through the hip joint and trunk, which is the same reference plane used in the CARP, then compatibility between the two systems can be retained. The advantage of the SARP is that it gives a direct visual representation of a person in the seated posture, which of course is consistent with the objectives of specialized seating.

8.2 DEFINITION OF MOVEMENT TERMS

It simplifies angular movement definitions if the joints about which the segmental movement takes place are considered to be uniaxial, so that movement takes

DEFINITIONS OF BASIC BODY SEGMENT MOVEMENT TERMS
(from SARP in sagittal plane)



Flexion:	Bending a joint.
Dorsiflexion:	Bending the wrist so the dorsal surface of the hand points toward the forearm.
Planter Flexion:	Bending the ankle so the foot points downward.
Extension:	Straightening a joint.
Hyperextension:	Extending a joint beyond the anatomical zero position.
Pronation of forearm:	Turning the forearm (radioulnar joints) so the palm faces downward.
Supination of forearm:	Turning the forearm (radioulnar joints) so the palm faces upward.

Fig 8.2 Definitions of classical body movement terms in the sagittal plane.

place in only one reference plane at a time. In reality, most joints are multi-axial and segmental movement can be a combination of sliding, multi-axial rotation, or gliding/rolling over changing articular surfaces. As a result, uniaxial rotation of joints rarely occurs.

The anatomical reference planes in which angular motion is observed and measured are defined in figure 8.1 as: a) mid-sagittal, b) frontal, and c) transverse. With reference to a person in the seated posture these three planes are defined as follows:

a) mid-sagittal plane - a vertical plane passing through the mid-line of the body dividing it equally into left and right halves;

b) frontal plane - a vertical plane passing through the hip joint and the erect trunk dividing it into front and back halves;and

c) transverse plane - a horizontal plane passing through the hip and knee joints.

All movements are described with reference to the conventional standing posture (CARP) which is considered the neutral or reference position (fig 8.1). Movements from the reference position are defined as motion taking place in one or more of the reference planes. Figures 8.2-8.4 define the more common movement terms used in specialized seating.

8.2.1— Pelvic Movement

Pelvic movements require more detailed consideration since changes in pelvic position relative to the SARP are so critical to specialized seating and this study. The joints in which movements of the pelvis occur are the two hip joints and the joints of the lumbar spine, particularly the lumbosacral articulation. Since pelvic position depends on the joints of the lower spine and the action at the hip joints for its movement, its motion is often associated with the motion of the trunk or spine, and sometimes with that of the thighs; especially if rigid (fixed) deformity or contractures exist in tissues acting across the hip joints.

The articulation between the spine and the pelvis occurs at the lumbosacral joint. The position of the pelvis is traditionally defined by the lumbosacral angle (fig 8.5a). The lumbosacral angle is defined as the angle between a line drawn through the superior plateau of the first sacral vertebrae and the horizontal. In the SARP this angle for normals is approximately 20-25° as measured in the mid sagittal plane. Motion of the pelvis can be described in terms of: a) pelvic tilt, b) pelvic obliquity, and c) pelvic rotation.

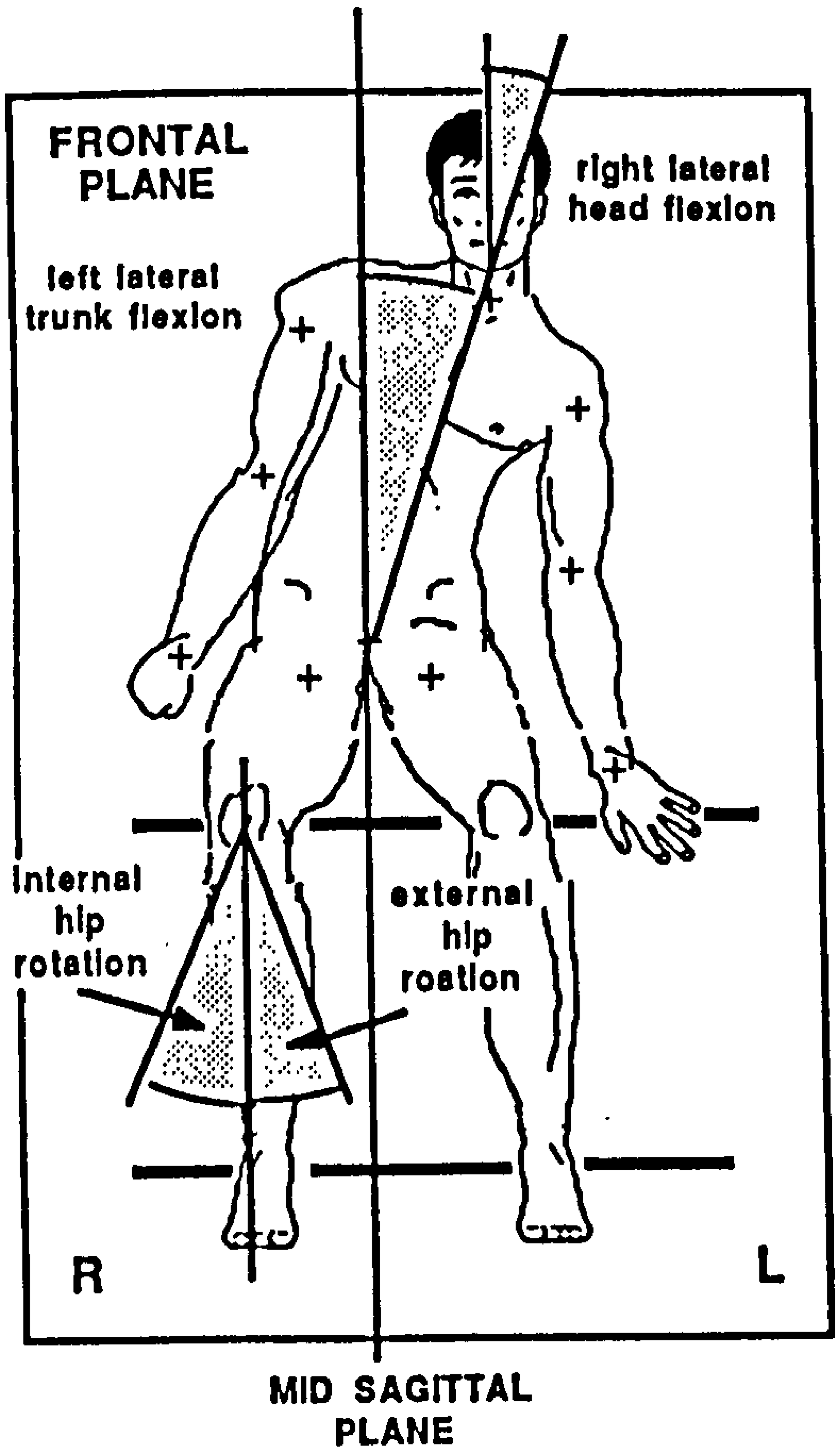


Fig 8.3

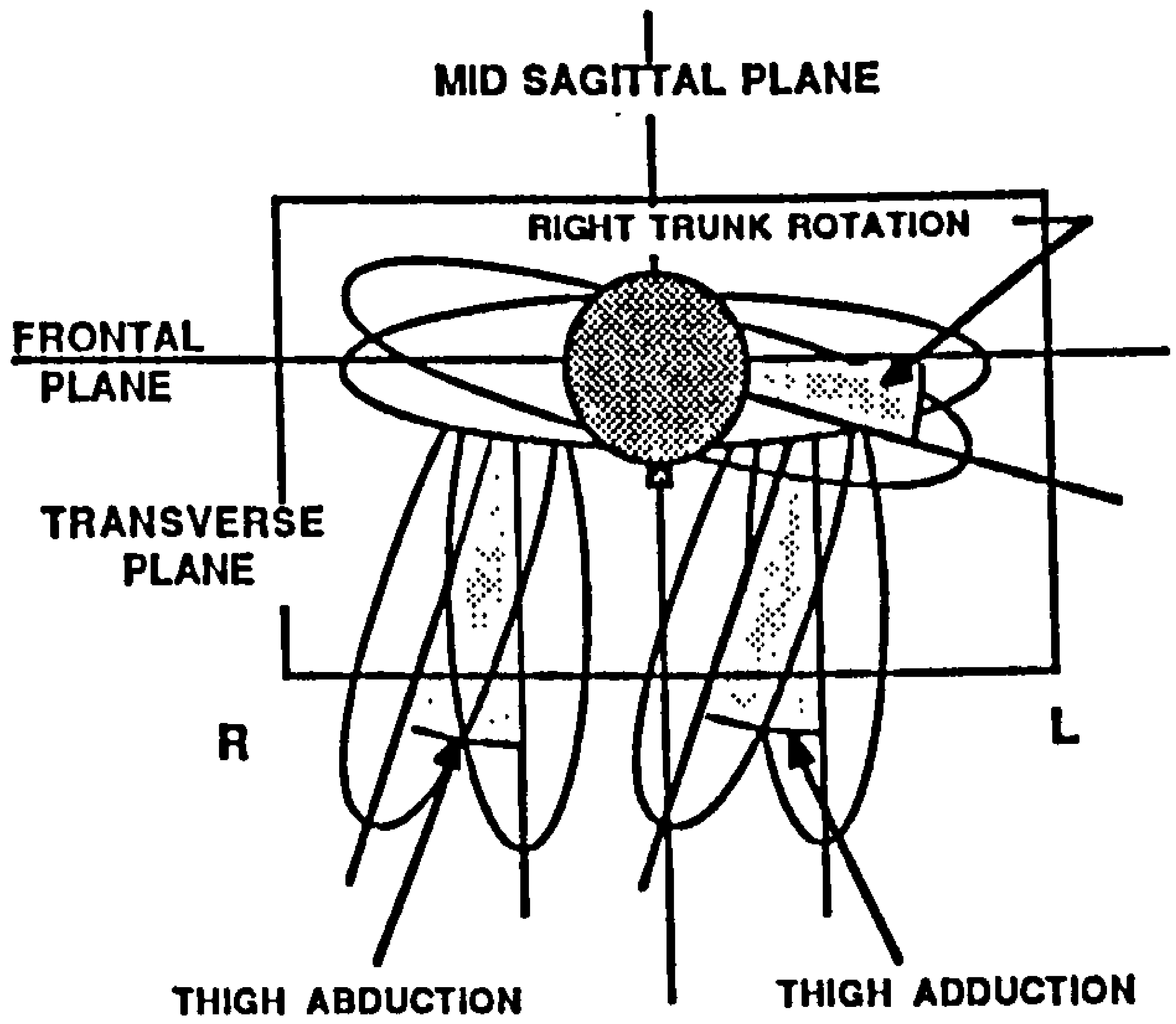


Fig 8.4

Internal rotation:	Twisting a joint inward (toward the body).
External rotation:	Twisting a joint outward (away from the body).
Supination of the foot:	Turning the medial border of the foot upward so the sole faces inward, on weight bearing. (Varus foot on weight bearing).
Eversion of the foot:	External rotation of the subtalár (Talocalcaneal) joint in the foot.
Pronation of the foot:	Turning the medial border of the foot downward so the sole faces outward, on weight bearing. (Valgus foot on weight bearing).
Abduction:	Moving a joint from the midline of the body.
Adduction:	Moving a joint toward the midline of the body.
Trunk rotation:	Twisting of the trunk in the transverse plane around the long axis of the trunk.

Figs 8.3 & 8.4 Definitions of body movements in the frontal and transverse planes, respectively.

a) Pelvic Tilt - movement in the sagittal plane from the neutral position is described as a forward or backward pelvic tilt (fig 8.5 b&c).

b) Pelvic Obliquity - rotational movement from neutral in the frontal plane about a horizontal axis passing through the lumbosacral joint measures the lateral obliquity of the pelvis. Obliquity is named in terms of the side which moves downward. A right pelvic obliquity means that the right side of the pelvis is lowered and the left side is raised (fig 8.6).

c) Pelvic Rotation - rotation of the pelvis in the transverse (horizontal plane) measured about a vertical axis passing through the lumbosacral joint. The movement is defined in terms of the direction towards which the front of the pelvis turns (fig 8.7).

PELVIC MOVEMENT IN THE SAGITTAL PLANE

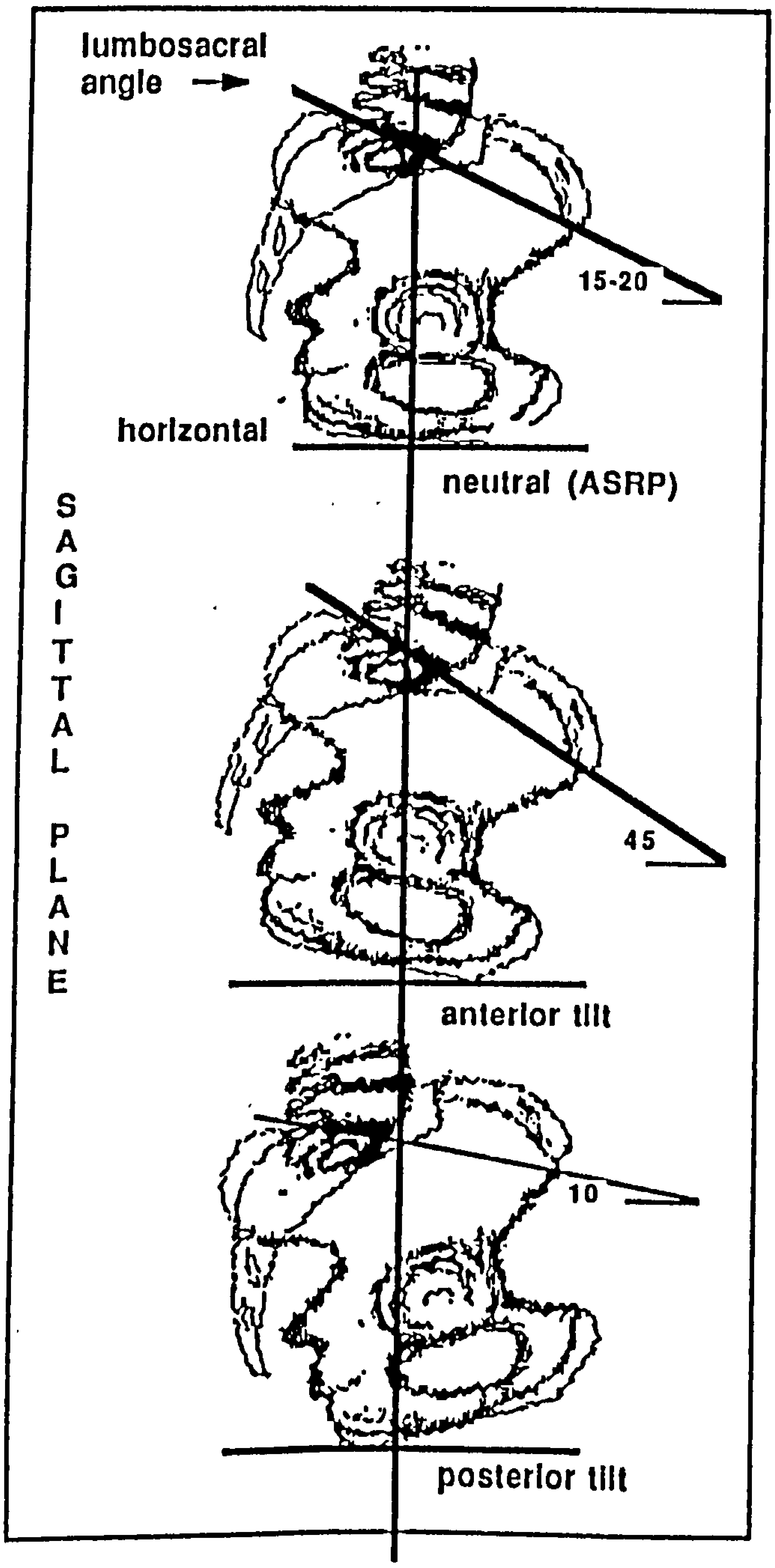


Fig 8.5(a) Neutral Pelvis: The position of the pelvis in the erect seated position with the lumbosacral angle approximately 15-20°.

Fig 8.5(b) Forward Tilt: A rotation of the pelvis from the neutral position in the sagittal plane about a frontal - horizontal axis, in such a manner that lower pelvis moves downward and backward and the upper pelvis moves forward.

Fig 8.5(c)Backward Tilt: A rotation of the pelvis from the neutral position in the sagittal plane about a frontal - horizontal axis, in such a manner that the lower pelvis (symphysis pubis) moves forward and upward and the upper pelvis moves backward.

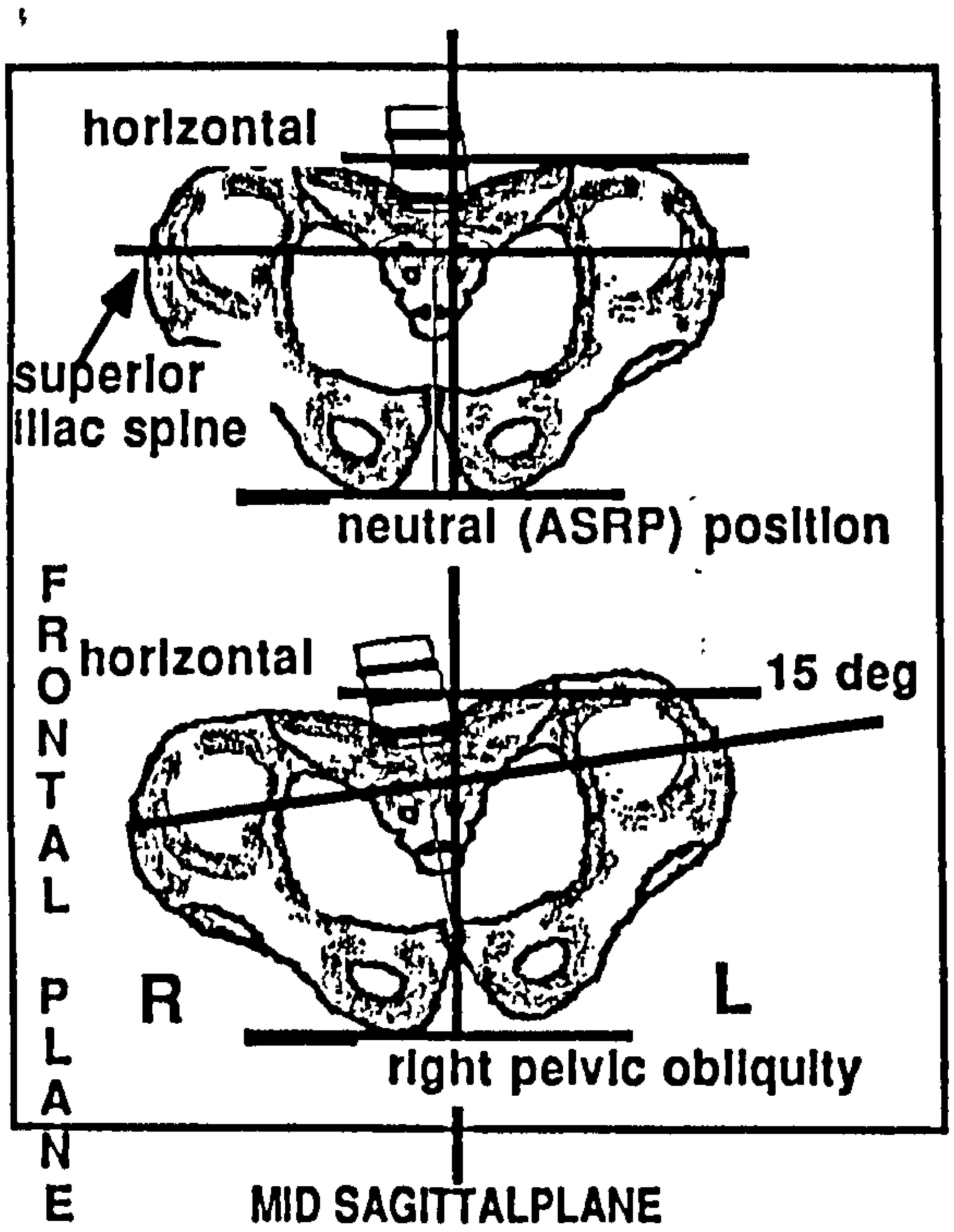


FIG 8.6 **Lateral Obliquity:** A rotation of the pelvis in the frontal plane about a sagittal-horizontal axis in such a manner that one iliac crest is lowered and the other raised. The obliquity is named in terms of the side which moves downward. Thus a right pelvic obliquity means the right iliac crest is lowered and the left is raised.

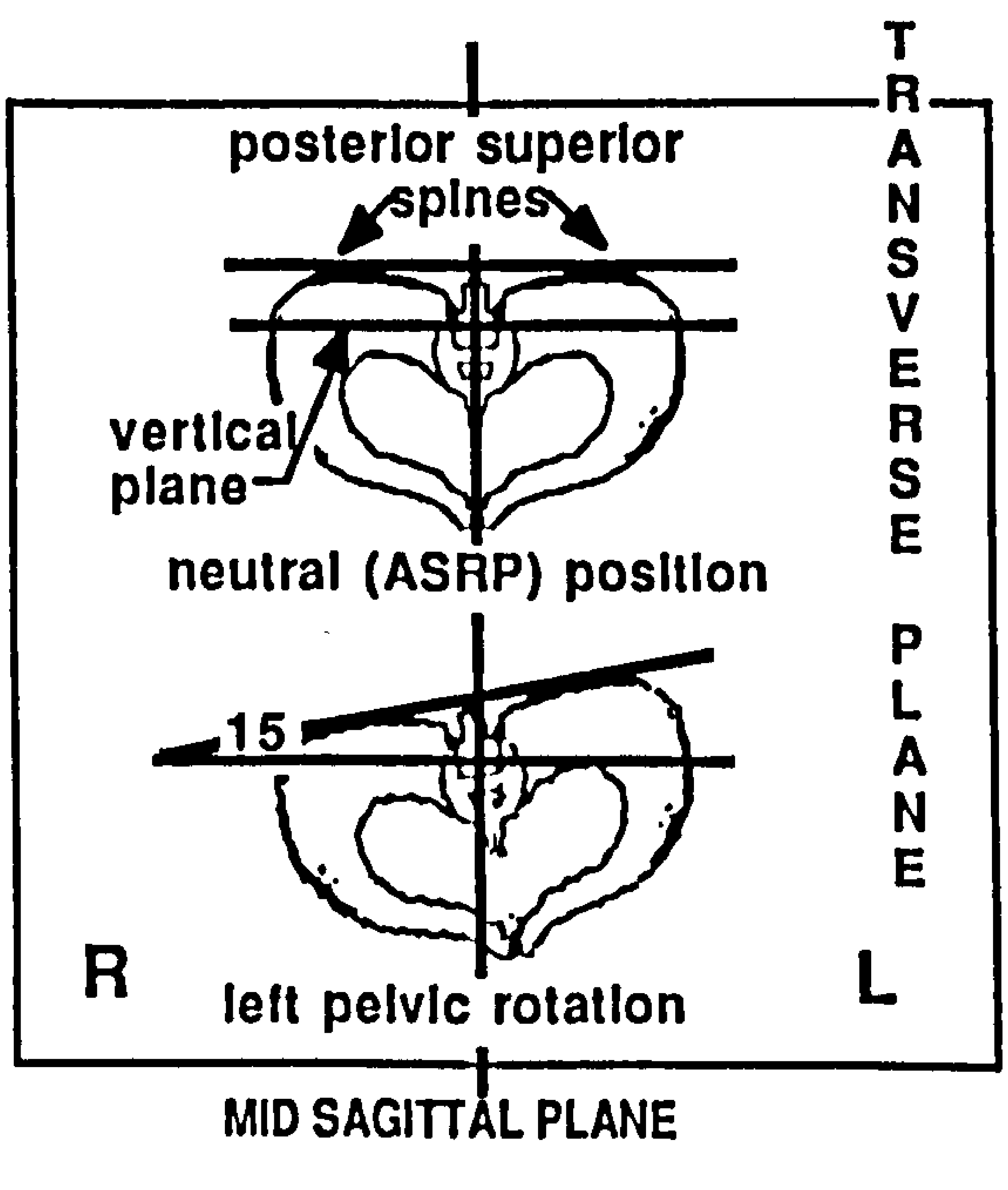


FIG 8.7 **Rotation:** A rotation of the pelvis in the transverse (horizontal) plane about a vertical axis approximating the longitudinal axis of the spine. The movement is defined in terms of the direction which the front of the pelvis turns. The illustration shows a left pelvic rotation of about 15° in the transverse plane.

DEFINITIONS - LINEAR MEASUREMENTS

M1 Sitting Height - Seat plane to top of head.

M2 Back Plane to Back of Head - Distance from the extension of the back plane to back of head.

M3 Thigh Length - Back plane to popliteal space.

M4 Leg Length (L/R) - Distance from the under knee popliteal space to foot plane.

M5 Foot Length - Overall foot/shoe length.

M6 Shoulder Width - The maximum width of the shoulders.

M7 Chest Width - Width of the chest taken at the axilla level

M8 Waist Width - Measured at the narrowest dimension across the waist.

M9 Pelvic Width - Measured at the level of the greater trochanters.

M10 Knee Width - Maximum outside dimensions across the knees.

M11 Occipital Protuberance to Center Line - Center line is described by a vertical line on the sagittal plane passing through the midpoint of the pelvis.

M12 Acromion Height L/R - Seat plane to acromion both left and right sides.

M13 Overall Width - The maximum width occupied by the individual.

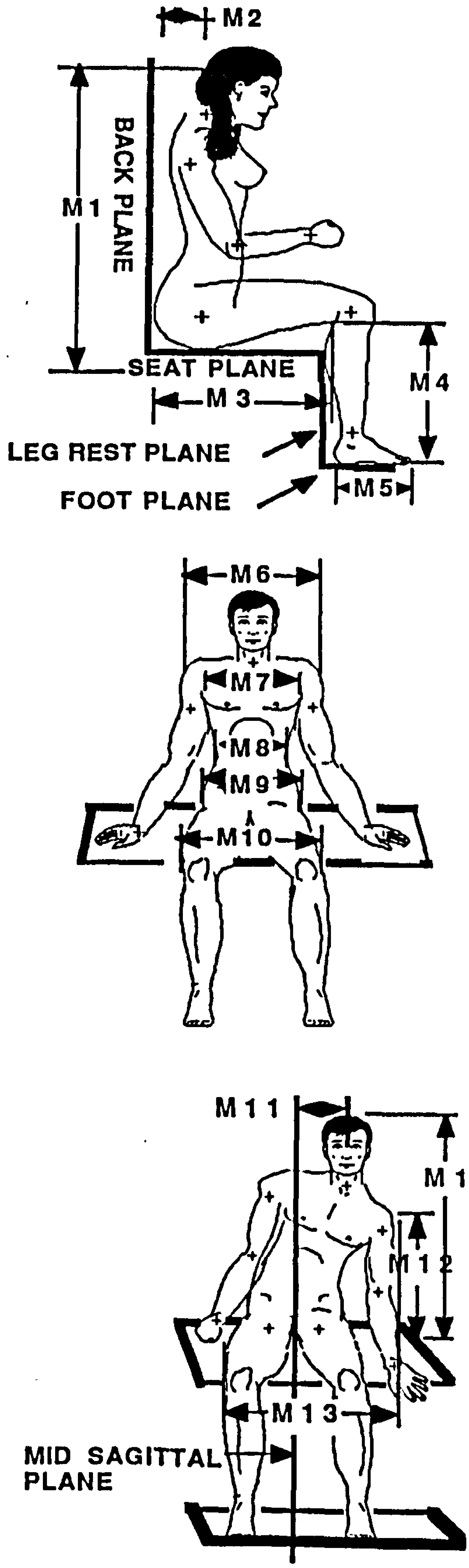
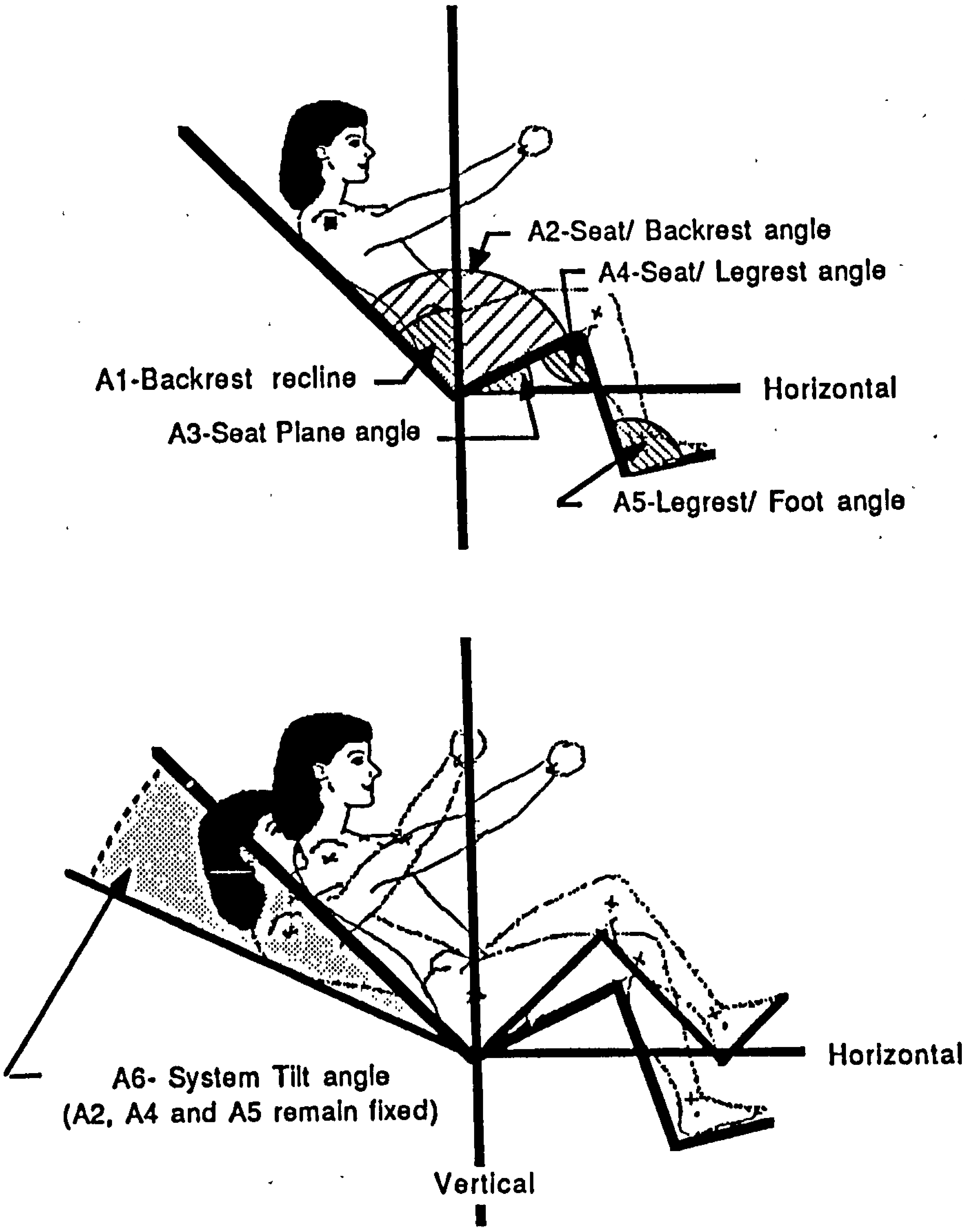


Fig 8.8 Definitions of commom linear dimensions used in specialized seating.

DEFINITIONS - ANGLES OF SEAT PLANES



A1 Recline Angle - Inclination of the back plane with respect to (wrt) the vertical

A2 Seat to Back Angle - The angle between the seat and back planes

A3 Seat Plane Angle - The seat plane angle wrt the horizontal reference axis

A4 Seat/Legrest Angle - The angle of the leg support wrt the seat plane

A5 Legrest/Footrest Angle - The angle of the footrest wrt the plane of the legrest.

A6 System Title Angle - A description of the angle in space of the complete system wrt the vertical axis. By definition a change in tilt angle causes an equal change in orientation of all support surfaces, i.e., angles A2, A4, and A5 remain unchanged. Changes in A1 and A3 are equal to the change in tilt angle A6.

FIG 8.9 Definitions of seat plane angles which define the angular posture of a seated person.

8.3. DEFINITIONS OF MEASUREMENTS

8.3.1. Datum Planes

It is usually difficult to accurately determine linear and angular measurements of the body without x-ray analysis. Also the angular positions of the body are rarely identical to the mechanical angles of the seating surfaces that support the body. For these reasons seating researchers and clinicians have developed a **seating datum** (reference) system that can be used clinically to measure and record the position of the seated person in space. This datum system is based on four seat planes or surfaces which are illustrated in figure 8.8, top diagram. They may be defined as follows in the neutral or SARP position:

i. Back Plane

A vertical reference plane which defines the posterior or back support surface of a person seated in the seated anatomical reference position (SARP).

ii. Seat Plane

A horizontal reference plane which defines the seat support surface of a person in the SARP.

iii. Leg Support Plane

A reference plane which defines the location of the posterior surface of the legs in the SARP.

iv. Foot Support Plane

A reference plane which defines the location of plantar surface of the feet in the SARP.

8.3.2— Linear Seating Measurements

Linear measurements of body segments with reference to the above reference planes are used to obtain certain body measurements. Angular measurements of the reference surfaces with respect to each other and with respect to the horizontal and vertical defines the angular position (seated posture) of the person.

Figure 8.8 defines thirteen linear measurements that are used in specialized seating, especially when custom fabrication of total body supports are required.

8.3.3— Angular Seating Measurements

Naturally, the seating support surfaces do not remain in the neutral position as defined above. In fact, its the angular positions of the support surfaces relative to the neutral position that clinically defines the seated posture. Figure 8.9 defines six angular measurement terms that define the seated posture in the sagittal plane. Similar measurements in the other two planes can completely define a person's position in space.

8.4. OTHER MEASUREMENT METHODS

Several investigators have developed more detailed methods for measuring and recording spinal and pelvic orientations. Andersson et al (1979) describes a method of analyzing the influence of lumbar support and backrest inclination on the shape of the lumbar spine. During and Goudfrooij, (1985) used a similar analytical method to compare the standing postural parameters of the lumbar and pelvic alignment between normal subjects and a group of spondylolysis patients. Bethune, (1986) proposes a statistical method using external spinal tracings obtained using a draftsman flexible rule and transferring the curves to graph paper. Merritt et al (1986) compares the reproducibility of three clinical methods for measuring trunk flexibility in normal subjects. Although this approach showed that two of the methods had good reproducibility, their application is limited to trunk movements only. Adams et al (1986) describe an inclinometer method for measuring the curvature of the lumbar spine. They demonstrated good agreement with subsequent radiographic measurements (r=0.91). However, his method did not include measurements of pelvic alignment which is essential for this study.

In summary, a variety of methods ranging from the classical orthopaedic X-ray evaluation to external reproduction of spinal curves have been used to measure and record the orientation of the spine and pelvis. A method modified from those used by During (1985) and Andersson (1979) has been adopted for use in this study. This method will be described in detail in a following section.

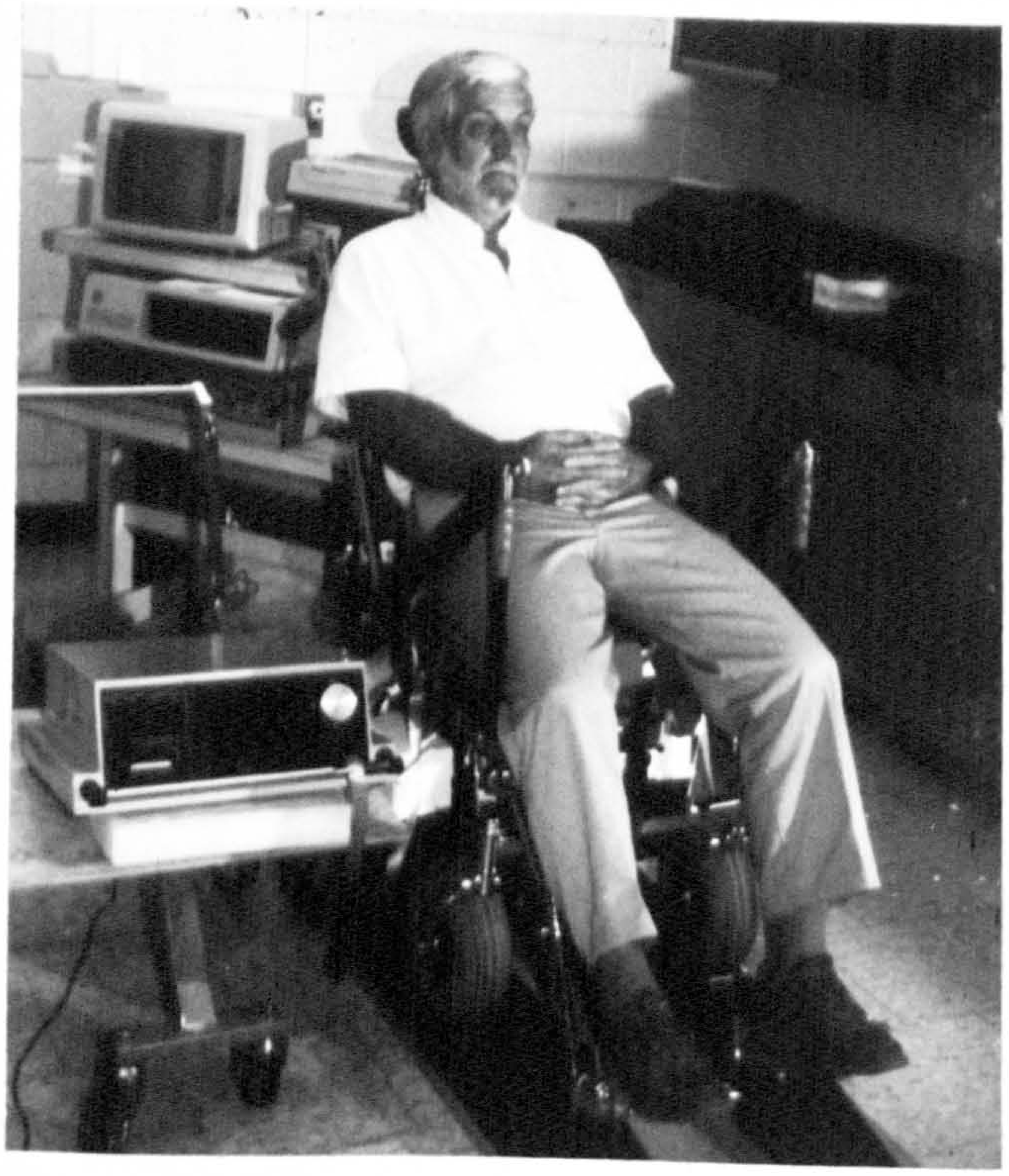


Fig 9.1 Labratory set-up with subject positioned on the body positioning chair (BPC). The BPC permits postioning of subjects in the nine postures studied. Two test positions are shown; forward trunk flexion 50°(upper) and backrest recline 20° (lower). Load cells mounted under the four wheels permit calculation of the centre of gravity of each subject. Load cells under the seat measure the seat friction in the fore and aft directions. Pneumatic transducers on the seat measure normal pressures.

CHAPTER 9. DESCRIPTION OF STANDARD POSTURES AND POSITIONING INSTRUMENTATION

9.1. THE POSITIONING INSTRUMENTATION

It is evident that able-bodied people assume a wide spectrum of sitting postures that could be described as a range of 'normal' sitting postures. In contrast, the sitting postures of people with disabilities is largely dictated by the nature of their disability and the configuration of the wheelchair seating arrangement. This spectrum of postural differences presents a basic problem when undertaking a study to compare differences in interface variables that may arise as a result of changes or differences in sitting postures of the two groups.

To resolve the problem a body positioning chair (BPC) was constructed (fig 9.1). The purpose of the BPC is to position the test subjects in reproducible sitting postures. The BPC uses a Fortress Scientific model 655 powered wheelchair with a universally adjustable seat frame as the base unit. The model 655 offers built-in independent adjustment of the backrest using a locking air spring. The complete seat and back assembly can be tilted in space with an electrically powered actuator. Built on the base unit are independent seat and back components. The seat component is mounted on an aluminum platform. The platform in turn contains four axial ball bearing assemblies mounted on highly polished and hard ground steel rods. This permits the seat to move with relatively low friction in the fore and aft directions. Two load cells mounted at either end of the platform measure the seat forces tending to move the seat platform in the fore or aft direction.

The seat itself is constructed of three layers of 25 mm (1") Sunmate(b) foam (type, soft blue, 5 lb/ft^3). The Sunmate foam can be described as a polyurethane foam with viscoelastic mechanical properties. The three layers of foam are mounted on a 12.5 mm (1/2") plywood board and covered with a lightweight, two-way stretch fabric. The back component is constructed in the same manner, but using only one layer of 12.5 mm (1/2 ") polyurethane foam. Both seat and back components can be adjusted linearly to ensure adequate thigh and back support for the seated subject.

The footrests are of standard wheelchair design and can be adjusted in height to meet individual differences in leg length. The armrests are designed to serve as guides or stops during lateral trunk bending trials. An inclinometer is mounted on the backrest upright so that back recline and tilt angles can be accurately determined. A vertically adjustable headrest provides a consistent reference point for A/P and mid-line

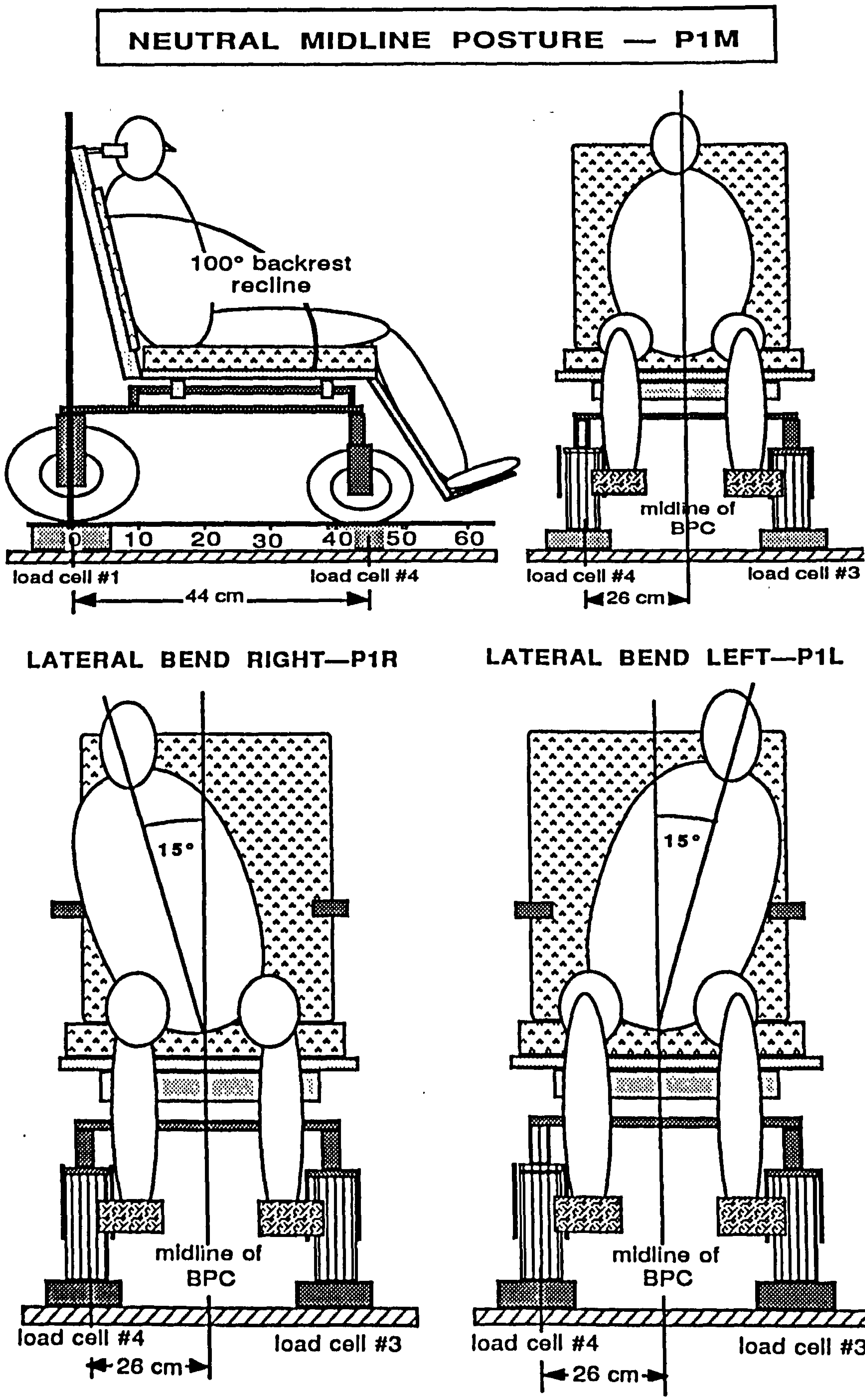


Fig 9.2 Definition of postures P1M, P1L and P1R.

positioning of the upper trunk and head. A centre line marked on the seat surface provides a consistent reference for pelvic positioning.

9.2. THE STANDARD STUDY POSTURES

An objective of the study is to detect differences in interface variables that may exist within and between the two study groups. And also how these differences may change as a person moves from one body posture to another, i.e., from upright sitting to 30° of forward trunk flexion. During the course of the study nine body postures were investigated using the BPC. Definitions of the nine standardized postures are as follows.

Position (P1M): Defined as the posture in the BPC with the pelvis placed as far posteriorly on the seat as possible, trunk and pelvis on the mid-line, and head in contact with the headrest. The seat surface is horizontal with the backrest reclined 10° to the vertical (100°). Arms are comfortably placed on the thighs and footrests are adjusted to take approximately 10% of the body weight (fig 9.2 upper).

Upright Trunk Bending - Left (P1L): Defined as the posture in the BPC the same as P1M above, except that the trunk is flexed to the left until the left side contacts the elevated armrests (approximately 15° of lateral trunk flexion (fig 9.2 lower right).

Upright Trunk Bending - Right (P1R): Defined as the posture in the BPC the same as P1M, except the trunk is flexed to the right side (fig 9.2 lower left).

Forward Trunk Flexion 30° (P2): Defined as the posture in the BPC the same as P1M, except the head and trunk are flexed forward 30° along the A/P mid-line (fig 9.3 upper left).

Forward Trunk Flexion 50° (P3): Defined as the posture in the BPC the same as P1M, except the head and trunk are flexed forward 50° along the A/P mid-line (fig 9.3 upper right).

Back Recline 110° (P4): Defined as the posture in the BPC the same as P1M, except the backrest is reclined to the 110° position (fig 9.3 lower left).

