THE UNIVERSITY OF MANITOBA

AN ANALYSIS OF TRUNK KINEMATICS IN THE SAGITTAL PLANE AND OF THE ROLES OF ERECTOR SPINAE AND RECTUS ABDOMINIS MUSCLES IN FIVE TYPES OF LOCOMOTION

BY

JULIETTE E. COOPER

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES IN PARTIAL

FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE OF

DOCTOR OF PHILOSOPHY

DEPARTMENT OF ANATOMY

WINNIPEG, MANITOBA

FEBRUARY, 1987. ©

Permission has been granted to the National Library of Canada to microfilm this thesis and to lend or sell copies of the film.

The author (copyright owner) has reserved other publication rights, and neither the thesis nor extensive extracts from it may be printed or otherwise reproduced without his/her written permission.

L'autorisation a été accordée à la Bibliothèque nationale du Canada de microfilmer cette thèse et de prêter ou de vendre des exemplaires du film.

L'auteur (titulaire du droit d'auteur) se réserve les autres droits de publication; ni la thèse ni de longs extraits de celle-ci ne doivent être imprimés ou autrement reproduits sans son autorisation écrite.

ISBN 0-315-37439-X

AN ANALYSIS OF TRUNK KINEMATICS IN THE SAGITTAL PLANE AND OF THE ROLES OF ERECTOR SPINAE AND RECTUS ABDOMINIS MUSCLES IN FIVE TYPES OF LOCOMOTION

BY

JULIETTE E. COOPER

A thesis submitted to the Faculty of Graduate Studies of the University of Manitoba in partial fulfillment of the requirements of the degree of

DOCTOR OF PHILOSOPHY

© 1987

Permission has been granted to the LIBRARY OF THE UNIVERSITY OF MANITOBA to lend or sell copies of this thesis, to the NATIONAL LIBRARY OF CANADA to microfilm this thesis and to lend or sell copies of the film, and UNIVERSITY MICROFILMS to publish an abstract of this thesis.

The author reserves other publication rights, and neither the thesis nor extensive extracts from it may be printed or otherwise reproduced without the author's written permission.

038068

For

John, Pilar and Elizabeth

ACKNOWLEDGEMENTS

This dissertation is the result of the contributions of many individuals. In particular I would like to thank the following persons:

Dr. R. E. Grahame for allowing me the privilege of her patient guidance and constant encouragement throughout this study.

Dr. T. V. N. Persaud, Head, Department of Anatomy, for the opportunity to study in the Department.

Arthur O. Quanbury, Head, Biomedical Engineering Services, Rehabilitation Centre for Children, for sharing generously his knowledge and his time, and for his friendship.

Dr. W. Dahlgren and Dr. H. I. C. Dubo, members of my advisory committee, for their constructive criticism and helpful recommendations at all stages of this study.

Karen Ross, Kinesiologist, Rehabilitation Centre for Children, for her many suggestions and her expert assistance during the course of the study.

Rick Borchert, who wrote the "LRS" computer program and spent countless hours solving programming and operator problems. His enthusiasm and encouragement eliminated my computer anxiety and created a convert.

Frank Hreczuch, Engineering Technologist, Rehabilitation Centre for Children, for on-the-spot solutions to electronic problems and for maintenance and repair of the equipment.

Mary Cheang, Biostatistical Consultant, Medical Biostatistical Consulting Unit, for her advice on appropriate statistical tests during the design of the study.

Dr. M. Alexander, Faculty of Physical Education and Recreation Studies, for the loan of equipment during the final stage of data extraction.

Roy Simpson, for his expertise in preparing the photographs and diagrams.

Dr. J. E. K. Cooper, for his cheerful assistance during the final experiments.

Prof. Ann Booth, Head, Division of Occupational Therapy, for her commitment to research and her constant support throughout my graduate career.

The subjects who participated in the study and offered their enthusiasm for the project.

Judy Anderson, Donna Collins, Emily Etcheverry, Gerard Gouw, Jean Hay, Mary Pflueger and Anne Strock for caring, encouragement, questions and discussion.

The Manitoba Health Research Council, for financial support of this study.