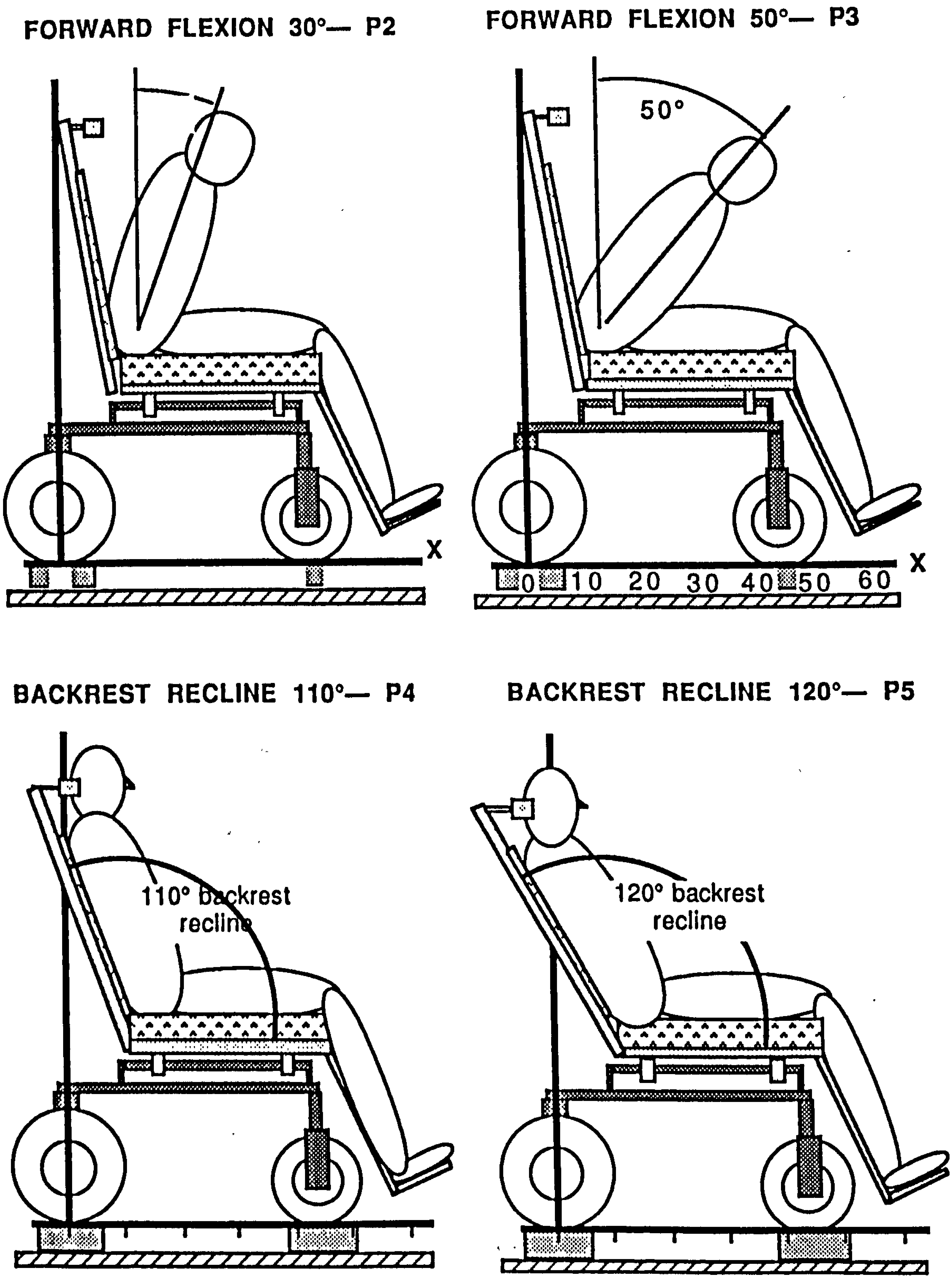


Fig 9.3 Definition of postures P2, P3, P4 and P5.

Back Recline 120° (P5): Defined as the posture in the BPC the same as P1M, except the backrest is reclined to the 120° position (fig 9.3-lower right) .

Body Recline 10° (P6): Defined as the posture in the BPC the same as P1M, except the whole seat assembly (seat and back) is tilted 10° in space (fig 9.4 left).

Body Recline 20° (P7): Defined as the posture in the BPC the same as P1M, except the whole seat assembly (seat and back) is tilted 20° in space (fig 9.4 right).

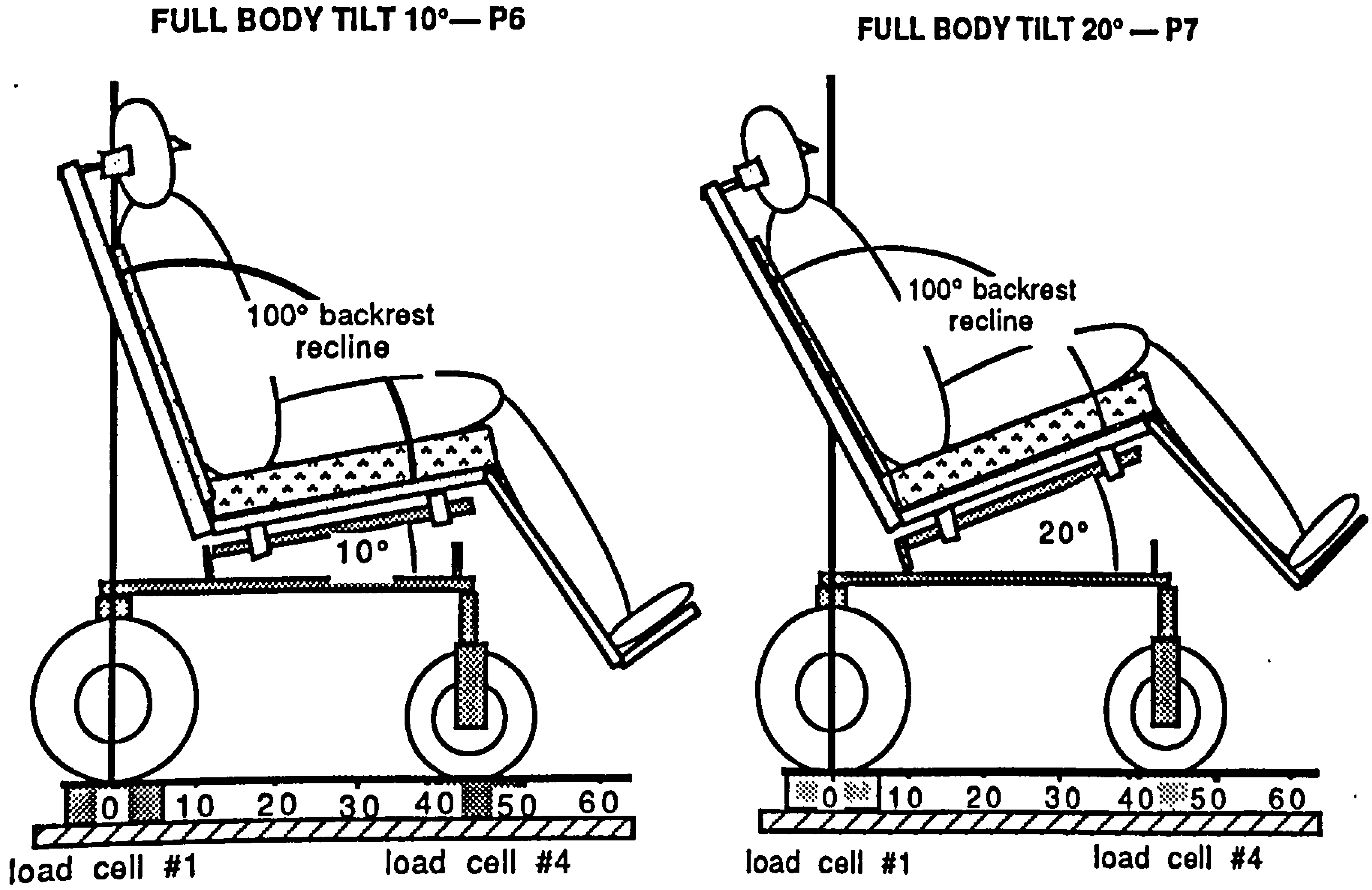


Fig 9.4 Definition of postures P6 and P7.

The essence of the study is the measurement and identification of the differences that may exist within and between normal and disabled subjects in the above nine standard postures.

CHAPTER 10. RADIOGRAPHIC COMPARISONS OF PELVIC AND SPINAL ALIGNMENT

10.1. THE OBJECTIVE

The hypothesis states that "---abnormal postures and deformities exhibited by individuals with spinal cord injuries can have a significant influence on body-seat variables---". This first component of the study addresses the question of whether or not abnormal postures and deformities can be shown to exist within the disabled population.

10.2. METHODS AND MATERIALS

10.2.1. The General Approach

Previous studies involving measurement of spinal deformities (vide 2.2) offers a number of possible methods by which spinal and pelvic alignment can be measured. Most of the methods are applicable to measurements of spinal deformity or range of motion, very few are applicable to pelvic alignment. The most accurate method appears to be a radiographic (x-ray) series containing the pertinent anatomical views. Unfortunately, most of the studies in the literature are retrospective in nature and little consistency has been observed in terms of body positioning and alignment while taking the radiographs. As a result, useful comparative values could not be drawn from these sources. Therefore, a modified radiographic procedure was developed and a series undertaken involving both the normal and disabled subjects.

The radiographic series includes two sagittal views (A/P) and two frontal views (M/L) of all subjects. Since the neutral posture (P1M) was common to both views only three unique postures were studied. The postures are three of the nine postures studied in the other three components of the study. The three radiographic positions were selected as being typical of seated postures in a wheelchair. A series involving all nine postures would have subjected the candidates to levels of radiation exposure which were deemed unethical and unnecessary. This phase of the study was conducted with the guidance and approval of Dr. Robert Tooms, an orthopedic surgeon.

Each subject was first placed in the upright seated posture (P1M) in the body positioning chair (BPC). The BPC was then carefully located relative to the x-ray head and two sagittal views taken, one in the P1M posture and the other in the P2 posture. The P1M posture is defined as the neutral reference posture (fig. 9.2). P2 is the posture of 30° forward trunk flexion. The positioning chair was then rotated 90° and two more radiographs taken in the frontal plane. The first was a repeat of the upright mid-line position (P1M) and the second was in a position of lateral trunk bending to the right

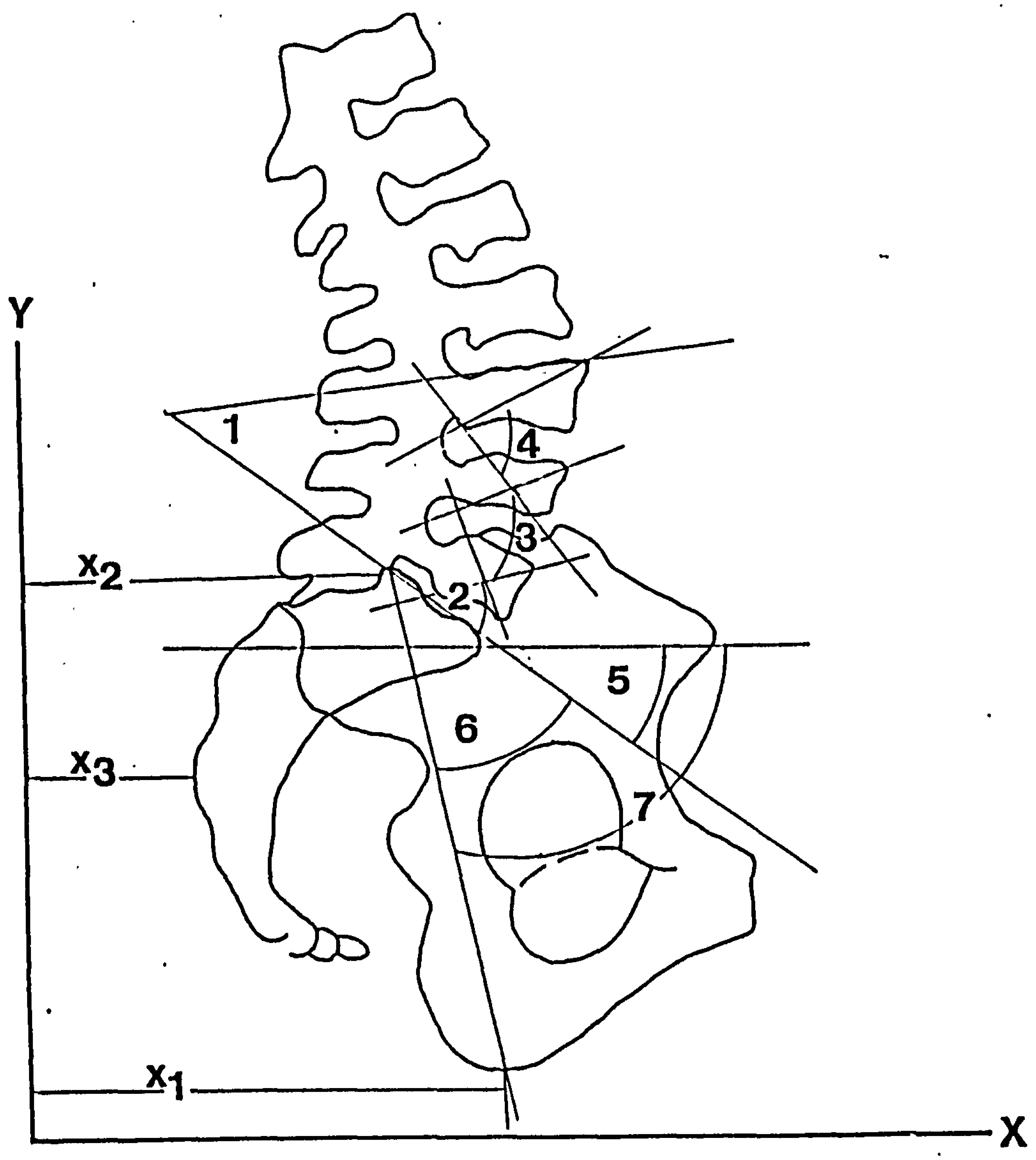


Fig. 10.1 Definitions of lumbar, sacral and pelvic angles and distances as measured from sagittal radiographs.

(P1R). The modified armrest of the BPC provided a consistent stopping point during lateral trunk bending. The lateral trunk bending in the frontal plane was approximately 15° (fig. 9.2).

Careful attention was given to both the position of the subjects in the BPC and the orientation of the chair relative to the radiographic equipment. After several experimental trials a setting was obtained for the positions of the x-ray head and this was maintained throughout the series. Markings on the floor assured accurate repositioning of the chair placement so that all subjects maintained the same distance from the film tray. The film tray was mounted on the wall and the chair moved up against it. This arrangement placed the film tray approximately 15 cm from the subjects trunk. Although parallax and other optical errors exist, they can be essentially ignored since they are the same for all subjects. In other words, comparisons of changes in anatomical alignment between subject groups rather than use of absolute values means that consistent optical errors essentially cancel out. However, this approach does not resolve the potential errors associated with interpretation and measurement of skeletal landmarks from the radiographs. This problem was resolved by placing a reference measurement (ruler) in the x-ray view so measured distances could be compared to known distances. Measurements were also made between bony landmarks and a fixed (metallic) reference on the BPC, both before and after changes in posture. Summation of distance measurements to the reference and directly between the bony landmarks permitted a check on the accuracy of bony landmark interpretation and distance measurements. It is estimated that an accuracy of ±0.5 cm on distance measurements and ±2° on angular measurements was achieved.

10.2.2. Analysis of Radiographs

A method modified from the procedures developed by Anderson et al (1979) and During and Goudfrooij (1985) has been used to identify and measure the critical variables that define lumbar, sacral, and pelvic alignment in the sagittal plane. The author developed the complete method for measurement of the pelvis in the frontal plane. In total eight angles and six distances were measured and recorded from each of the twelve disabled and ten normal subjects. The measured variables are defined as follows (the numbers refer to those in figure 10.1).

1- <u>Lumbar Angle (S1-L3)</u>: The angle defined by lines drawn along the superior plateau (sacral end plate) of S1 and the superior plateau of L3. Changes in angle S1-L3 describe

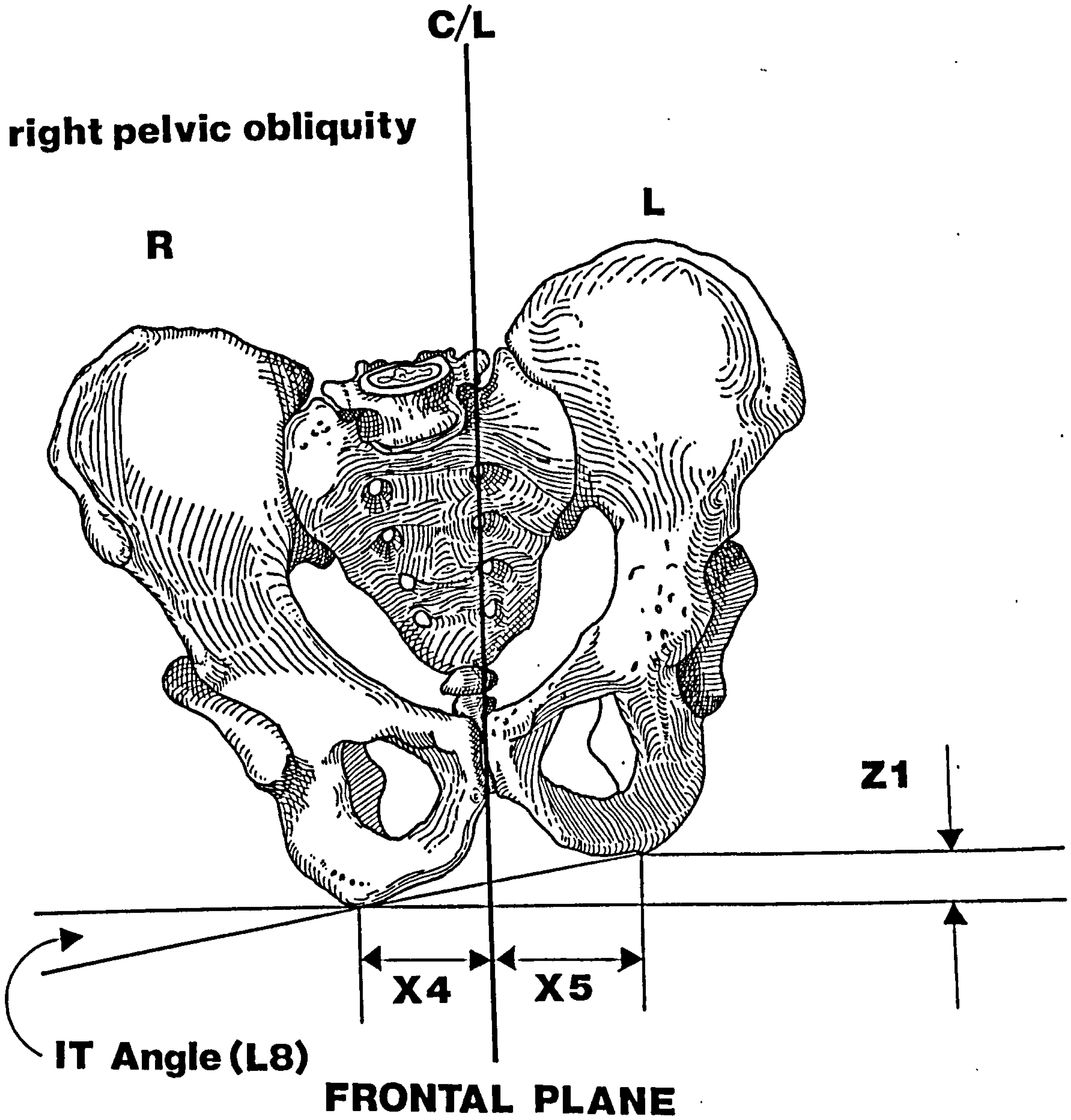


Fig. 10.2 Definitions of pelvic measurements in the frontal plane.

the total angular movement occurring in the lumbar segments of the spine between S1 and L3 in the sagittal plane.

2 - Sacral Angle (S1-L1): The angle defined by lines drawn through the anterior-superior and posterior-inferior aspects of the spinal segments L1 and the superior plateau of the sacral segment S1. Changes in angle S1-L1 describes the angular spinal movements between segments S1-L1 in the sagittal plane.

3 - Angle (L1-L2): The angle defined by a line drawn through the inferior-anterior and superior-posterior aspects of segment L1 and a line drawn through the inferior-posterior and superior-anterior aspects of segment L2. Changes in angle L1-L2 describes the angular spinal movement between lumbar segments L1 and L2 in the sagittal plane.

4 - Angle (L2-L3): The angle defined by a line drawn through the inferior-anterior and the superior-posterior aspects of lumbar body L2 and the inferior-posterior aspect and the superior-anterior aspect of lumbar body L3. Changes in angle L2-L3 describes the movement occurring between lumbar bodies L2 and L3 in the sagittal plane.

5 - Sacral Angle (SEP-HOR): The angle defined by a line drawn through the sacral end plate (SEP) and the horizontal reference line. This angle is often termed the "lumbosacral" angle and describes the angle or movement of the sacrum relative to the horizontal reference in the sagittal plane.

6 - Sacral - Pelvic Angle (SEP - IT): The angle defined by the line through the sacral end plate (SEP) and a line drawn between the superior-posterior aspect of the sacral end plate (S1) and the inferior aspect of the ischial tuberosity as determined in the upright mid-line position (P1M). Changes in angle SEP-IT describes the movement that occurs between the sacrum (S1) and the pelvis in the sagittal plane.

7 - Pelvic Angle: Defined by the sum of angles SEP-IT and SEP-HOR. The pelvic angle describes the angle of the pelvis relative to the horizontal which includes the movement of the sacrum relative to the pelvis in the sagittal plane. The movement of the sacrum (angle SEP-IT) is limited to 5° which decreases with age.

8 - IT Angle (LIT): Defined by a line joining the inferior aspects of the left and right ischial tuberosities and the horizontal reference line drawn through the lowest ischial

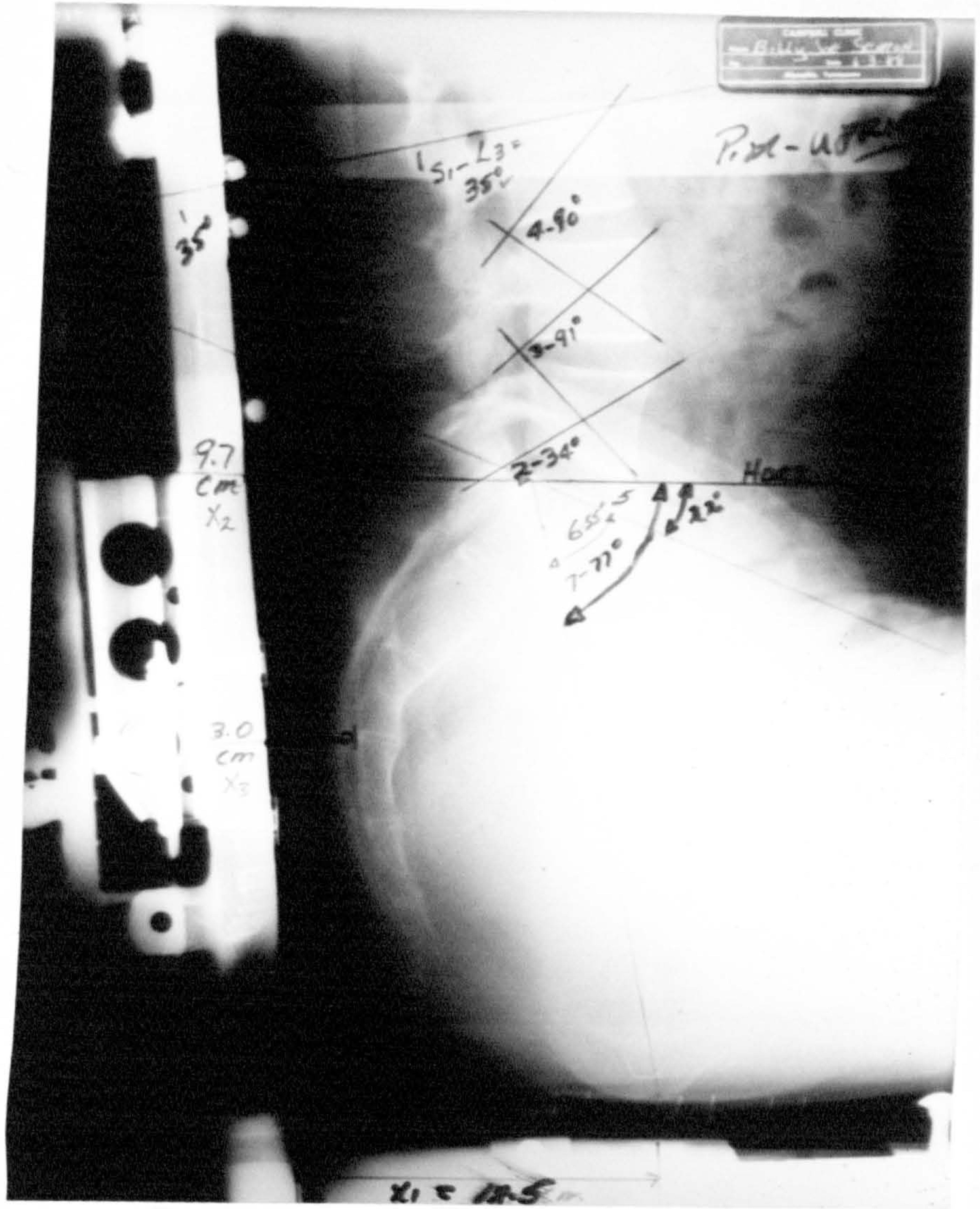


Fig 10.3 Illustrative photograph of an A/P x-ray indicating the seven angles and three distance measurements used to define the orientation and displacement of the pelvis (see definitions in fig 10.1). Posture shown is the neutral P1M position.

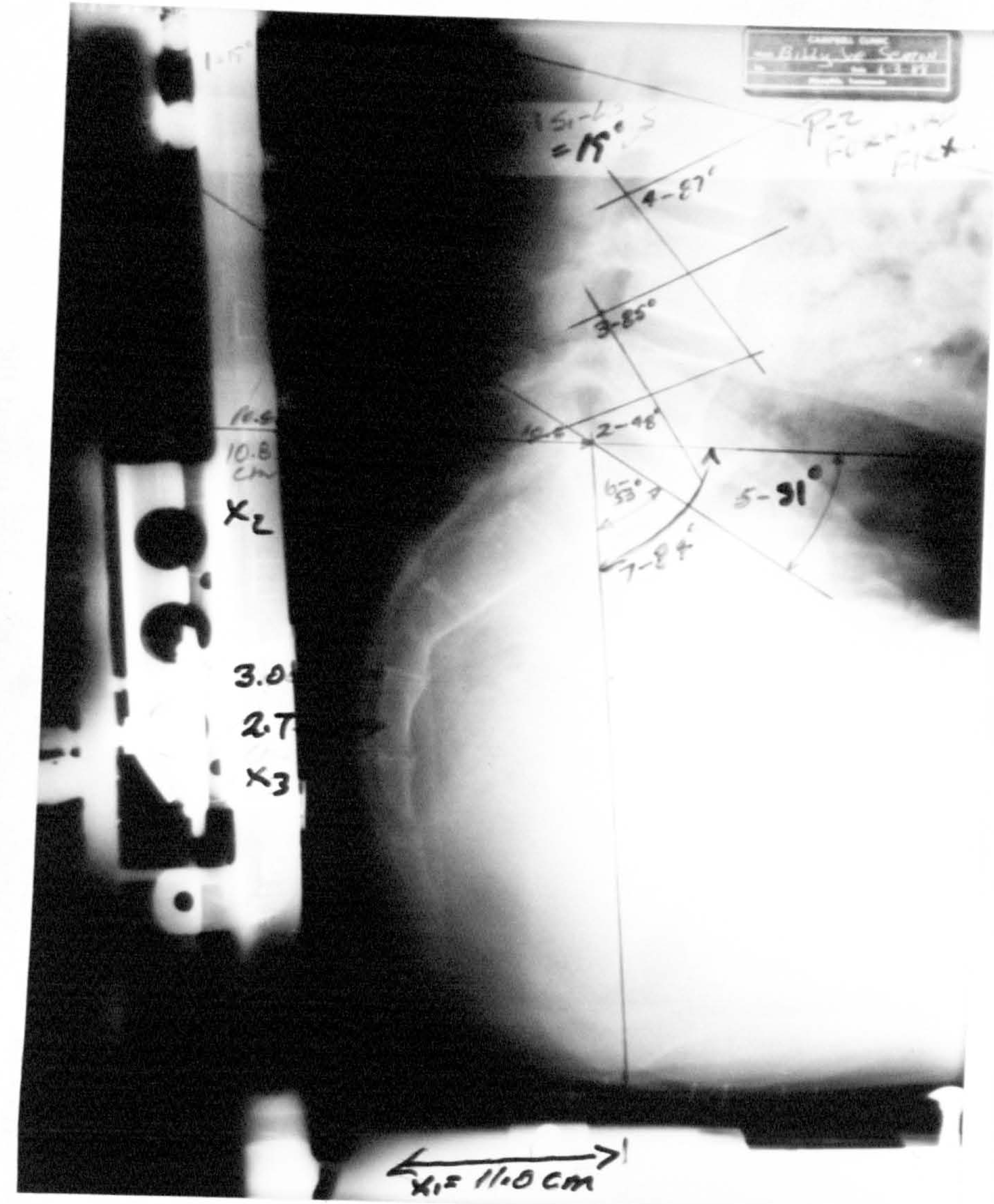


Fig 10.4 Illustrative photograph of same subject as in figure 10.3 after trunk flexion to 30°, P2 posture.

tuberosity. This angle defines the angle of the pelvis in the frontal plane (obliquity or lateral tilt angle)(fig 10.2).

X_1 - <u>IT - Reference Distance</u>: A measurement of the horizontal distance from the inferior aspect of the ischial tuberosity to the reference axis. The distance X1 describes the location or movement of the ischial tuberosities in the X direction as measured in the sagittal plane (fig.10.1).

X_2 - <u>SEP-Reference Distance</u>: A measurement of the horizontal distance from the superior-posterior aspect of S1 to the reference axis. The X2 measurement describes the location or movement of S1 in the X direction as measured in the sagittal plane.

X_3 - <u>S3/4 - Reference</u>: A measurement of the horizontal distance from the apex of sacral bodies S3/S4 to the reference axis. X3 describes the movement of the apex of the sacrum (S3/S4) in the X direction.

Z_1 - <u>IT VERT</u>: A measurement of the vertical position of the inferior aspect of the left ischial tuberosity as viewed in the frontal radiograph. Z1 describes the vertical position and movement of the left ischial tuberosity during mid-line sitting (P1M) and during right lateral trunk flexion (P1R) (fig 10.2).

X_4 - <u>RIT-C/L</u>: A measurement of the horizontal position and movement of the right ischial tuberosity relative to the mid-line reference axis in the frontal plane. X4 describes the position and lateral movement of the right ischial tuberosity during mid-line sitting (P1M) and during right lateral flexion (P1R) (fig 10.2).

X_5 - <u>LIT-C/L</u>: A measurement of the horizontal position and movement of the left ischial tuberosity relative to the mid-line reference axis in the frontal plane. X5 describes the position and lateral movement of the left ischial tuberosity during mid-line sitting (P1M) and during right lateral flexion (P1R) (fig 10.2).

As indicated the above angles and distances were measured for three body positions; the mid-line neutral position (P1M), 30° of forward trunk flexion (P2), and right lateral trunk bending (P1R). A smaller sample of subjects also had radiographs taken in the 50° forward trunk flexion position (P3). Mean values, standard deviations and standard errors were computed for the subjects within the two samples. Figures 10.3

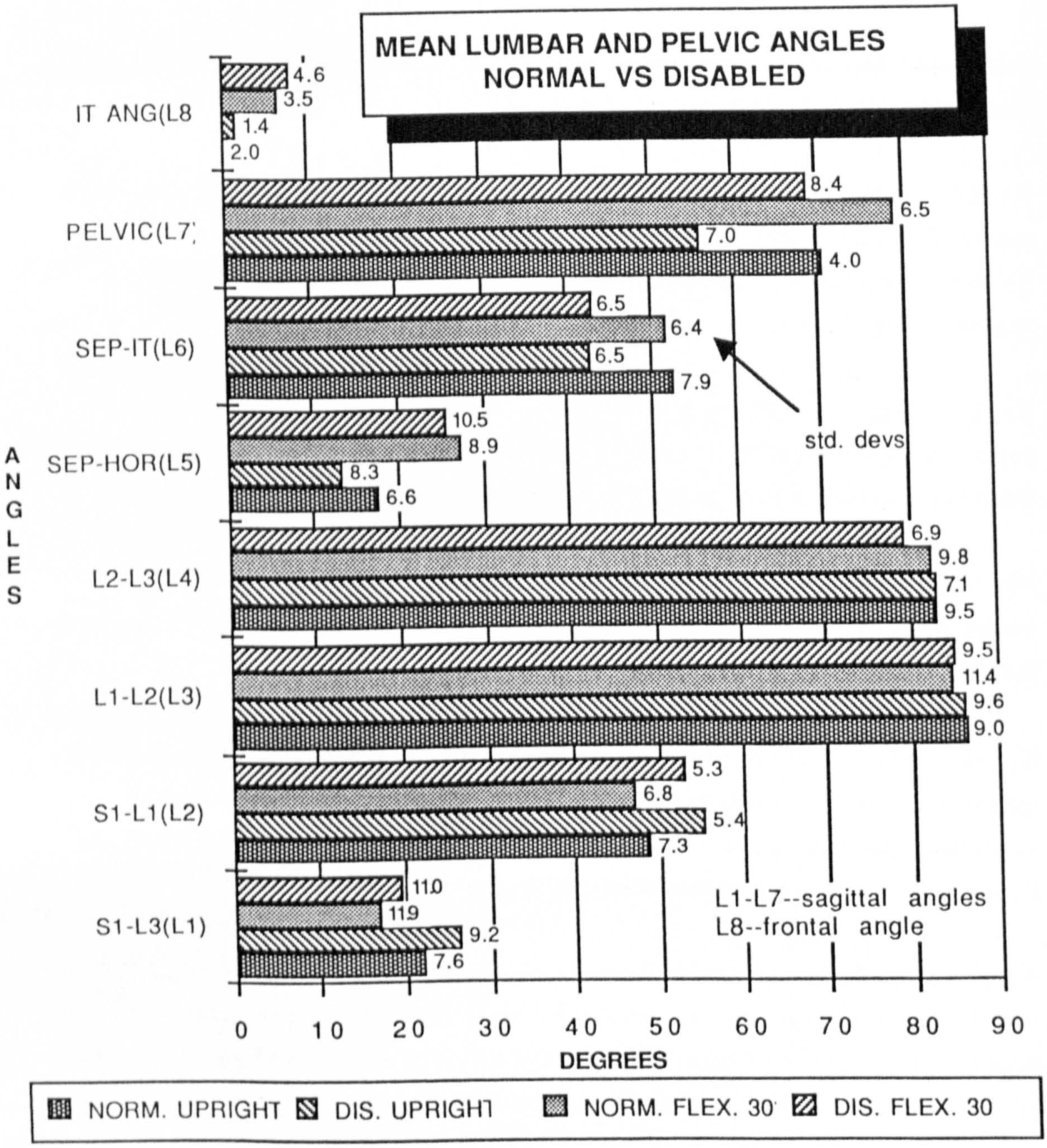


Fig 10.5 Nine mean lumbar, lumbosacral and pelvic angles measured from A/P (sagittal) and M/L (frontal) radiographs. Angles (L1) to (L7) are measured in the A/P plane and (L8) in the M/L. Values shown are for the upright posture (P1M) and the 30° trunk flexion posture (P2), for both disabled (n=12) and normal (n=10) groups. Values shown at the end of each bar are values for one standard deviation. Tabular values are provided in Table 10.1

and 10.4 are photographs of representative x-rays; showing both angular and distance measurements in the sagittal plane, the P1M erect posture (fig10.3) and the flexed posture (fig 10.4).

10.3. RESULTS OF RADIOGRAPHIC COMPARISONS

Table 10.1 and figures 10.5 to 10.8 contain the results of the radiographic measurements. The means and standard deviations of the angles and distances for both sample groups were computed and charted. The results of the angular spinal and pelvic measurements in the sagittal plane and the pelvic obliquity in the frontal plane (IT ANG) can be seen in table 10.1 and figure 10.5. The mean values plotted are for both groups in two body positions, neutral (P1M) and forward flexion to 30° (P2). Standard deviations provide an indication of the scatter of the data points within each group (n=10 for normal, n=11 for disabled).

The objective is to determine differences in spinal and pelvic alignment that may exist within and between the two groups as the subjects move from one sitting posture to another (P1M to P2).

Table 10.1 Tabulation of mean angular values (M) and standard deviations (SD), from A/P and M/L planes taken in two body positions, P1M and P2. The tabular values were used to plot the chart in fig 10.5.

		NORMAL ANGLES				DISABLED ANGLES			
VARIABLE	REF	P1M(Erect)		P2(30°)		P1M(Erect)		P2(30°)	
		M	SD	M	SD	M	SD	M	SD
S1-L3	L1	22.1	7.6	17.0	11.9	29.5	9.2	19.6	11.0
S1-L1	L2	48.9	7.3	47.1	6.8	55.4	5.4	53.1	5.3
L1-L2	L3	86.2	9.0	84.6	11.4	85.9	9.6	84.9	9.5
L2-L3	L4	83.1	9.5	82.3	9.8	83.1	7.1	79.5	6.9
SEP-HOR	L5	17.8	6.6	27.5	8.9	13.4	8.3	25.7	10.5
SEP-IT	L6	53.0	7.9	51.7	6.4	42.9	6.5	43.2	6.5
PELVIC ANG.	L7	70.8	4.0	79.2	6.5	56.3	7.0	68.8	8.4
				P1R(15°)				P1R(15°)	
IT ANG.	L8	1.4	2.0	6.4	3.5	3.2	1.4	7.9	4.6

Angles L1-L7 in figure 10.5 give four mean values per series for each of the seven angles measured. The first set in each series shows the angles for both normal and disabled subjects in the neutral (P1M) position. The second set in each series shows the values for the same angles after the trunk has been flexed forward 30° (P2). IT ANG (L8) shows the mean pelvic obliquity differences between the neutral P1M and lateral bending (P1R) positions in the frontal plane. A preliminary summary of the results can be made by focusing on angles (L1), (L7), and (L8) in fig 10.5. Lumbar angle, (L1), is a

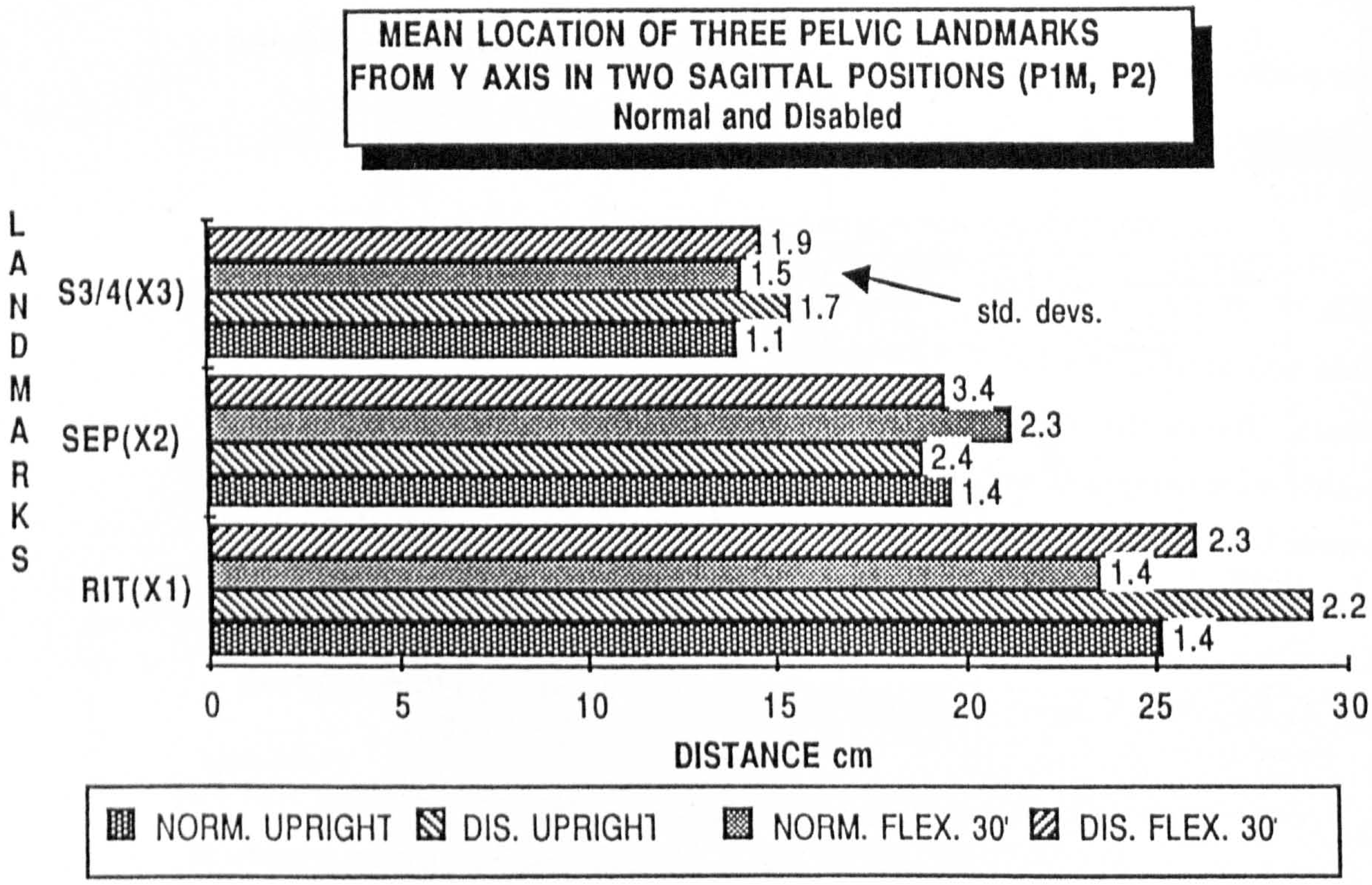


Fig 10.6 The mean x distance location in the sagittal plane of three pelvic landmarks; the right ischial tuberosity (X1),the posterior sacral end plate (X2) and the apex of S3-4 (X3) in two postures, upright and flexed forward 30°.

Table 10.2 Tabular values used to plot above chart (fig 10.6).
Mean values (M) and standard deviations (SD).

		NORMAL cm				DISABLED cm			
VARIABLE	REF	P1M(Erect)		P2(30°)		P1M(Erect)		P2(30°)	
		M	SD	M	SD	M	SD	M	SD
RIT	X1	25.2	1.4	23.6	1.4	29.1	2.2	26.1	2.3
SEP	X2	19.7	1.4	21.3	2.4	18.9	2.4	19.5	3.4
S3/4	X3	14.0	1.1	14.2	1.7	15.5	1.7	14.7	1.9

result of lumbar segment angle values (L2), (L3), and (L4). Pelvic angle (L7) is a result of the sum of angular values (L5) and (L6).

Regarding the lumbar angle (L1) the normal group shows an apparent mean value of 22° (std. dev. = 7.6), and the disabled 26°, (std. dev. = 9.2); with a mean difference of 4°. When flexed forward 30° (P2) the mean values change to 17° and 20° respectively, with the mean difference reducing to 3°. This suggests that the disabled group on the average have more lumbar lordosis in the upright sitting position compared to the normal group. Upon forward trunk flexion to 30° the lumbar curve (L1) reduces on the average of 5° in the normal group and 6° in the disabled, while retaining an apparent average difference of 3° in the flexed position.

Pelvic angle, (L7), shows the normal group to have a mean pelvic angle of 71° in the neutral (P1M) position, whereas the disabled group has a mean pelvic angle of 56°; a 15° mean difference. A reduced pelvic angle means that the pelvis has rotated posteriorly. A change in body position to P2 shows mean pelvic angle change to 79° for the normal group and 69° for the disabled group. As expected, the pelvic angle increases with forward flexion, an average of 8° for the normal and 13° for the disabled.

L8 shows the pelvic obliquity in the frontal plane as measured according to figure (10.2). L8 can be either positive or negative but only absolute values were used in the computational comparisons. A left tuberosity higher than the right side is defined as a positive angle. Figure 10.2 shows the mean value of (L8) in the frontal plane in both the neutral P1M and right bending P1R positions. As expected, the mean (L8) values for normal subjects in the P1M position is very close to 0°, i.e., 1.4 degrees, which is too small to show on the plot. In contrast, the disabled group show an mean pelvic obliquity to the right of 3.2° in the P1M position, i.e., an elevated left side. Upon lateral bending to the right (P1R) the mean normal values increase to 6° and the disabled to 8°; retaining an apparent difference of about 2°. This difference is insignificant based on the estimated of measurement accuracy and the standard deviations computed.

A closer examination of the lower four plots in figure 10.5 provides insight into the relative movement that takes place in the lumbosacral-sacral area. Angle (L1) gives the lumbar angle between spinal segments S1-L3. Plots (L2), (L3) and (L4) provides angular displacement information between segments S1-L1, L1-L2, and L2-L3, respectively; both in the neutral posture and after 30° of forward trunk flexion. It can be seen that most of the motion takes place at the lumbosacral-sacral joint (S1-L1(L2)). Relatively little angular motion takes place between segments L1-L2 or L2-L3.

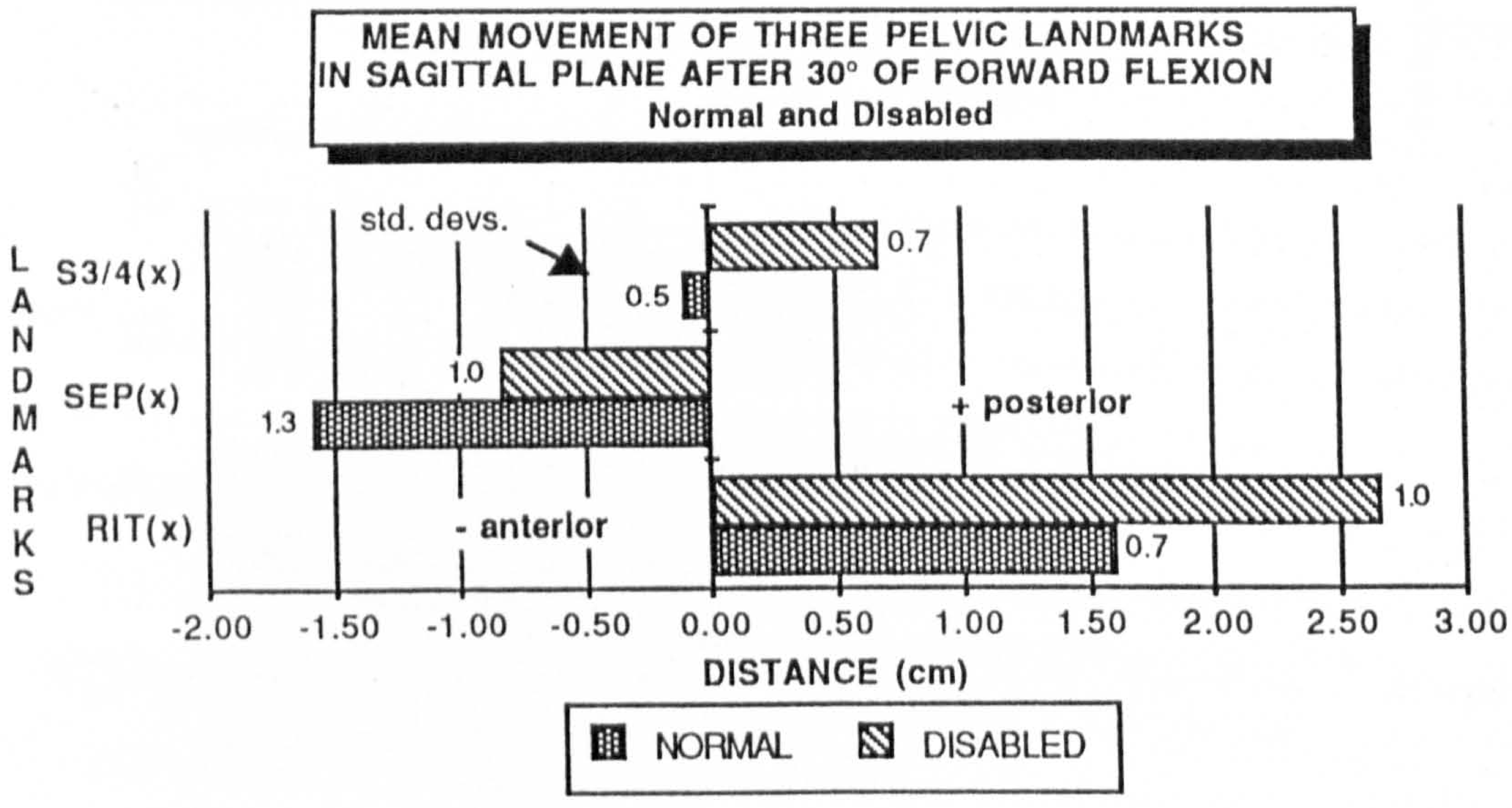


Fig 10.7 The mean movement in the sagittal plane of three reference pelvic landmarks (RIT, SEP, S3/4) upon forward trunk flexion of 30° (P2) from the neutral posture P1M.

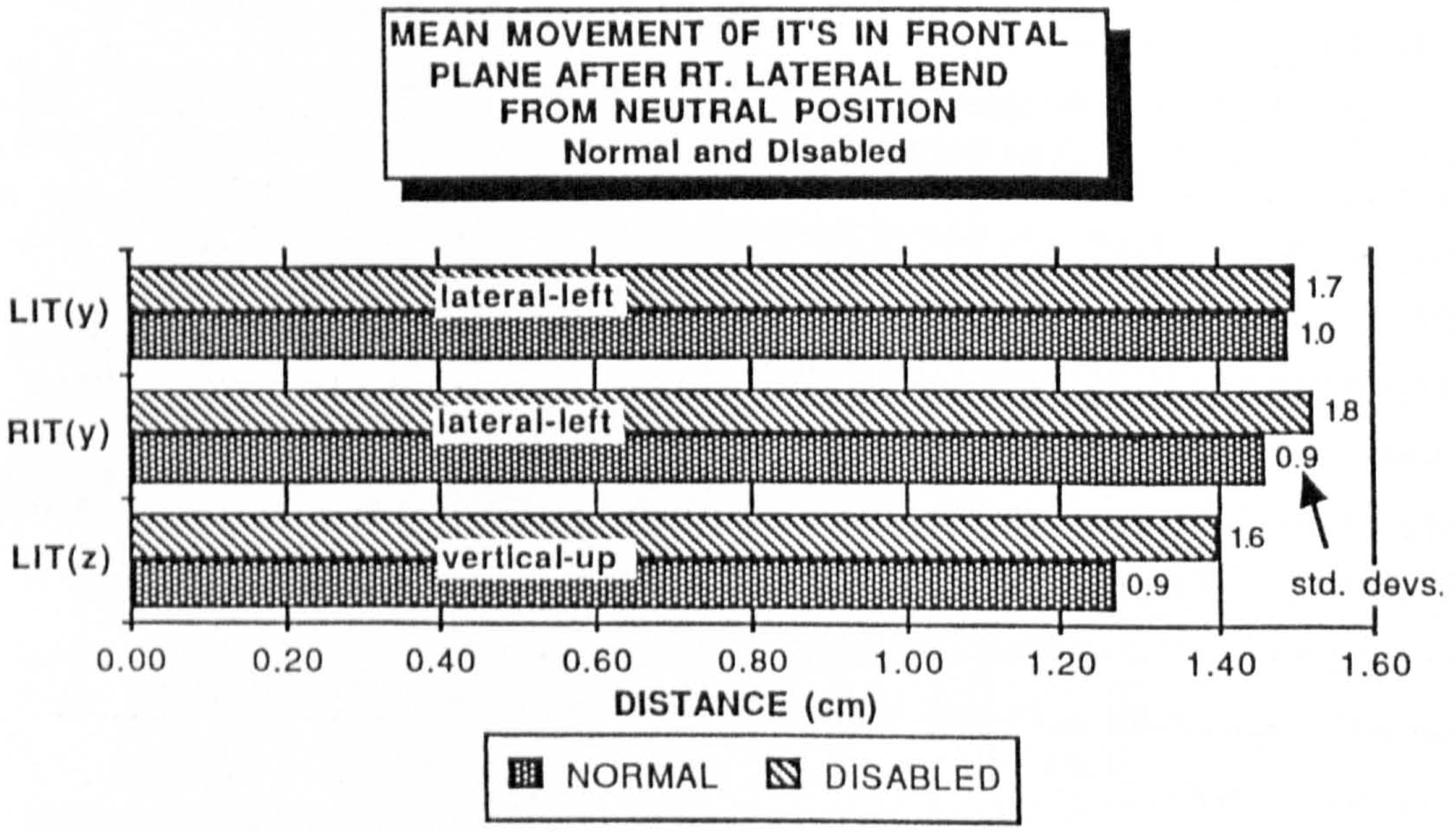


Fig 10.8 Mean movement of left and right ischial tuberosities (LIT, RIT) in the left lateral direction (y) and of the LIT in the vertical (z) direction upon right trunk bending.

The results shown in figures 10.6-10.8 and table 10.2 are mean linear displacements of the chosen pelvic landmarks when movement takes place from the P1M position to the P2 and P1R positions. Figure 10.6 shows the actual forward position of the pelvic landmarks RIT, SEP, and S3/4, in the sagittal plane relative to a vertical reference located approximately in the plane of the backrest of the BPC. Examination of the mean values for the right ischial tuberosity (RIT(X1)) indicates mean values of 25 cm for the normal group in the upright position (P1M) and 29 cm for the disabled group. After changing the posture to 30° of forward flexion (P2) the normal group's mean RIT location reduces to 24 cm and the disabled to 26 cm. In literal terms it appears that the ischial tuberosities actually move posteriorly by the differences between the above mean values within the two groups. Results for the two positions of the other pelvic landmarks; the sacral end plate (SEP) and the sacral segments (S 3/4) can also be seen in figure 10.4.

Figure 10.7 contains the same information as figure 10.4, except it emphasizes the displacement of the pelvic landmarks that takes place during a change of body position from P1M to P2. The plots show the apparent mean displacement both within and between the normal and disabled groups. Of particular interest is the movement of the right ischial tuberosity (RIT). The mean RIT posterior displacement of the normal group is about 1.6 cm, whereas the disabled group displaces about 2.6 cm in the posterior direction. Similar analysis shows that the sacral end plate (SEP) moves forward, with the normal group moving approximately twice as far as the disabled. Also, the apex of the sacrum (S 3/4) on the average moves slightly anteriorly in the normal group, but moves posteriorly an average of about 0.7 cm in the disabled group. This difference between the two groups is probably insignificant given the accuracy of measurement and computed standard deviations.

Figure 10.8 gives the results of mean lateral movements of the two pelvic landmarks in the frontal plane; the left ischial tuberosity (LIT) and the right ischial tuberosity (RIT). Movement in the Y direction denotes lateral movement and movement in the Z direction vertical movement. The two body positions plotted are for upright neutral sitting (P1M) and right lateral bending (P1R). Upon lateral bending to the right the left ischial tuberosity (LIT(z)) moves vertically upward an average of 1.25 cm for the normal group and 1.40 cm for the disabled. Also, both the right (RIT) and left (LIT) ischial tuberosities move laterally to the left approximately 1.5 cm. Again the values shown at the end of each plot are standard deviations.

The above results have been stated without statistical analysis, which is discussed in the chapter on conclusions (chapter 14).

CHAPTER 11. COMPARISONS OF PRESSURE DISTRIBUTIONS

11.1. THE OBJECTIVE

This component of the study has been designed to evaluate the pressure distributions that occur at the body-seat interface surface. The aim is to identify pressure distribution differences that may exist within and between the normal and disabled populations and how the distributions may be affected as a result of deformity and/or alterations in sitting postures.

11.2. METHODS AND MATERIALS

11.2.1. The General Approach

The same subjects, 10 normal and 12 disabled, participated in this phase of the study. Two Oxford Pressure Monitors (OPM) were used to measure the pressure values under the buttock area. Four, 3 x 4 cell matrix transducers (48 cells), were fixed to the seat surface so that their positions relative to a reference axis would remain constant. The transducer array was placed so that the ischial tuberosities of all subjects would fall on the back two arrays (24 cells). The front two transducer arrays were used only to assure that the maximum pressure values actually occurred within the back two arrays.

The subjects were placed in the P1M position on the BPC and pressure recordings taken in the P1M and all remaining 8 standard positions (P1L to P7), (vide 9.2). Maximum pressure values, location of maximum values, mean average pressure values and mean peak pressure gradients were all computed from the recorded data. The seat cushion used was the same as previously described (vide 9.1). All subjects were given a fifteen minute free movement period at the beginning of each recording series in order to allow the seat foam and measurement equipment to reach steady state. All pressure recordings were taken by the same two people, each using the same OPM. The properties of the seating foam could be described as being firm which meant that relatively small indentations of the seating surface occurred. This approach minimized the curvatures to which the transducers were subjected, but also increased the average peak pressure values higher than those which would be clinically acceptable. The subjects were not removed from the BPC between changes in body positions, thereby minimizing errors that could result from inaccuracies associated with reseating subjects (Bader, 1982). Successive pressure recordings were taken in one posture in order to conduct repeatability and variability checks on the pressure measurements.

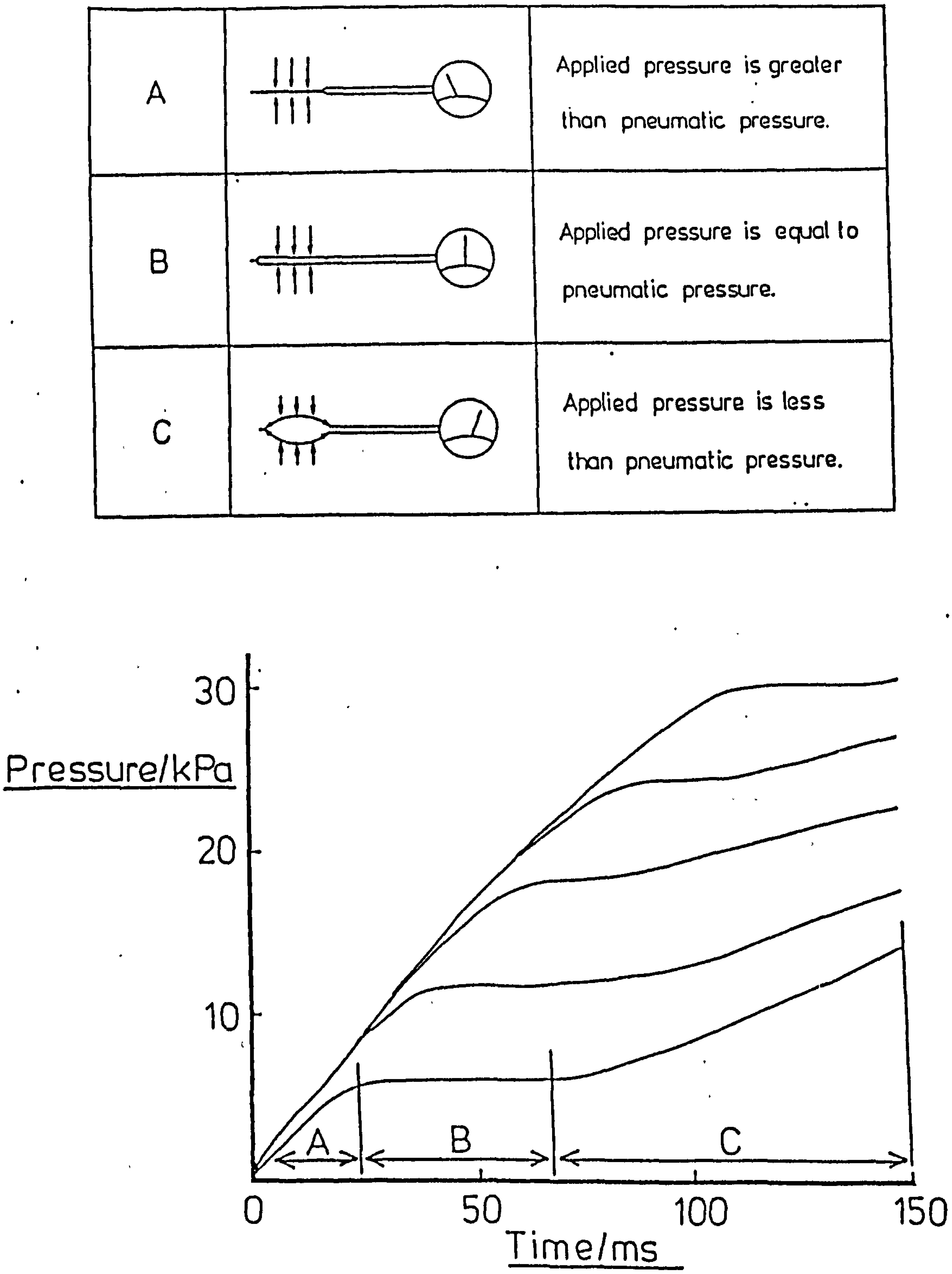


Fig 11.1 Schematic of operation of the Oxford Pressure Monitor (OPM) (from Bader and Hawken, 1986).

Positioning of the arms and support of the forward flexed trunk required a procedural compromise. First, the neutral P1M posture calls for the hands to be placed on the thighs so upper body weight is not transmitted via the arm rests. Postures P2 and P3 are defined as 30° and 50° of forward flexion from the P1M posture, respectively. However, since most of the disabled subjects could not support their torso in the flexed P3 posture without arm support the following procedure was used for all subjects in the forward flexed posture, P3.

A padded bar was secured across the armrests at the anterior location that yielded a trunk flexion of 50° (P3). Subjects rested their full trunk weight against the pad with hands placed on their knees (fig 9.1). This compromise would give erroneous results if absolute values were used. However, since relative values were used in making the comparisons this compromise was deemed acceptabled.

Calibration of the equipment was done as per the developer's recommendations at the beginning of each trial, and again at the end of those trials in which any comparatively high or low values were observed. Each of the twelve pressure cells on each of the OPM units had to be monitored manually through rotation of a manifold dial. This means that its a decision of the operator as to when pressure equilibrium has been achieved in the cell being monitored and its time to change to the next cell. The two operators agreed to wait until such time that it was clear that a maximum value had been achieved and only minor changes were occurring around the maximum observed value. This time was approximately 5 to 10 seconds. It was further noted that each operator finished the scanning of their 12 cells in about the same time, thereby indicating that the desired consistency was retained throughout the series.

11.2.2. Description of the Oxford Pressure Monitor (OPM)

The OPM was developed at the Oxford Orthopedic Engineering Centre, Nuffield Orthopedic Centre, Oxford, England. The OPM is being manufactured and marketed by Talley Medical Ltd. (a), London, England. The OPM consists of three basic components; the pneumatic transducer array, the monitor and the calibration jig.

The transducers are supplied in either a 3 x 4 matrix configuration, a 1 x 6 matrix, or as single cells. An important feature of the OPM development is that there is no electrical contacts necessary within the pneumatic transducer itself. Also, a transducer cell is only inflated when a pressure reading is taking place thereby minimizing the perturbation effect on surrounding body contact areas. The 3 x 4 matrix configuration used in the study is comprised of 12-20 mm diameter cells mounted on 28 mm centres. This matrix provides an effective measurement area of approximately 8,000 mm^2. Each cell in the matrix is fed by a small diameter flexible tube which plugs into the monitor manifold switch. The thickness of the deflated matrix is approximately 2-3 mm.

The OPM works on the principle first described by O'Leary and Liddy, (1978) and shown in figure 11.1. Initially the pressure in the system increases linearly up to the applied pressure (phase A). During this phase the cell remains flat at the interface and the incoming air only pressurizes the connecting air lines. As the applied pressure is reached and exceeded, the cells start to inflate, the system volume increases and the rise in pressure is smaller for a given injected volume of air (phase B). The exact slope of the curve is largely determined by the compliance of the interface materials. In the final phase (phase C) the slope of the curve increases dramatically, as the cell has become fully inflated and the resulting tension in its walls resists further changes in volume. Although fig 11.1 shows a family of curves up to 30 kPa the working range generally used is within the lower three curves (0 -20kPa, 0 -150 mm Hg).

The OPM uses the minimum injected volume of air to measure an applied pressure and therefore does not depreciably deform the interface under investigation. This has been achieved by pressurizing the system with a constant flow rate, as controlled by a high pressure pump and needle valve. The pressure/time characteristics of this system are continuously monitored by an electronic circuit. The circuit is programmed to detect the point of changeover from Phase A to Phase B (fig 11.1), when the system pressure equals the applied pressure and the cell starts to open. At this instant of time, a static measurement of system pressure is obtained using a semiconductor strain gauged pressure monitor (range 0 - 100 kPa; 0 - 750 mmHg). The air flow is then reversed and the cell is exhausted. The frequency of the total cycle necessarily depends upon the applied pressure, which is in excess of 2 Hz at the maximum pressure limit of 250 mmHg. The OPM is battery-powered by a 12 volt (6 Ah) rechargeable Ni-Cd battery. The stated pressure range is from 0 - 300 mmHg with a resolution of 1 mmHg. Calibration data published by the developers (Bader and Hawken, 1986) show a maximum deviation from linearity of 3 percent over the optimum range of 0-250 mmHg.

The calibration jig is composed of three components; a pressure guage, a double sided vinyl bladder and a slotted compression box. The procedure involves placing the cell transducer matrix between the two halves of the inflatable bladder. The bladder/transducer assembly is then placed in the compression box and the bladder is inflated to 100 mmHg as indicated by an interconnected external pressure guage. The manifold of the OPM is then rotated through the various cell settings and a variation from the 100 mmHg calibration value is noted. A calibration setting on the side of the OPM allows adjustment of the recorded values to the 100 mmHg calibration value. This procedure permitted calibration of the two OPM units used to within ±3% of the 100 mmHg calibration level across the cell matrix. This level of accuracy was achieved consistently throughout the recording session.

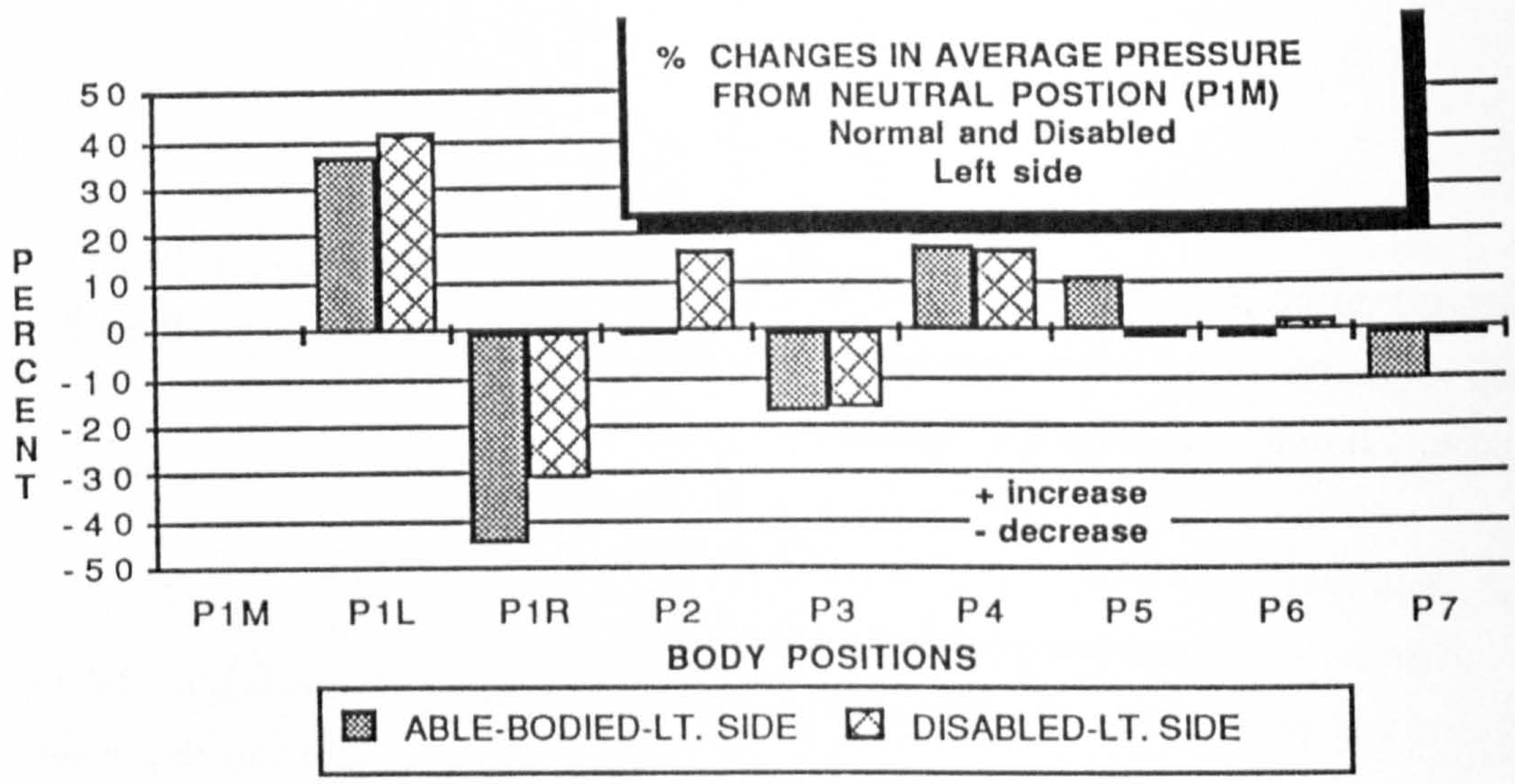
11.2.3. Rationale for the Selection of the Oxford Pressure Monitor

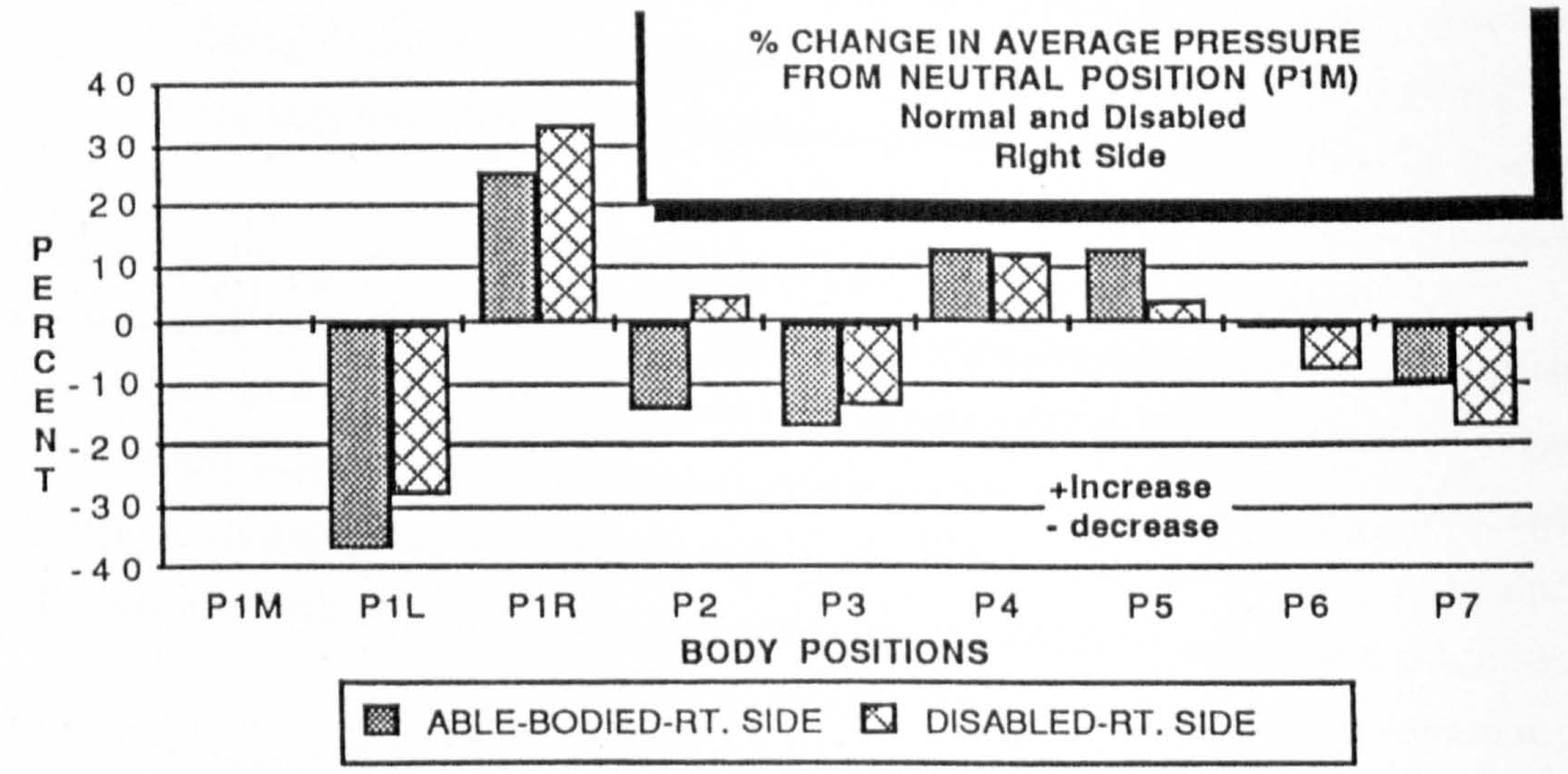
The review of previous work on quantification of pressure distribution (vide 3.4.2) revealed a wide selection of possible approaches. Each approach is not without its limitations. In general, miniature electronic transducers proved difficult to position accurately, are expensive, and are not readily available in matrix configurations that could be applied to the needs of this study. Also, reports of hysteresis problems, contamination by tangential forces and electrical interference discouraged use of miniature electronic devices.

Preliminary trials were conducted with the TIPE device developed at the Texas Institute for Rehabilitation Research in Houston, Texas (vide 3.4.2). The device uses a twelve by twelve pneumatic transducer matrix which is attractive because of the large surface area covered. However, the maximum pressure of 150 mm Hg was found inadequate. Also, consistent problems were encountered with the electrical contacts within the transducers. And finally, no calibration method is provided with the TIPE system making it difficult to conduct routine calibrations.

Suggestions have been made that accuracy is improved when a single transducer cell is attached directly to the skin over an ischial tuberosity, rather than placing an array on the test surface. The radiographic results show that the ischial tuberosities shift approximately 4 cm during the complete range of movements studied. Therefore, the single cell skin method is not appropriate for experiments in which large changes in trunk positions are to be studied.

Finally, three years of previous experience has shown the OPM to be a reliable and durable device that provides reasonably accurate results within known limitations. Linearity and variability tests have been done and published by the developers (Bader, 1982; Bader and Hawken, 1986). The OPM is commercially available and for this and the above reasons it was chosen as the most appropriate pressure measurement system to use in the study. Furthermore, the fundamental approach of using relative rather than absolute values with the same seat cushion means that the errors associated with the OPM will at least be consistent and therefore largely self cancelling (Reddy et al,1984).





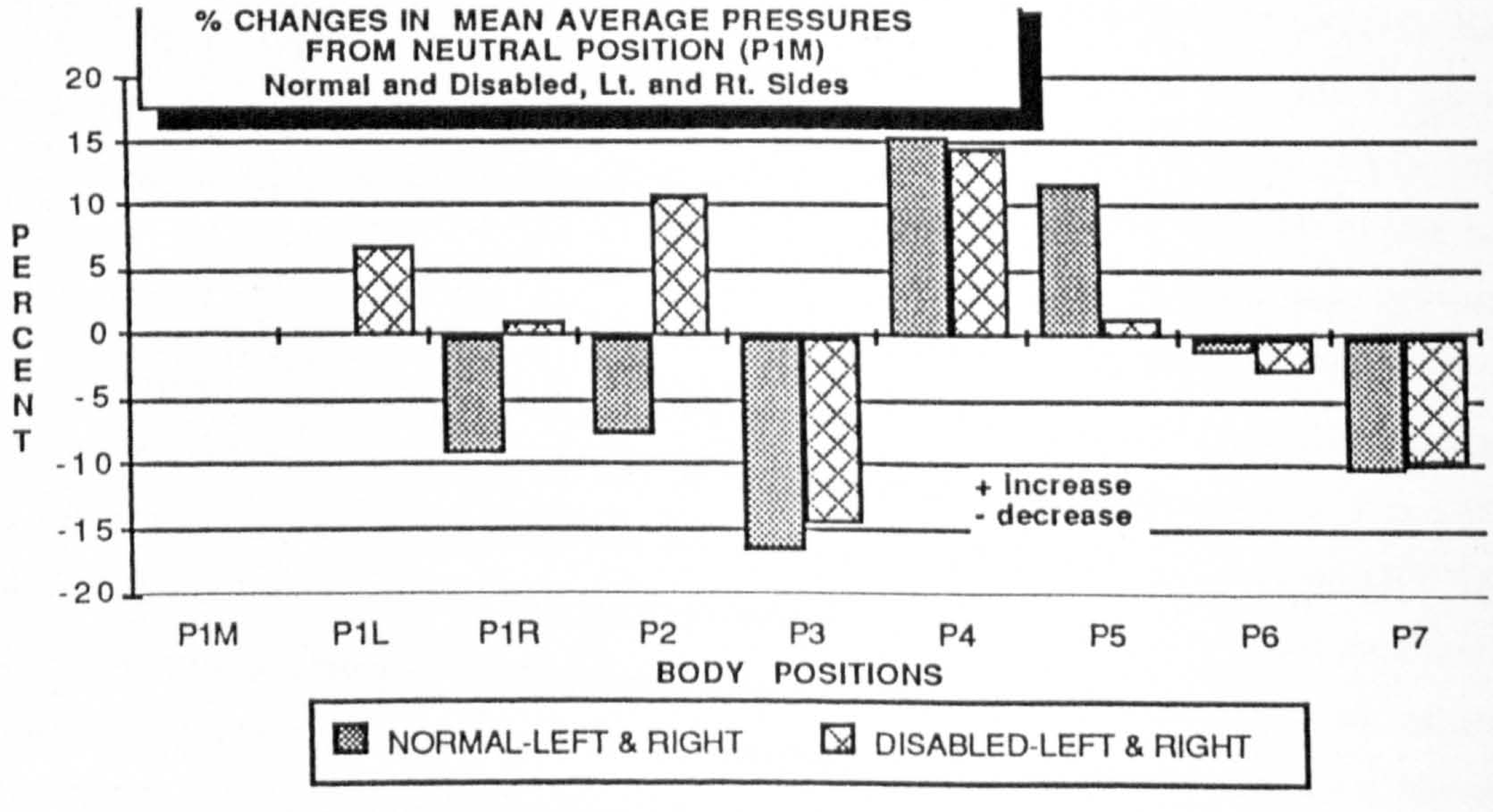


Fig 11.2 Percent change in average pressures from those in neutral position(P1M); left side, right side and mean of left and right combined.

11.3. RESULTS OF COMPARISONS OF PRESSURE DISTRIBUTION

Twenty four pressure data points were collected for the twenty-two subjects (10 normals and 12 disabled) in the nine predetermined seating postures (vide 9.1). Again, the emphasis in the data analysis is on detecting relative differences within and between the two samples rather than being overly concerned with the absolute values. Although some absolute values will be presented in graphical format, the conclusions will be based mainly on the significant differences or similarities within and between the study groups.

Data reduction has focused on three variables of pressure distribution; the average pressures, the maximum pressures and peak pressure gradients. Mean values were computed for both sample groups for the nine sitting postures. Standard deviations of the data for each group were also computed which gives a sense of the variability of the data points. Each variable is presented separately in the following sections with a summary of the composite results following.

11.3.1. Comparisons of Average Pressures

Average pressures across the OPM transducer matrix were computed by summing the recorded pressure values in each of the twelve cells and dividing the total by twelve. The mean values of all the subjects in each sample group were then averaged to give the weighted mean average for each of the body positions, P1M to P7 (vide 9.1). This analysis was done for both left and right sides separately and also in combination (average of left and right). The results are shown in figures 11.2 which give the mean average pressure values in mmHg for each body position, P1M to P7. It is helpful to visualize position P1M, the upright neutral posture, as the reference posture. All other eight positions are changes from the neutral posture, which were initially chosen as being representative of typical alternate postures assumed in a wheelchair.

Figure 11.2 indicates the average pressure distribution under the ischial tuberosity area both within and between normal and disabled groups. As expected the average pressure increases on the left side with similar reductions on the contralateral side when the trunk is flexed to the left (P1L) and the right (P1R)(upper two diagrams). Similar reductions can be noted in both groups with forward flexion of the trunk to 30° (P2) and 50° (P3). Similar trends in both groups can also be seen in the two positions of back rest recline, P4 and P5, and in full body tilt positions, P6 and P7. Although the differences in average pressures between the two groups are probably not statistically significant, the values do serve to confirm the expected results and the logic of the biomechanical assumptions.

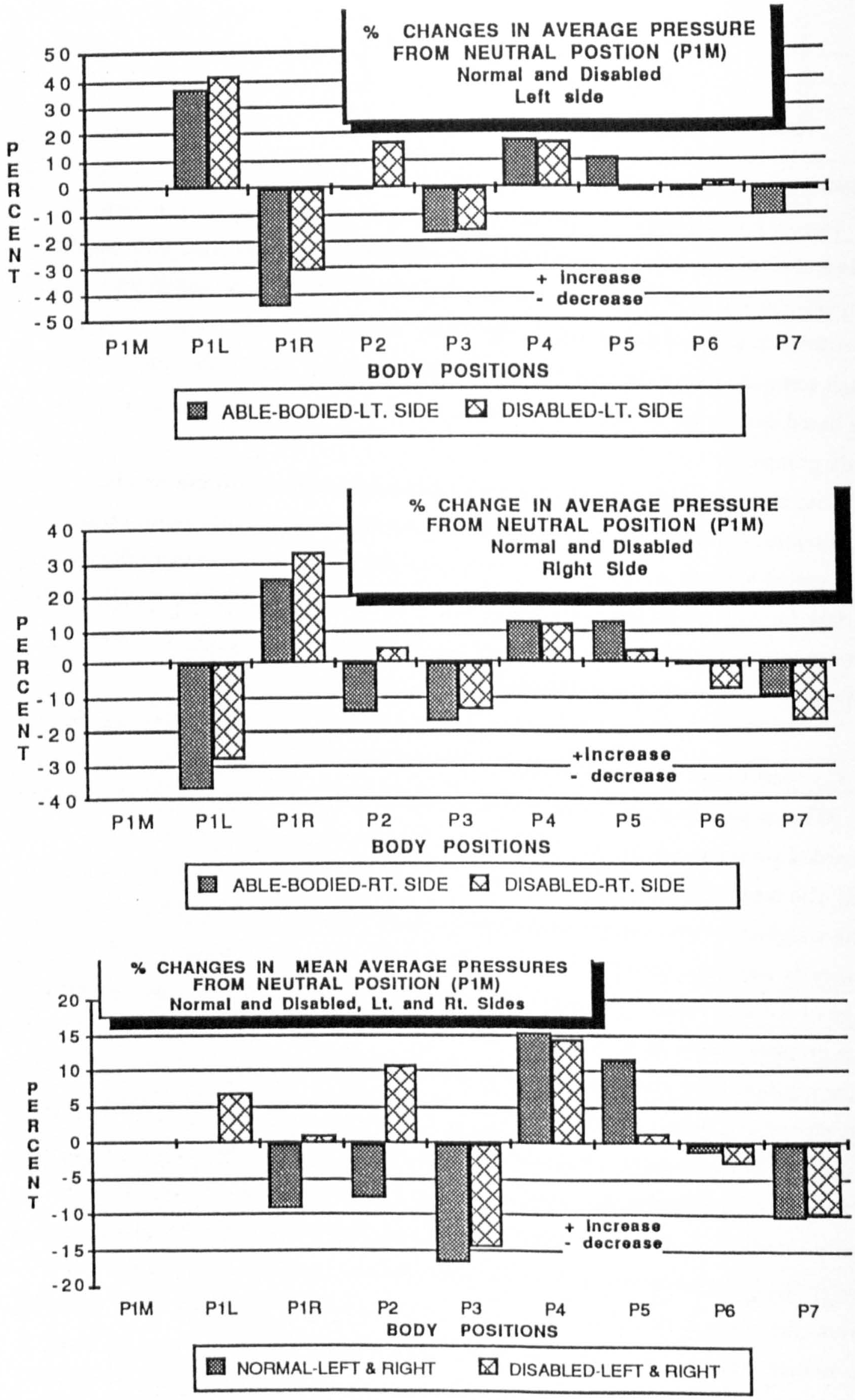


Fig 11.3 Percent change in average pressures from those in neutral position(P1M); left side, right side and mean of left and right combined.

Figure 11.3 emphasizes the intra and inter differences in the mean values of the groups. The differences are plotted as a percentage change from the neutral values (P1M). Of course, P1M has a 0 value on the chart. Similar contralateral trends are seen in P1L and P1R (upper plots) as a result of left and right lateral bending respectively. However, the results indicate that with 30° of forward flexion (P2) the average pressure apparently increases between 5 and 15% (L & R) for the disabled versus a decrease of between 1 and 15% for the normal group. Continued flexion to 50° (P3) results in a marked and almost equal decrease of approximately 15% for both groups. Reclining the backrest to 110° (P4) shows an increase in maximum values from the neutral position, which reduce somewhat when the backrest is further reclined to 120° (P5). Full body tilt of 10° (P5) begins to show pressure values approaching those of P1M with greater reductions taking place in the disabled group on the right side (8%). Clear indications of average pressure reductions from P1M values for both groups only begin to occur when the whole body is tilted in space between 10 and 20° (P7) or when the trunk is flexed forward to the P3 position.

The lower plot in figure 11.3 is an average of the percent changes which occur when left and right sides are combined and averaged. This provides an indication of the weighted mean average pressure changes that occur across the buttock in the critical area of the ischial tuberosities. As can be expected the plot does not show any appreciable differences from the upper two plots of each side.

11.3.2. Comparison of Maximum Pressures

Maximum pressure is defined as the maximum pressure values occurring in the vicinity of the left and right ischial tuberosities. Maximum values were determined for both the left and right sides independently. For purposes of analysis, the maximum value for either the left or right side was used. In the cases in which the maximum pressures occurred near the front edge of the back 3 x 4 transducer array additional readings were done using the forward set of transducers, thereby assuring that the maximum values did in fact occur within the back array. However, there was no way of confirming the exact location of the ischial tuberosities relative to the centre of the cell yielding the maximum value without moving the subject. It therefore can be assumed that some maximum values may be higher than those recorded, particularly in those cases in which the maximum values were not centred over the recording cell.

Examination of the results in figures 11.4 and 11.5 indicate marked differences between the normal and disabled samples. The upper two plots are for the left and right sides, with the lower plot being an average of the left and right sides. The mean

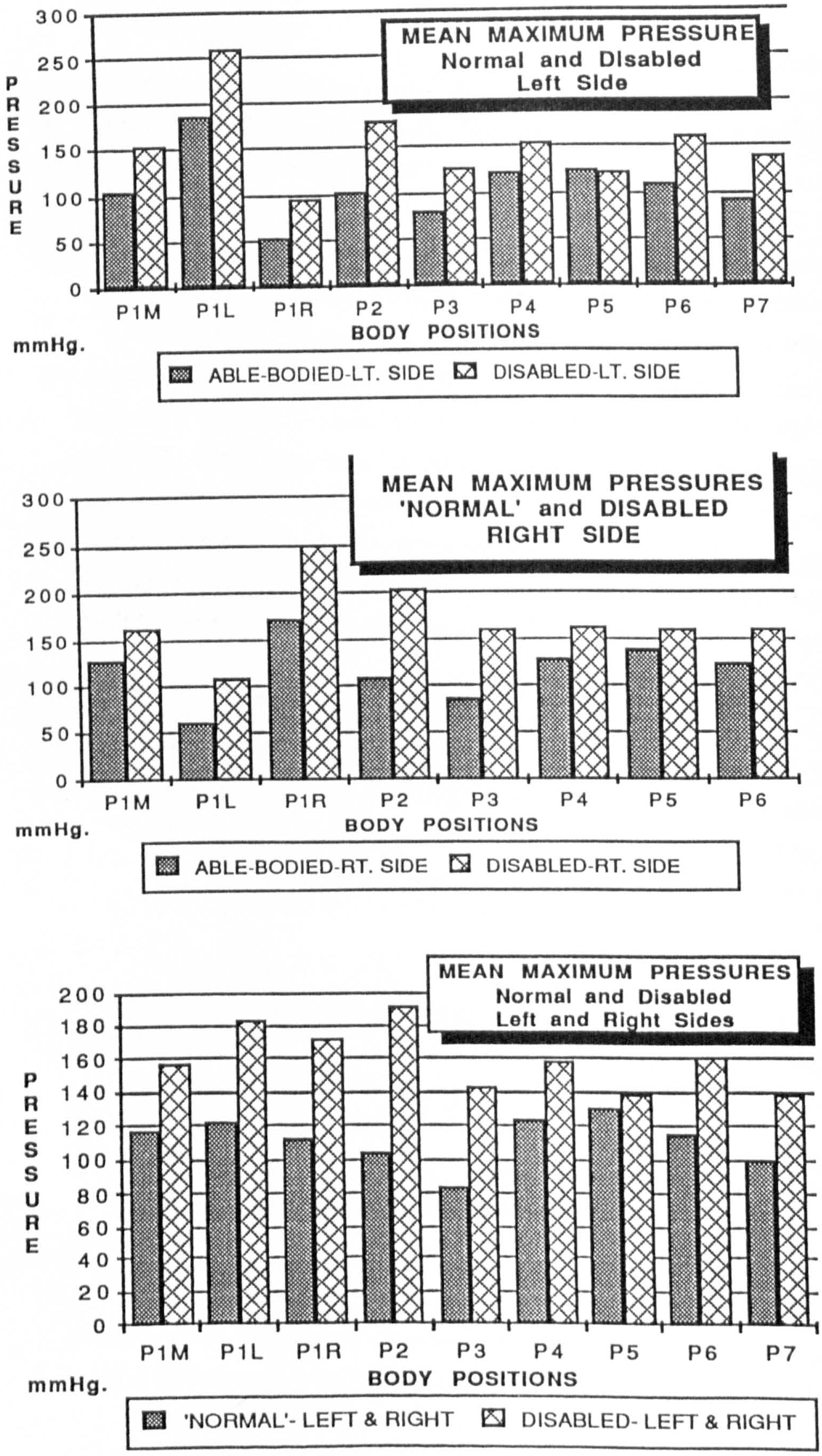


Fig 11.4 Plot of the mean maximum pressures on the left side, right side; and combined left and right sides for normal and disabled subjects

maximum pressures for the disabled are higher than those for the normal subjects in all nine body postures. This is particularly evident in the lateral bending (P1L and P1R) and in the 30° of forward trunk flexion (P2). Similar distinct differences can also be seen in the two positions of total body recline (P6 and P7).

Of special interest is the manner in which the differences in maximum pressures are reflected in postural changes from the neutral position (P1M). Figure 11.5 summarizes these differences in terms of percentage changes from the values recorded in the P1M posture. The lower plot shows the average of left and right percentage changes from the average(L&R) values recorded in the neutral position(P1M). Bending to the left (P1L) shows a maximum value increase for both groups with a larger per cent increase for the disabled group of approximately 6%. Bending to the right shows an average decrease for the normal group of about 7.5% with increase of the disabled group of about the same amount. Similar trends are observed in the P2 position with a rather large per cent increase in the disabled group (greater than 20%). Forward trunk flexion to 50° (P3) shows a percentage reduction in maximum values for both groups, with normals about 3 times those of the disabled (9% versus 27%). The reclined backrest positions (P4 and P5) indicate almost equally opposite trends in the P5 position. The disabled maximum values increase over those recorded in the neutral position by about 12% and the disabled reduce about the same amount. Again, percentage maximum pressure reductions from those in the P1M position only begin to exceed 10% when the total body is tilted in space by about 20° (P7). In this respect similar trends are seen for both groups.

The location of the maximum pressure relative to a known reference is of interest. Figure 11.6 is a plot of the A/P location of the X value of the maximum pressure. It can be noted that the location of the maximum pressure falls slightly further forward on the seat in all positions for the disabled group when compared to the normal group. Positions P4 and P5 appear to have the largest difference between the groups. And finally, it can be seen that the location of maximum pressure is apparently influenced by changes in sitting posture.

11.3.3. Comparisons of Peak Pressure Gradients

Peak gradients are defined as the rate of change of pressure between the cell recording the maximum pressure and any adjacent cell measuring the lowest value. Distance adjustments were made for diagonally-located cells. As with average and maximum pressures, mean values and standard deviations were computed for left and right sides for both groups.

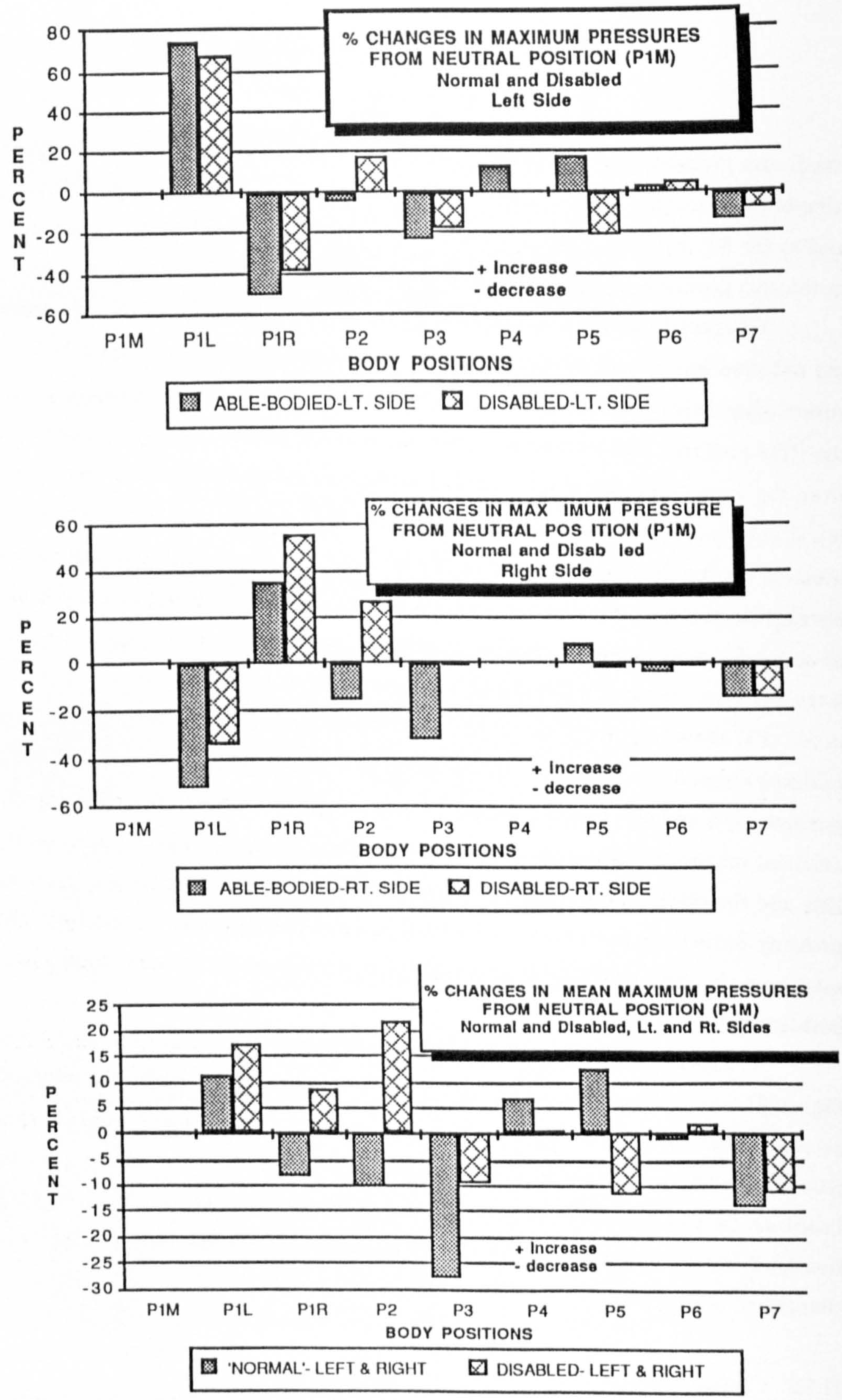


Fig 11.5 Plot of the percent changes of the mean maximum buttock pressures from those in the neutral posture P1M; left side, right side and combined left and right sides.

Figure 11.7 (upper plot) shows the peak gradient plots of the weighted mean average (left and right combined) for all nine positions. The most striking results are that the peak gradients for the disabled group are on average at least twice those of the normal group in all sitting postures. Of particular interest are the large apparent differences that occur in left and right bending (P1L and P1R) and in forward flexion (P2 and P3).