ABSTRACT

Little is known about the kinematics and trunk muscle activity of stair or ramp walking, types of locomotion that are required in many jobs. There is a high incidence of back injury in heavy industry, therefore it is important to understand the demands that are placed on the spinal mechanism by the locomotor requirements of a work situation. In this study electromyography was used to analyze the activity of erector spinae and rectus abdominis muscles in 18 normal male subjects during five different forms of locomotion: Level Walking, Stair Climbing, Stair Descent, Ramp Climbing and Ramp Descent. Foot switches were used to identify the temporal events in the gait cycle, and simultaneous high speed cinefilm was used to record the displacement of upper body segments during each type of locomotion. A custom-designed computer program was used to store and analyze the data. Duration of the gait cycle, cadence, body segment displacement in the sagittal plane and myoelectric activity were compared between the locomotion types. Significant differences were found. Cycle duration was longer in the two climbing locomotion types than in the descending types with the opposite being true of cadence. Anterior inclination and excursion of trunk and pelvic segments were greater in climbing than in descending, as was hip excursion. The myoelectric activity of erector spinae was also greater in the two climbing locomotion types than in the descending types; the myoelectric activity of rectus abdominis was low in all instances. Triphasic activity was observed in both groups of muscles and may be related to body segment displacement in all three planes. It was concluded that the load on the back is increased during stair and ramp climbing, therefore these activities should be restricted or modified for the back-injured worker.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER I

INTRODUCTION

Low back pain may be defined as pain or discomfort occurring in the area of the lumbosacral spine and can be classified as acute, subacute, chronic or recurring (Nachemson & Andersson, 1982; Snook, 1983). Although the exact cause of low back pain is unknown in the majority of patients, it is generally conceded that mechanical stress is a factor in the etiology of the condition and it is well accepted that mechanical stress, particularly that caused by forward bending, can aggravate existing low back pain (Berkson et al., 1977; Nachemson, 1978; Nachemson & Andersson, 1982; Andersson & Ortengren, 1984; Marras et al., 1984; Bendix et al., 1985). According to Tichauer (1971), a minor back injury may result in major disability because it interferes with the individual's ability to react promptly and effectively to physical work stress.

It is known that low back pain is associated with jobs that involve frequent bending, twisting and lifting (Frymoyer et al., 1980; Andersson, 1981). Stubbs (1981) stated that low back pain and back injury arising from materials handling jobs, which involve frequent bending, lifting and carrying, are constant factors in industry.

Bigos et al. (1986) studied 31,000 employees of the Boeing Company in the United States and found that materials handling jobs accounted for 56% of all back injuries with the most common type of injury being "strain" (84%). Spengler et al. (1986), reporting on the same sample, calculated that between 1979 and 1980 back injury cost the Boeing Company $1.8 million, and represented 41% of the total incurred cost of all types of injury. In Manitoba, back injury constituted over 18% of all compensated injury claims in 1985; when the average number of lost work days per accident is calculated, back injury accounted for a loss in wages of approximately $2 million (Workers Compensation Board of Manitoba, 1985). Langrana et al. (1984) estimated that in the United States back injury resulted in an average loss of 28.6 workdays per year per 100 subjects and, in 1975, in the United Kingdom, Benn and Wood stated that back injury accounted for a greater loss of work days than strikes.

However, in spite of numerous investigations into the possible relationship between work, mechanical loading of the back and back pain, it is not known which load factors are the most likely to give rise to back pain, and little is known about the load on the back produced by various work situations or occupations (Andersson & Ortengren, 1984; Nordin et al, 1984). According to Berkson et al. (1977), biomechanical analyses of the loads on the vertebral column created by different activities are needed.

In addition to work, many everyday activities involve frequent flexion of the vertebral column (Soderberg, 1986), a position that is known to aggravate and may be an etiological factor in low back pain (Bendix et al., 1985). Level walking, climbing and descending stairs are common activities of daily living (Andriacchi et al., 1980) and, with the addition of climbing and descending ramps, may be components of jobs in construction, shipping, mining, forestry and farming. It is known from studies by Thorstensson et al. (1982, 1984, 1985) and Thurston and Harris (1983) that flexion of the vertebral column occurs during level walking. Empirical evidence suggests that flexion of the vertebral column also occurs during stair and ramp climbing. Therefore, the locomotor component of a particular job may, by itself, generate a mechanical load on the back.

No direct means exist by which to measure loads upon the human lumbar spine *in vivo*, however the indirect measurements of intradiscal pressure, intra-abdominal pressure and the myoelectric signal from paraspinal muscles have been shown to be directly related to loading of the lumbar spine in sitting and standing postures (Marras et al., 1984).

Studies of human locomotion have concentrated on motion of the lower limb and Gracovetzky (1985) stated that the contribution of the spinal mechanism to walking has been almost totally ignored. However, Carlson and Thorstensson (1981) and Thorstensson et al. (1982) observed that the trunk, because of its large mass, plays an important role in equilibrium control, therefore smooth interaction between trunk and limbs is essential for efficient locomotion. Trunk muscles have the potential to control motion of the vertebral column, but little is known about the specific roles of the erector spinae and rectus abdominis muscles during functional activities, including locomotion (Soderberg & Barr, 1983; Basmajian & DeLuca, 1985).