Again, it is of clinical importance to determine how changes in body postures may affect the change in peak gradients from those experienced in the neutral positions (P1M). Figure 11.7 (lower plot) indicates the percent change in peak gradients from those measured in the P1M posture for the other eight postures. The results show a rather marked percentage increase in the disabled group in the two lateral bending positions, P1L and P1R. Increases occur in the disabled group, but decreases occur in the normal group when forward trunk flexion occurs (P2 and P3). Rather minor changes occur in the remaining postures, except in the position P5 which shows a apparent decrease in both groups of approximately 15%.

11.4. SUMMARY OF RESULTS FROM MEAN VALUE PRESSURE COMPARISONS

At this point it is useful to extract the salient results from each pressure analysis in order to construct an integrated picture of the pressure distribution characteristics at the body-seat interface. The following itemized summary statements should still be considered tentative, since definitive conclusions can not drawn until significance testing has been completed.

1) The average pressure distributions, although changing with alteration of body posture do not show marked differences between the study groups. The lowest average values occur during lateral bending (P1L and P1R) and full body tilt (P7). Comparing left and right values indicates a higher average pressure value in all positions on the right side of approximately 5 to 10%.

2) Changes in body posture from the neutral position appear to influence the average pressure distribution across the ischial/buttock area. Of the eight alternate postures studied, position P3, 50° of forward flexion, causes the largest percentage reduction in average pressure (15%); whereas 20° of backrest recline (P4) causes the greatest increase (15%). Both normal and disabled groups exhibit similar responses in positions P3 and P4. Positions P1L, P1R and P2 exhibit different trends between the disabled and normal groups. In general, a change of body posture from the P1M position causes an increase in average pressure for the disabled with no change or reduction in

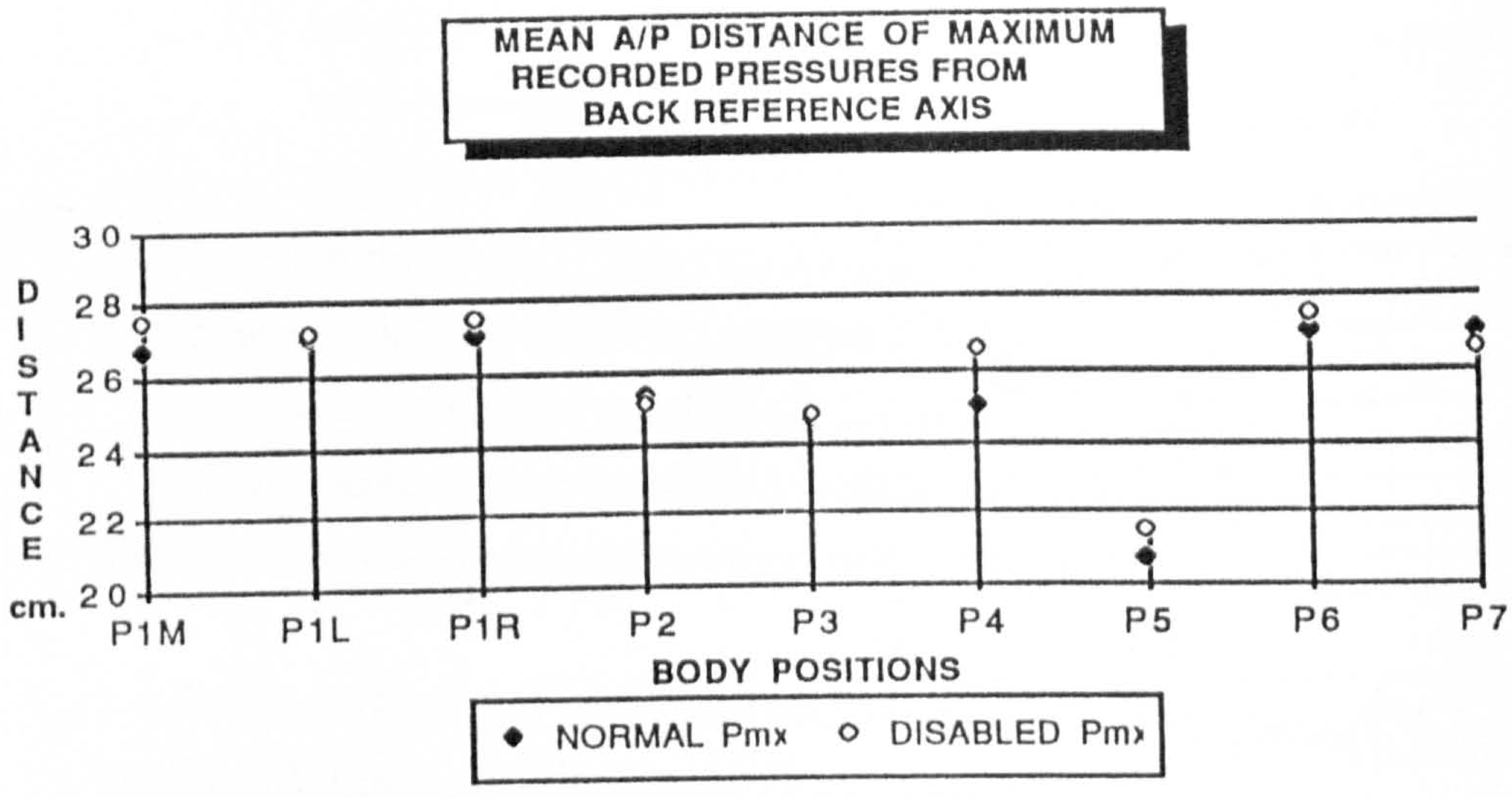
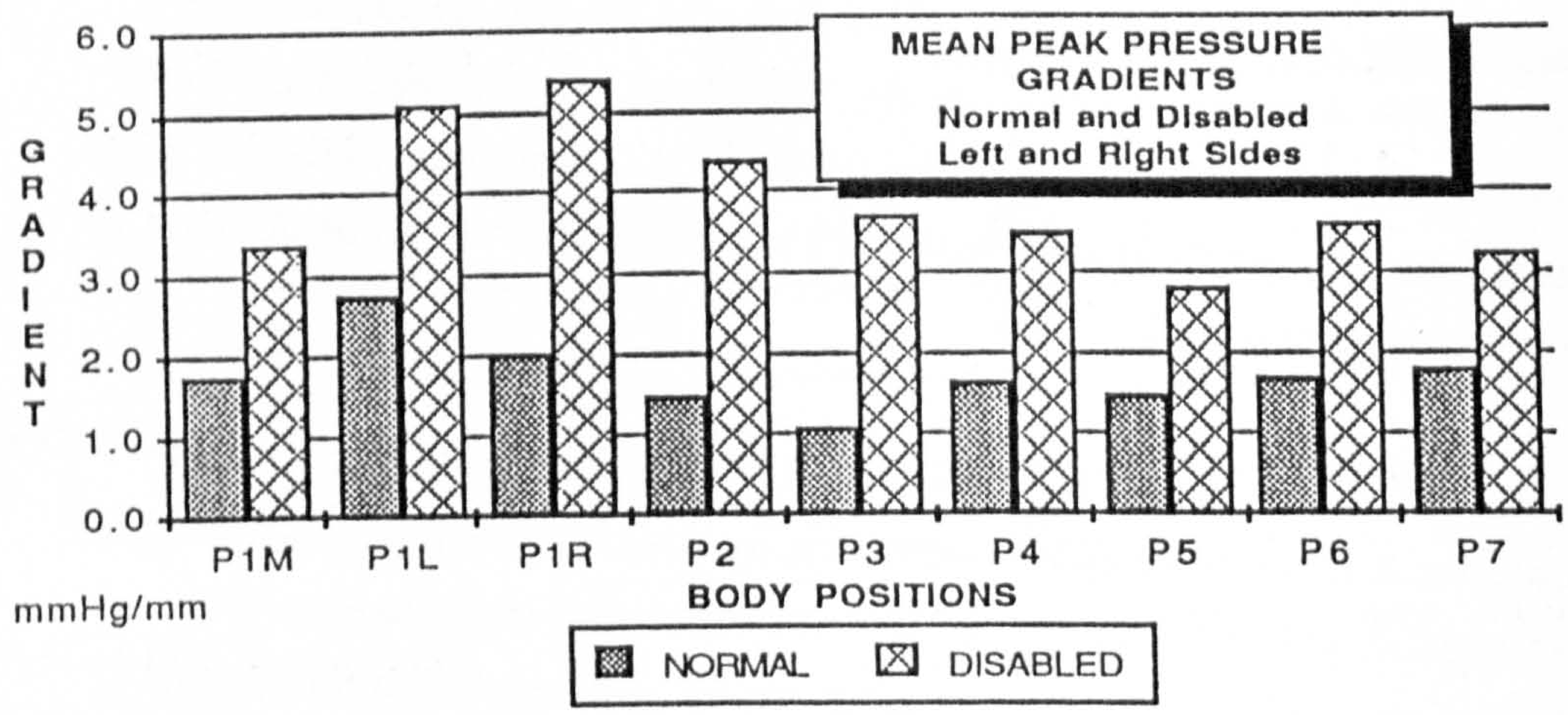


Fig 11.6 Plot of the 'x' location of the maximum pressure (Pmx) from the y reference axis.



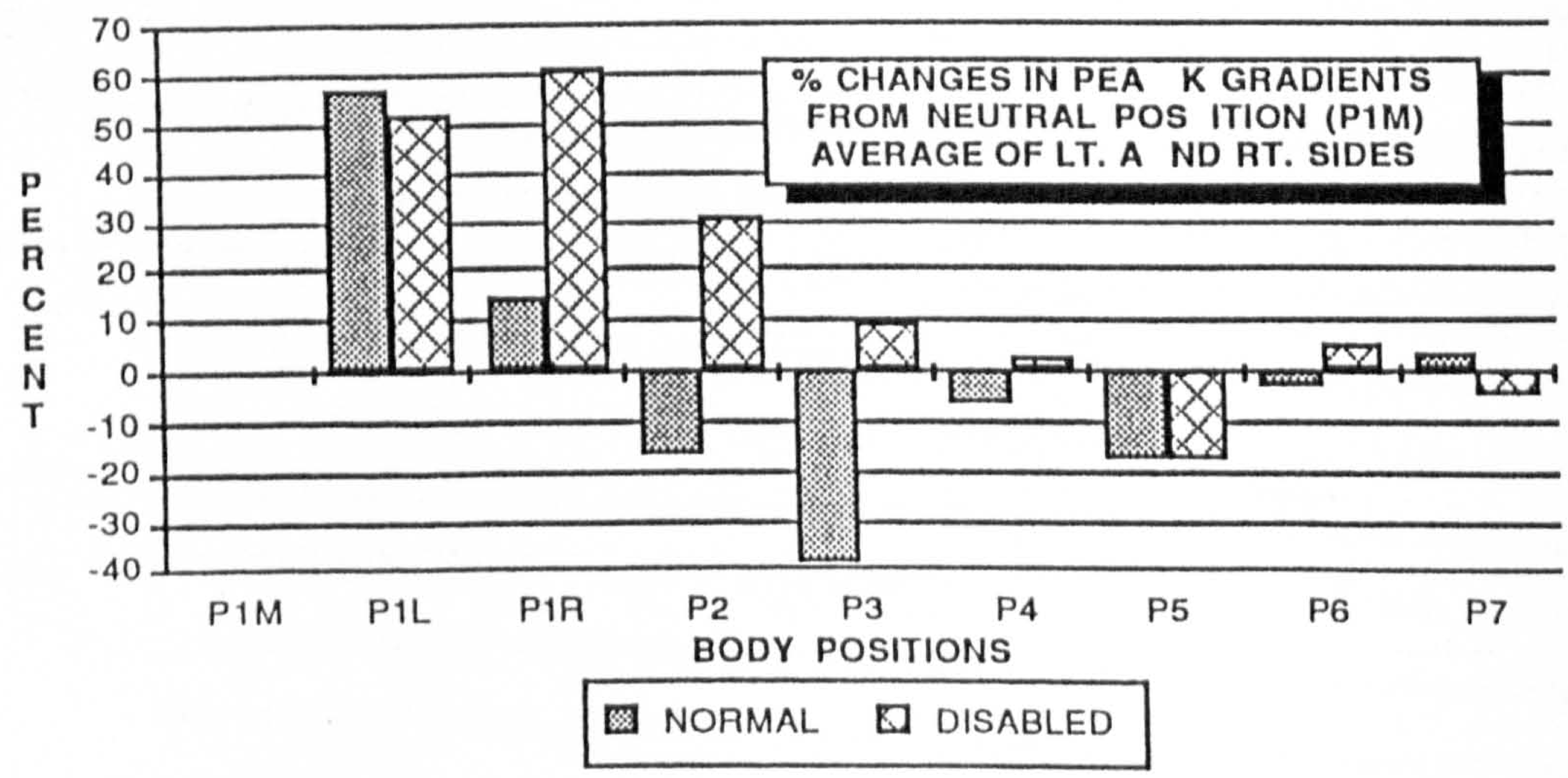


Fig 11.7 Plot of maximum pressure gradients ocurring on either the left or right side. Lower chart gives the percent changes in mean maximum gradients from the P1M values in the middle chart, for both normal and disabled subjects.

pressure for the normal group. After forward flexion (P3), the full body tilt to 30° (P7) offers the next largest decrease of average pressure from the neutral position (10%).

3) Mean maximum pressures for the disabled group are higher in every position measured; with the largest differences occurring in the lateral bending and forward flexion positions.

4) Changes in body position from the neutral position appear to influence the maximum pressure values. Percentage changes from the values experienced in the neutral posture vary widely between study groups and between body postures. As expected, similar trends in changes in maximum and average pressures are evident, although some distinct differences are apparent. A salient point is that the mean maximum pressure increases over 20% in the disabled group upon forward trunk flexion to 30° (P2), whereas it decreases 10° in the normal group. Forward flexion to 50° (P4) appears to cause a drastic decrease from the P1M values of greater than 25% in the normal group, whereas the disabled group only decreases by less than 10%. Also, similar trends of 10% reduction in maximum and average pressures are seen for both groups when the whole body is tilted 20° in space.

5) Mean peak pressure gradients for the disabled group appear markedly higher than the normal group in all body postures studied. The maximum differences occur in the lateral bending and forward flexion positions.

6) The percentage changes from the peak pressure gradients experienced in the neutral position show large variations between the study groups and between body postures (P1L to P7). The largest percentage increase occurs in the disabled group during lateral right bending (P1R), i.e. a 60% increase. Marked differences occur between the two groups in the P1R position, a 60% versus a 15% increase for the normal group. Forward trunk flexion (P2, P3) causes apparent percentage increases in peak gradients for the disabled, whereas decreases occur in the normal group; with a maximum decrease of about 40% occurring in the P3 position.

As indicated at the outset two additional variables can also have influence on the interface characteristics, namely, tangentially-induced shear forces and centre of gravity location.

CHAPTER 12. COMPARISON OF THE TANGENTIALLY-INDUCED SHEAR EFFECT AT THE BODY-SEAT INTERFACE

12.1. THE OBJECTIVE

The objective of this component of the study is to investigate the manner in which the tangentially-induced shear force changes with body posture, both within and between study groups.

A number of investigators have reported that shear stresses and the resulting tissue deformation is a major aetiological factor in ulcer formation (Dinsdale, 1973; Neumark, 1981; Bennett, 1979, 1984a; Patterson, 1984; Chow, 1974; Guttmann, 1976; Roaf, 1976). Chow and Odell make the point that shear stress is involved in uniaxial pressure, localized pressure, and any non-uniform pressure distribution or any pressure that causes tissue distortion. This makes the shear stress difficult to study because its a result of differential radial expansion between the buttocks and cushions or cushion covers (Chow and Odell 1978).

There are also shear stresses that are introduced to the interface surface as a result of forces acting tangential to the support surface, i.e. the friction forces. Riechel (1958) reported on the increase of tangential shear forces on the sacrum that result from raising the head of a hospital bed. If one differentiates between the types of shear stresses, i.e., pressure or normally induced and tangentially - induced, it becomes feasible to conceive of methods to measure the simpler tangentially-induced shear forces. That is, if the mechanical properties of the support surfaces remain the same and the body positions are reproducible, determination of the relative tangential shear force will give at least partial insight into this complex interface phenomenon. However, the manner in which the two shear components interact within the tissues will remain unknown.

12.2. METHODS AND MATERIALS

12.2.1. The General Method

The forces acting tangential to the seat surface were measured in both the fore and aft directions. The BPC seat surface (foam plus plywood substrate) is mounted on an aluminum substructure that moves freely in the sagittal plane. The free movement is limited by the load cells mounted at each end of the substructure (design details in Appendix 1). The cells measure the forces parallel to the seat surface that tend to move the seat in the fore and aft directions relative to the wheelchair frame. These forces are defined as the tangentially-induced shear (TIS) or friction force acting at the seat-body interface. The TIS was measured in all nine seating postures for all subjects. Mean

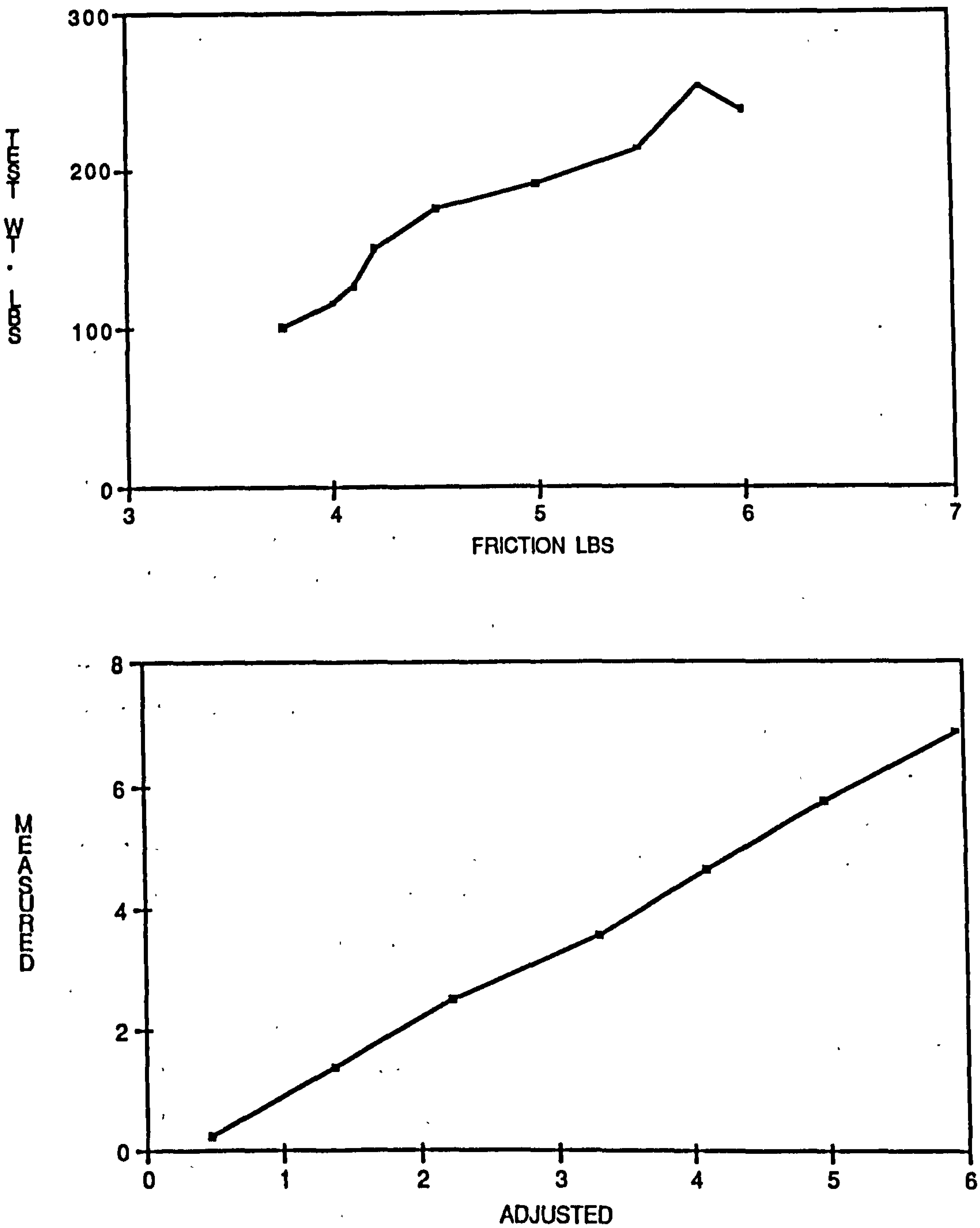


Fig 12.1 Plots showing variability of friction in linear bearings with increasing load(upper chart). Lower chart indicates adjustments made to measured friction values based on simple regression fitting. The adjusted values were used to reduce the measured TIS force to account for the bearing friction.

values and standard deviations for both groups were computed and charted in a manner similar to the pressure and radiographic studies.

12.2.2. Description of Shear Measurement Instrumentation

The load cells used were Sensortronics(c), model 60040, which are a low profile bi-directional version that is produced in a variety of load capacities. The capacity chosen was 0 to 22.6 kg. (0 - 50 lbs), which has a overload capacity of 200%. The transducer itself is of the loaded beam design with attached strain gauges in a 4-arm (Wheatstone Bridge) configuration. The cell has a full scale output of 3 mv/v and a non-linearity of 0.05% FS and hysteresis of 0.03% FS. The cells have a compensated temperature range of 0-150°F (0-65.5°).

The data acquisition system used was a Metrabyte(d) DAS-16 multi-functional high speed analogue-digital (A/D) I/O expansion board for the IBM PC computer. The board has a 12 byte successive approximation converter with a 12 μs conversion time. The channel input configuration is switch selectable on the board, providing a choice between 16 single ended channels or 8 differential channels with a 10 volt range. The board was configured to sample both the four centre of gravity load cells and the two shear load cells. The voltage range selected was 0 - 5 volts. A machine language driver for A/D functions provided by Metrabyte was called from a compiled Basic program developed using Microsoft Quick Basic 3.0. The conversion rate was approximately 615 samples per second. The conversion rate varied slightly because of the nature of the conversion process. Because of problems with noise appearing in the signal despite efforts to filter it, 200 sets of samples were taken at once and averaged to give readings for each cell. The machine language only cycled through the six channels once per machine language call. The Basic program, therefore, had to spend time outside the machine language routinely storing the samples in an array variable for further processing. The conversion time for 200 sets of data, including the machine language calls and the processing in Basic, range from 1.93 s to 1.97 s, or through-put of 622 to 690 samples per second, respectively.

The front end signal amplification and conditioning was done using a commercial unit from Action Instruments(e), Model AP4051. This unit is a plug-in module which contains a power supply, with adjustable excitation voltage. The power supply is regulated and electronically isolated from the signal conditioner circuit. The amplifier/signal conditioner has two stages: a fixed gain graded amplifier and noise filtering, and an output buffer stage with zero and span adjustments for in-field calibration. The internal power supplies for the bridge and signal conditioner circuits

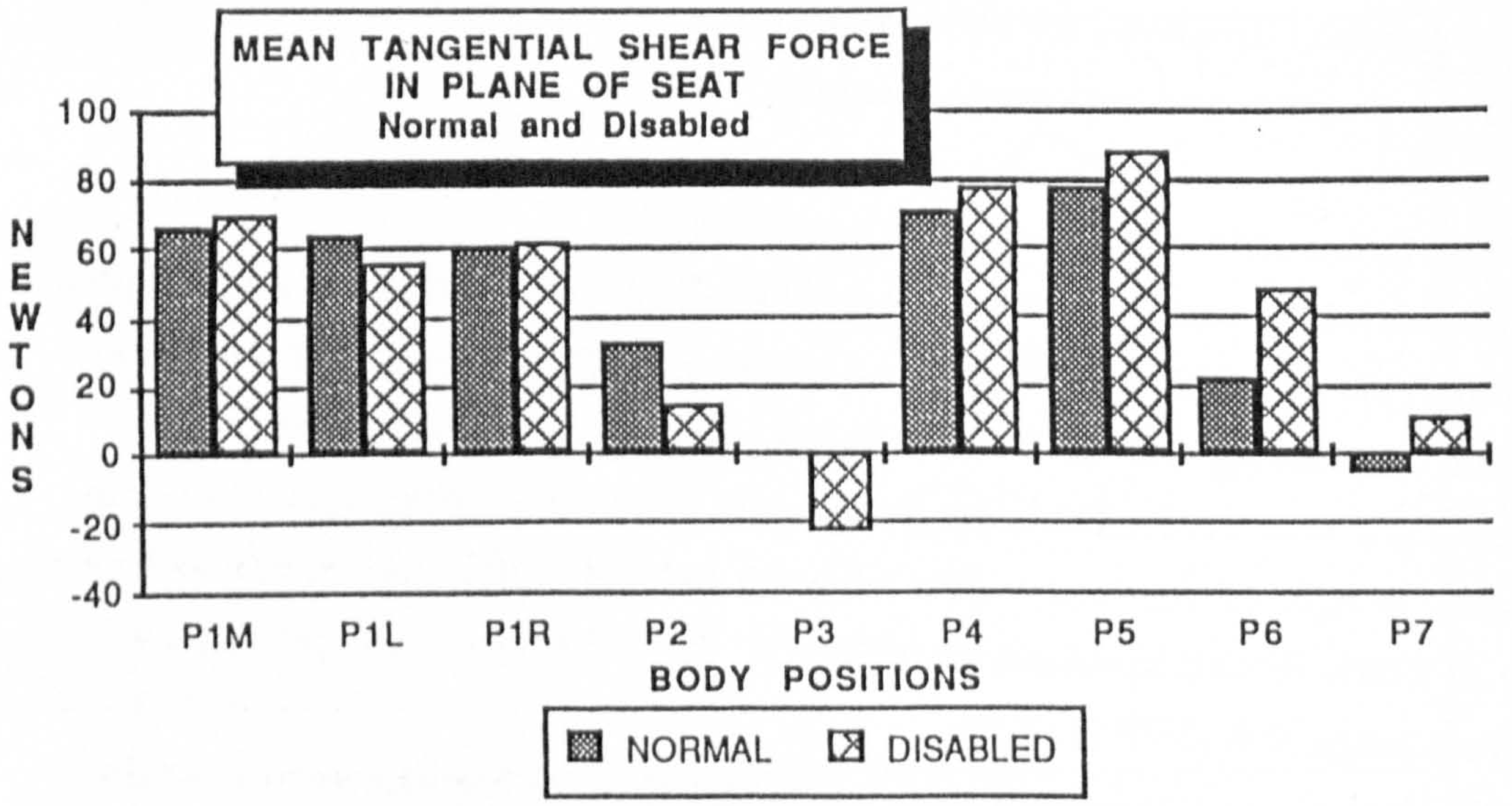


Fig 12.2 Mean tangentially - induced shear forces as measured by the A/P load cells in the nine body postures for both normal and disabled subjects.

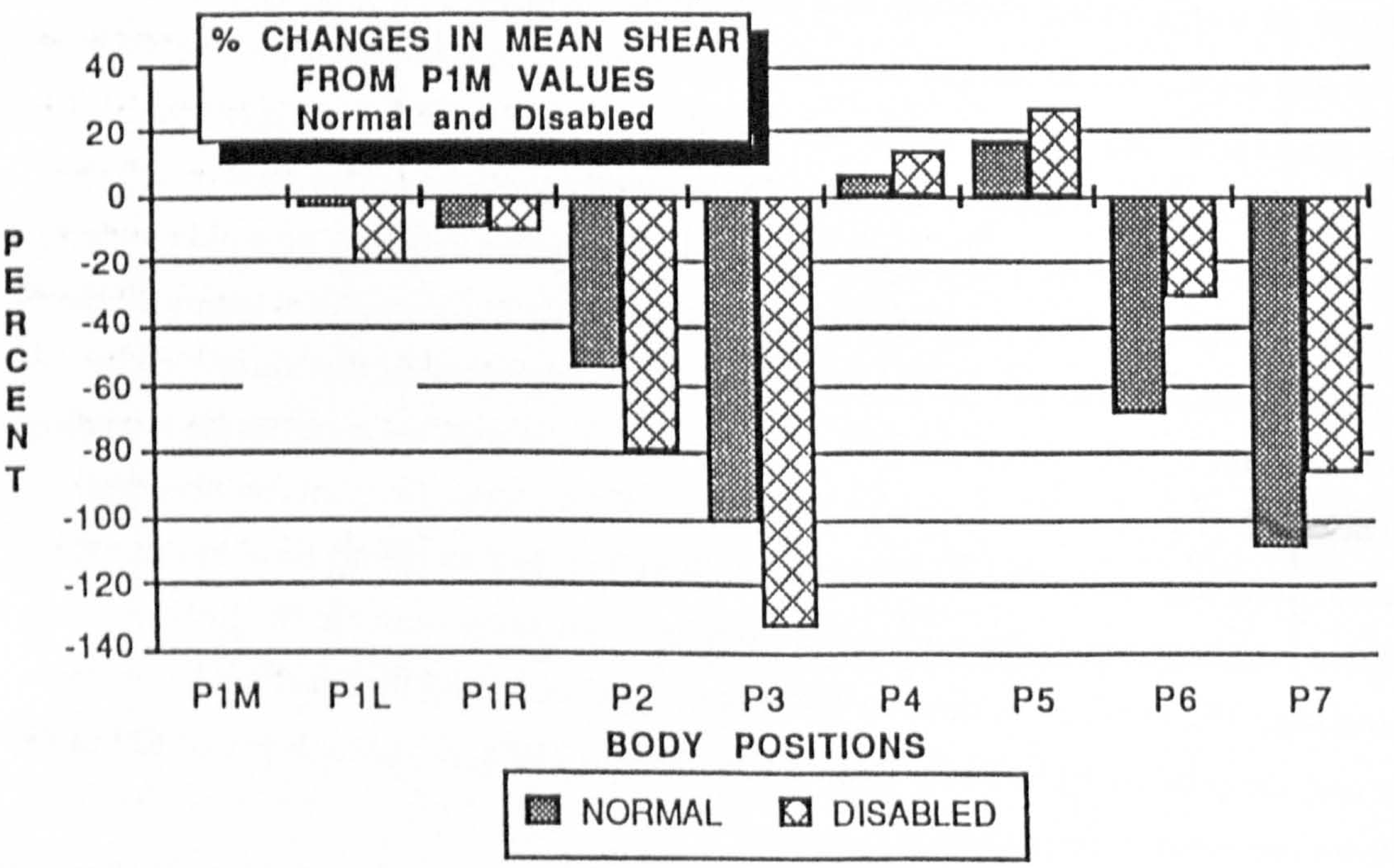


Fig 12.3 Percent changes in the mean values of the tangentially-induced shear (TIS) from values measured in the neutral position, P1M.

are isolated from each other and from line power and ground (earth). The Basic software program combined with the Metrabyte A/D board permitted calibration and adjustment of the baseline settings for the load cells prior to each run. Calibration of the load cells was done using a spring scale. A plot of the force versus voltage output indicates a linearity of approximately 2% within the operating range of the scale (0 - 25 lbs., 0 - 11.3kg). A previous accuracy check of the same load cells indicated a 2% linearity over a range of 0-22.6kg (50lbs).

Tests were done to determine the the friction force in the linear bearings, which in turn was used to adjust the values measured by the loads cells accordingly. Regression computations were necessary because the friction values varied slightly with body mass (fig 12.1). Further compensations were not made for reductions in transmitted mass due to body tilting (10&20°) in space since the error was considered negligible.

12.3. RESULTS FROM SHEAR FORCE MEASUREMENTS

The composite results of the tangentially-induced shear (TIS) forces are charted in figure 12.2. The TIS values for body positions P1M, P1L, and P1R are about the same. This is to be expected, since the TIS values are only measured in the fore and aft (A/P) directions. Therefore, lateral trunk bending, left or right, should have little influence on the A/P shear values. However, trunk flexion to the 30° (P2) and 50° (P3) positions show marked reductions in the A/P shear values. Again, this is to be expected since the subject moves away from the upper backrest which is the primary reaction point causing the TIS forces. It is speculated that only 30° of forward flexion still keeps the sacral area in contact with the backrest due to the posterior shift of the ischial tuberosities. Further flexion to 50° (P3) reduces the TIS to near zero, suggesting that the sacral area has moved away from the backrest.

As expected, reclining the backrest to the P4 and P5 positions increased the TIS values to their maximum values at the P5 position. The TIS forces are reduced upon full body tilting (P6, P7), approaching zero at the 20° tilt position (P7). It can be seen that the TIS forces have reversed direction for the normal subjects between 10° and 20° body tilt, but they are still acting in the posterior direction for the disabled group.

Again, it is of clinical importance to analyze how the TIS forces can be minimized from those values experienced in the neutral seating posture (P1M). Figure 12.3 indicates the percentage change of the TIS values from those measured in the P1M posture. Forward flexion to P2 and P3 positions causes marked reductions in the TIS force, which reaches more than a 100% reduction in the P3 position. Reclining the backrest causes increases in TIS values to a maximum of 28% in the P5 position.

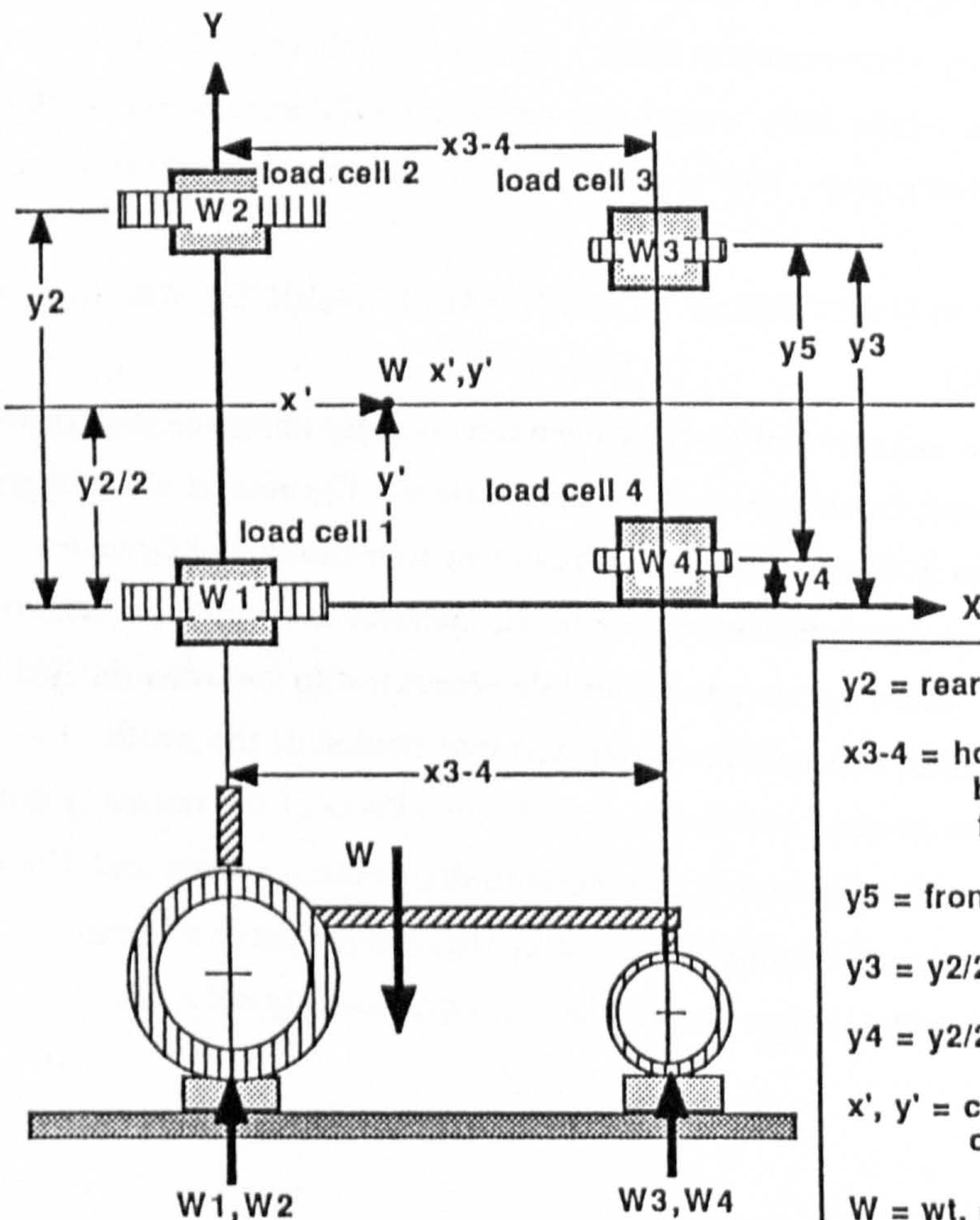
Increases appear notably larger for the disabled versus the normal group (28% versus 17% in P5). Full body reclining (P6/P7) causes marked decreases in TIS force. The disabled group show an apparent reduction of 85% in the 20° tilt posture (P7); with the normal group reducing by approximately 108%, a reversal of TIS force. It can be assumed that continued tilt of the whole body would also cause the TIS forces to eventually reduce to 0 in the disabled group. This is estimated to occur at a tilt angle of about 25°.

12.4. DISCUSSION OF RESULTS FROM TANGENTIALLY-INDUCED SHEAR FORCE COMPARISONS

The above results suggest that body posture can strongly influence the magnitude of the tangentially-induced forces acting at the seat surface. The results also suggest that there are differences in the magnitudes and percent reductions of TIS values between the two study groups, particularly when the backrest is reclined or the whole body is tilted in space. There appears to be an inter-relationship between the TIS force and the movement of ischial tuberosities during forward flexion of the trunk.

The above results do not provide insight into the effects of the normally-induced shear. However, if it can be shown that the pressure distribution values and TIS values are reduced simultaneously, an assumption could be that the net shear stresses experienced by the supporting tissues should also be comparatively reduced.

A) CENTRE OF GRAVITY COMPUTATIONS:--BODY POSITIONING CHAIR



y2 = rear wheel base (cm)

x3-4 = horizontal distance between back and front wheels (cm)

y5 = front wheel base (cm)

y3 = y2/2 + y5/2 (cm)

y4 = y2/2 - y5/2 (cm)

x', y' = cordinates of CG of wheelchair (cm)

W = wt. of wheelchair (kg) = (W1+W2+W3+W4)

1) Taking moments about x axis

$$Wy' = y2(W2) + y3(W3) + y4(W4)$$

$$y' = (y2(W2) + (y2/2 + y5/2)W3 + (y2/2 - y5/2)W4) / W \qquad \text{--- (1)}$$

2) Taking moments about y axis

$$Wx' = x3\text{-}4(W3+W4)$$

$$x' = (x3\text{-}4(W3 + W4)) / W \qquad \text{-- (2)}$$

Note: Any 4-wheeled device can be used by simply measuring dimensions y2, y5, x3-4 and inserting the values in equations (1) and (2).

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Fig 13.1 Computations for determining the centre of gravity of the body positioning chair (BPC).

CHAPTER 13. COMPARISONS OF THE CENTRE OF GRAVITY LOCATIONS AND DISPLACEMENTS

13.1. THE OBJECTIVE

The objective of this final component of the study is to compare the location of the centre of gravity (CG) and observe differences in displacement of the CG within and between the two study groups.

Bardsley (1977) compared the motility and mobility of the CG of both normal subjects (adults and children) and disabled children. Pen tracings of the CG displacements were compared. Assumptions were made that displacement of the CG were directly related to pressure re-distribution. However, no pressure measurements were taken so that direct comparisons between the CG displacement, pelvic alignment (deformity) and maximum pressure locations were not possible. This aspect of the study attempts to add the variable of CG location and displacement to the previously measured variables of pelvic alignment, pressure distribution and tangentially-induced shear force.

13.2. METHODS AND MATERIALS

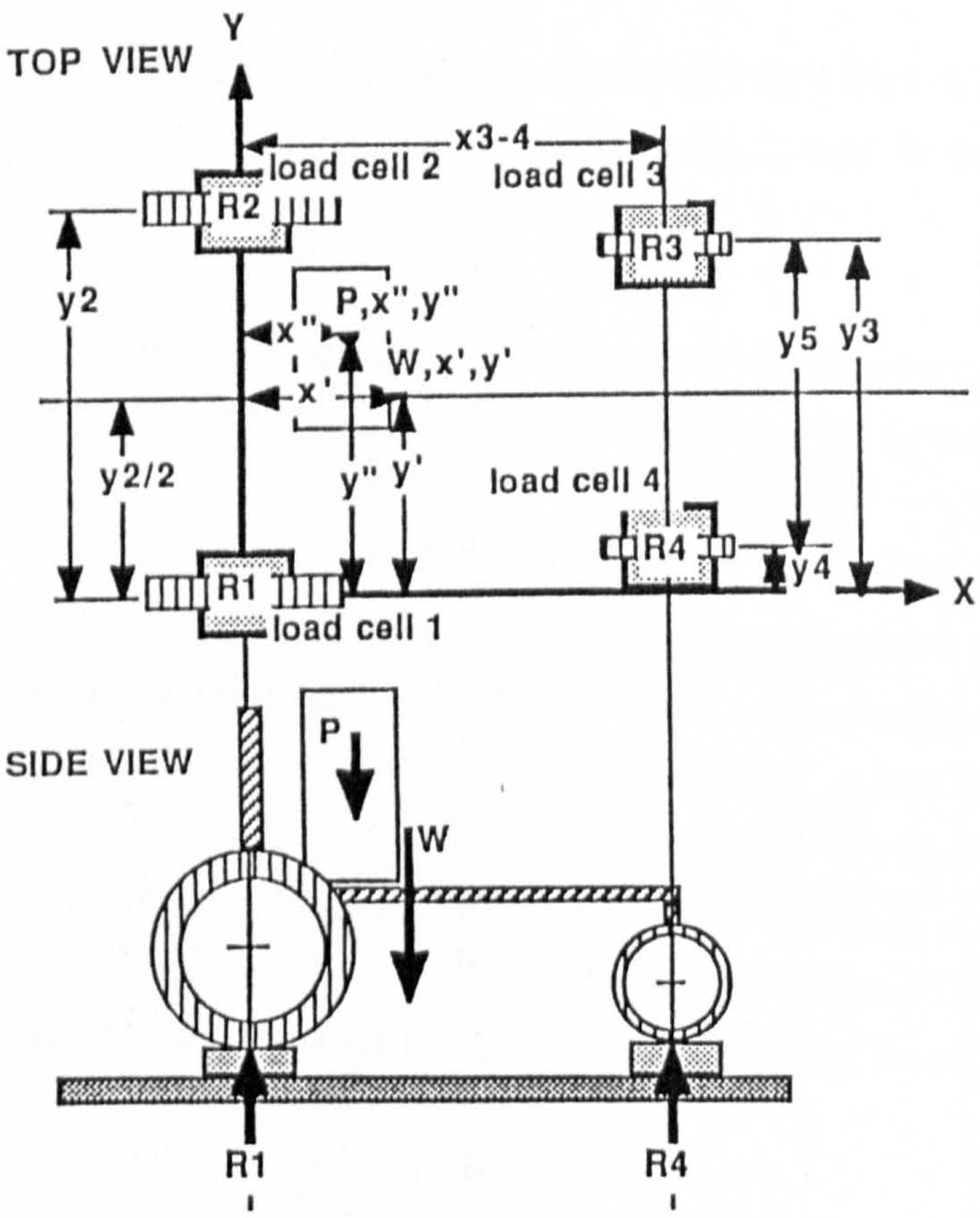
13.2.1. The General Approach

The approach taken has been to measure the location of the CG of all subjects in both groups, while seated in the body positioning chair (BPC). Again, the identical nine seated postures (P1M to P7) were used, so that direct comparisons to the pressure distribution, shear and radiographic results could be made. Load measurements were made by four load cells located under the wheels of the BPC. Computer computations permitted real-time calculations of the X" and Y" components of the CG location of each test subject in each of the nine sitting postures. Again, coalescing of the data involved calculations of mean values and standard deviations of the CG for each group. Results are based on comparisons of mean values of the two groups in the nine postures.

13.2.2. Method for Quantification of CG Location

Four load cells were designed and constructed to measure the vertical loads of the BPC and its occupant. Each cell was designed to withstand a maximum value of 226 kg (500lbs). The voltage outputs from the load cell strain gauges were amplified and fed into the same Metrabyte Data Acquisition Board (vide 12.2.2) that was used for sampling the shear load cells. Both the shear and CG values were recorded in the same

B) CG COMPUTATIONS: BPC AND OCCUPANT



y2 = rear wheel base (cm)

x3-4 = horizontal distance between back and fronwheels (cm)

y5 = front wheel base

y3 = y2/2 + y5/2 (cm)

y4 = y2/2 - y5/2 (cm)

x', y' = cordinates of CG of wheelchair (cm)

x", y" = cordinates of CG of occupant (cm)

W= wt. of wheelchair (kg)

P= wt. of wheelchair (kg)

R = W+P

1) Occupied Wheelchair

$$Mx = R2y2 + R3y3 + R4y4 - Py'' - Wy' = 0 \qquad (3)$$

$$My = R3x3 + R4x4 - Px'' - Wx' = 0 \qquad (4)$$

2) Unoccupied Wheelchair

$$Mx = W2y2 + W3y3 + W4y4 - Wy' = 0 \qquad (5)$$

$$My = W3x3 + W4x4 - Wx' = 0 \qquad (6)$$

3) Equate (3) and (5)

$$Py'' = W2y2 + P2y2 + W3y3 + P3y3 + W4y4 + P4y4 - W2y2 - W3y3 - W4y4$$

$$Py'' = P2y2 + P3y3 + P4y4 \text{ (Substituting as in (1))}$$

$$Py'' = P2y2 + P3(y2/2 + y5/2) + P4(y2/2 - y5/2)$$

$$y'' = (P2y2 + P3(y2/2 + y5/2) + P4(y2/2 - y5/2)) / P \qquad (7)$$

4) Equate (4) and (6)

$$R3x3 + R4x4 - Px'' = W3x3 + W4x4$$

$$Px'' = (W3 + P3)x3 + (W4 + P4)x4 - W3x3 - W4x4$$

$$Px'' = P3x3 + P4x4$$

$$x'' = (X3\text{-}4(P3 + P4)) / P \qquad (8)$$

Fig 13.2 Computations for the centre of gravity (CG) of the body positioning chair and occupant, which permits calculation of the occupant's CG independent of the BPC.

sampling series. Calculations software combined with the data acquisition software computed the CG location of the seated occupant independent of the wheelchair (BPC).

The CG measurements were incorporated into the same trials as the pressure distribution measurements. That is, while pressure and shear forces were being measured, in each of the nine body postures, CG measurements and computations were being done. Three data samples were taken for each body posture for each subject; one at the beginning of the pressure measurement, one during the pressure measurement, and one at the end. Each recorded CG measurement is an average result of 200 scans of the load cells. Very small variances (<1%) occurred between the three measurements per body position.

The sequence of CG measurements was to first establish the baseline value (null value) for each load cell. Then the unoccupied BPC was placed on the load cells so that the weight of the BPC could be determined in terms of the loads on each cell. The subject was then placed in the BPC and the combined weight of the subject and the BPC were determined. This sequence permitted the determination of the weight of the BPC, the mass of the occupant and the computation of the CG of the occupant independent of the weight of the BPC. Once the baseline data had been entered into the computer program rapid computations of CG locations were possible for each of the nine body positions studied.

Data analysis involved combining the X and Y values of the three data samples for each subject in both groups, so that mean values and standard deviations could be computed in each of the nine postures.

13.2.3. Method for Determining the CG of the Seated Occupant

Figures 13.1 and 13.2 contain the mathematical computations for determining the CG of the subject based on measurements from the four load cells. The procedure involves moment calculations of the load cell values about the X and Y axes. Computations are first done for only the BPC (fig 13.1) and then for the BPC plus the occupant (fig 13.2). The computation cancels out the weight for the PBC, thereby permitting the X and Y components of the CG to be expressed in terms of the weight of the occupant only (X", Y"). Formulae (7) and (8) in figure 13.2 are used in the software program which permits the rapid calculation and recording of the CG coordinates for each subject in each of the nine postures.

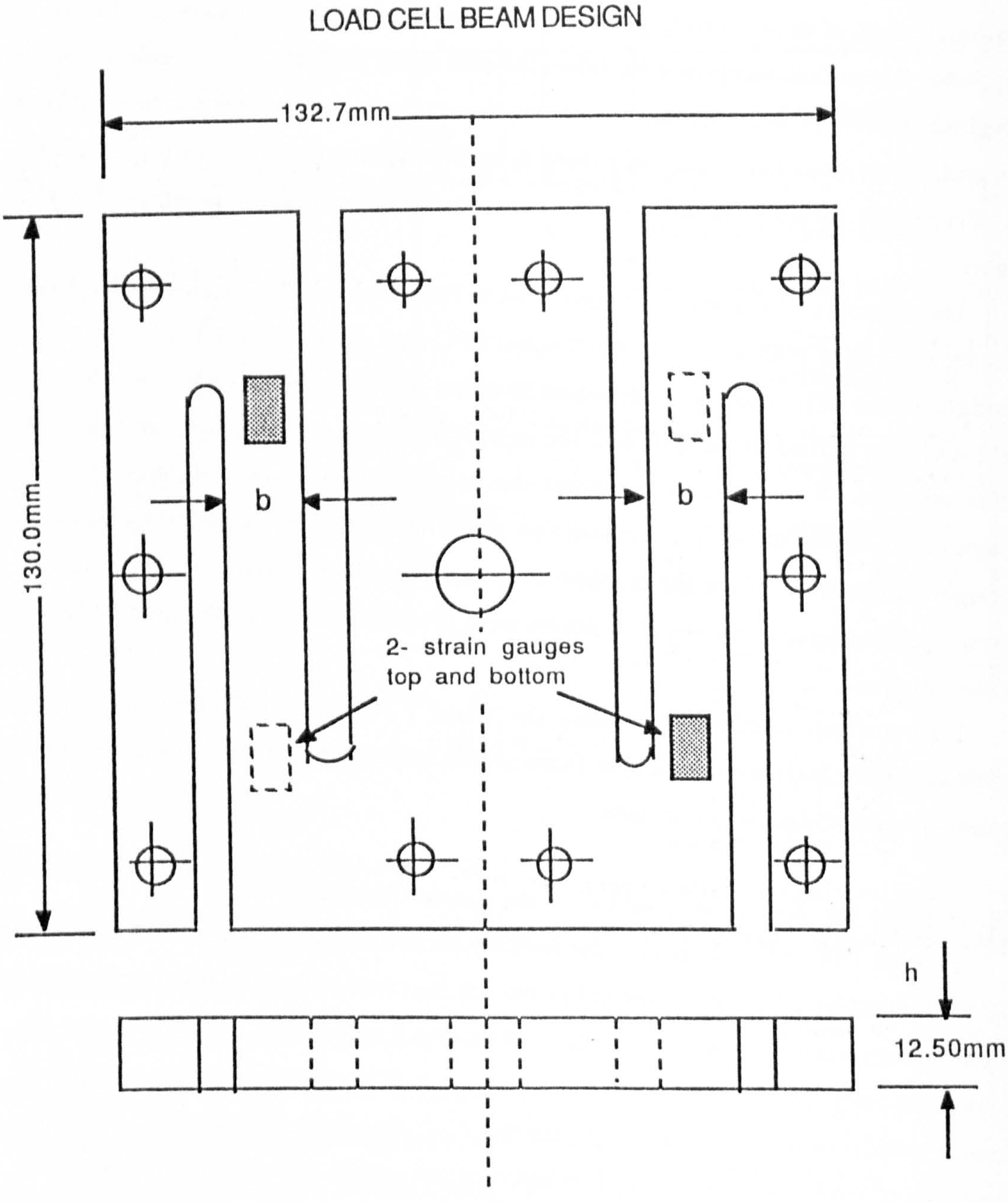


Fig 13.3 Schematic of load cell beam design. Complete shop and assembly drawings are contained in Appendix .

13.2.4. Specifications and Design of Load Cell Instrumentation

a) Specifications and Calculations

Each of the four load cells should be able to withstand a maximum design load of approximately 500 lb(225 kg); which includes the weight of the BPC plus occupant, plus a safety factor of approximately 100 lb (45 kg), the assumption being that during the alignment process on the C.G. platform that any one cell could receive the majority of the load under light impact conditions, up to 225 kg. The instrumented (stressed) members of a load cell were to be fabricated from 1/2" (12.5 mm) thick aluminum (24S-T4) flat stock. To be useful each cell must provide a linear response over the range of 10-135 lbs (4.5-61 kg), and not exceed the tensile yield strength (S_u) of aluminum at the maximum rated load of 500 lbs.

Calculations for beam width 'b', given height 'h'= 0.500"

For a rectangular cantilevered x-section

$$smax = su = \frac{6M}{bh^2} \qquad (1)$$

where, M = FL, Maximum bending stress

F = Maximum Force

L = Maximum length through which F acts

b = width of X-section

h = height of X-section

s_u = tensile yield strength = 46 x 10^3 lb/in2

Let F = 500 lbs., L = 4.00", h = .500", (See fig 13.3)

Solving for 'b', from (1)

$$2b = \frac{6FL}{s_u h^2}$$

$$= \frac{6 \times 500 \times 4.00}{46 \times 103 \times (.500)^2}$$

2b = 120/115 = 1.04"

b = 0.502", i.e. 0.500" on each side.

The value 'b'=.500" (12.5 mm) was used to design the two cantilevered members in the the load cell shown schematically in figure 13.3. The x-sectional dimension of the beam is 12.5 x 12.5 mm.

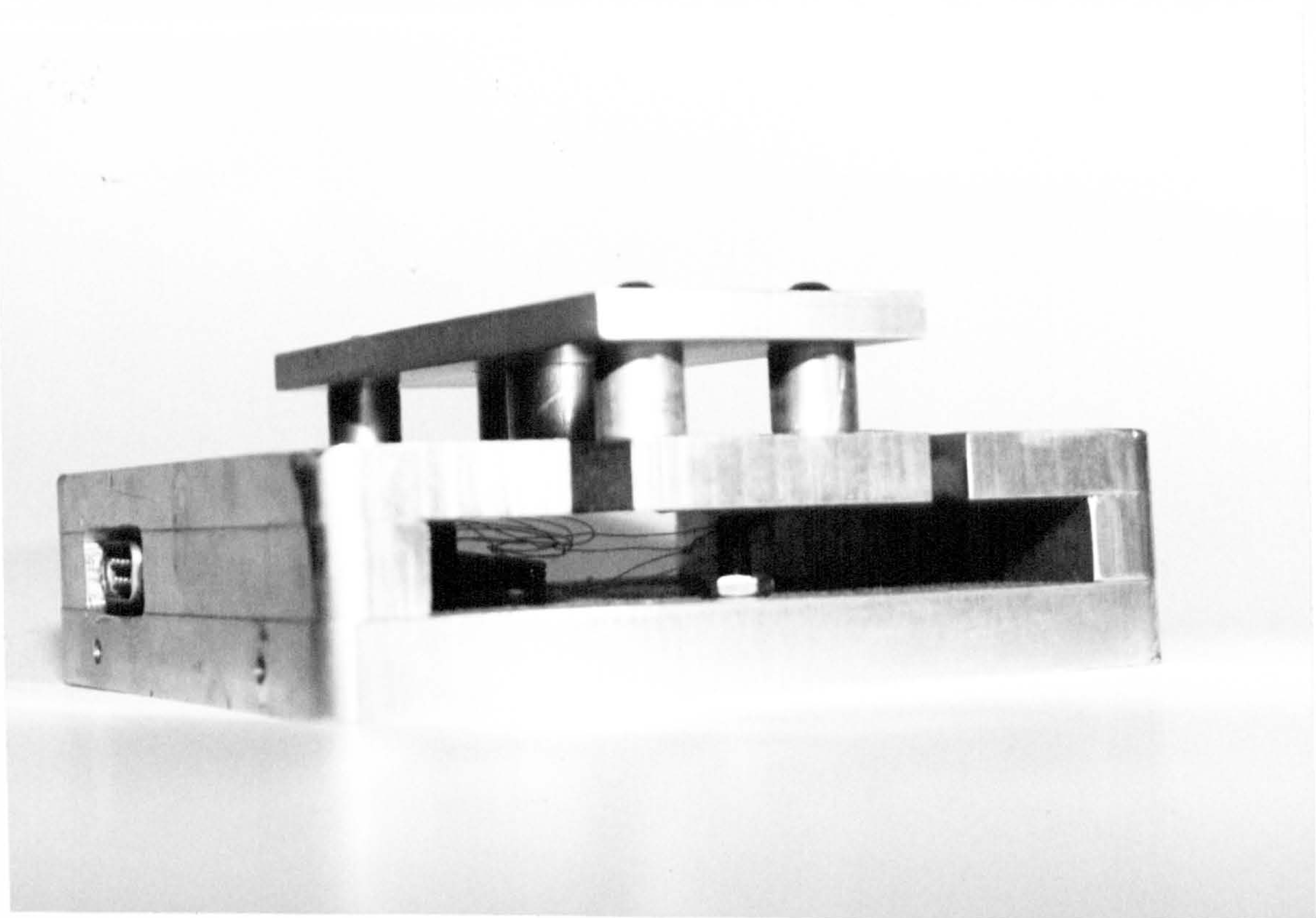


Fig 13.4 Assembled load cell with dust cover removed. Four load cells are used to compute the centre of gravity of the seated subject.

b) Load Cell and CG Platform Design

The details for the mechanical design of the load cells are in appendix 2. The design was based upon original work done by Wheatley (1982) with modifications by Reger et al (1985). The modifications done for the study consists of changes to the beam configuration, as detailed above, to give performance values consistent with the above specifications. The method for loading the cells was changed to a flat load bearing plate (Part C, Appendix 2). Also, vacuum formed plastic covers were added to protect the strain gauge wires from inadvertent damage. Figure 13.4 is a photograph of an assembled load cell with the dust cover removed.

The four load cells are mounted in the CG platform which consists of two parallel rails. The width between the rails and the load cell locations within each rail can be altered. In this manner the platform and load cell locations can be adjusted so that any wheelchair can be mounted on the CG platform. A ramped surface at one end of the rails allows the wheelchair plus occupant to be rolled onto the platform (fig 13.7).

c) Design of the Electronic Instrumentation

The load cell design called for four strain gauges, one mounted on each side of each cantilever beam. One gauge measures compression strain and the other tension strain. This approach reduces the accuracy requirements of placement of the wheels on the load cell. The gauges used are Micromeasurement 350 ohm, 6.35 mm in length and self-temperature compensated (EA-06-27BF-350 option LE).

Both the strain gauge mounting and the analogue circuit board were obtained from the Centre for Rehabilitation Technology, Helen Hayes Hospital, West Havestraw, New York. The analogue circuit is designed to provide four channels of output based on four Burr-Brown INA102 amplifiers and a Texas Instruments TLC25L4 quad. op. amplifier. The gain of the INA102 amplifier is dip switch selectable for 1, 10, 100, and 1000. A single trim pot provides nulling. The TLC25L4 is used as a non-inverting op. amplifier with a pot adjustable gain from 1 to 10.

Considerable difficulty was encountered in the implementation of the electronic instrumentation system. A major problem was the electronically noisy location of the laboratory. However, problems were systematically overcome by additional grounding, shielding, and auxiliary circuitry which permitted improved trimming, balancing, plus drastic noise reduction. The schematics for the circuit board modifications are included in Appendix 3.

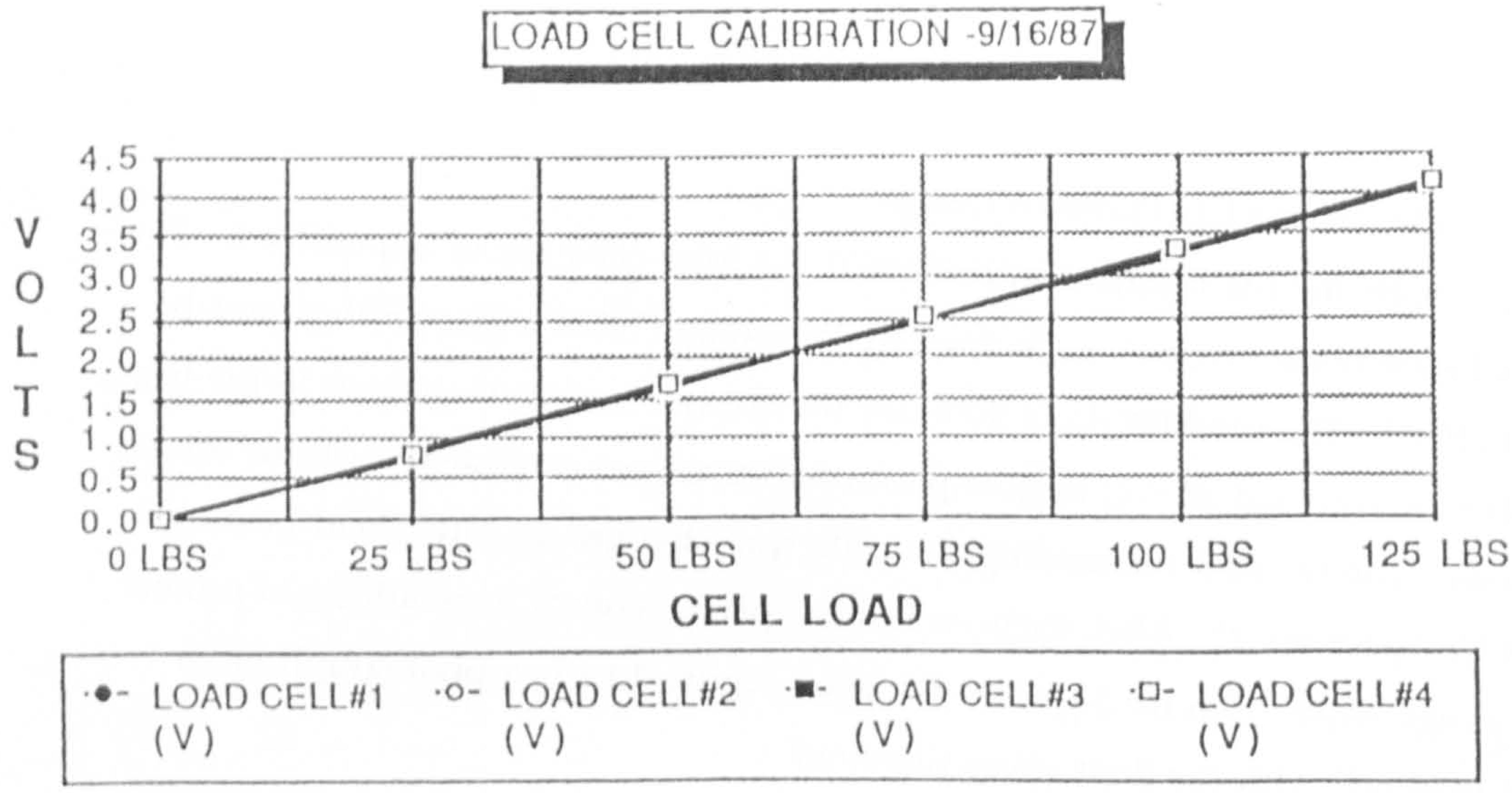


Fig 13.5 Plot of load versus output voltage for four load cells showing good linearity and consistency between cells.

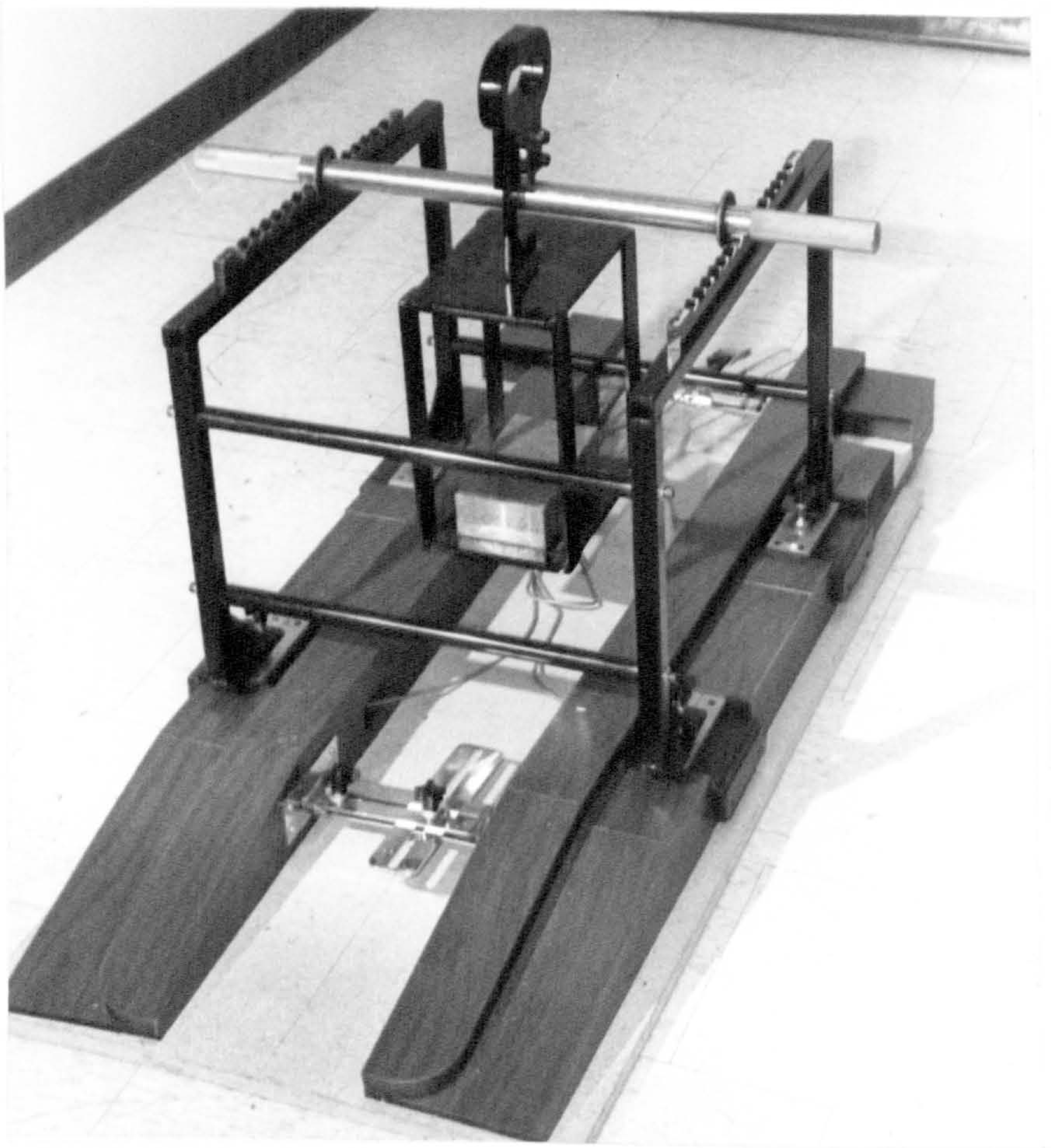


Fig 13.6 Centre of gravity calibration frame on four load cells with suspended test mass. The calibration frame permits measured CG coordinates to be compared to computed values prior to trials with subjects.

The analogue signals from the above circuitry were fed to the Metrabyte A/D system (vide 12.2.2) installed in an IBM PC computer. The Basic program working in conjunction with the Metrabyte board computes the CG coordinates and the body mass. The results were displayed on the computer screen, stored on disc, and printed out in hard copy for each of the nine body positions.

13.2.5. Calibration of the CG Platform

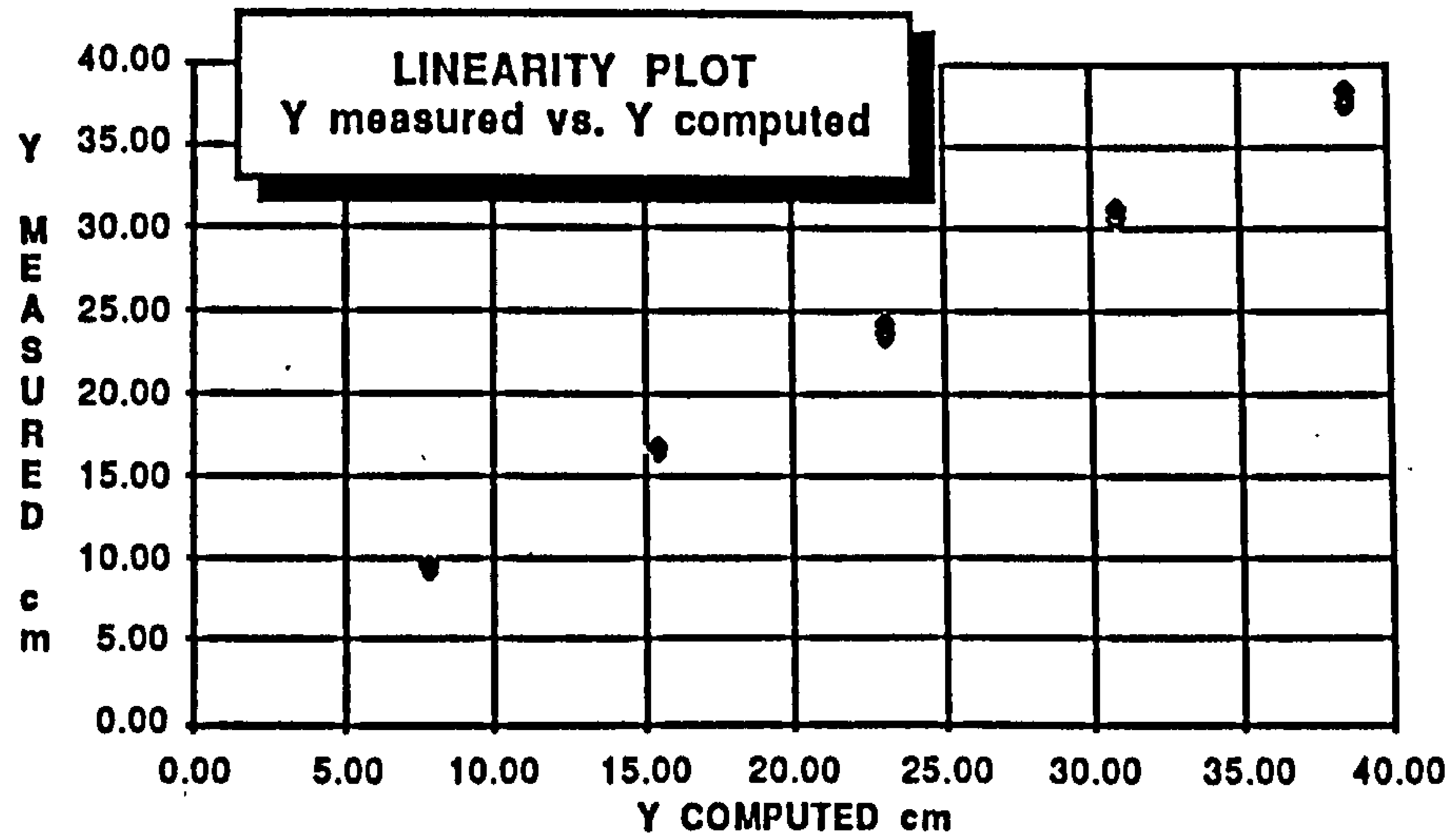
Calibration and accuracy estimations of the load cell platform were carried out at three levels. First the load cell transducers (cantilever beam/strain gauge components) were checked for linearity and accuracy over an operating range of 4.5-68kg. Figure 13.5 is a plot of output voltage versus incremental load for the four load cells. Accuracy of 1-2% over the operating range was achieved on all four load cells as can be seen from the linearity plot. After a ten minute warm-up period very little drift was measured due to RFI noise or temperature effects (< 0.5kg).

The second level of calibration involved the CG computations. This was achieved by constructing a calibration frame, a weight pan and calibrated weights up to a maximum load of 250 lbs (113kg). Figure 13.6 is a photograph of the calibration frame and weight pan.

The calibration procedure involved placing the calibration frame on the four load cells with a fixed weight of 250 lbs (113 kg.) in the weight holder. The weight holder was moved to measured (known) X and Y coordinates on the calibration frame. Measured X and Y coordinates and computed X and Y coordinates were then compared. A linearity plot of the compared results for X and Y values is shown in figure 13.7. A scatter plot of measured CG versus computed CG is shown in figure 13.8. The schematic and computations used to calculate the calibration CG locations X' and Y' are given in figure 13.9.

It can be noted in figure 13.8 that consistent errors in the lower values of Y' of 2 to 3 cm were obtained, when measured CG values are compared to computed values. In spite of many checks the reasons for this difference could not be determined. Therefore a correction factor was added to the computed values to adjust them to the measured values in the lower ranges of Y'. However, this was not considered a serious problem since the majority of the CG locations in the sample data occur in the 15 - 30 cm range of Y' and 5 - 20 cm range of X; ranges which have reasonably good correlation between measured and computed values.

The third and final level of calibration was to measure the CG location of the BPC loaded with a weight of 250 lbs (113kg). A knife-edge balance platform was con-



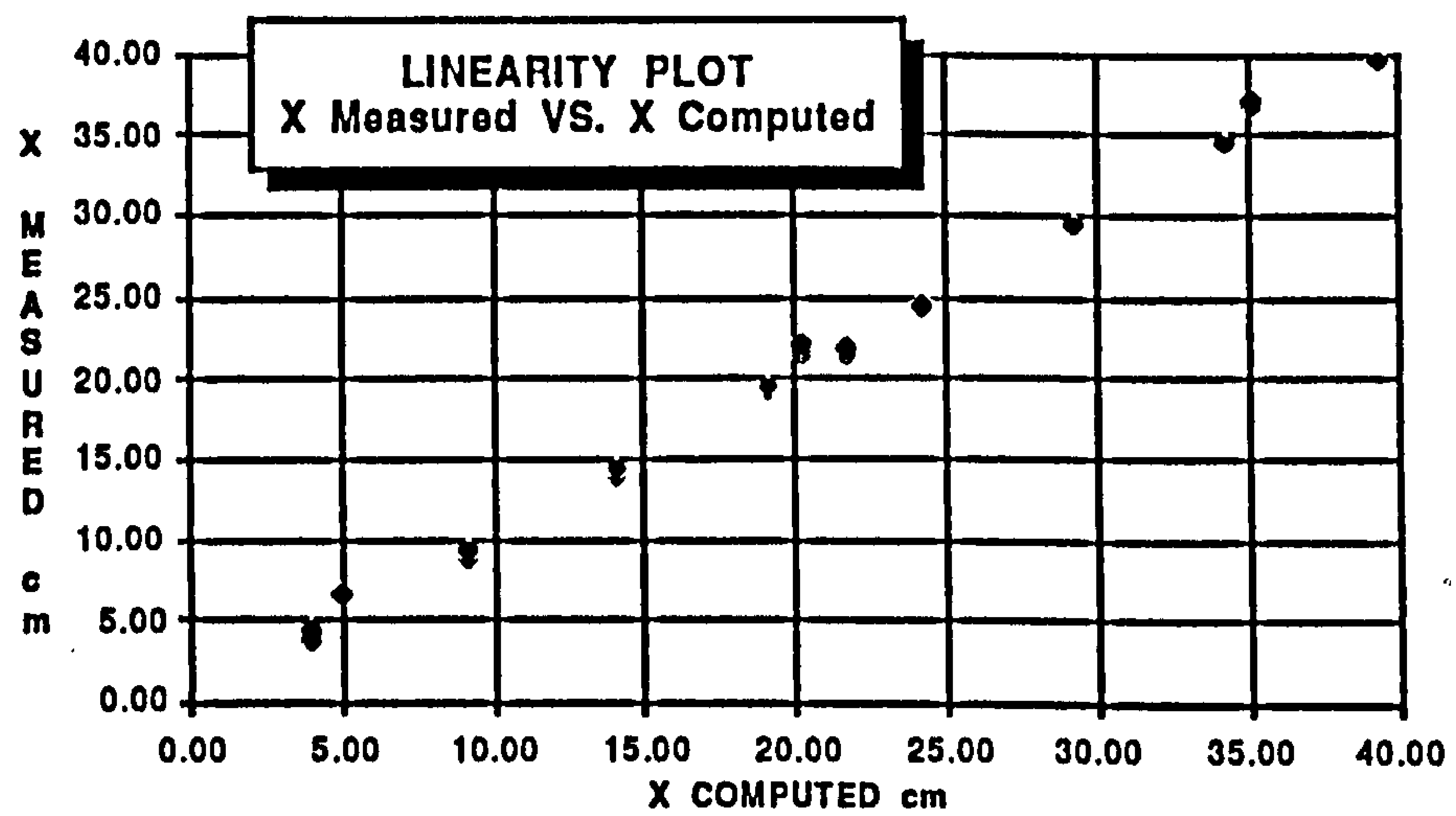


Fig 13.7 Linearity plots of X' and Y' coordinates of CG of test mass on calibration frame (fig 13.6)

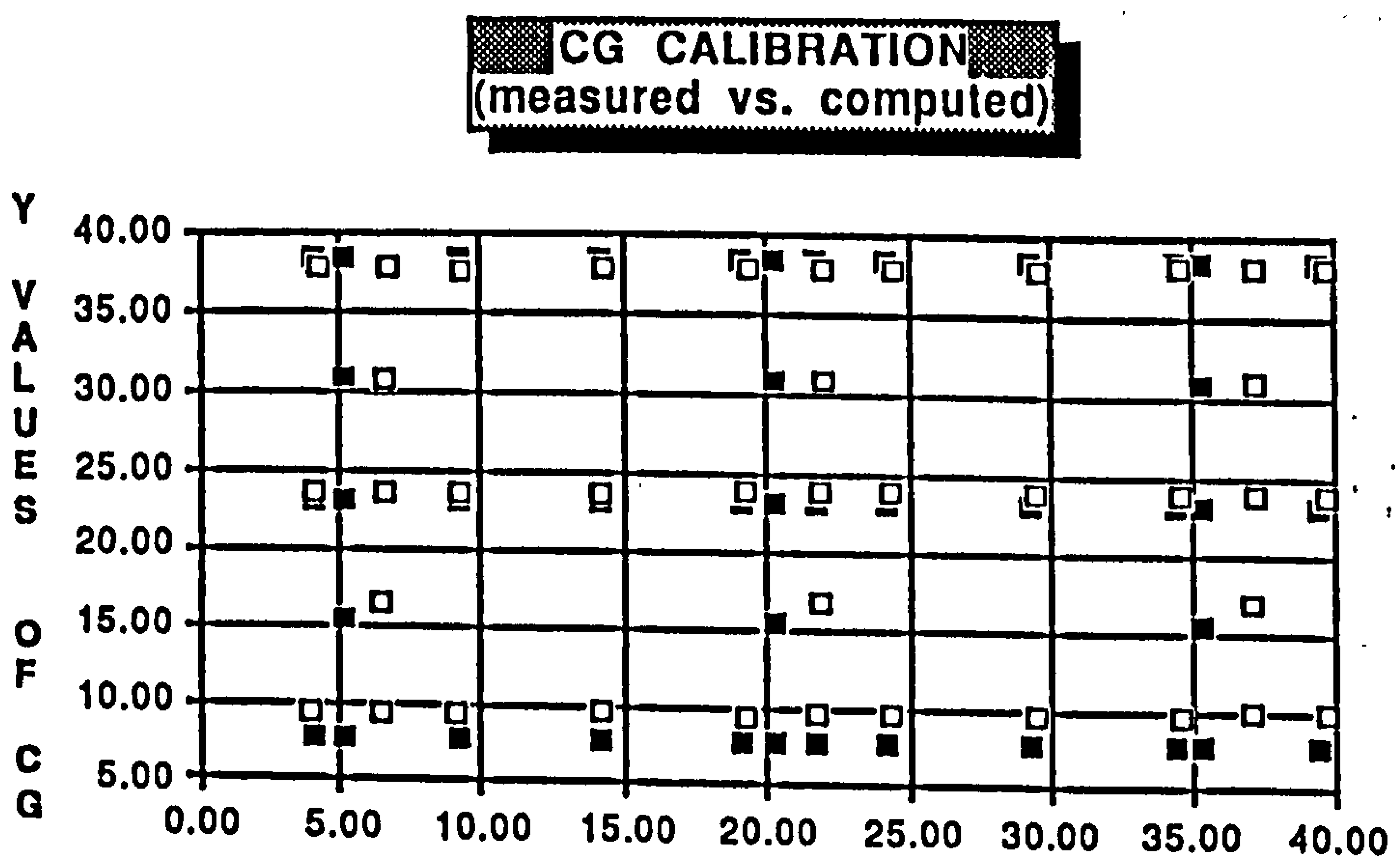
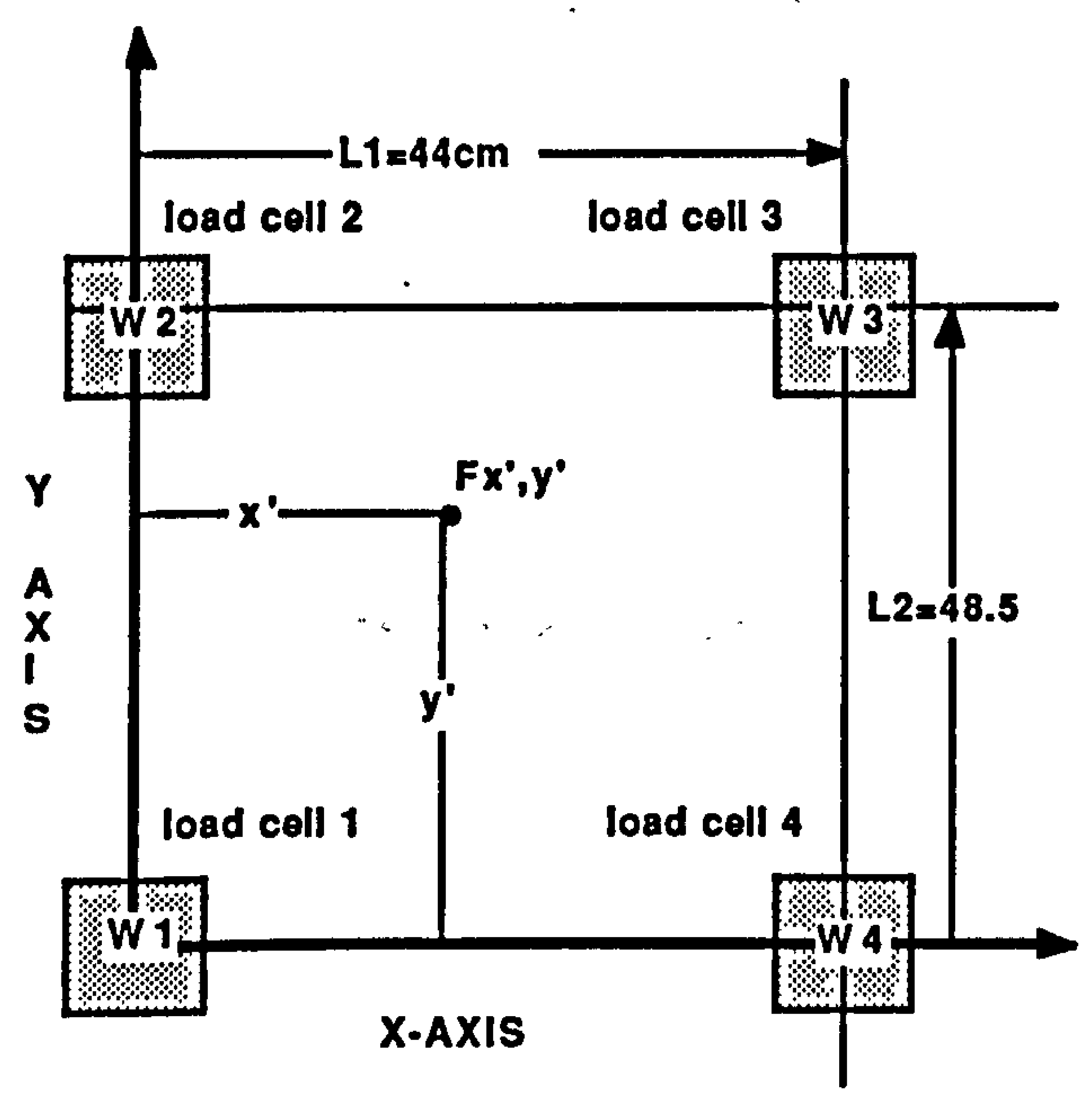


Fig 13.8 Scatter plot of CG coordinates, measured versus computed.

structed which allowed determination of X' and Y' for the loaded BPC by recording the balance point in the side to side (Y') and the fore and aft (X') directions. These values were compared to the X' and the Y' values obtained by using the load cell and computation method described previously. An approximate error of 0.5 cm in the X value and 1 cm in the Y value were noted. However, it was difficult to precisely determine the balance point on the balance platform, so differences between measured and computed values can be attributed to this difficulty.

In summary, it is estimated that the CG computations are accurate to +/- 0.5 cm over the range of concern.

CENTRE OF GRAVITY CALIBRA TION



x', y' = cordinates of centre of gravity (cm)

W1,W2,W3,W4 = wt. recorded by load cells1,2,3 and 4, respectively (kg).

F = total wt. = W1+W2+W3+ W4 (kg)

1) Taking moments about x axis

$$Fy' = L2(W2+W3)$$

$$Y' = \frac{L2(W2+W3)}{F}$$

2) Taking moments about y axis

$$Fx' = L1(W3+W4)$$

$$X' = \frac{L1(W3+W4)}{F}$$

During calibation L1=44.0 cm and L2= 48.5 cm., and F = sum(W1-W4), Therefore,

$$\boxed{Y'' = \frac{48\ (W2+W3)}{W1+W2+W3+W4}} \quad \text{and} \quad \boxed{X' = \frac{44\ (W3+W4)}{W1+W2+W3+W4}}$$

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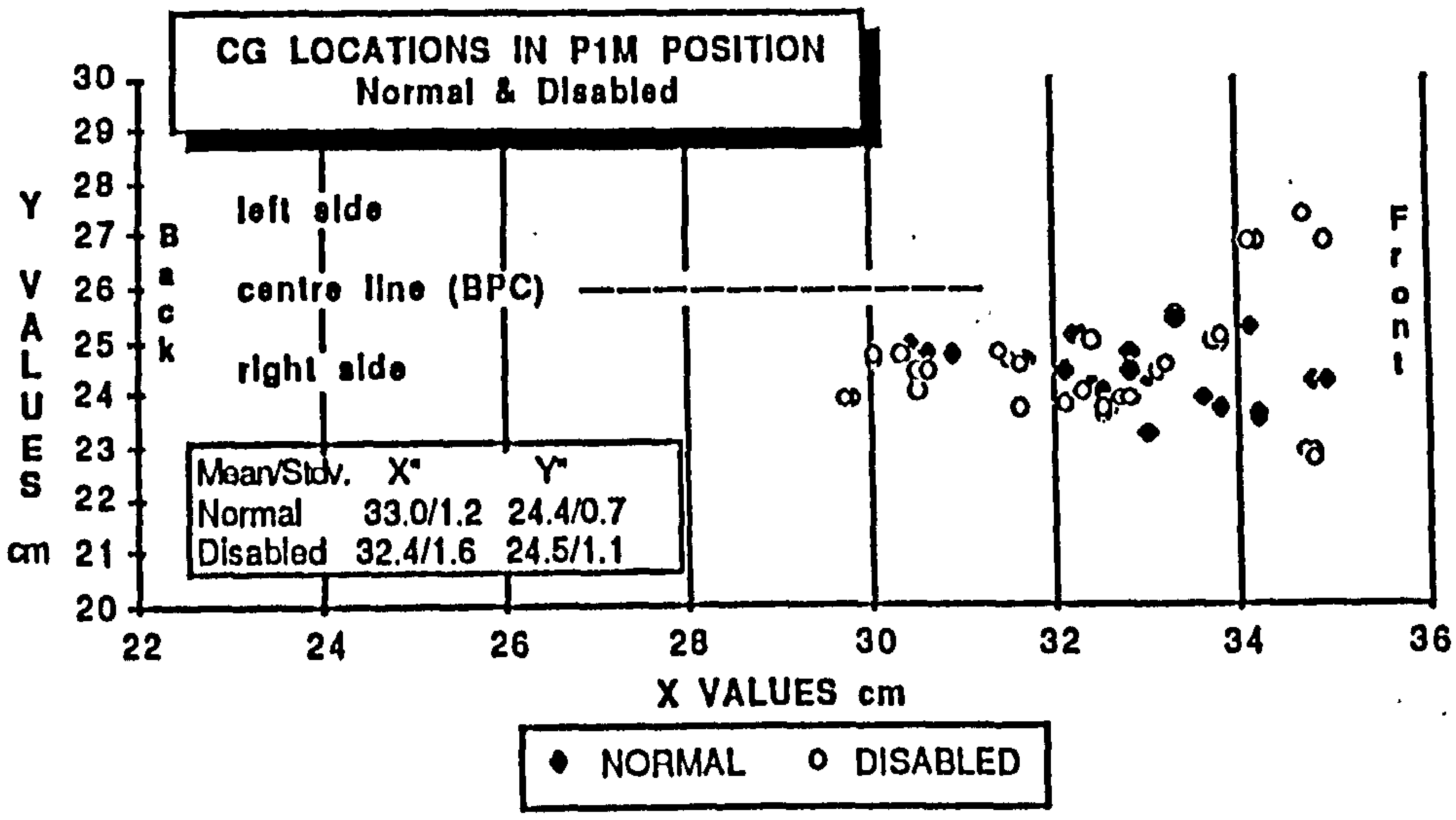
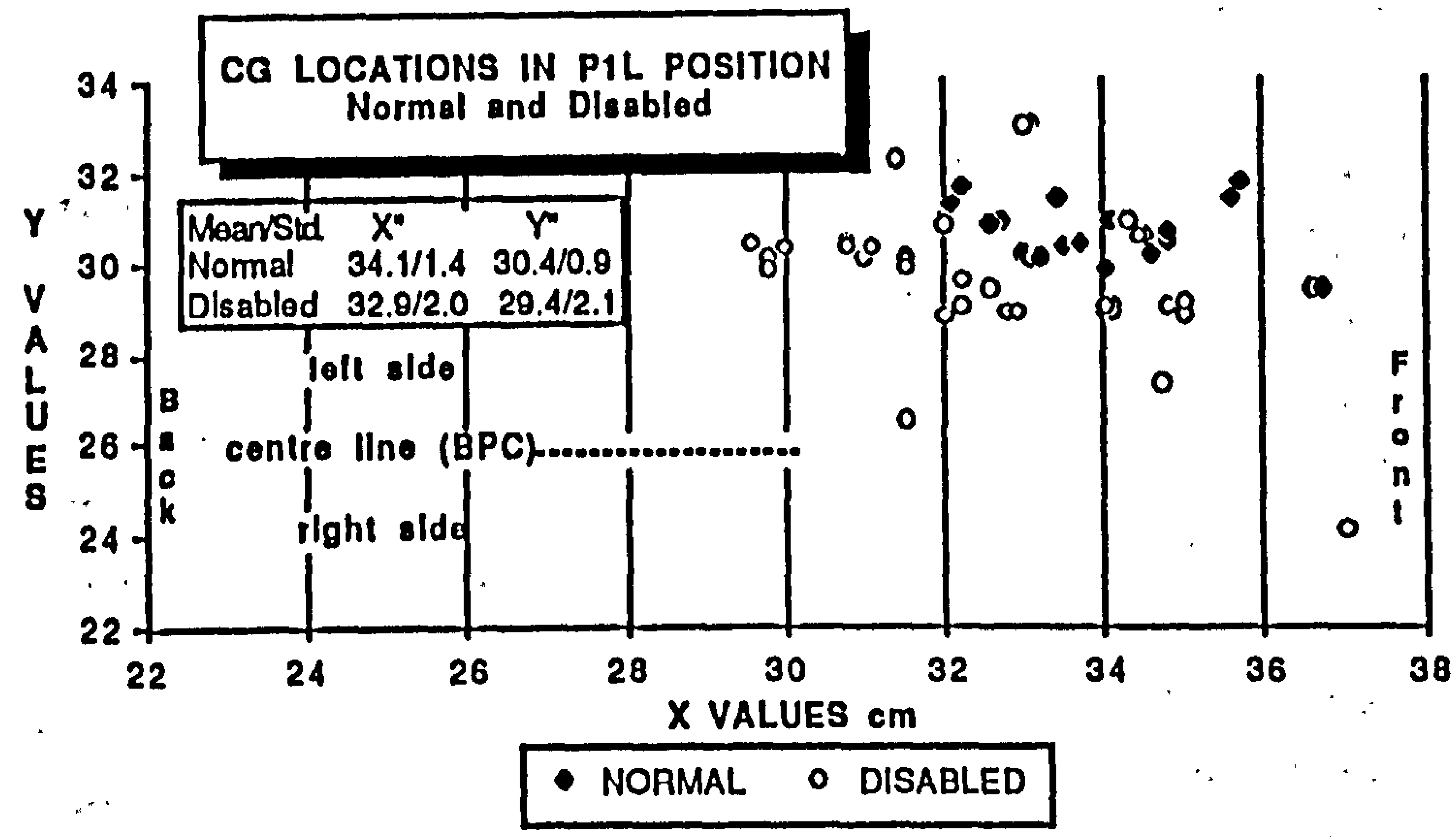
Fig 13.9 Calculations and reference locations for CG comparisons using calibration frame (fig 13.6).





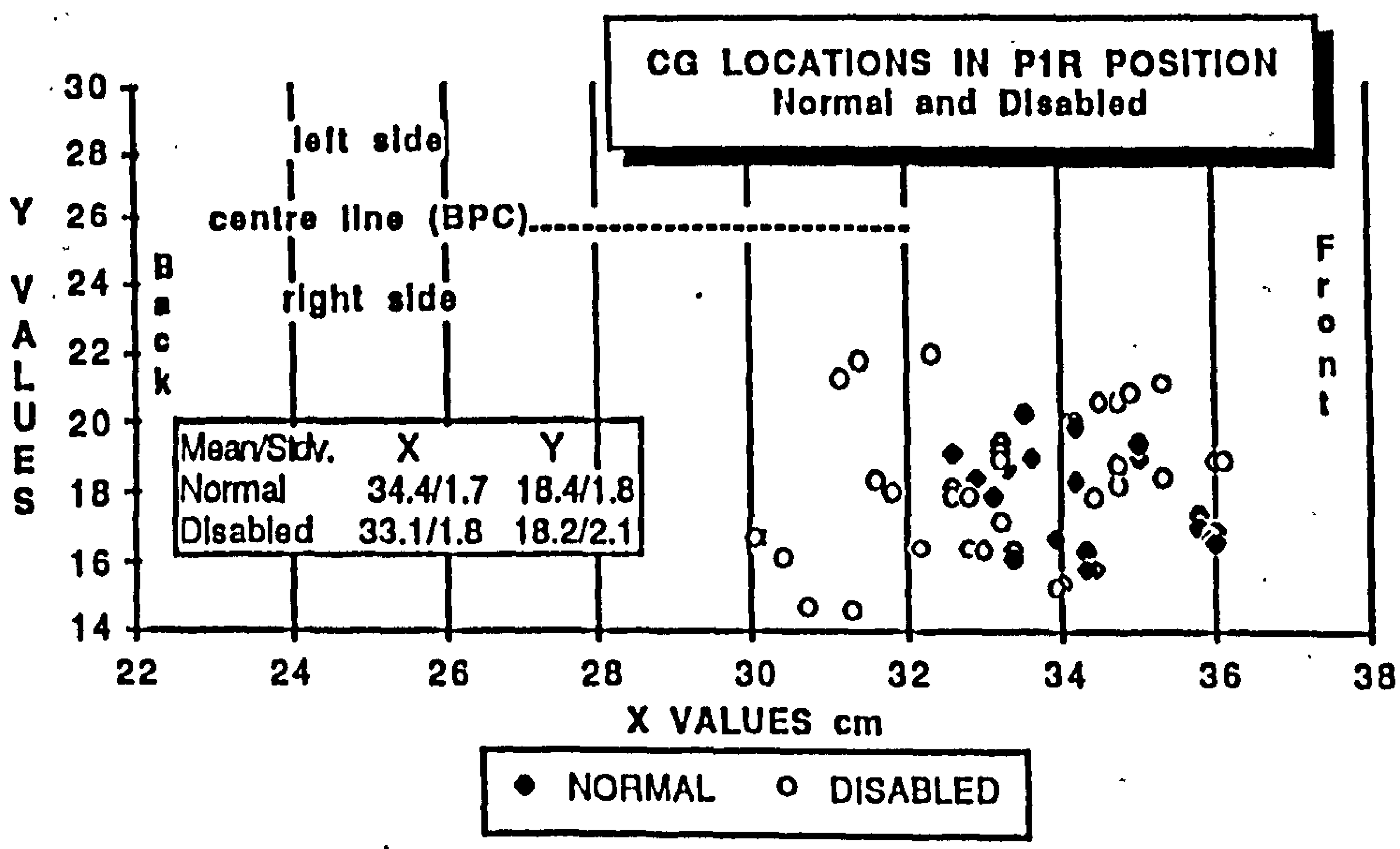


Fig 13.10 Mean centre of gravity locations for normal and disabled subjects for three body positions; neutral (P1M), left lateral bending (P1L) and right lateral bending (P1R).