Although some investigations of trunk muscle activity during level walking have been carried out, only one study was found that reported activity of trunk muscles during stair climbing (Joseph & Watson, 1967). Only one study was found that reported on the activity of trunk muscles in ramp climbing (Waters & Morris, 1970). Likewise, motion of the trunk during level walking has been studied (Thorstensson et al., 1982, 1984, 1985; Thurston & Harris, 1983; Bejjani et al., 1984), but no reports were found of quantification of trunk motion during stair and ramp climbing. Therefore, there is a need to gather basic information about trunk motion and trunk muscle activity during locomotion.

The present study was conducted to quantify trunk motion in the sagittal plane and to clarify the role of erector spinae and rectus abdominis muscles during the types of locomotion commonly found in the workplace - level walking, stair climbing and descent, and ramp climbing and descent.

Cinefilm was used to record the position of the trunk, and electromyography (EMG) was used to analyze the activity of the erector spinae and rectus abdominis muscles during the five different types of locomotion. The temporal events in the gait cycle were defined by footswitches whose signals were stored and displayed simultaneously with the EMG signals. The cinefilm was synchronized with the EMG tracing by means of a signal light in the field of view of the camera.

From the accumulated data, comparisons between the different types of locomotion were made. These included duration of the gait cycle, cadence, amount of total muscle activity during the gait cycle, points of peak muscle activity during the gait cycle, excursion of the trunk and pelvic segments and excursion of the hip angle in the sagittal plane.

Because of the limitations of the study, the investigation was confined to sagittal plane motion of the body. Only angular displacement data for trunk, pelvis and hip are reported in this thesis.

The data on trunk motion and muscle activity occurring in normal subjects during different types of locomotion can serve as a baseline for investigations of individuals with low back injury sustained in the workplace, or may be used in studies of locomotor requirements in specific jobs. Data from this study may also be used in workplace modification to enable back injured workers to return to their jobs, a stated goal of the Workers Compensation Board of Manitoba (Workers Compensation Board of Manitoba, 1982). Finally, data from this study might also be used in workplace modification to prevent low back injury.

CHAPTER II

REVIEW OF THE LITERATURE

EMBRYOLOGY

FORMATION AND DIFFERENTIATION OF SOMITES

By the end of the second week of intrauterine development the inner cell mass of the human embryo has differentiated into a bilaminar germ disc consisting of an epiblastic layer and a hypoblastic layer. The epiblast will give rise to the embryonic ectoderm and mesoderm; the embryonic endoderm is formed when epiblastic cells migrate ventrally and displace the cells of the hypoblast. At the beginning of the third week the primitive streak appears on the caudal surface of the embryonic disc and epiblastic cells begin to migrate toward it. These cells invaginate the primitive streak and move laterally to form the mesoblast or mesodermal germ layer. Initially the mesoblastic cells form a thin sheet on each side of the midline, however as the notochord and neural tube develop, the cells adjacent to these structures proliferate to form a longitudinal column of paraxial mesoderm. By day 20 this paraxial mesoderm begins to divide into paired segmental blocks or somites. The first pair of somites develops adjacent to the cranial area of the notochord and subsequent pairs develop in a craniocaudal sequence until, by the end of the fifth week, there are 42 to 44 pairs of somites (Hamilton & Mossman, 1972; Williams & Warwick, 1980; Crelin, 1981; Moore, 1982; Sadler, 1985).

Each somite consists of tightly packed epitheliod cells which begin to differentiate by the end of the fourth week. Those cells forming the ventral and medial walls of the somite lose their epithelial characteristics and become polymorphous. Known now as the sclerotome, the cells then shift their positions with respect to adjacent structures such that they surround the notochord and developing neural tube, and extend into the body wall. The sclerotome will give rise to the connective tissue, cartilage and bone of the axial skeleton. The cells of the dorsal and lateral walls of the somite constitute the dermomyotome. Cells from its medial aspect proliferate to form a closely packed mass, the myotome, which will ultimately form the striated musculature of the trunk. The remaining cells, constituting the dermatome, spread out beneath the overlying ectoderm and give rise to the dermis and subcutaneous fascia (Hamilton & Mossman, 1972; Crelin, 1981; Moore, 1982; Sadler, 1985).

DEVELOPMENT OF THE AXIAL SKELETON

The notochord is the primitive axis of the embryo. However, because it is a flexible cellular rod, it is inadequate as a supporting structure and is retained only as a central axis about which the sclerotomes will organize to form the vertebral column (Sensenig, 1942; Moore, 1982).