13.3. RESULTS OF COMPARISONS OF CG DISPLACEMENTS

Again, the objective is to investigate the differences in CG locations within and between the two study groups and how the displacements of the CG may vary as the body posture is altered from one sitting position to another. The X" and Y" values of the CG of all test subjects in both groups were computed. The X" and Y" values are in centimeters measured from their respective reference axis, the origin of which is located at load cell 1, the load cell underneath the right rear wheel (W1) of the BPC (fig 13.9). Positive X" values are distances from the origin (rear wheel axles) in the forward or anterior direction. Positive Y" values are distances measured across the BPC from the right to the left side. The rear wheel base of the BPC is 52 cm, therefore the theoretical centreline of the BPC seat, assuming a perfectly mid-line orientation on the BPC, will be at Y"=26 cm from the X reference axis (fig 13.9).

The X" and Y" values for each subject in both groups were plotted on scattergrams, one plot for each of the nine body postures studied. Mean values and standard deviations of X" and Y" were computed for both groups. Figures 13.10-13.13 contain the summary results. The mean values and standard deviations of X" and Y" are given in tabular format within each scattergram. The conceptual centreline of the BPC has been drawn at Y = 26 cm which provides a visual reference for interpretation of Y values. The X and Y axis scales have been adjusted as necessary in each diagram so that the scatter plots would contain all the data points and the tabular information. The tabular information provides a quick summary of the scatter data for both study groups (mean values), as well as an indication of the variability of the data (standard deviations). A brief summary of the CG results will be presented for each of the nine scattergrams (figs 13.10-13.13).

First, it can be noted that in all diagrams, except in the P1L position (fig 13.10), that the CG of subjects in both groups is displaced to the right side of the conceptual mid-line of the BPC. This can be a result of a number of factors; a consistent measurement and computation error, a slight malignment of the seat on the BPC frame, or deformities that shift the CG from the mid-line. Since the mean offset has a consistent value of about 2 cm for both groups, it is assumed to be the result of an alignment error versus any variable that can be attributed to postural differences.

The following discussion will emphasize the differences in the location of the mean values of X" and Y". However, the scattergrams provide additional information in terms of the variability of the CG locations between the normal and disabled groups. The variability or spread of the CG data points is attributed to; postural deformities,

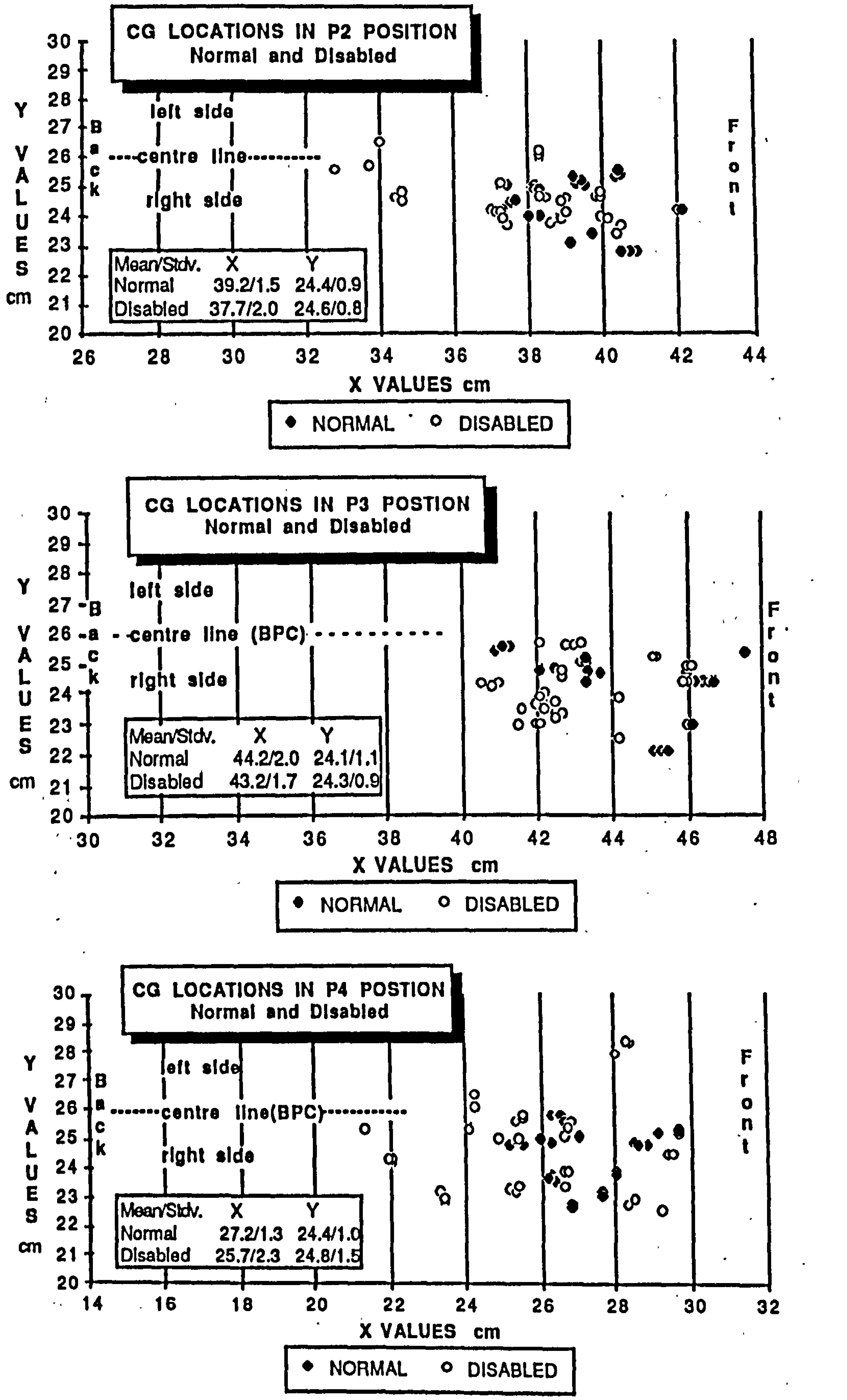


Fig 13.11 Mean centre of gravity locations for normal and disabled subjects for three body positions; forward trunk flexion 30°(P2), 50° (P3), and backrest recline 110°(P4).

varying ability to reposition the trunk and variability in body mass distributions exhibited by the subjects in the disabled group.

Larger X" values indicate a CG location further from the back plane of the BPC. Larger Y" values indicate CG locations increasing in distance from the right to the left side, with the conceptual mid-line located at Y = 26 cm, or the midpoint of the BPC wheel base.

1) Neutral Position (P1M)

Visual inspection of the upper plot in figure 13.10 indicates a larger spread of CG locations for the disabled group with a mean value at X" = 32.4, Y" = 24.5. The normal group mean value of X" = 33.0, Y" = 24.4 suggests a mean CG location closer to the front of the BPC, however the difference is probably not statistically significant. The approximately 1 cm difference does tend to suggest that reduced lower body mass may displace the mean CG posteriorly in the disabled group.

2) Left Lateral Bending (P1L)

Inspection of the middle plot in figure 13.10 indicates the CG locations upon lateral bending to the left. It can be noted that the normal group is able to achieve a greater lateral shift (mean Y" = 30.4 cm) than the disabled group (mean Y" = 29.4 cm). One disabled subject was apparently unable to even shift their CG beyond the mid-line. It is also evident that several normal subjects shifted forward as well as sideways during the lateral bending position, i.e., X" > 36 cm. In general, the mean X" values for both groups remain similar to those measured in the P1M position. The most notable difference between the groups is the difference in the variability of their data points (0.9 vs. 2.1)

3) Right Lateral Bending (P1R)

The lower plot in figure 13.10 shows the expected displacement to the right side of both groups, with almost identical mean Y" values, i.e., 18.4 and 18.2. It is also noted that all subjects were able to displace their CG to the right side of the mid-line. In reviewing the radiographic data it was noted that the majority of the lateral spinal deformities were concave to the right which possibly accounts for the mean differences in CG location upon left and right lateral bending for the disabled group.

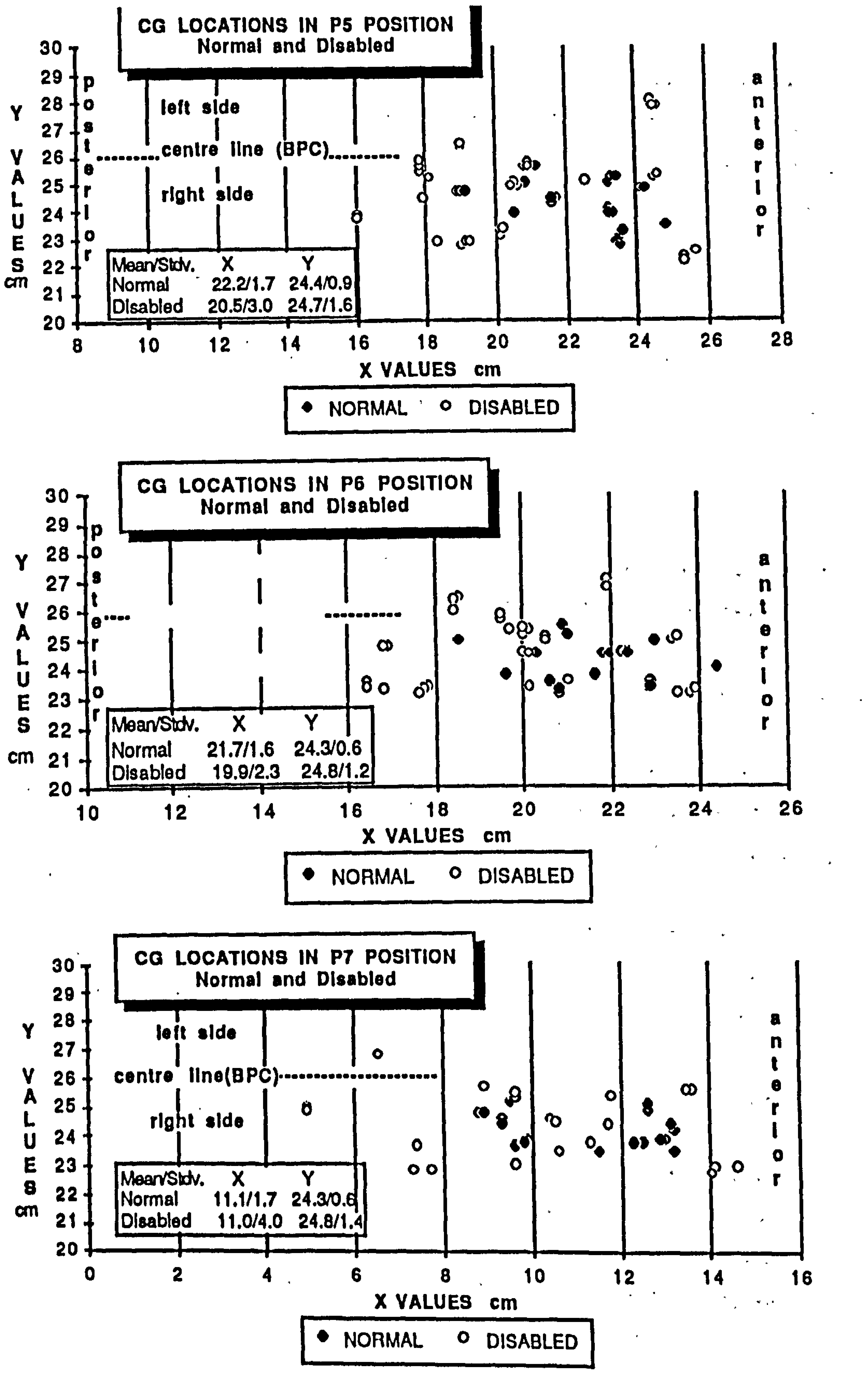


Fig 13.12 Mean centre of gravity locations for normal and disabled subjects for three body postures; 120° backrest recline (P5), and 10° (P6) and 20°(P7) full body tilt for normal and disabled subjects.

4) Forward Trunk Flexion to 30° (P2)

The upper plot in figure 13.11 suggests that trunk flexion forward from the neutral position effectively moves the CG anteriorly, resulting in the mean X" values of 39.2 for the normal group and 37.7 for the disabled group, with little difference in side to side Y" locations. It can also be noted from the scattergram that at least six disabled subjects were unable to effect a forward CG shift beyond 35 cm. It was noted that several subjects had difficulty maintaining this posture during the measurement session. They compensated by placing their hands on the armrests of the BPC which may have partially negated the forward shift of their CG.

5) Forward Trunk Flexion to 50° (P3)

The middle plot in figure 13.11 shows the CG locations after forward flexion position to 30°. Both groups appear to have similar mean values with the disabled group unable to effectively shift quite as far forward.

6) Backrest Recline to 110° (P4)

Reclining the backrest to the 110° position expectedly shifts the CG posteriorly. It is interesting that a shift of only the trunk causes a greater displacement to the rear for the disabled group (X" = 25.7 cm) than it does for the normal group (X" = 27.2 cm). This can be confirmed visually by the data spread in the scattergram.

7) Backrest recline to 120° (P5)

The P5 plot in figure 13.12 shows that increased backrest recline causes a further shift of the CG posteriorly, while maintaining a similar difference between the X" values for disabled and normal groups (20.5 vs. 22.2). As expected, mean Y" values remain similar for both groups.

8) Full Body Tilt of 10° (P6)

Whole body tilt of 10° causes an even further posterior displacement of the CG, when compared to position P5 (fig 13.12). Again, posterior shifts appear greater for the disabled group, Y" = 19.9 cm versus Y" = 21.7 cm with the difference remaining approximately the same for P4 and P5 (2 cm).

9) Full Body Tilt to 20° (P7)

The lower plot in figure 13.12 suggests that further whole body tilt to 20° causes even further posterior displacement with the mean X" value for both groups being

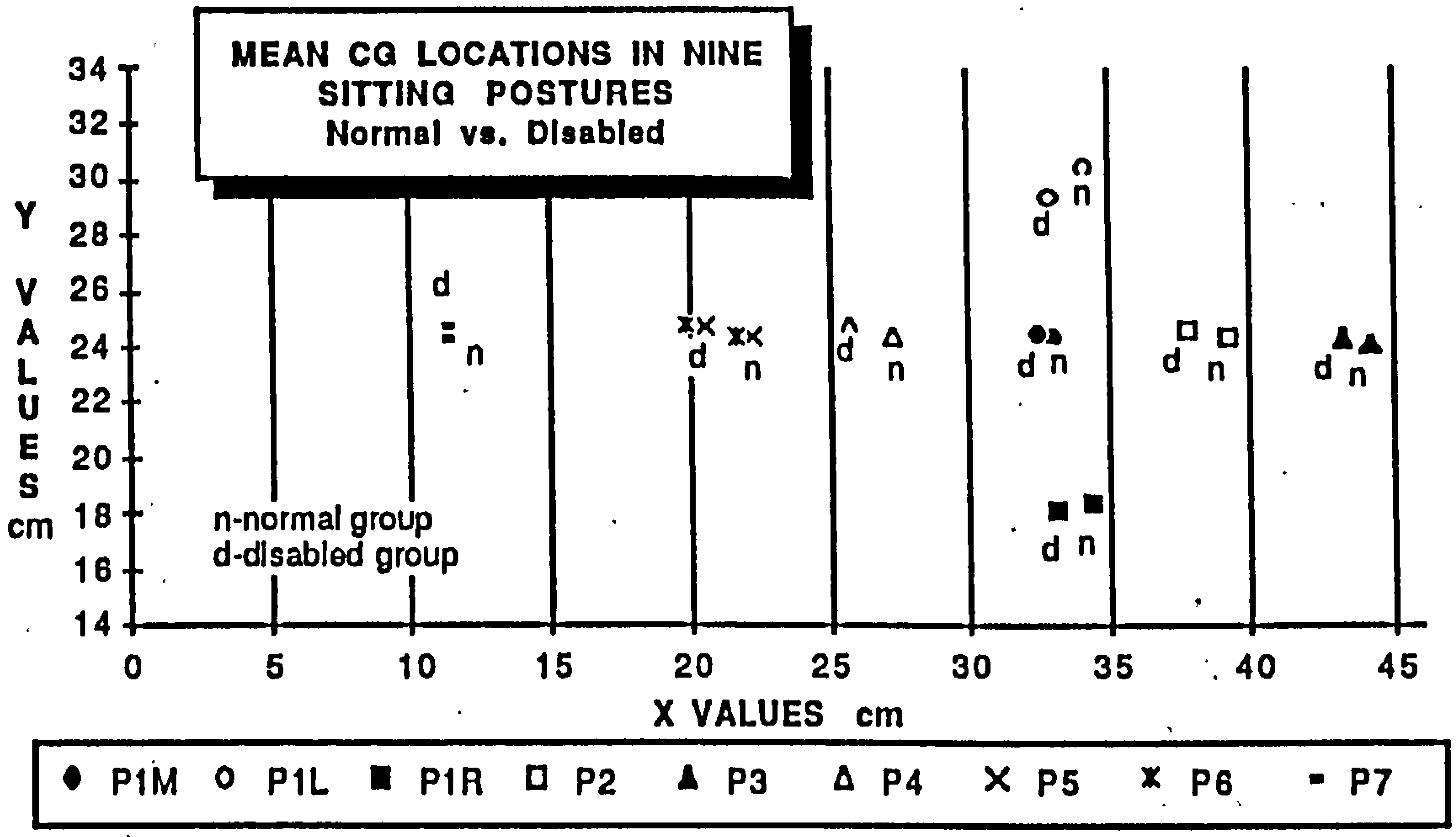


Fig 13.13 Mean centre of gravity locations from references axes for disabled and normal groups in the nine sitting postures.

essentially equal, i.e., X" = 11 cm. There is a significant spread in the Y" values of the disabled group, i.e., stdv. = 4.0 cm. This could be due to the wide variability of the trunk weight within the disabled group. Since it is the trunk weight that mostly effects the CG displacement within the disabled group, this also accounts for the lesser spread of data for the normal group; whose body weight is more evenly distributed between the trunk and lower limbs. As expected, the mean Y" values of both groups remain essentially identical at approximately Y" = 24.5 cm.

10) And finally, figure 13.13 is a summary plot of the mean CG values for both groups.

First, it is evident that the actual centreline, as indicated by the Y" value of the CG, is approximately equal to 24 cm for both groups. Secondly, the rather large A/P displacement of the CG resulting from the postural changes is more evident in the composite plot. The CG locations range from a rearward value of about Y" = 11 cm in the P7 (full body recline - 20°) to a forward value of about Y" = 44 cm in the P3 (forward flexion - 30°) posture. And finally, it can be noted that mean CG values of the disabled group are located more posteriorly than the normal group in all nine sitting postures. However, in contrast to figures 13.11-13 the composite plot of the mean CG values in figure 13.14 gives no indication of the spread of the data between the two groups.

13.4. DISCUSSION OF RESULTS FROM CG COMPARISONS

Measurement of CG locations in the nine study postures has indicated several apparent differences between the normal and disabled groups. The mean CG values indicate that the disabled group has a more *posterior* location of their CG in all postures. This is consistent with the biomechanical rationale that the weight distribution of the disabled group due to lower limb wasting causes a relative posterior CG displacement. However, when the scatter data are reviewed for each posture (figs 13.11-13), a number of the disabled subjects exhibited CG locations considerably outside the normal range. Also, several disabled subjects appeared unable to attain the same degree of *lateral CG shift* upon left lateral trunk bending when compared to normal subjects.

Possibly the most important confirmation is the wide variability in the CG location of the disabled. This suggests an inability of spinal cord injured people, on the average, to retain a mid-line upright posture as consistently as a similar group of able-bodied people. It is proposed that this is a result of varying skeletal deformities and/or muscle dysfunction that results in asymmetrical sitting postures, hence larger variances in their CG locations.

It is now important to assimilate the CG findings with those from the three other aspects of the study (radiographic, surface shear, and pressure) in an attempt to better understand how the various parameters may be interrelated. However, before assimilating the individual components into a composite perspective that can lead to final conclusions, its necessary to examine the statistical inferences of the above observations.

CHAPTER 14. ASSIMILATION OF RESULTS

14.1. THE GENERAL APPROACH

The previous four chapters contain the objectives, procedural details and results pertaining to each of the four components of the study. Differences within and between the disabled and the normal samples were identified in each of the study components. It is now important to attempt to understand how the various interface variables may be interrelated. For example, it would be of value to know if there is a posture in which both tangentially-induced shear and maximum pressure are minimized, thereby minimizing the deleterious effects of both factors; or it would be of value to know how a centre of gravity change may interact with the location of maximum pressure; or if certain sitting postures cause the maximum pressure (Pm) to move to buttock locations other than those at high risk of ulceration. This chapter attempts to investigate these complex inter-relationships between the study variables, as the last step towards drawing the final conclusions from the study.

14.2. IMPLICATIONS FROM STATISTICAL ANALYSES

A statistical sample can never give us more information than if the whole population was measured. Since in most cases it is impractical to measure the whole population, a representative sample is chosen. But sampling introduces error. Statistical sampling theory permits an estimation of the error and indicates how it may be reduced by increasing the sample size. The theory is based on the concept that some form of random or probability sampling method was used to select the sample(s). The various sampling techniques are designed to remove selection bias so that the sample will be as representative of its population as possible. It has been assumed that the selection process briefly outlined in Chapter 7 reasonably meets these criteria, and therefore basic sampling statistics and the Central Limit Theorem can be used to describe the the characteristics of the two chosen populations.

Confidence limits provide a way of estimating the probability that the results from the random samples are good estimates of the population parameters. Tests of significance are useful when decisions are needed as to whether observed differences are likely real or only due to sampling errors. These tests are particularly valuable and necessary when the sample sizes are small. Both techniques rely on the assumption that the distribution of the sample data reasonably approximates a normal (Gaussian) distribution (Ehrenberg, 1986).

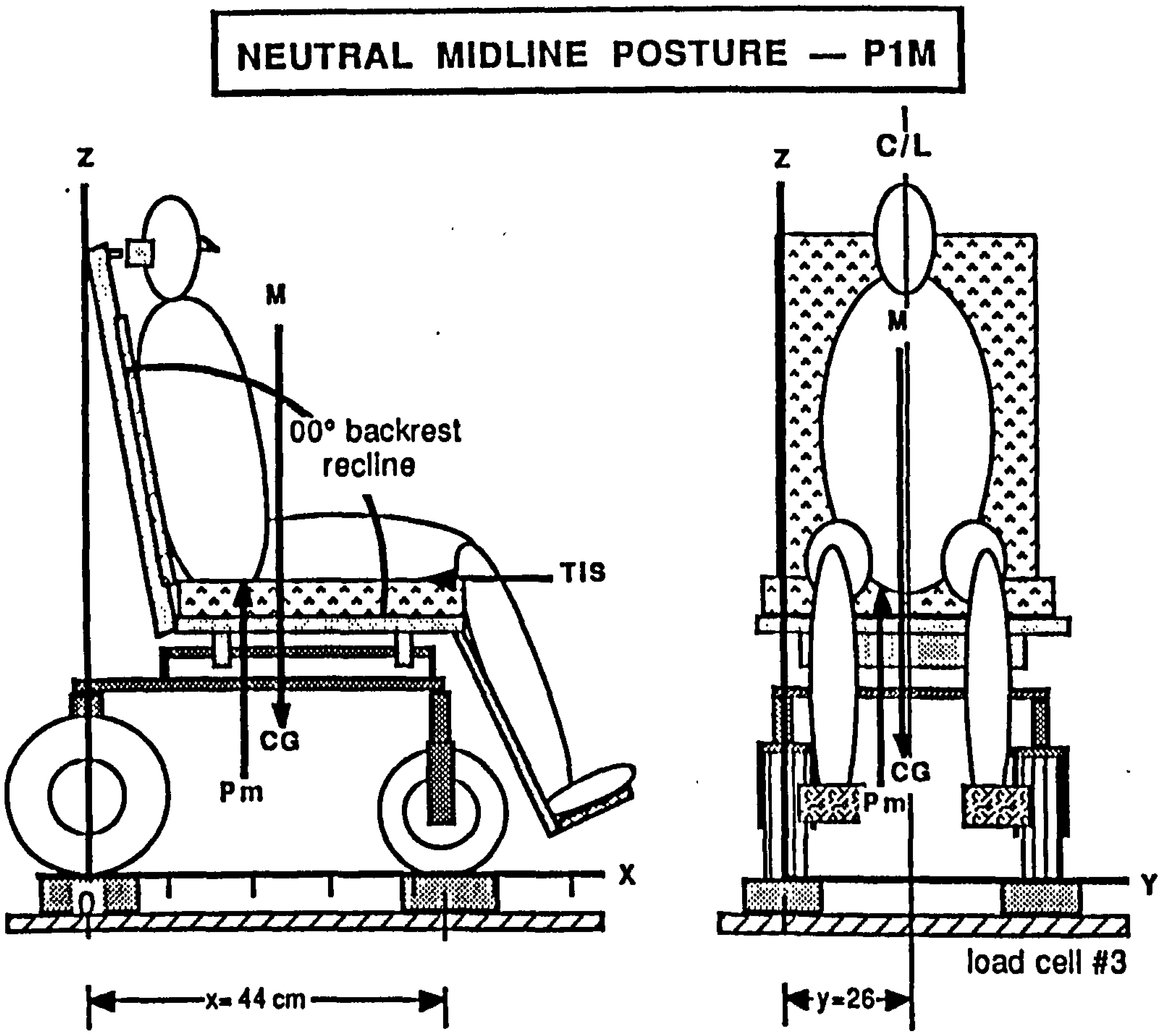
Tests for normality have been done on all the raw data (pressure, CG, radiographic and surface shear). The tests have been done using the Kolmogorov-Smirnov one-sample test because of its sensitivity, even with small sample sizes (Rafferty, et al, 1985). The Kolmogorov-Smirnov test used gives a goodness-of-fit statistic between the sample and the theoretical normal distribution. If the probability is low it is evidence that the sample came from something other than a normal distribution. The two-tail level of significance test was also used. This also indicates the probability of obtaining such a sample from a population having a normal distribution. For example, a value less than 0.05 is evidence that the population was not normal (Rafferty et al, 1985). The skewness and kurtosis were also examined. Skewness values give an indication as to how far to the left or right the sample data curve may be displaced compared to the theoretical normal distribution. Kurtosis gives an indication of the 'peakedness' of the sample data curve, compared to a theoretical normal distribution. For example, excessively low curves will have more data points within the tails and therefore the descriptive aspects of the Central Limit Theorem may no longer apply.

Based on the above analysis, combined with a visual inspection of the distribution of the sample data, the conclusion was made that within reasonable limits of skewness, and with a few exceptions, the recorded data approximates a normal distribution.

The hypothesis calls for the determination of differences in the mean values derived from the two samples studied. Specifically, comparisons both within (paired) and between (unpaired) the two groups have been done on the data derived from the nine postures and four variables studied. Tests of significance have been done. The Student's T-test was used to test the paired and unpaired samples because of its increased sensitivity to the errors associated with small samples (Spiegel, 1975). The significance given by the test is the two-tailed probability of observing the t-value if the samples were taken from the same population. A significance level of 0.05 was chosen as the level about which significance would be declared.

Comparative statistical theory, although a powerful tool, does contain a grey area when declaring significance. The concept of 'a' level tends to place exaggerated importance on a precise level of significance. For example, having declared a level of 0.05 it makes little sense to declare a result of 0.055 insignificant. Therefore, the terms insignificant, marginally insignificant, marginally significant and significant will be used to statistically describe the results of the study.

The results of the descriptive statistical analyses have been added to most of the plots and tables associated with presentation of the results in chapters 10-13. The



MEAN P1M COMPOSITE VALUES

Variable	Normal Group Value	Normal Group % Chge. from P1M	Disabled Group Value	Disabled Group % Chge. from P1M
Lumbar Angle S1-L3° (L1)	22	0	27	0
Lumbosacral Angle° (L5)	18	0	13	0
Pelvic Angle° (L7)	71	0	56	0
IT Location (X1), cm	25	0	29	0
IT Angle° (L8)	1.4	0	3.2	0
RIT to C/L (X4), cm.	6.3	0	5.8	0
LIT to C/L (X5), cm.	6.5	0	7.1	0
LIT Location (Z1), cm.	- 0.1	0	0.1	0
Avg. Pressure (Pa), mmHg.	73	0	70	0
Max. Pressure (Pm), mmHg.	117	0	158	0
Loc. of P(m), (Pm_x,Pm_y), cm.	26.7, 22.1	0	27.5, 23.5	0
Peak Gradient (Pg), mmHg mm^{-1}	1.7	0	3.4	0
Tangent. Shear (TIS), newtons	67	0	69	0
CG Location (X",Y"), cm.	33.0, 24.4	0, 0	32.4, 24.5	0, 0

Fig14.1 Composite results in the neutral posture (P1M).

following section addresses the comparative analysis in the course of discussion of the composite results and final conclusions.

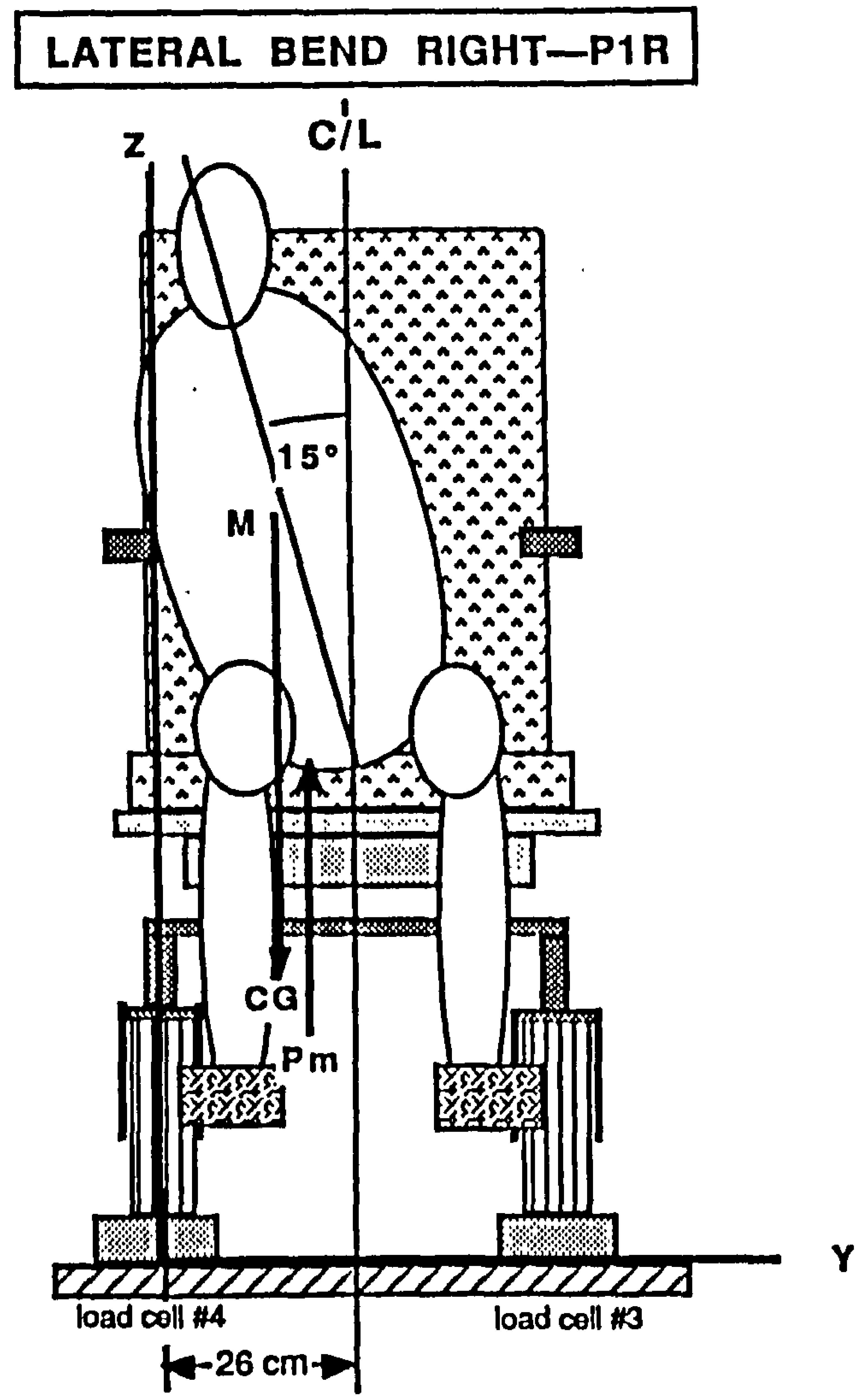
14.3. THE COMPOSITE RESULTS

The method used to present the composite results will be to summarize the salient results from each aspect of the study (pressure, shear, skeletal alignment and CG location) with emphasis on their inter-relationships. As done previously, comparisons between the disabled and normal groups will be made in each of the nine study postures with particular emphasis on changes from the neutral posture (P1M). Figures 14.1-9 contain the composite results derived from the four aspects of the study in each of the nine postures. These figures will be referred to throughout the following discussion. Significance has been declared at the 0.05 level.

14.3.1. Composite Results in the Neutral Posture (P1M)

The neutral posture P1M has been defined as the reference posture to which all other postures are compared. It is defined as the posture which simulates the typical sitting posture in a wheelchair. Figure 14.1 contains the spinal/pelvic, pressure distribution, tangential shear (TIS) and CG location values for both groups. Of course, the percent differences from the P1M posture are zero.

Regarding the spinal and pelvic alignment, a salient finding is the difference in the mean pelvic angle (L_7) between the two groups; 71° for the normal versus 56° for the disabled. This finding is further denoted by the linear displacement values of the IT location (X_1) from the posterior reference axis. The disabled group has an X1 value of 29 cms versus a value of 25 cm for the normal group. These results support the clinical observation that individuals with spinal cord injury tend to sit with a posterior pelvic rotation (L7), which in the sample studied is 15° (71° - 56°) less than the normal group. This posture displaces the ischial tuberosities forward by an average amount of four cms (29-25 cm). Differences in pelvic angle (L7), sacral angle (L6) and the linear displacement of the ITs (X1) were found significant; differences in all other angles and distances insignificant. At the same time, it can be seen that deformities tend to shift the pelvis away from the mid-line as viewed in the frontal plane. The mean RIT to C/L (X_4) and LIT to C/L (X_5) distances for the normal group show very little lateral offset from the mid-line (0.1 cm). Whereas, the disabled group have a mean offset from the mid-line of 1.6 cm as indicated by the RIT and LIT to C/L distances of 5.8 cm and 7.1 cm, respectively. However, the large standard error due to the spread of the data points renders this observation insignificant.



MEAN P1R COMPOSITE VALUES

Variable	Normal Group Value	Normal Group % Chge. from P1M	Disabled Group Value	Disabled Group % Chge. from P1M
IT Angle° (L8)	6.4	78	7.9	59
RIT to C/L (X4), cm.	4.9	-29	4.3	35
LIT to C/L (X5), cm.	8.0	19	8.3	14
LIT Location (Z1), cm.	0.3	133	1.5	93
Avg. Pressure (Pa), mmHg.	68	-9	73	1
Max. Pressure (Pm), mmHg.	172	47	204	29
Loc. of P(m), (Pm_x,Pm_y), cm.	27.0, 19.9	1, -10	27.5,19.5	0, -17
Peak Gradient mmHg mm^{-1}	2.0	15	5.4	61
Tangent. Shear (TIS), newtons	60	-10	62	-11
CG Location (X",Y"), cm.	34.4, 18.4	4, -25	33.1, 18.2	2,-26

Fig 14.2 Composite results in the right lateral bending posture (P1R)

Regarding the pressure distribution, a significant difference exists between the mean maximum (Pm) values (117 versus 158 mmHg). It can also be noted that the mean X location of Pm (Pmx) is only about 1 cm (27.5 - 26.7) further forward for the disabled group, even though the mean location of the ischial tuberosity is about 4 cms forward of the normal group. The results also indicate a significant difference in the peak pressure gradients (Pg) between the two groups (1.7 versus 3.4 $mmHg^1$). This is to be expected due to the large difference in the maximum pressure values.

Although differences exist in other variables recorded in the P1M posture, i.e, shear, CG location, and other spinal/pelvic angles, the differences are much smaller and therefore were judged insignificant.

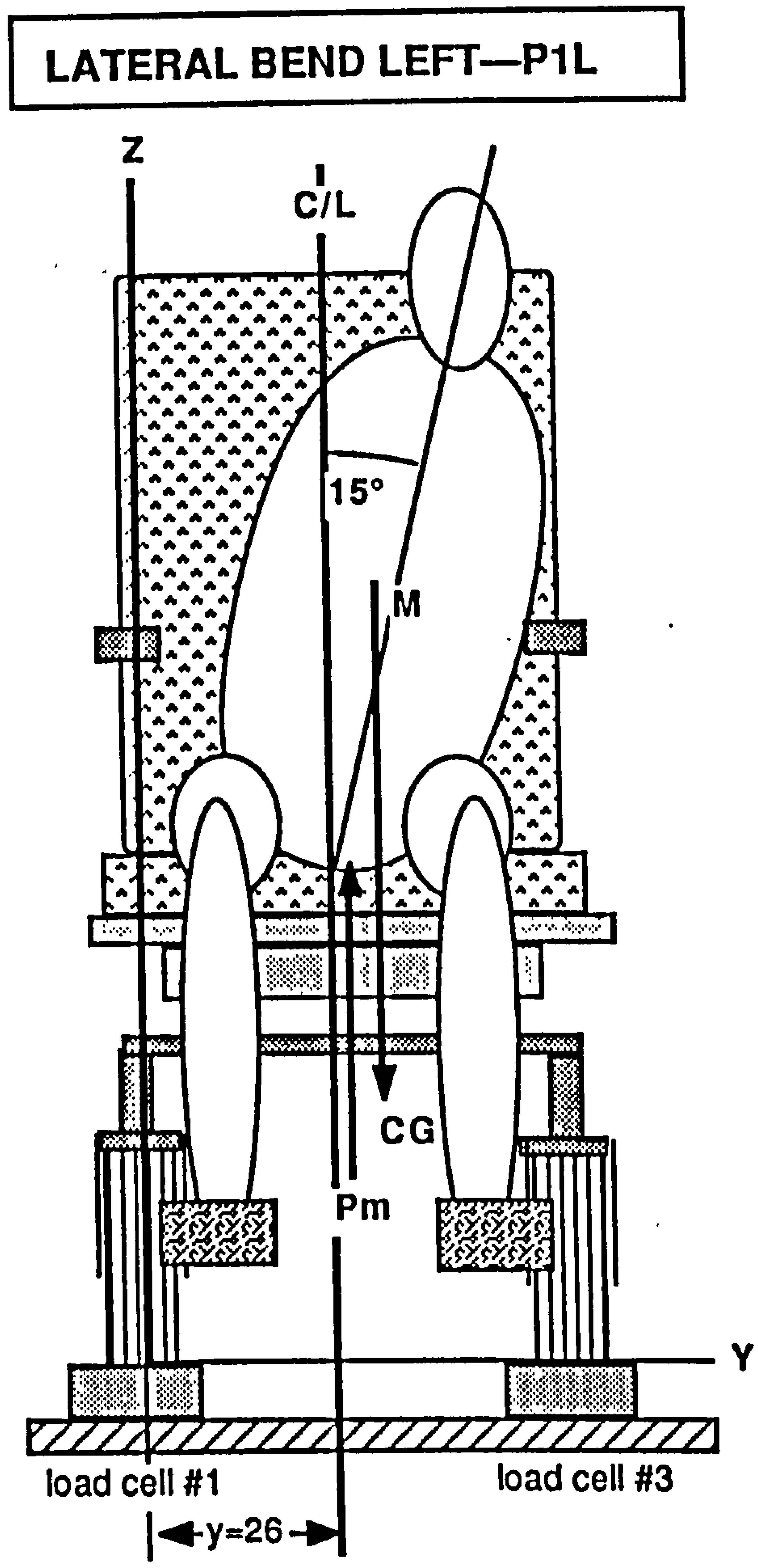
It is now of research and clinical interest to determine how the above mean values recorded in the P1M posture change with changes in seated body posture. Figures 14.2-9 contain the composite results derived from measurements taken in the remaining eight postures studied.

14.3.2. Composite Results in the Lateral Right Bending Posture (P1R)

Radiographic measurements were also made in the P1R posture, lateral trunk bending to the right of approximately 15°. The first and third columns in figure 14.2 give the mean composite values for both groups, and the second and fourth columns give the percentage change in values from the mean P1M values contained in figure 14.1. Positive values of percent change are increases and negative values decreases from P1M values. In this manner it is possible to gain an understanding as to how the mean values change as a direct result of shifts in sitting posture from the neutral position. Since the P1R posture is a trunk shift to the right from the P1M posture, only minimal differences in the sagittal plane of the spine and pelvic measurements were observed. Therefore, the values were not included in figure 14.2.

As can be anticipated lateral shifting from a neutral position does little to alter the mean average pressure (Pa) across the ischial area. However, maximum pressure (Pm) on the right side increases by about 29% for the disabled group and about 47% for the normal group. Of particular interest is the significant difference between the two groups, D=204, N=172 mmHg.

The results suggest that the location of the maximum pressure (Pm) shifts further to the right side (-17%) for the disabled group than it does for the normal group (-10%). It can be speculated that this is due to less flexibility in the spine of the disabled group which would tend to cause more lateral pelvic tilting (right obliquity) on right trunk bending. For example, a right pelvic obliquity would cause an increase in the buttock



MEAN P1L COMPOSITE VALUES

Variable	Normal Group Value	Normal Group % Chge. from P1M	Disabled Group Value	Disabled Group % Chge. from P1M
Avg. Pressure (Pa), mmHg.	72	0.2	72	7
Max. Pressure (Pm), mmHg.	185	58	258	63
Loc. of P(m), (Pm_x,Pm_y), cm.	27.7, 26.6	—	27.2, 28.2	—
Peak Gradient (Pg), mmHg mm^{-1}	2.7	57	5.1	52
Tangent. Shear (TIS), newtons	65	-3	55	-20
CG Location (X",Y"), cm.	34.1, 30.4	—	32.9, 29.4	—

Fig 14.3 Composite results in the lateral left bend posture (P1L).

pressure distribution on the right side, when compared to individuals with normal spinal/pelvic alignment and movement. This observation appears to be consistent with the measured difference in the pelvic tilt angle (IT angle L_8) of 7.9° versus 6.4° for the normal group. However the mean differences are small and upon statistical analysis have been judged insignificant.

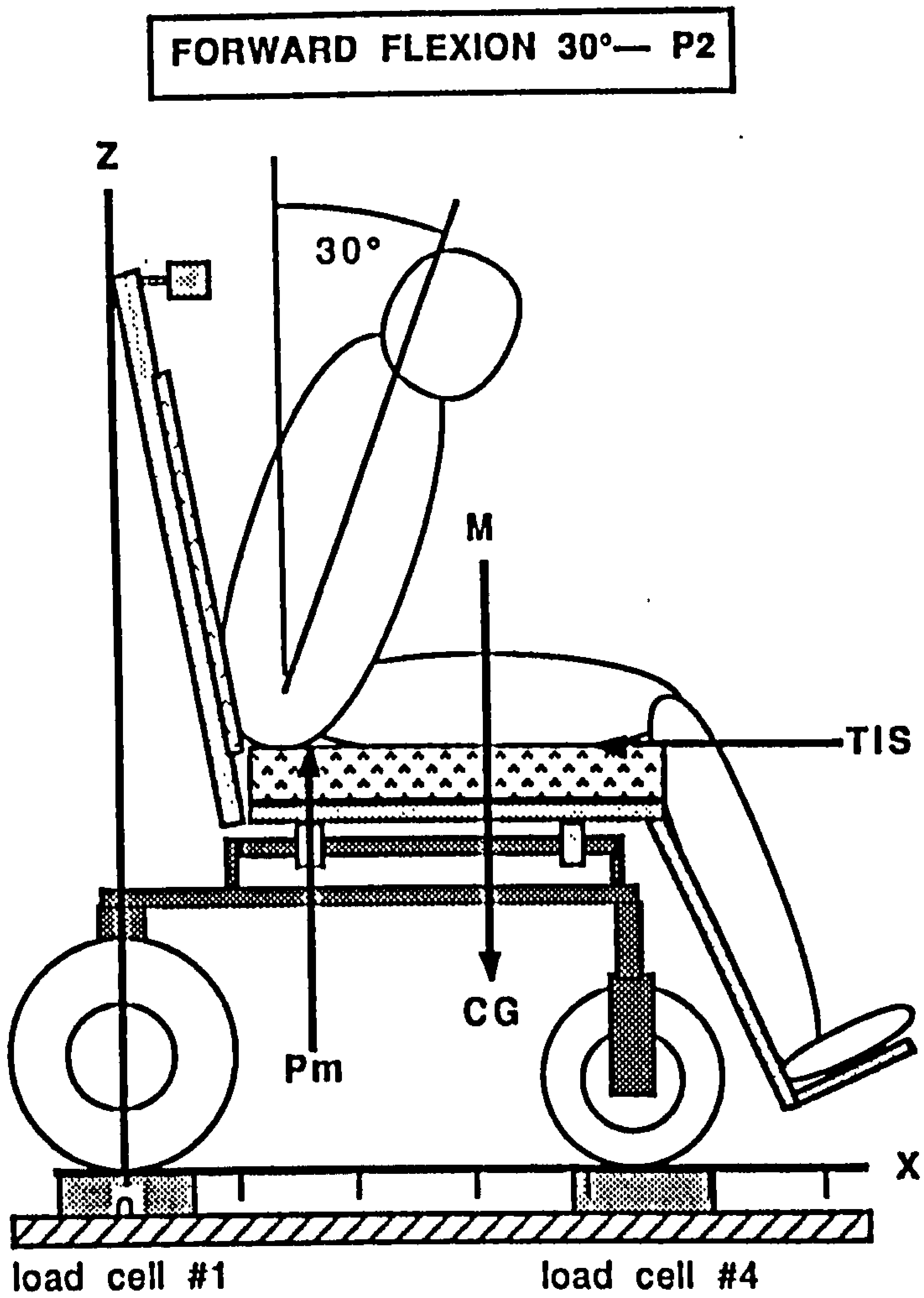
Right lateral bending causes little change in the A/P shear (10-11%). Upon 15° of lateral bending the CG for both groups falls, on average, about 1 cm lateral to the Y location (Y" = 18.4 and 18.2 cm) of the maximum pressure (Pmy = 19.9 and 19.5 cm). It can be postulated that if the location of the maximum pressure (Pm) is already located laterally due to a pelvic obliquity that lateral trunk bending to that side will elevate the Pm value due to the shift of the CG directly over or close to the Pm location. This may account, in part, for the higher mean Pm values of the disabled over the normal group on right lateral bending.

14.3.3. Composite Results in Lateral Bending Left (P1L)

The results in fig 14.3 indicate that lateral bending to the left of 15° from the neutral posture causes insignificant changes to the average pressure. Large increases in maximum pressure over P1M values occur for both groups (58% and 63%). Again, an even larger difference occurs between the groups, D=258, N=185 mmHg.

Similar left shifts in the location of the CG are recorded for both groups (20%). The mean CG line falls closer to the location of maximum pressure for the disabled group (Pmy = 28.2 cms) than it does for the normal group, Pmy = 26.6 cms. However, this difference in mean values was not found to be significant.

Probably the most important clinical finding is that upon lateral trunk bending to either the left or right side the maximum values on the contralateral side do not drop to zero. In the two groups studied, lateral bending to the right only reduced the maximum pressure on the left side to D=94 (39%), N=52 (51%) mmHg. Lateral bending to the left reduced the right side to D=107 (33%), N=62 (51%) mmHg. Again the significant differences in the contralateral values between the groups can also be noted, as well as the observation that a greater percentage reduction appears to occur in the normal group. Lateral bending of 15° is approximately the limit of lateral bending allowed by the armrests of most conventional wheelchairs, assuming that the width of the chair is appropriate for the individual.



MEAN P2 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge. from P1M	Disabled Value	Disabled % Chge. from P1M
Lumbar Angle S1-L3° (L1)	17	-30	20	-35
Lumbosacral Angle° (L5)	27	35	26	48
Pelvic Angle° (L7)	79	11	69	18
IT Location (X1), cm	24	-13	26	-20
Avg. Pressure (Pa), mmHg.	67	-8	77	-11
Max. Pressure (Pm), mmHg.	104	-10	192	22
Loc. of P(m), (Pm_x,Pm_y), cm.	25.3, 22.5	-5, 2	25.1, 23.4	-9, 0
Peak Gradient (Pg), mmHg mm^{-1}	1.5	-16	4.4	30
Tangent. Shear (TIS), newtons	32	-53	14	-80
CG Location (X",Y"), cm.	39.2, 24.4	19, 0	37.2, 24.6	16, 0

Fig 14.4 Composite results in the 30° forward flexion posture (P2)

14.3.4. Composite Results in Forward Flexion 30° (P2)

The results in figure 14.4 indicate that forward flexion of the trunk to 30° from the neutral posture (P1M) causes changes in the spinal and pelvic alignment. The lumbar angle(L1) reduces (-30% and -35%), the lumbosacral angle(L5) increases (35% and 48%), the pelvic angle(L7) increases (11% and 18%) and the ischiae move posteriorly (-13% and -20%). It is of interest that all of the percentage changes are larger for the disabled group than they are for the normal group. Of particular interest is the significant difference in pelvic angle between the groups, D=79°, N=69°, a mean difference of 10°.

Average pressure moderately reduces as the body weight is transferred to the thighs and feet (N=-8% and D=-11%). Mean maximum pressure (Pm) appears to decreases for the normal group (-10%), but surprisingly increases for the disabled group (22%). As a result the difference between the groups reaches its largest value (46%) of all the postures studied. Rather sharp decreases are seen in the peak gradients (N=-59% and D=-77%), with the difference (N=1.5 vs. D=4.4) between the groups remaining significant.

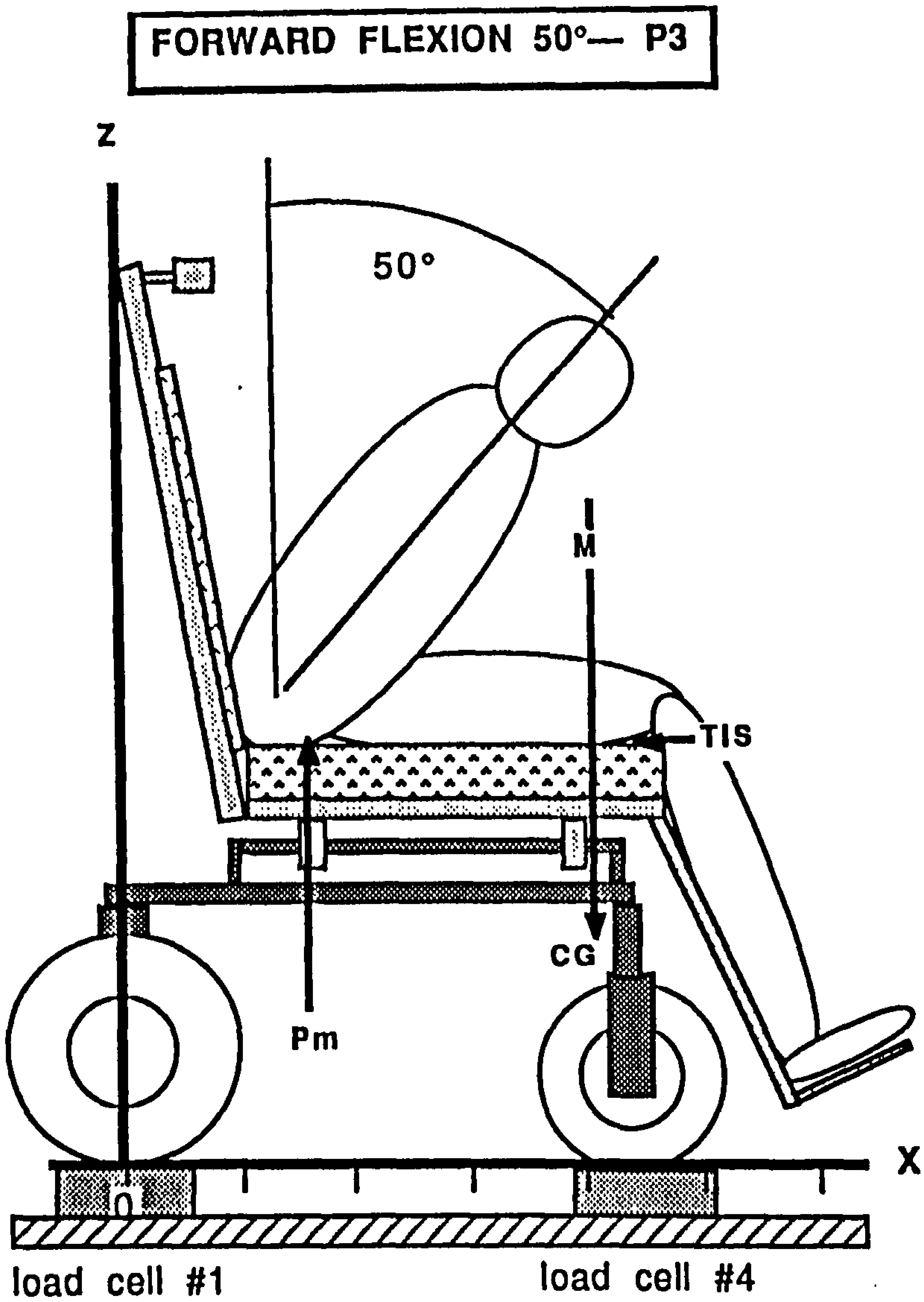
Tangential shear (TIS) decreases significantly but still not to zero. It is rationalized that the posterior movement of the lower sacrum maintains a reactive force against the backrest which is counter-balanced by the TIS force. Visual observation during measurements further supports this rational.

The "X" location of the mean CG line shifts forward approximately 19% for the normal and 16% for the disabled, with little or no change in the Y" values.

14.3.5. Composite Results in Forward Flexion 50° (P3)

The results in figure 14.5 indicate that forward flexion to 50° from the neutral P1M posture causes further reductions in most of the values. Average pressure decreases by N=-17% and D=-15% as more upper body weight is shifted to the posterior thighs and footrests. Maximum pressure also decreases, but more in the normal group (-28% versus -9%). The difference in maximum values between groups remains large (41%). Location of maximum pressure moves posteriorly (-8% and - 10%), most likely caused by still further posterior movement of the ischial tuberosities.

Marked reductions now occur in the TIS force with a reversal of direction taking place in both groups (D=133%, N=102%). This suggests that the sacrum has moved away from the backrest causing the tangential shear force to approach zero. The reversal of 33% by the disabled group is a result of the reactive force from the armrests



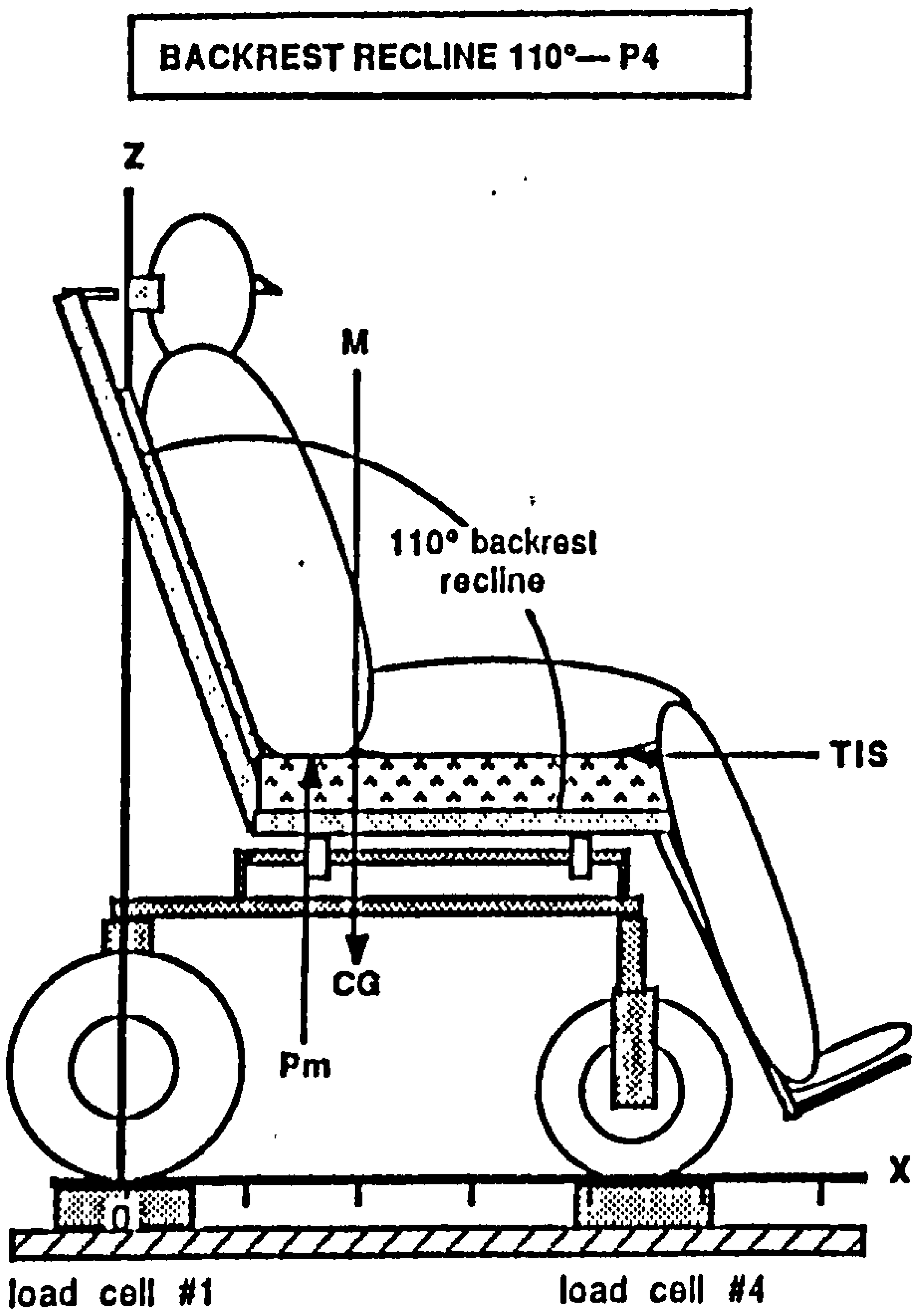
MEAN P3 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge. from P1M	Disabled Value	Disabled % Chge. from P1M
Avg. Pressure (Pa), mmHg.	61	-17	60	-15
Max. Pressure (Pm), mmHg.	84	-28	143	-9
Loc. of P(m), (Pm_x,Pm_y), cm.	24.7, 22.8	-8, 3	24.8, 21.8	-10,-7
Peak Gradient (Pg), mmHg mm^{-1}	1.1	-39	3.7	9
Tangent. Shear (TIS), newtons	-1	-102	-23	-133
CG Location (X",Y"), cm.	44.2, 24.1	34,-1	43.2, 24.3	33, -1

Fig 14.5 Composite results in the 50° forward flexion posture (P3).

as this group required more external trunk support in order to maintain the 50° flexed trunk posture.

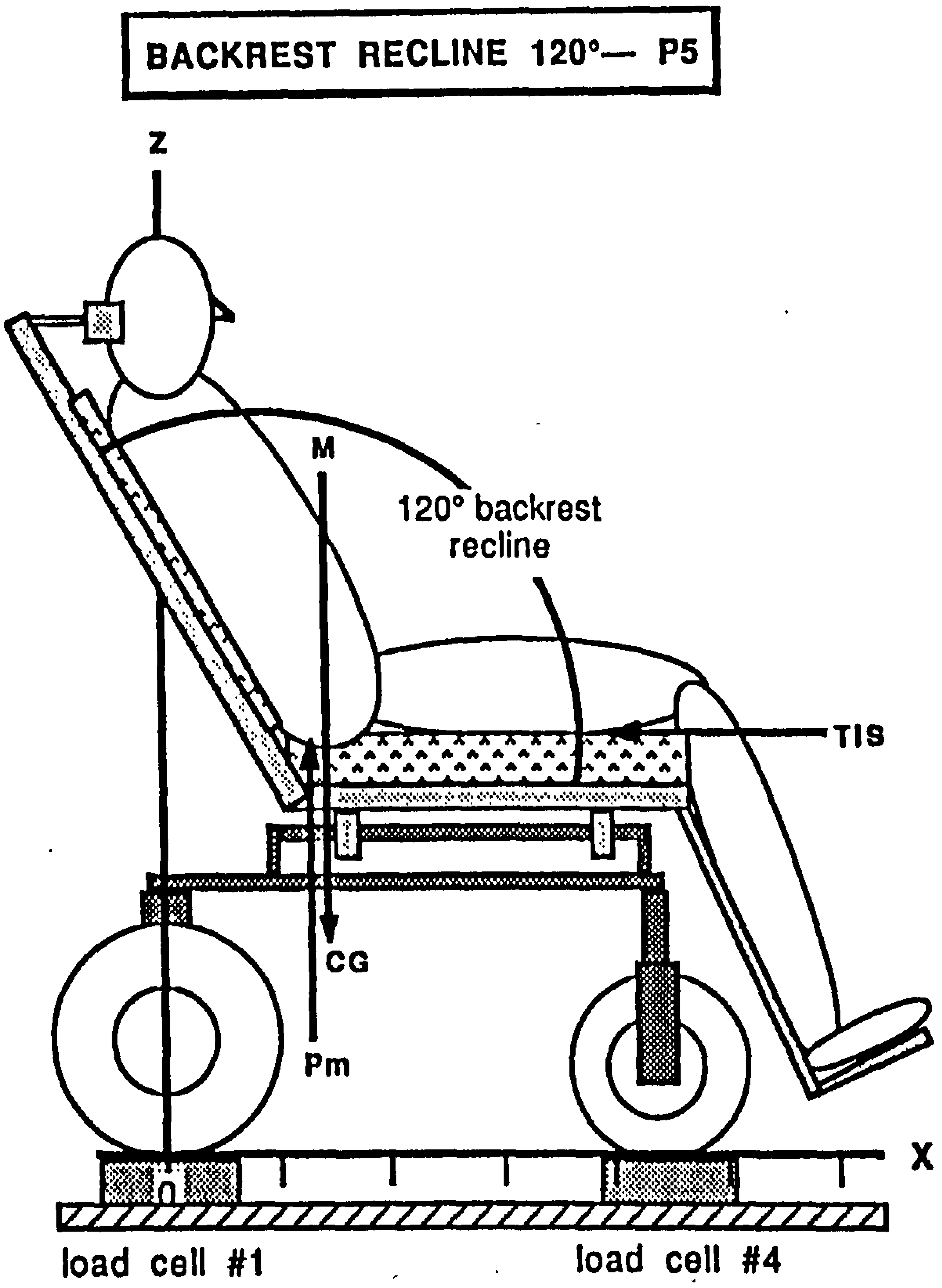
The mean X" value of the CG line is now located about 20 cm anterior to the location of maximum pressure, which is about the same for both groups.



MEAN P4 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge.	Disabled Value	Disabled % Chge.
Avg. Pressure (Pa), mmHg.	84	15	80	14
Max. Pressure (Pm), mmHg.	124	7	158	0
Loc. of P(m), (Pm_x,Pm_y), cm.	20.7, 24.7	-6, 11	26.6, 22.6	-3, -4
Peak Gradient (Pg), mmHg mm^{-1}	1.6	-6	3.5	3
Tangent. Shear (TIS), newtons	72	8	79	14
CG Location (X",Y"), cm.	27.2, 24.3	-18, 0	25.7, 24.8	-21, 1

Fig 14.6 Composite results in the 110° backrest recline posture P4.



MEAN P5 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge. from P1M	Disabled Value	Disabled % Chge. from P1M
Avg. Pressure (Pa), mmHg.	82	12	71	1
Max. Pressure (Pm), mmHg.	131	13	140	-12
Loc. of P(m), (Pm_x,Pm_y), cm.	20.7, 23.8	-23, 8	21.5, 22.5	-22, -4
Peak Gradient (Pg), mmHg mm^{-1}	1.4	-17	2.8	-18
Tangent. Shear (TIS), newtons	79	17	89	28
CG Location (X",Y"), cm.	22.2, 24.4	-33, 0	20.5, 24.7	-37,1

Fig 14.7 Composite results in posture P5.

14.3.6. Composite Results in Backrest Recline 110° (P4)

The results in figure 14.6 indicate that reclining the backrest to 110° from the P1M position causes several unanticipated changes in the interface parameters. Average pressure distribution (Pa) increases approximately 15% for both groups, whereas the maximum pressure (Pm) remains about the same. The Pmx location of the maximum pressure does displace slightly posteriorly (-6% and -3%), while maintaining an apparent difference in the Pmx values of about 6 cms (26.6 - 20.7 cm). However these differences are assumed to be insignificant.

The peak pressure gradients (Pg) remain significantly higher for the disabled group (3.5 vs. 1.6 mmHg mm^{-1}).

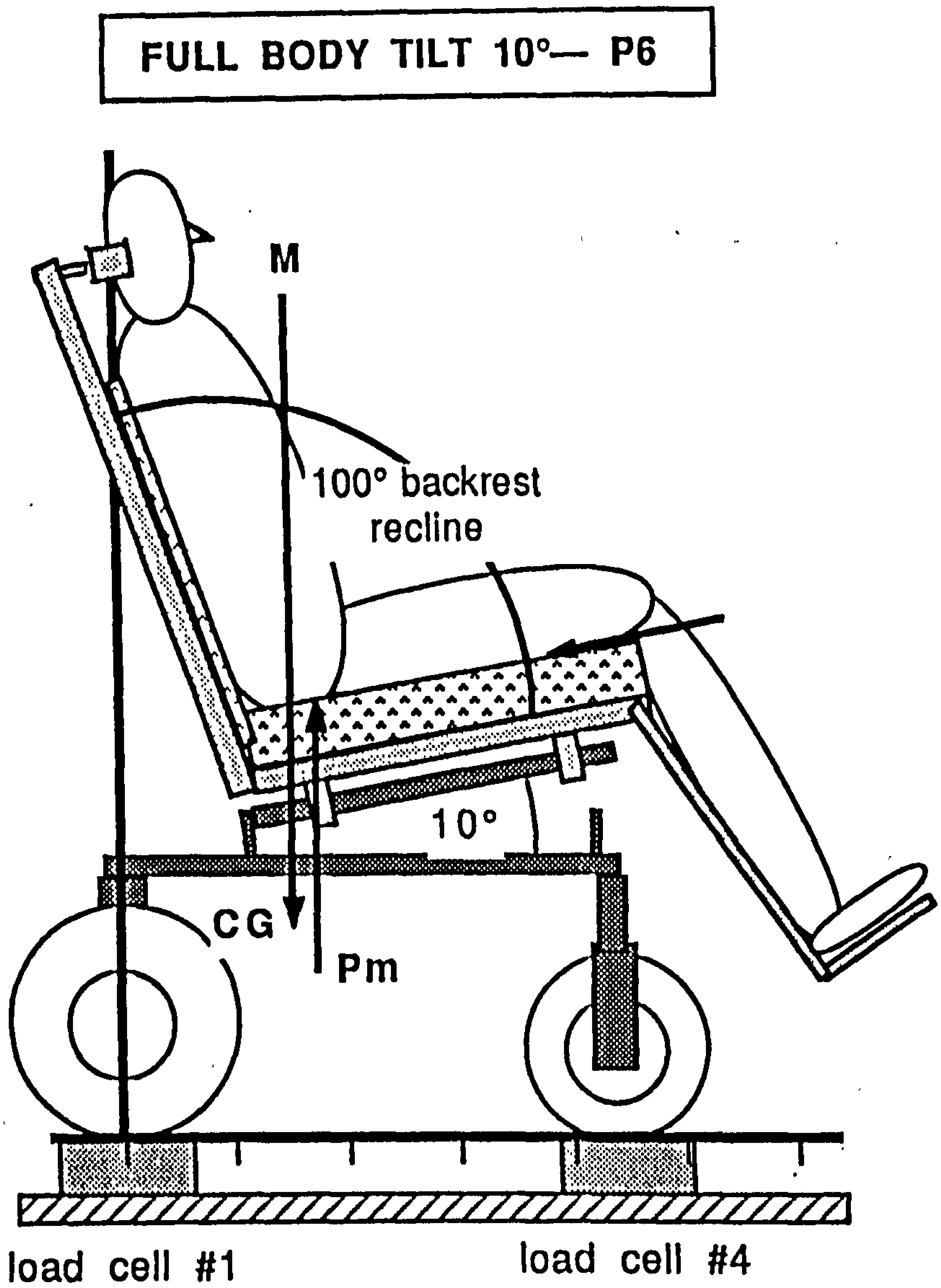
As anticipated, the CG line shifts posteriorly (X" = -18% and -21%) while remaining approximately on the same mid-line as P1M (Y'=24.3, and 24.8 cm). It can be noted that the mean X" value of CG now falls 7.5 cm (27.2 - 20.7) anterior to the Pmx location for the normal group, whereas the CG falls about 1 cm (25.7 - 26.6) anterior to the location of the maximum pressure Pmx for the disabled group. This could explain the reason for the measured increase in both average and maximum pressures when a decrease was expected as a result of the partial shifting of the trunk weight to the backrest. A slightly lower increase in the disabled group is consistent with the fact that a larger percent of their body weight is transferred to the backrest due to the relative difference in the distribution of their body weight compared to the normal subjects.

14.3.7. Composite Results in Backrest Recline 120° (P5)

The results in figure 14.7 indicate that further reclining of the backrest to 120° causes continued changes in most of the interface parameters. Average pressure remains the same or possibly continues to increase slightly (N=12% and D=1%). Again, the relative difference in mass distribution most probably accounts for the difference between the groups.

Surprisingly, maximum pressure appears to continue to rise for the normal group (13%) and decrease for the disabled group (-12%). Statistically these changes are marginally insignificant, suggesting that no real change has occurred from the neutral P1M values.

Location of the CG continues to move posteriorly approximately (22% and 23%). Insignificant changes occur in the peak pressure gradients, with the values for the disabled group remaining about twice those of the normal group (2.8 versus 1.4 mmHg mm^{-1}).



MEAN P6 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge. from P1M	Disabled Value	Disabled % Chge. from P1M
Avg. Pressure (Pa), mmHg.	72	-1	67	-3
Max. Pressure (Pm), mmHg.	115	-1	161	2.3
Loc. of P(m), (Pm_x,Pm_y), cm.	26.5, 22.5	1, 2	27.0, 21.5	0, -8
Peak Gradient (Pg), mmHg. mm^{-1}	1.7	-3	3.5	5
Tangent. Shear (TIS), newtons	22	-67	48	-31
CG Location (X",Y"), cm.	21.7, 24.3	-34, 0	19.9, 24.8	-39, 1

Fig 14.8 Composite results in the P6 posture.

As anticipated, posture P5 causes marked increases in TIS values, with increases in the disabled group being about 1.5 times those in the normal group (N=17%, D=28%).

Expectedly, the mean CG line moves posteriorly approximately 35% from the P1M position (N=-33%, D=-37%). It can be noted that in the P5 posture the CG line appears to fall 1.5 cm anterior to the location of the maximum pressure (Pmx) for the normal group, and 1 cm posterior to the Pmx for the disabled group. However, this difference (2.5cm) is small and was found to be insignificant.

14.3.8. Composite Results in Full Body Tilt 10° (P6)

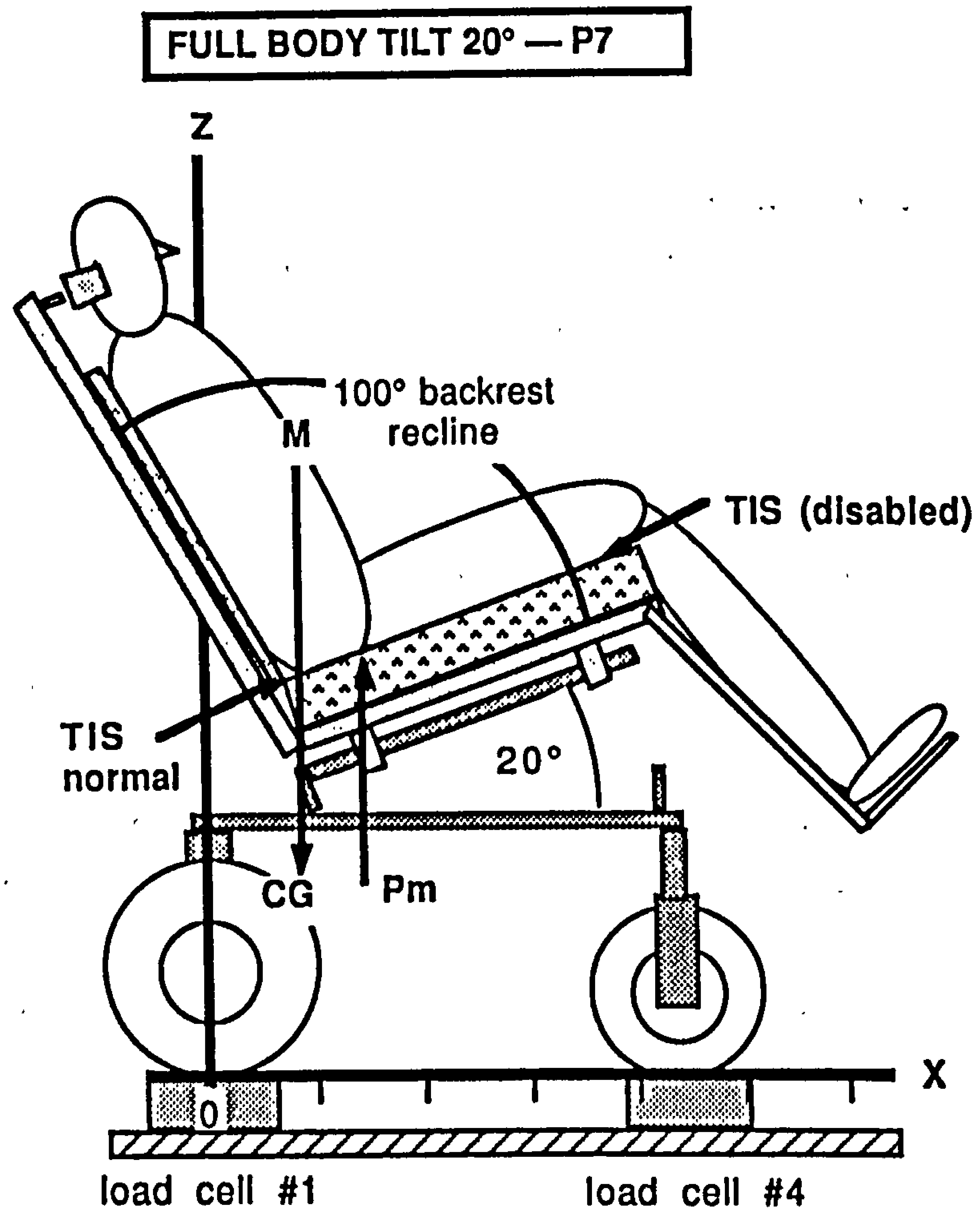
By definition, a full body tilt of 10° implies a rotation of the complete body in space 10° while seated in the neutral P1M configuration. Biomechanically, this causes changes to the interface parameters that appear different from those that occur due to reclination of the backrest (P4 and P5).

The results in figure 14.8 indicate that insignificant changes from P1M values occur in average and maximum pressures and peak pressure gradients within the samples. The average pressure distribution remains approximately the same with insignificant differences between the samples (N=72, D=67 mmHg). In contrast, the mean maximum pressure (Pm) for the SCI group remains 28% higher than the normal group (161 vs. 115 mmHg). The location of the maximum pressure (Pmx) undergoes minimal change in both groups. The peak pressure gradients Pg remain approximately 2 to 1 disabled over normal (1.7 versus 3.5 mmHg mm^{-1}).

Expectedly, significant decreases in tangential shear values (TIS) are recorded (N=-67%, D=-31%), while retaining an apparent 2 to 1 ratio of disabled over normal (N=1, D=2.2). The mean CG line moves posteriorly about 35% (N=-34%, D=-39%), resulting in a CG line that is about 5cm posterior to the average location of the maximum pressure (Pmx), (N=4.8 cms, D=6.1 cms).

14.3.9. Composite Results in Full Body Tilt 20° (P7)

The results in figure 14.9 indicate that further tilting of the whole body to 20° in space from the neutral P1M posture increases the trends observed in the P6 position. Significant reductions in both average and maximum pressures now occur in both groups of between 10 and 14 percent, as more trunk weight is shifted to the backrest. Insignificant differences occur in the location of the maximum pressure (< 3%). The peak pressure gradients (Pg) remain approximately the same as in the P1M posture, with the disabled values being approximately 1.5 times those of the normal group (3.0 versus 1.8 mmHg mm^{-1}).



MEAN P7 COMPOSITE VALUES

Variable	Normal Value	Normal % Chge. from P1M	Disabled Value	Disabled % Chge. from P1M
Avg. Pressure (Pa), mmHg.	66	-10	63	-10
Max. Pressure (Pm), mmHg.	100	-14	140	-11
Loc. of P(m), (Pm_x,Pm_y), cm.	25.4, 22.5	1, 2	25.0, 22.9	-3, -3
Peak Gradient (Pg), mmHg mm^{-1}	1.8	4	3.0	-4
Tangent. Shear (TIS), newtons	-6	-108	10	-85
CG Location (X",Y"), cm.	11.1, 24.3	-66, 0	11.0, 24.8	-66, 1

Fig 14.9 Composite results in P7 posture.

Expected significant reductions occur in the tangential shear force (TIS) (N=-108%, D=-85%) with reversal of direction possibly taking place in the normal group.

The mean CG line now falls far posterior to the X location of maximum pressure (N=14.3 cm, D=14 cm).

Tilting the whole body in space is more effective in shifting the body weight posteriorly than is reclining the backrest. Tilting results in a proportionately larger shift of body weight to the backrest, thereby causing the reductions in maximum pressures observed in positions P6 and P7. Also, the posterior displacement of the CG line is greater in full body tilting thereby maximizing the distance between the location of maximum pressure and the line of gravity.

Reductions of TIS forces in the disabled group to near zero at 20° of body tilt (P7) is dramatically opposite to the increase in TIS values to their maximum value in the 120° backrest recline position (P5). It is evident that if it is indeed important to minimize tangential shear force at the seat-body interface full body tilting must replace the current practice of backrest reclining.

CHAPTER 15. STUDY CONCLUSIONS

The following conclusions are drawn from both the analysis of the individual study components (Chapters 10 - 13) and from the composite results in the preceding chapter (Chapter 14). For purposes of brevity only the more salient findings will be presented in itemized format. No implication of importance is intended by the ranking. All values stated are mean computed values derived from either the normal or disabled samples studied.

1) In general, a person with a spinal cord injury (SCI) will sit in their neutral posture with a posteriorly tilted pelvis. On average, the pelvis will be tilted 15° more than for able-bodied individuals.

2) Forward trunk flexion to 30° causes forward rotation of the pelvis, with angular motion also taking place mainly at the lumbo-sacral joint, SEP-HOR (L5). The mean pelvic angles (L7) change from N=71°, D=56° in the neutral posture to N=79°, D=68° in the flexed posture. Minimal motion takes place at the sacroiliac joint (L6).

3) Sitting in a neutral posture with 100° backrest recline causes the lumbar spine, as defined by spinal segments S1-L3, to assume a lordodic angle of about 26° for people with SCI and about 22° for the able-bodied. The difference of between the groups (4°) was found to be insignificant.

4) A posterior pelvic tilt causes the ischial tuberosities of the person with a SCI to be displaced anteriorly. On average, the ischial tuberosities are displaced 4 cm anterior to those of individuals without a spinal injury.

5) Forward flexion of the trunk to 30° from a neutral upright posture causes the ischial tuberosities to move posteriorly. On average, the posterior displacement is 2.7 cms for individuals with a SCI, and only 1.6 cm for individuals without a spinal injury, an average difference of about 1cm. This small difference tested to be insignificant.

6) Average pressure distribution between populations is effected insignificantly by alterations of sitting posture. Forward flexion of the trunk causes the largest decrease within both groups from the average pressures in the neutral position. On average, a reduction of approximately 15% from the values in the neutral position occurs at 50° of trunk flexion (P3).

7) Mean maximum pressures of individuals with a SCI are significantly higher than able-bodied individuals in all nine sitting postures studied. In the neutral posture, on average, the maximum pressure for a person with a SCI is 26% higher. The differences range from a low of 6% in the P5 position to a high of 46% in the P2 position.

8) The maximum pressures can be reduced by postural changes from those values occurring in the neutral P1M posture. These reductions are; lateral trunk bending 15° (32-38%), forward flexion to 50° (-9%), backrest recline to 120° (-12%), and full body tilt to 20° (-11%).

9) Alterations to sitting postures causes similar changes to the A/P location of the maximum pressure (Pmx) for both disabled and able-bodied. With respect to the neutral posture, forward trunk flexion of 30° and 50° results, on the average, in a posterior shift of 2.4 and 2.7 cm, respectively. The largest posterior shift of 6 cm occurs in the backrest recline 120° (P5) posture.

10) On average, individuals with spinal cord injury have peak pressure gradients that are 1.5 to 2.5 larger than non-injured people. Among the postures studied maximum reduction of the peak gradients from those measured in the neutral posture occur after backrest recline to 120° (P5) (-8%).

11) Tangential shear force (TIS) acts at the body-seat surface in all nine postures studied. With respect to the neutral posture, maximum reductions of TIS force occur upon trunk flexion of 50° (133% or a reversal of 33%) and upon full body tilt to 20° (-85%). It is reasonable to assume that full body tilt beyond 20° will also reduce TIS to zero, and thereafter cause a reversal and an increase in TIS force in the opposite direction. The manner in which the TIS interacts with the normally induced shear stresses to produce potentially deleterious effects in the supporting tissues has not been clarified by this work.

12) In general, the centre of gravity of individuals with a SCI is displaced further posteriorly when compared to a normal sample. The difference in posterior displacement is dependent on the posture. On average, the mean difference ranges from an insignificant amount in the P7 posture to a maximum of 1.8 cms in the P6 posture. Alteration of posture from the maximum forward flexed position of 50°(P3) to full body tilt of 20°(P7) causes the CG to move posteriorly a total distance of about 30 cm, which is approximately the same for both groups.

13) The relative location of the maximum pressure (Pmx) and the location of the CG line exhibit major differences as a result of postural changes. The exact relationship between these two variables, and how they may effect the physiological responses in the supportive tissues remains unknown.

CHAPTER 16. DISCUSSION OF SUMMARY CONCLUSIONS

16.1 SPINAL AND PELVIC ALIGNMENT

When sitting mid-line in a wheelchair with a seat to back angle of approximately 100°, the typical posture assumed by individuals with spinal cord injury (SCI), as represented by the study sample, on average, will be different from able-bodied individuals. The pelvis will be rotated further posteriorly by about 15°, with the ischial tuberosities displaced further forward by about 4 cm. These findings are in support of the clinical observations reported by Zarkarkow (1984) and Hobson (1984). Surprisingly, the results indicate a slightly larger lordodic angle (S1-L3) in the SCI group (26°) than in the normal group (22°). These values compare favorably with Andersson et al (1979) who reported a total lumbar angle of 25° with backrest inclination of 100°. However, these findings do not support the observations of Lancourt (1981) and Pope (1985) who reported flattening of the lumbar spine in their disabled subjects.

These results suggest that the kyphotic sitting deformity observed in the spinal injured population occurs mainly above the lumbar level, i.e., in the thoracolumbar or thoracic spine with compensation in the cervical spine. This finding has implications as to the height of a dynamic backrest or lumbar pads in wheelchair seats.

A postural change of trunk flexion to 30° causes significant changes to both the angle of the pelvis (P7) and the location of the ischial tuberosities relative to the support surface (X1). The new pelvic angle increases from 71° to 79° in normals and increase from 56° to 68° in spinal injured people; a relative increase of about 8° versus 12°. Most of the pelvic angle change is due to rotation of the pelvis as measured by the change in the lumbosacral angle (L5). This rotation of the pelvis can cause the ischiae to slide posteriorly about 2.6 cm for paralytic individuals and about 1.6 cm for able bodied individuals.

Forward trunk flexion to 30° is the range that would be typical when a person is actively propelling a wheelchair, working at a desk, or undertaking many manual tasks. Since many individuals conceivably spend considerable periods of time in the flexed trunk positions, pressure profiles and other interface variables should be evaluated in these postures in order to identify conditions of increased risk. These findings have implications regarding the location and size of cushion cut-outs for pressure relief, as well as upon the drawing of conclusions regarding cushion typography from data derived from able-bodied subjects.

The radiographical results also indicate that, on average, individuals with a spinal cord injury will have lateral spinal/pelvic deformities that displace the pelvis laterally from the mid-line. The laterally displaced pelvis will most likely also have a pelvic obliquity angle of about 1.5°. Lateral trunk bending from the neutral posture will have different effects on pelvic displacement. The displacement will be dependent on the direction and magnitude of the pelvic deformity in the frontal plane. Lateral trunk bending of 15° to the right side produces a mean vertical movement of the left ischial tuberosity of about 1.5 cm, with the difference between the two populations being insignificant. Lateral trunk bending of 15° also causes a lateral displacement of the ischiae in the opposite direction, a distance of about 1.5 cm.

The importance of these latter findings is that the location of the pelvis, and in particular the critical bony areas are not static. That is, the bony prominences are constantly moving with every change in sitting posture. Spinal/pelvic deformities are present in varying degrees in the SCI population that can cause an asymmetrical alignment of the pelvis with respect to the seat surface in the frontal plane. Although not studied, a reasonable assumption is that the greater the lateral pelvic deformity (obliquity) the greater will be the asymmetrical distribution of the weight bearing forces. This assumption is consistent with the findings of Drummond, et al (1985) who demonstrated asymmetrical loading of the ischiae in 16 paralytic subjects.

16.2. PRESSURE DISTRIBUTION

16.2.1. Average Pressure Distribution

The average pressure distribution across the critical buttock area varies little between normals and individuals with paralytic injuries. Furthermore, typical alterations in postures assumed while sitting in a wheelchair have minimal effect on the average pressure distribution. Forward flexion of the trunk to 30° is required to obtain a reduction of about 15% from the average pressure values in the neutral posture.

16.2.2. Maximum Pressure Distribution

People with a spinal cord injury have maximum pressures that are significantly higher than able-bodied individuals in most postures typically assumed in a wheelchair. The differences in mean maximum pressure range from a minimum of 6% in the backrest recline 120° posture to a maximum of 45% in the forward flexed posture 30° (P2). The difference in the neutral posture (P1M) is about 25%. It is concluded that these inherent differences in mean maximum pressures are largely due to asymmetrical ischial loading caused by spinal/pelvic deformities and atrophy of tissue over the ischiae.

Changes in posture can effectively reduce the maximum pressures of SCI people from those values that are present in their neutral sitting posture. On average, recline of the backrest to 120° reduces the neutral values by 12% as does the full body tilt to 20° (11%). Forward trunk flexion to 50° yields a reduction of 9%. Lateral bending to 15° reduces the neutral values by 30-40%.

In contrast to Drummond et al, (1985), who found a posterior redistribution of maximum pressures in a paralytic study group, these findings indicate a slight anterior displacement of the location of maximum pressure in all nine postures (fig 11.5). However, the Drummond study only analyzed pressures in one posture in a predominantly pre-adolescent group, so direct comparisons may not be appropriate. Anterior displacement of the maximum pressure is in keeping with the measured anterior displacement of the ischiae.

The implications of these findings are possibly more important from a research perspective. They suggest that caution must be exercised when using normal subjects and absolute pressure values to make judgements about the pressure distribution characteristics of commercial cushions or generic seating materials. Also, shape and pressure contour derived from normal subjects can yield profiles that are misleading regarding the characteristics of paralytic buttock contours.

16.2.3. Peak Pressure Gradients

A peak pressure gradient has been defined as the maximum gradient between any two adjacent pressure recording cells. The definition of peak gradient in this study (vide 11.3.3, page 108) results in peak gradients being identified in association with peak pressure measurements: the values of which correspond to the difference between peak pressure and the lowest pressure measured at an adjacent measurement point. Therefore, the peak gradients are also inherently higher for the paralytic group in all sitting postures (fig 11.6). On average, peak gradients are higher by 1.5 to 2.5 times depending on the assumed posture. These values can be reduced through reduction of maximum pressures through alterations of posture discussed above (16.2.2).

The implications from these findings are several. First, shear stress theory implies that the higher the force gradient between two points the higher the induced shear stress. The body-support interface is a complex arrangement of bony prominences, overlying tissue, clothes, transducer, cushion cover, and supporting materials which are all subjected to both normal and shear loading. Therefore, it is very difficult to make accurate determinations as to the exact nature of the shear stresses in the tissues. However, it seems reasonable to assume that if high pressure gradients are recorded by the transducer matrix, these gradients should, in some manner, be related to elevated shear stresses in the supporting tissues.

Secondly, if shear stress acting in combination with high normal loads presents a critical combination of events as proposed by Bennett et al (1979) and Chow and Odell, (1978) then knowledge of cushion materials and postures which minimizes these occurrences is vitally important. It follows that a simple measurement of maximum pressure in one static posture may not yield the information necessary to minimize the critical combination of normal pressure and shear interacting at a pelvic location under high risk of ulceration.

16.3. TANGENTIALLY- INDUCED SHEAR FORCE (TIS)

The TIS has been defined as the force acting parallel to the seat surface that is required to maintain A/P equilibrium. It can be thought of as the friction or shear force acting at the seat surface that prevents the pelvis from sliding forward. It varies with body posture and is largely dependent on the horizontal reaction force applied to the body by the backrest. Horizontal reaction forces at the footrest can also affect the tangential shear values.

It is evident that the TIS force exists in all sitting postures. Similar TIS values exist for both normal and disabled individuals in most postures. The highest values occur when the backrest is reclined. With respect to the neutral posture, substantial reductions of TIS force occur due to changes in posture. Largest reductions occur when the trunk if flexed forward 50° (-133%) and upon full body tilt to 20° (-85%).

Identification of only the TIS component of shear acknowledgedly yields only partial insight into the complex shear stress phenomenon. Since the seat surface on the BPC, or any supporting cushion used in practice, is not hard and flat but soft and contoured, components of the measured horizontal reaction force will not be acting precisely tangential to the buttock surface. However, given this limitation to the approach the comparative values recorded do exhibit changes that are in keeping with the empirical biomechanical rational. If this relatively simple measurement provides useful information, then an enhanced clinical practice may result. It is proposed that the TIS measurements will be no less accurate than the pressure measurements currently being obtained clinically.

16.4. CENTRE OF GRAVITY

A useful confirmation is the finding that the CG location in the A/P plane of individuals with lower limb wasting and well developed torsos shifts posteriorly compared to values derived from the able-bodied population. The average mean difference in CG location is about 1.5 cm, a value which varies with body posture. The full range of body movement in the sagittal plane from forward flexion of 30° to full body recline of 20° causes the location of the line of the CG to traverse a mean distance of about 30 cm.

A possibly more important finding is the wide variation exhibited by SCI individuals. Examination of figs 13.11-14 clearly indicates a wide spread between individual values when compared to the normal data points. This suggests that mean values of CG location may not be very useful clinically and that determinations will need to done on an individual basis. The magnitude of the CG shift throughout the postures studied also suggests that CG determinations be done with specific functional postures in mind.

16.5. INTER-RELATIONSHIPS BETWEEN VARIABLES

This study and specifically this chapter has just touched upon the complex task of explaining how the extrinsic factors of pressure, tangential shear, centre of gravity, deformity, and posture interact to create conditions of increased stresses in supporting tissues. Based on the findings of this study and relevant preceeding work, it is postulated that all of the above factors, to varying degrees, affect the events occurring at the body-seat interface and in turn the underlying tissues. Further research efforts should more thoroughly examine these inter-relationships in an attempt to better quantify both the nature and rank order of importance of composite effects on supporting tissue. One can speculate that a composite indicator scale may emerge that would provide vastly improved guidelines for clinical decision-making. This scale could ultimately provide quantitative information far beyond the information currently available in clinical settings; which is largely limited to gross indications of maximum pressure values in one static posture.

16.6. DISCUSSION OF HYPOTHESES

The study hypotheses were initially stated in terms of two null hypotheses (Ho);

1) That sitting posture has no effect on the seat/body interface variables, and

2) That individuals with spinal cord injuries have body/seat variables that are no different from the able-bodied population.

Based on the evidence presented above and at the declared significance level of 0.05, it seems reasonable to reject both null hypotheses. That is, maximum pressure, peak pressure gradients, tangentially-induced shear and location of maximum pressures *are significantly effected* by alterations in sitting postures. Secondly, it has been shown that the above variables, in addition to spinal/pelvic alignment, *do exhibit significant differences* between the two populations studied.

PART III. IMPLICATIONS FOR FUTURE RESEARCH AND CLINICAL PRACTICE

CHAPTER 17. LIMITATIONS OF FINDINGS

The estimated errors associated with measurement of each of the variables has been discussed in each of the relevant sections in Chapters 5-13. Theoretical estimations as to how closely the study sample groups are representative of their general populations is embedded in the statistical analysis. A level of 0.05 was used for declaring any findings significant. However, this till leaves some uncertainties, ie. a 5% probability that significance was declared in error (type I). Extrapolation of these results to populations of disabled people that fall outside the population inferred by the study sample can only be done with extreme caution. A description of the characteristics of the study samples and their inferred populations is given in Chapter 7. The samples sizes of 10 and 12 people are small and in that the lower end of reliability of Gaussian and central limit statistical theory. The greatest confidence should be placed in the trends observed in the results rather than on the actual values themselves. For example, the finding that maximum pressure and peak pressure gradients are higher than normal values for most sitting postures, is a more reliable statement than saying disabled values are 46% higher than the normal values in 30° forward trunk flexion.

Since comparisons between the two sample groups have formed the basis for measurement and the data analysis approach, this places a fairly narrow window on how far generalizations can be made. Although statistical theory gives guidance in this regard it is no substitute for common sense. For example, extrapolation of spinal/pelvic and possibly tangential shear data to post-polio populations or some elderly groups may be appropriate. However, use of pressure and CG data would probably be inappropriate because of the differences in body mass distributions and tissue wasting associated with the study sample.

All the pressure, TIS and radiographic readings were taken on one type of cushion material. The viscoelastic type used (3 layers x 25 cm thick) may be representative of several materials in common use. In as much as the approach provides consistency across subjects from the research perspective, it does place limitations on the generalization of findings to other types of support surfaces. For example, the outcomes from several aspects of the study may have been quite different had a balloon-type cushion been chosen instead of the viscoelastic foam.