During the fourth week sclerotomal cells surround and enclose the notochord. Each sclerotome is divided into a more dense caudal portion and a less dense cranial portion, both areas being briefly separated by a sclerotomic fissure. The cells in the caudal portion proliferate and spread into the intersegmental tissue until eventually the caudal portion of one somite fuses with the cranial portion of the adjacent somite to form the precartilaginous vertebral body. The mesoderm adjacent to the sclerotomic fissure condenses to form the perichordal disc which will give rise to the intervertebral disc (Hamilton & Mossman, 1972; Langebartel, 1977; Crelin, 1981; Moore, 1982; Sadler, 1985).

The notochord degenerates and ultimately disappears from the area of the vertebral bodies, but in the area of the intervertebral disc it persists, enlarges and undergoes mucoid degeneration to form the nucleus pulposus. The mesodermal cells surrounding the nucleus pulposus differentiate to form the fibrocartilaginous anulus fibrosus (Crelin, 1981; Sadler, 1985).

The sclerotomal cells surrounding the neural tube form the neural arch of the primitive vertebra; those in the body wall form the costal processes. In the cervical region these processes form the anterior tubercle of the transverse processes, in the thoracic region they extend ventrally in the body wall to form the ribs, in the lumbar region they form part of the transverse processes, and in the sacrum they form the lateral masses (Crelin, 1981; Moore, 1982; Sadler, 1985).

During the sixth week, two centres of chondrification appear in the primitive vertebral body to form the cartilaginous centrum. Centres also appear in the neural arch and extend ventrally to unite with the centrum, dorsally to fuse behind the the neural tube forming the laminae and spinous processes, and laterally between the myotomes to form the transverse processes (Hamilton & Mossman, 1972).

Ossification occurs in approximately the same sequence, beginning during the embryonic period and ending at approximately the twenty-fifth year (Hamilton & Mossman, 1972; Langebartel, 1977; Moore, 1982).

DEVELOPMENT OF MUSCLES OF THE TRUNK

At the beginning of the fourth week the cells of the myotome are mononucleate myoblasts. By the fifth week the myoblasts have elongated, divided and fused with each other to form myotubes, with centrally located nuclei. Myofilaments begin to appear in the cytoplasm and, as increasing numbers are laid down, the nuclei and mitochondria are displaced to the periphery of the myotube which is now known as a muscle cell (Crelin, 1981).

All trunk musculature is derived from the myotomes, with the exception of certain head and neck muscles that develop from branchial arch mesenchyme. Because of the fusion of adjacent sclerotomes, the centre of each myotome lies opposite an inter-vertebral disc. This position will allow the muscles derived from the myotome to move adjacent vertebral bodies (Langebartel, 1977; Williams & Warwick, 1980; Sadler, 1985).

During the fifth week of development there is rapid growth of the myotome such that it comes to lie adjacent to the neural tube dorsally and extends into the somatopleure ventrally. Axons from the ventral roots of the developing spinal nerves reach their respective myotomes at this time. Between the fifth and sixth week a longitudinal constriction develops in the myotome, dividing it into a small dorsal portion - the epimere, and a large ventral portion - the hypomere. These portions will later be permanently separated by the developing transverse process. The spinal nerve also divides into a dorsal primary ramus which innervates the epimere, and a ventral primary ramus which innervates the hypomere (Hamilton & Mossman, 1972; Sadler, 1985).

The epimeric portion of the myotome will form the extensor muscles of the vertebral column. It splits into a medial division, from which will develop spinalis, semispinalis, multifidus and rotatores, and a lateral division from which will develop longissimus and iliocostalis (Hamilton & Mossman, 1972; Sadler, 1985).

The hypomeric portion of the myotome extends ventrally into the somatopleure to form the lateral and ventral trunk musculature. The lateral portion splits into three layers to form the intercostal muscles in the thorax and the external oblique, internal oblique and transversus abdominis muscles in the abdomen. A longitudinal muscle column forms at the ventral tip of the hypomeres to form the rectus abdominis muscle in the abdominal region (Hamilton & Mossman, 1972, Sadler, 1985).