And finally, although every effort has been made to correctly analyze and present the voluminous data, the probability exists that errors have been made. In this event there appears to be no solution but to offer an apology in advance.

CHAPTER 18. IMPLICATIONS FOR CHANGES TO CLINICAL AND RESEARCH PRACTICES

18.1. CLINICAL PRACTICE

Progressive management of the pressure sore problem is much broader than the selection and application of appropriate cushions and related mobility devices. It is the implementation of a philosophy of management that begins with the staff in the primary care facility, continues throughout the post-acute phase and eventually permeates the life of a person who has insensitive weight-bearing tissues. This management philosophy espouses the need for pressure relief regimens, personal hygiene, adequate nutrition, and mainly individual responsibility and common sense in the pursuit of activities of daily living. Within this broader management approach the progressive rehabilitation team has relatively small window of opportunity to assist in the selection of appropriate pressure relieving and other devices, and to impart practical concepts of self care. It is within this narrow window that the following suggestions are made for improvement to clinical practice.

As outlined in Chapter 4 clinical facilities are not, in general, applying advanced concepts of pressure management. Furthermore, decision making is often carried out in a rather haphazard and unstructured manner. The following concepts are intended to complement existing practise of pressure management, which is presently based largely on minimizing maximum pressure, usually measured in one static sitting posture.

18.1.1. Enhancement of Pressure Monitoring

If one accepts the research evidence that shear stress is a deleterious interface factor, second only to normal pressure, then clinical monitoring of pressure should be expanded to gain information on peak pressure gradients. Measurement tools exist commercially that readily allow monitoring of pressures at multiple sites across the buttock. Computation of pressure gradients is simple. For example, knowledge of the location and magnitude of the gradients can assist in delineating between different cushions options, assuming all other factors being approximately equal.

It is evident that most active people in wheelchairs do not remain in one static posture (Bardsley, 1983; Patterson, 1984). Results of this study indicate that changes in posture have a direct bearing on maximum pressure, pressure gradients, and pressure distribution across the supporting tissues. Evaluation practices should be adopted which can determine an estimate of the frequency of specific postures assumed by individuals. Then, pressure assessments should be done in these postures. If possible, conditions of dynamic loading should also be determined during the evaluation process. It may not be possible to reduce the dynamic loads to safe thresholds, but it can have the benefit of making the person aware of daily situations that create peak loads on the tissues.

18.1.2. Management of Deformity

It is evident from this work and other studies that asymmetrical alignment of the pelvis, particularly in the frontal plane, can cause elevated interface pressure and indirectly shear stress. To date little emphasis has been given to preventing deformities of the spine and pelvis in the spinal cord-injured population. The conventional wheelchair seat and the interface materials used to distribute pressure have done more to foster deformity than to prevent it. Assessment of cushion materials and wheelchair designs should include a critical analysis of their capabilities to prevent lateral and anterior drifting of the pelvis.

Seat backs should be analyzed for height and possible dynamic features, which will allow a person to periodically extend their spine and realign their trunk and pelvis to postures that can periodically neutralize the effects of gravity.

The need for lateral trunk stability should be evaluated in order to provide the mid-line orientation necessary to minimize the onset of lateral spinal curvatures. In situations of existing deformities, postures should be identified that will both reduce the interface stresses as well as counteract the deforming influences of gravity. Patients should be given the means and encouraged to assume these postures whenever it is practical to do so.

18.1.3. Tangentially-Induced Shear Force

As indicated throughout the study alterations of sitting posture can influence the nature of the stresses experienced by the supporting tissues. Presently designed wheelchairs provide trunk stability, especially for tetraplegic patients, by reclining the backrest. Most newer powered wheelchairs also have a reclining backrest as a powered option. Elaborate designs have been marketed that prevent relative movement between the person and the backrest when reclining takes place. Based on the results of this study, reclining the backrest creates the highest increases in tangential shear forces at the seat surface.

Insufficient friction or lack of a stabilizing horizontal force at the seat surface allows the lower pelvis to drift forward while the superior aspect rotates posteriorly. It is common practice to recline the back rest of the wheelchair in order to use gravity to prevent involuntary trunk flexion, especially for tetraplegic patients. The results of this study indicate that backrest reclining drastically increases the tendency for a person to slide forward on the seat surface. To the extent that this sliding is resisted by the seat surface friction (TIS force), potentially damaging tangentially-induced shear stresses can be experienced by the supporting tissues. Also, backrest recline up to 120° has relatively little effect on reducing maximum pressure. A better approach to provide trunk and pelvic stability and reduce TIS force and maximum pressure is to tilt the whole body in space. On average, a tilt angle of 20-30° will reduce the TIS value to zero, while reducing maximum pressure values by 10-15% from those values in a neutral upright posture (horizontal seat with 100° of backrest recline).

18.1.4. Centre of Gravity Information

Clinical tools for measuring and rapidly displaying the centre of gravity of the seated person in various postures are not readily available. It is not clear from this study how CG measurements can be cost-effectively used in a clinical environment. Until it can be clearly demonstrated how the centre of gravity interacts with the other interface variables, it is probably best that it remains primarily a research tool. One can speculate that one useful application may be the quantification of the stability of the wheelchair and occupant, when the occupant assumes various postures within the wheelchair. This information could be particularly useful when specialized seating inserts are placed in existing wheelchair frames. Monitoring of occupant CG may give clues regarding postural asymmetry and how this may change with time. This information could then be used to develop indicators of increasing deformity or altered postures.

18.1.5 Management Systems and Decision Making Tools

Several outstanding clinical settings have documented the processes they use to provide high quality state of the art pressure management services. It cannot be over-stressed that a system involving evaluation, selection, provision and rigorous follow-up is essential for success. One promising development is that the knowledge being acquired in these leading centres can be placed on computers in the form of Expert Systems. These computerized tools can be cautiously used as guides for improved clinical decision making, as well as provide an organized method for patient tracking and follow-up. The potential of Expert Systems is not in replacing the human decision making process, but rather supplementing it by presenting options that can be drawn from a broader base of knowledge than that possessed by an individual clinician. Expert Systems are now being used in clinical medicine to support diagnostic procedures and prescription of drugs.

Pioneer efforts on Expert System development in specialized seating has been initiated, and is currently being evaluated under the name CUSH-FIT (Tractman and Ferguson-Pell, 1984). CUSH-FIT offers a systematic approach to decision-making related to prescription of cushions for wheelchairs. Clinicians are encouraged to become knowledgeable about the potential benefits of newly developing decision-making tools in order to better exploit their advantages on behalf of the patients they serve.

18.2. RESEARCH PRACTICES

During the course of this study several points regarding past research practices are worth noting. Considerable resources have been expended on conducting pressure-related studies involving both animal and human subjects. Various models and laboratory procedures have been developed, complete with a variety of methods for analyzing and communicating the results. Unfortunately, the variability between the procedures and methods of data analysis and presentation is such that repetition of studies or comparing of results is virtually impossible. This author makes a plea for standardization of measurement models and disclosure practices, so that results can be readily compared. It is only through cooperative efforts of this nature that consensus will be achieved and guidelines produced that are meaningful in clinical environments.

Secondly, it was noted in several studies reviewed that normal subjects were used as readily available substitutes for individuals with spinal cord injuries. This study has demonstrated that rather marked differences exist between the two populations, particularly when the body-seat interface variables are being investigated. It is recommended that these types of substitutions be done with extreme caution since the results may not readily extrapolate across populations.

CHAPTER 19. SUGGESTIONS FOR FUTURE RESEARCH ACTIVITIES

This closing chapter provides an opportunity to highlight areas for continued research work that were identified in the course of this study. The following suggestions are biased towards research activities that will yield practical guidelines and solutions for clinicians, rather than on the acquisition of basic knowledge as the defined end point.

The role of shear stresses and how they interact with normal forces to produce damaging responses within the buttock tissues, remains largely unknown. We must know how important it is to minimize shear stress in the tissues, and be able to readily determine when it has been minimized. This study has only nibbled at the problem and much remains to be done.

Seating surface configurations and materials should be developed that minimize shear stresses; but still provide the lateral and A/P stability to prevent drifting of the pelvis into asymmetrical deformities.

Measurement of extrinsic factors like interface pressure and tangential shear force are still one step away from information about the physiological responses actually taking place in the underlying tissues. Clinical tools that will allow the practical and direct determination of tissue responses to external loading are highly desirable. Present studies using oxygen tension, biochemical analysis and advanced blood flow monitoring offer exciting promise and should be highly encouraged.

In spite of the recent availability of improved interface pressure monitoring tools, further improvements are still required. More reliable, accurate, and durable devices are needed. They should be able to measure and record both dynamically and statically.

Measurement and recording of daily activity levels of people and the various postures in which they carry out their activities are of direct clinical interest. It should then be possible to simulate these conditions in clinical settings so that more informed determinations can be made about the combination of variables that may lead to conditions of high risk. It follows that the developement of practical tools to accomplish this goal are also required.

This study only partially examined the deformities associated with the spinal cord injury and how they may effect the variables at the interface surface. The study was limited to the pelvis and lower lumbar spine. Biomechanical analysis of the complete paralytic spine would provide valuable insight into the nature of the exhibited deformities, and thereby provide guidelines as to how seat designs may improve to better compensate for loss of upper body function.

A complete review of wheelchair seating design for use by people with spinal cord injury is highly recommended. Independent dynamic posturing should be possible even by individuals with high levels of spinal lesions. Both manual and powered systems that have independently actuated backrest recline and full body tilt should be evaluated in the light of their pressure, shear and body alignment characteristics.

And finally, increased efforts should be made by those engaged in research endeavors to ensure that new knowledge and measurement tools are carefully placed in the hands of clinicians. A commitment should be made at the onset of any research endeavor that at least equal emphasis will be given to facilitating the transfer process. Publications, video tapes, educational seminars, collaborative research projects, Expert Systems, and hands-on problem solving are all methods that can be effectively used to transfer research findings to clinical practice. It is only through commitments of this kind that the field of specialized seating will mature from an art to a science, and thereby assume its rightful and respectable place in the health-care delivery system of tomorrow.

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APPENDICES

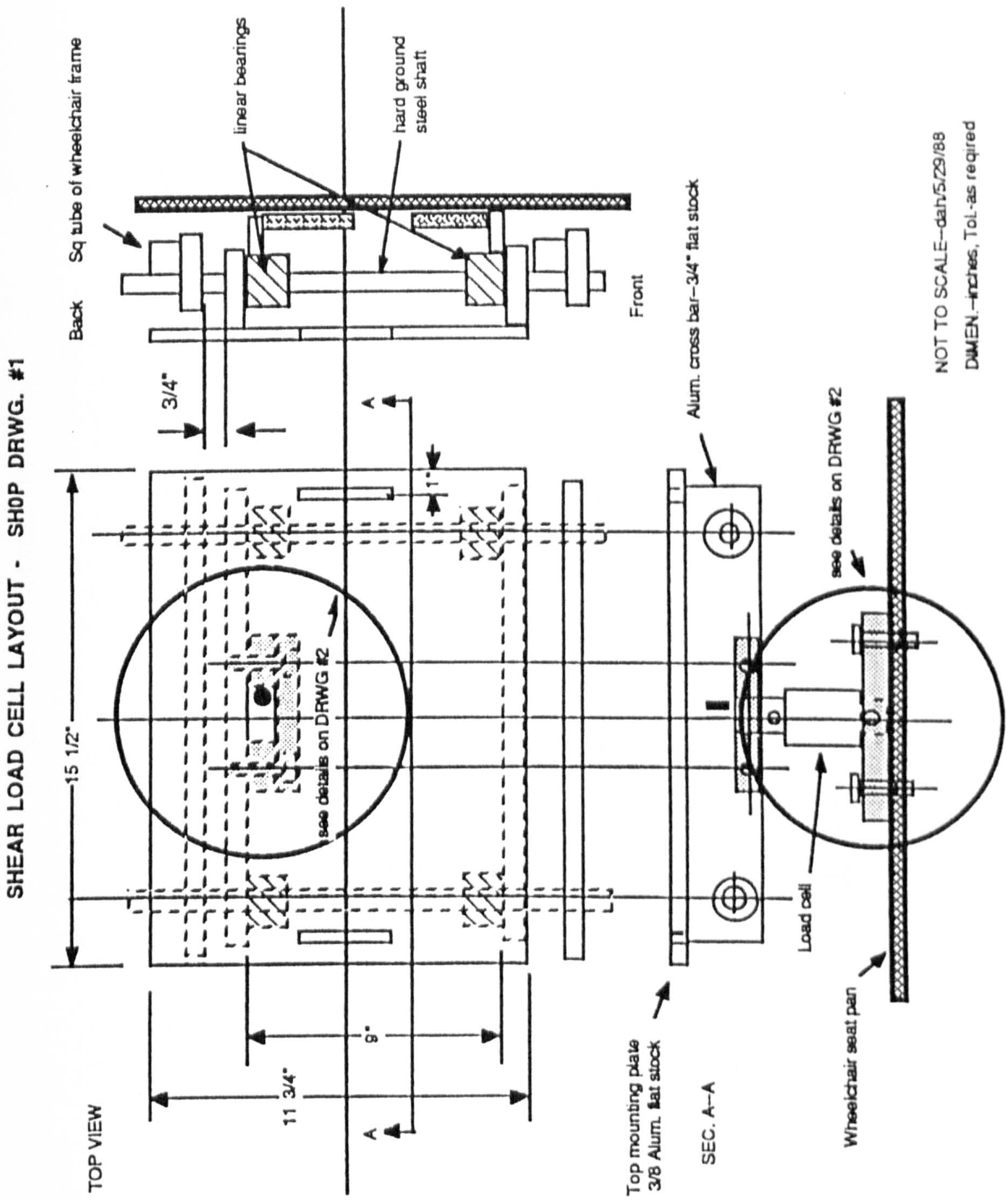
I-Shear load cell design and layout

II-Centre of gravity load cell design

III-Modifications to electronic circuitry

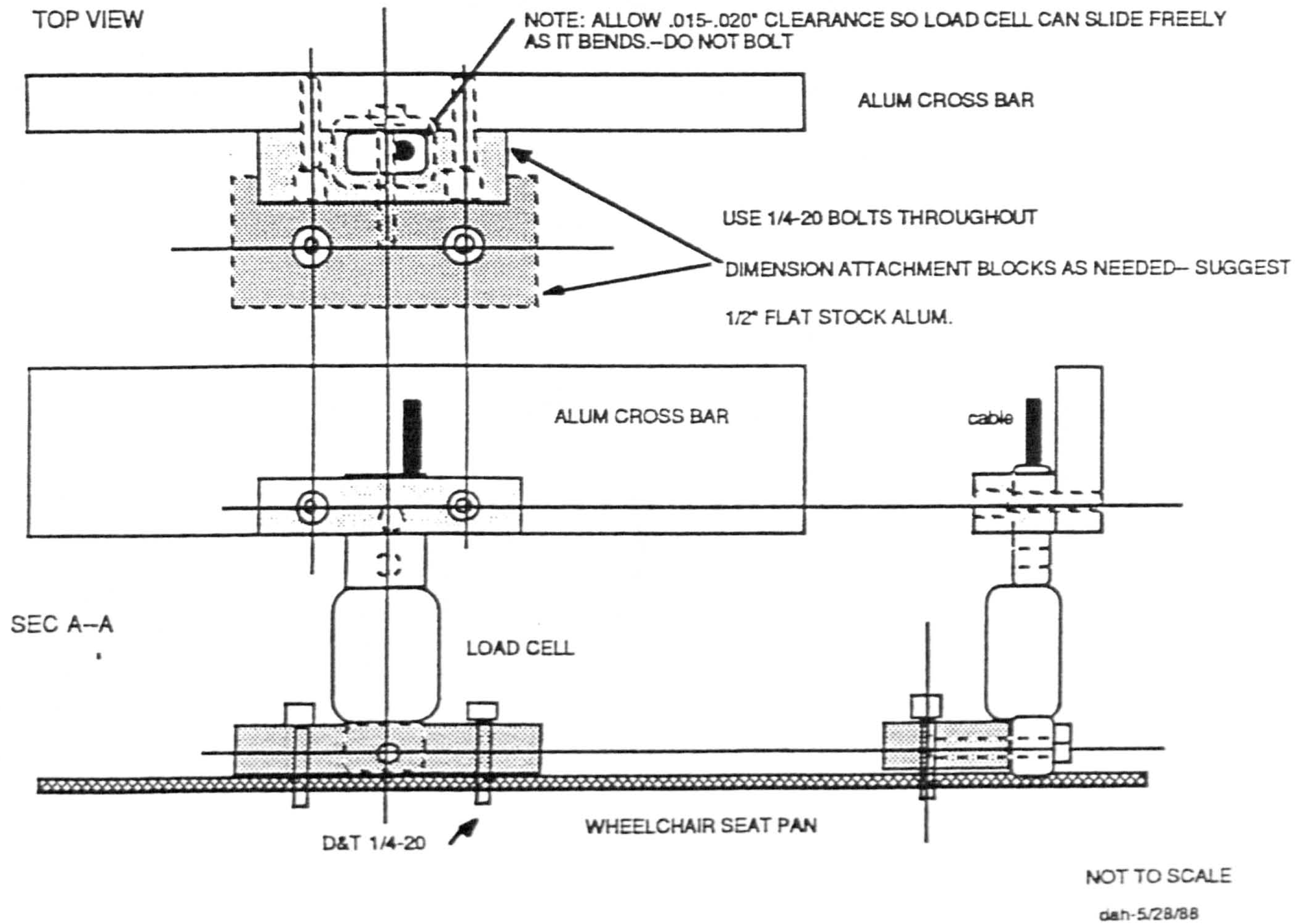
IV-List of commercial suppliers

APPENDIX I-A


SHEAR LOAD CELL LAYOUT - SHOP DRWG. #1
TOP VIEW
15 1/2"
11 3/4"
9"
3/4"
1"
A
A
see details on DRWG #2
Back
Front
Sq tube of wheelchair frame
linear bearings
hard ground steel shaft
Alum. cross bar–3/4" flat stock
Top mounting plate
3/8 Alum. flat stock
SEC. A–A
Load cell
see details on DRWG #2
Wheelchair seat pan
NOT TO SCALE–dah/5/29/88
DIMEN.–inches, ToL-as reqired

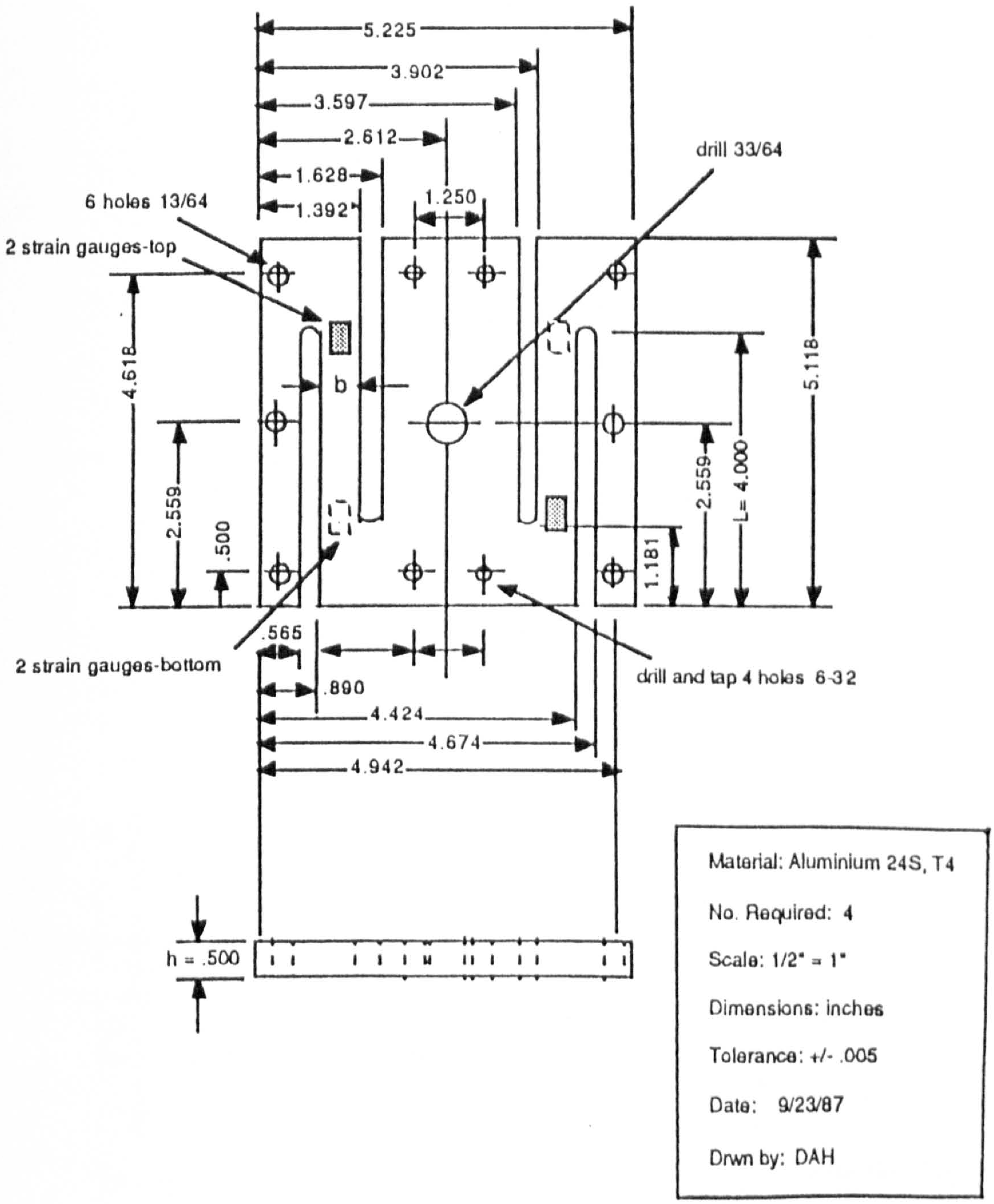
APPENDIX I-B

SHEAR LOAD CELL ATTACHMENT---Drw:



APPENDIX II-A

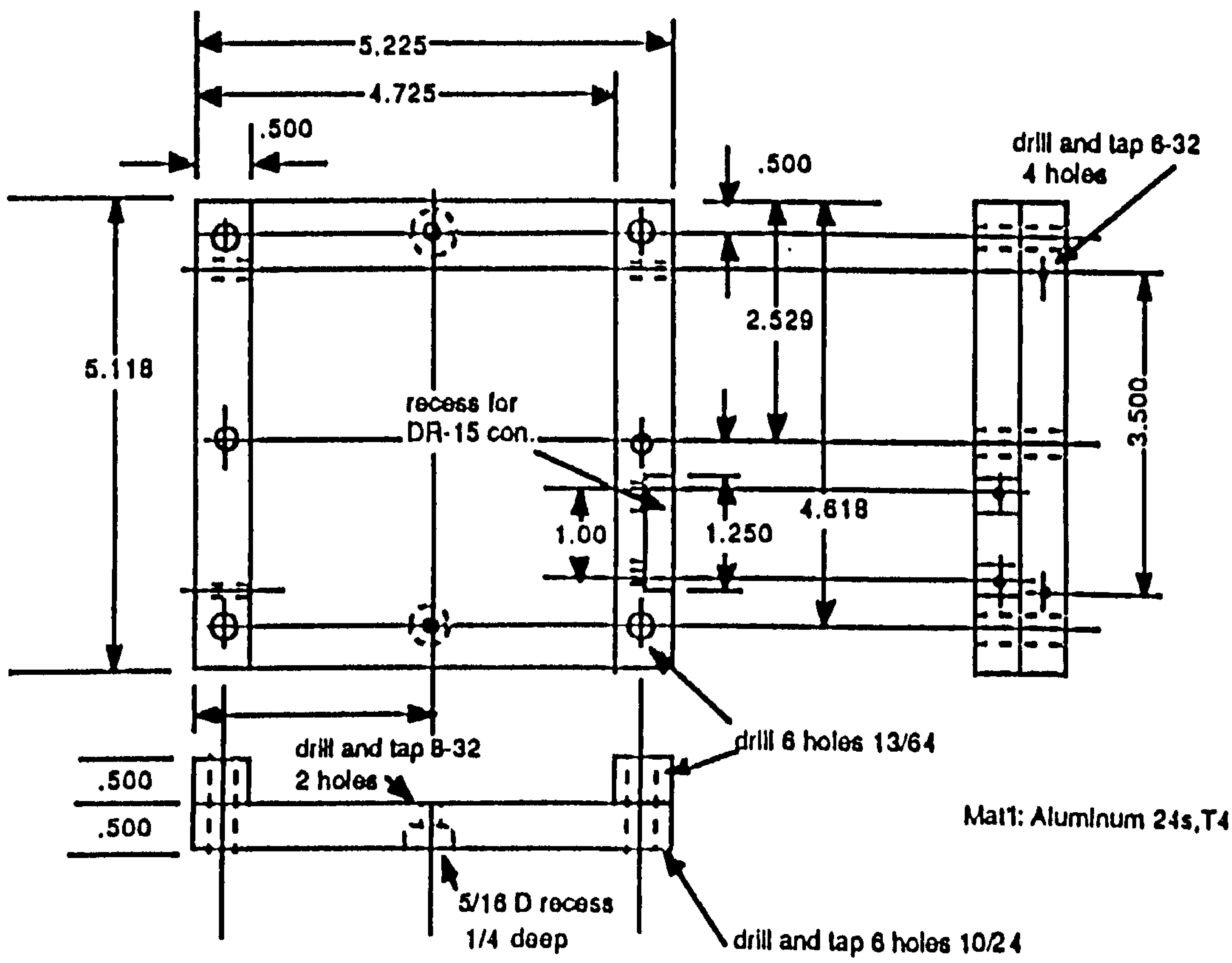
A) LOAD CELL DESIGN - TRANSDUCER PLATE



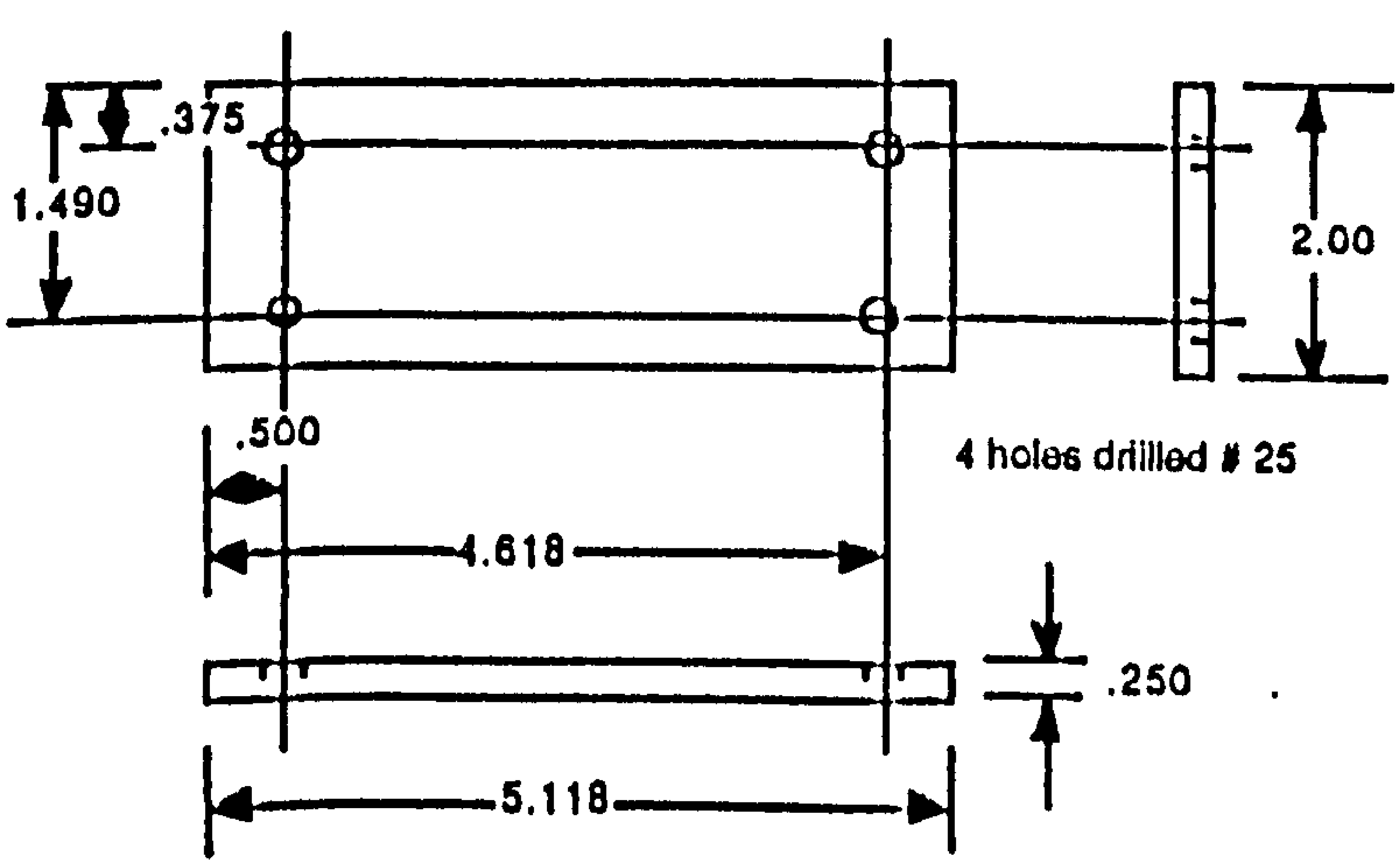
APPENDIX II-B

LOAD CELL DESIGN

B) BASE PLATE PARTS



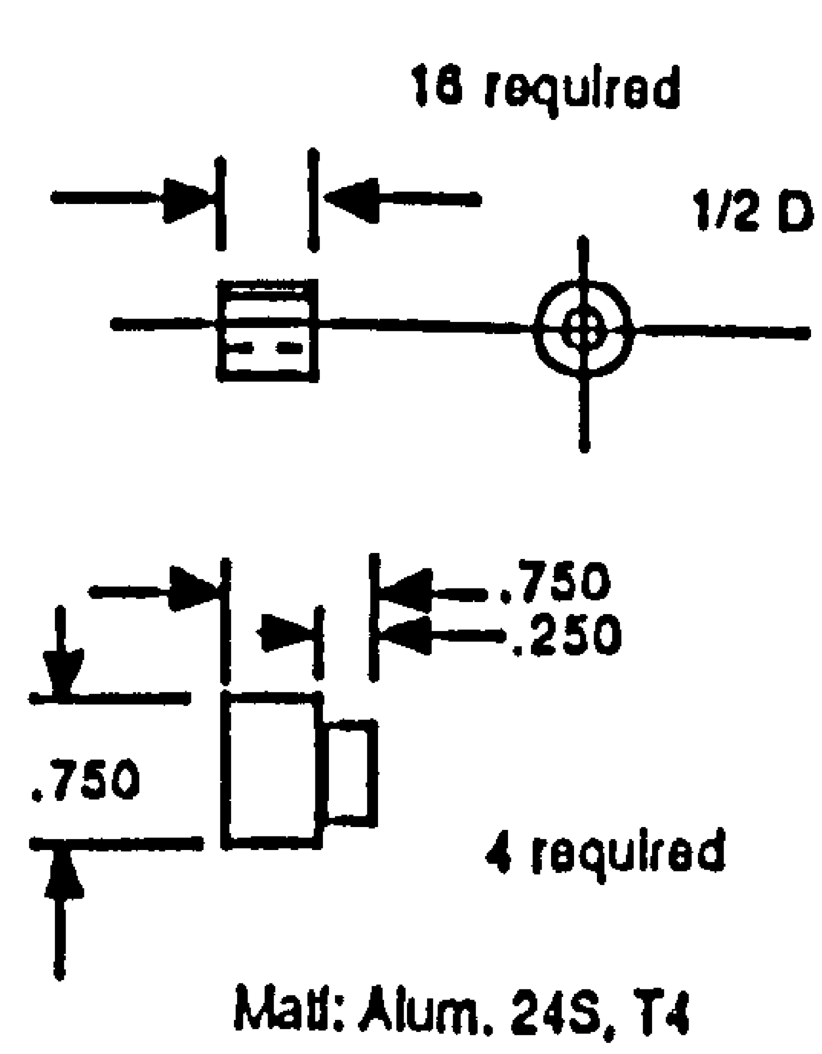
C) TOP LOAD BEARING PLATE



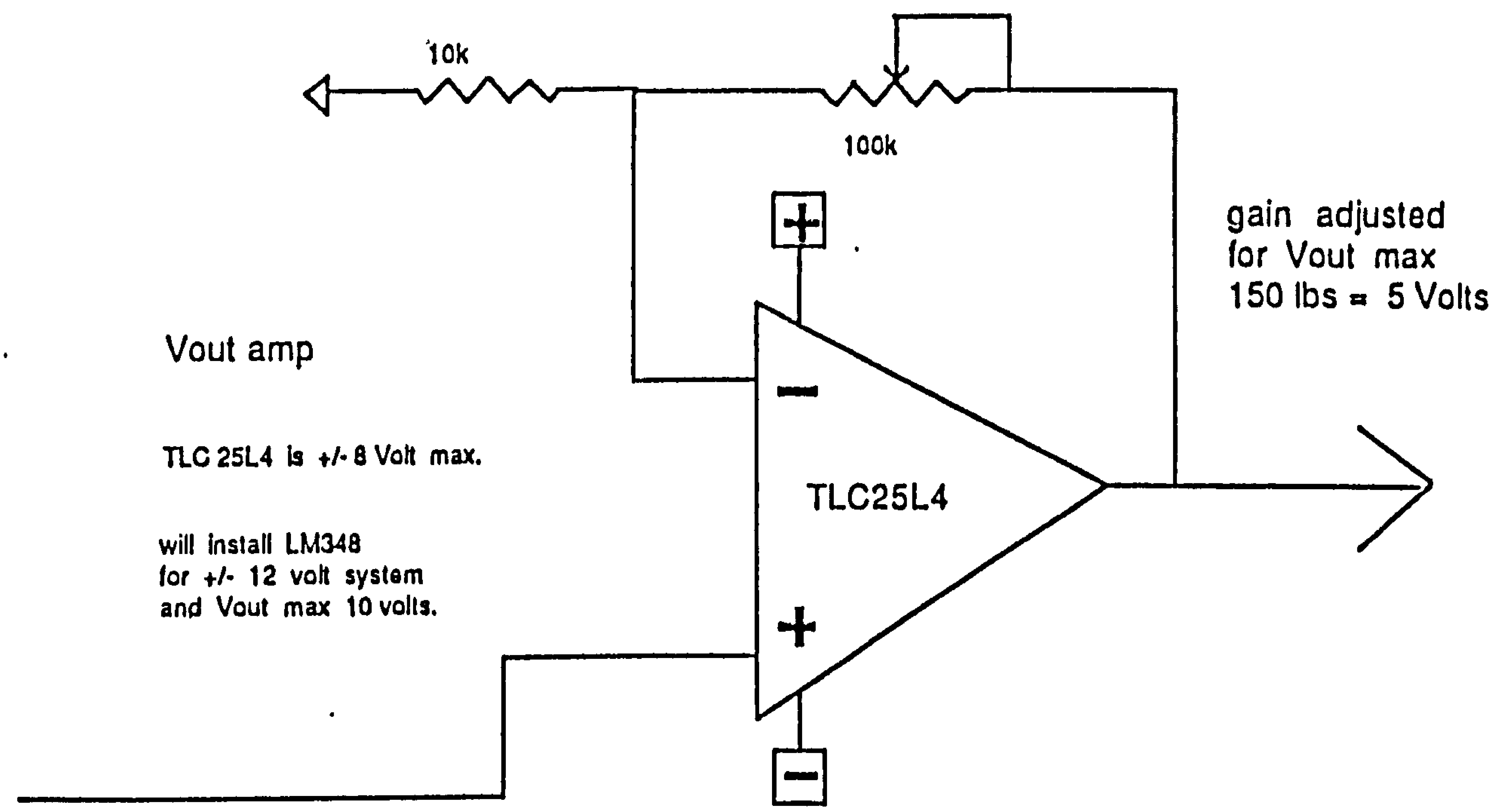
Mat,l: Aluminum 24s,T4

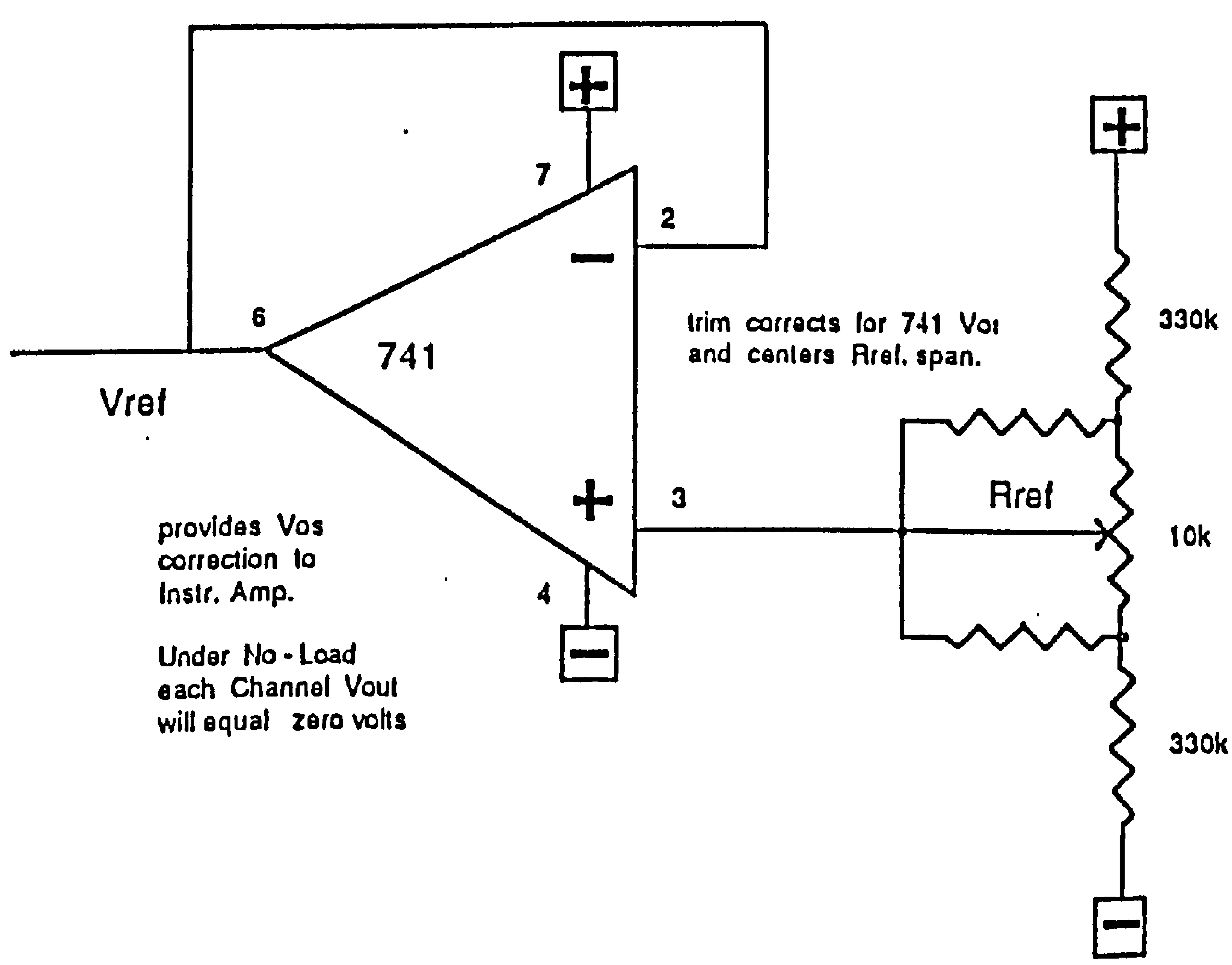
4 required

C) SPACERS

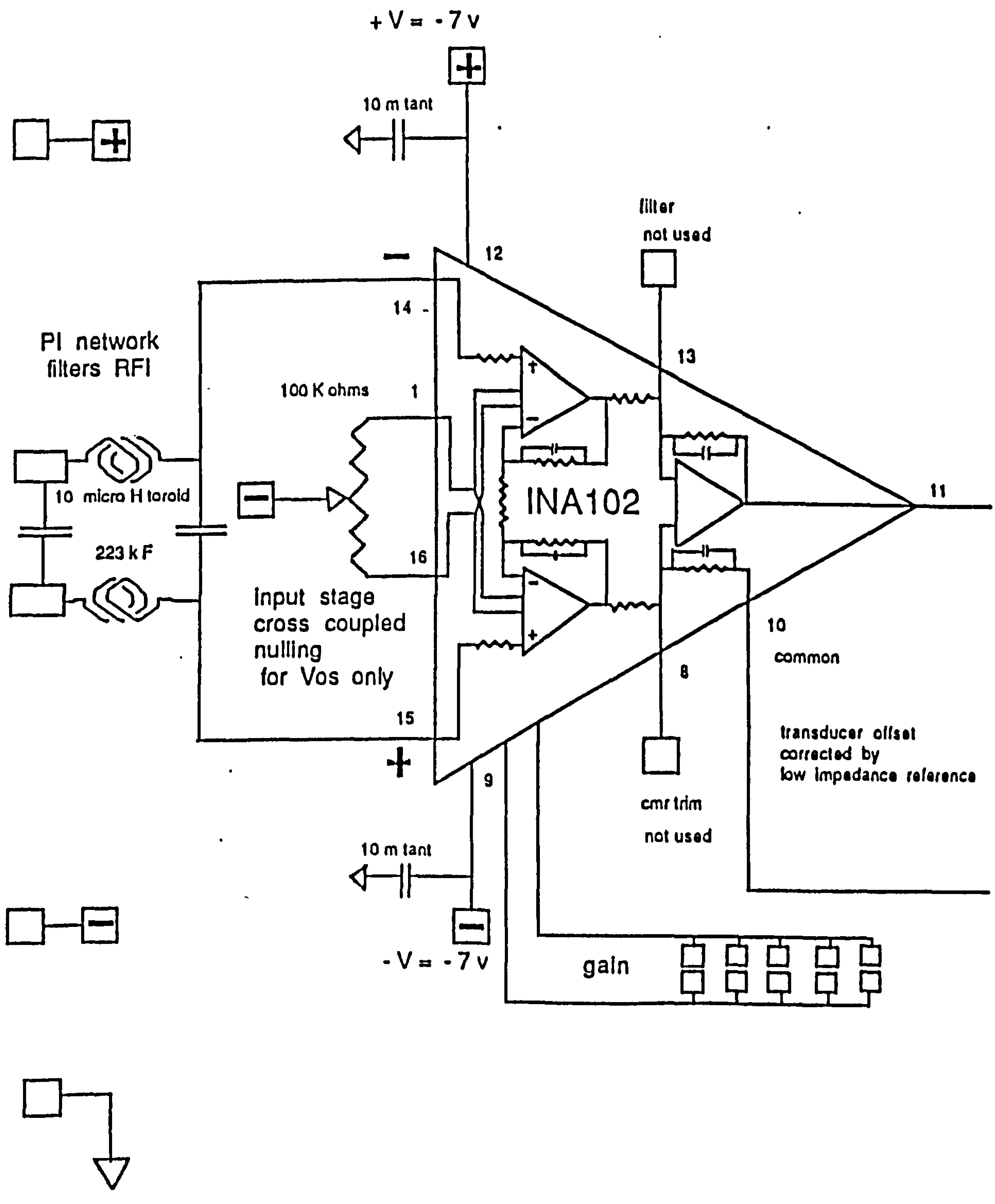


APPENDIX III-A

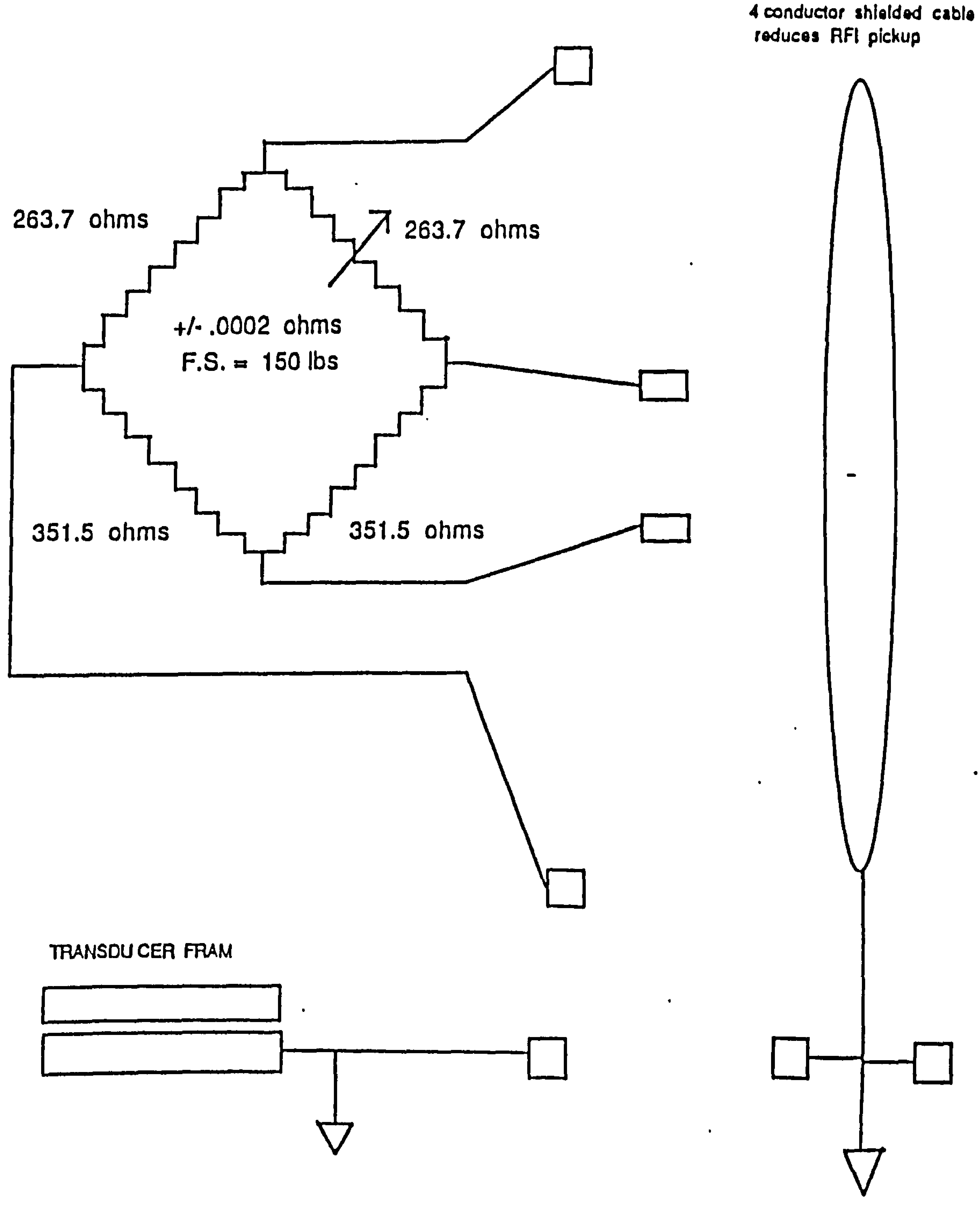




APPENDIX III-B



APPENDIX III-C



APPENDIX IV

a) Talley Medical Ltd.
Oxford pressure Monitor
47 Theobald St.
Borehamwood Herts
WD6 4RT, England

b) Dynamic Systems
Sunmate, Type, 1" blue (5 lbs/ft^3)
Rt. 2, Box 182 B
Leicester, NC, 28748
USA.

c) Sensotec Inc.
Memphis Micro Systems
P.O. Box 18630
Memphis , Tn.,38181-16030
USA.

d) Metrabyte Inc
440 Myles Standish Blvd.
Taunton, MA, 02780
USA.

e) Action Instruments Inc.
Model AP4051/AP4251 Strain- guage amplifier/
conditioner
8601 Aero Dr.
San Diego, CA, 92123
USA