DEVELOPMENT AND CLOSURE OF THE BODY WALL

During the fourth week the rapid growth of the somites results in lateral folding of the flat embryonic disc to establish the ventral body wall of the embryo. This consists of a thin layer of somatopleure which is subsequently invaded by myoblasts from the hypomeres, causing an advancing thickening of the primitive body wall. By the late embryonic period this muscular thickening has advanced to the point that only a broad, diamond-shaped area of somatopleure remains around the attachment of the umbilical cord. Fusion of the edges of the muscular body wall begins in the upper thoracic region of the embryo, then in the suprapubic region. From these two areas fusion extends towards the umbilicus, the line of fusion being the linea alba. By week 12 the fusion of the definitive body wall is usually complete (Hamilton & Mossman, 1972, Sadler, 1985).

NORMAL ANATOMY

VERTEBRAL COLUMN

The vertebral column forms the central axis of the body and is the central pillar of the trunk (Kapandji, 1974; Clemente, 1985). It is a strong yet flexible bony and ligamentous structure comprised of 33 vertebrae and interposed intervertebral discs (Last, 1978). The vertebrae show regional variation and are differentiated into seven cervical, twelve thoracic, five lumbar, five sacral and five coccygeal vertebrae. The caudal bones in the column fuse and, by approximately age 25, the sacral and coccygeal vertebrae have united to form the sacrum and coccyx, respectively. The average length of the vertebral column in the male is 71 centimeters (cm), of which the cervical portion contributes 12.5 cm, the thoracic portion 28 cm, the lumbar portion 18 cm and the sacrum and coccyx 12.5 cm to the total length. The intervertebral discs account for one fifth of the total length of the column (Lockhart et al., 1965; Clemente, 1985).

Because of its rigidity, the vertebral column gives static support to the head and trunk, attachment to the ribs and limbs, and protection to the neuraxis (Last, 1978; Cailliet, 1981). Since it is also a flexible structure, it permits locomotion and purposeful movement (Kapandji, 1974; Cailliet, 1981).

Cailliet (1981) considered the vertebral column to be an aggregate of superimposed segments, each being a self-contained functional unit. The functional unit itself comprises an anterior element consisting of two vertebral bodies with their intervening intervertebral disc, and a posterior element consisting of the neural arch and two synovial articulations. The functions of the anterior element are to support, bear weight and absorb forces, while those of the posterior element are to guide and limit movement between the two adjacent vertebrae (White and Hirsch, 1971; Andersson, 1983; Bogduk, 1983).

During fetal life the vertebral column is curved into one continuous anterior concavity which is designated the primary curvature (Lockhart et al., 1965; Williams & Warwick, 1980). As fetal development proceeds, the lumbosacral angle appears. After birth, as the child raises and balances its head, the cervical part of the vertebral column becomes concave posteriorly. Kapandji (1974) stated that during evolutionary development the transition from quadrupedal to bipedal stance led first to straightening and then to inversion of the lumbar curvature and that similar changes are recapitulated during ontogeny. Therefore, at birth the lumbar column is convex posteriorly, but by 13 months the convexity disappears and from three years of age onward posterior concavity is evident (Asmussen & Klausen, 1962).

By early childhood the vertebral column has two primary curvatures, in the thoracic and sacral regions, and two secondary curvatures, in the cervical and lumbar regions (Williams & Warwick, 1980; Clemente, 1985). The cervical curve is considered to extend from the first cervical to the second thoracic vertebra, the thoracic curve from the second to the twelfth thoracic vertebra, the lumbar curve from the twelfth thoracic vertebra to the lumbosacral angle, and the sacral curve from the lumbosacral angle to the apex of the coccyx (Williams & Warwick, 1980).

The spinal curvatures that result from normal development are termed "physiological" and are produced partly by the wedge-shape of the vertebral bodies, but chiefly by the intervertebral discs (Last, 1978). Because they decrease the longitudinal stiffness of the vertebral column, these normal curves increase its shock absorbing capacity (Adams & Hutton, 1985).

VERTEBRAE

Each of the 24 presacral vertebrae has certain common elements which vary somewhat in specific regions according to function. A typical vertebra has a cylindrical body which consists of a dense bony cortex surrounding a spongy medulla. The cortex of the superior and inferior aspects is thickened at the periphery to form a distinct rim which is derived from the epiphyseal plate and which fuses to the vertebral body by age 25 (Williams & Warwick, 1980). Attached to the posterior aspect of the body is the vertebral arch, composed of a pair of cylindrical pedicles laterally and a pair of flattened laminae which fuse in midline to complete the arch posteriorly. Seven bony processes take origin from the vertebral arch. The single spinous process arises from the junction of the two laminae and is directed backwards. The two transverse processes originate at the junction of pedicles and laminae and are directed laterally, while the four articular processes arise from the same region with two directed superiorly and two inferiorly. In addition, each vertebral arch has associated with it costal elements which become independent units, the ribs, only in the thoracic region. In the other divisions of the vertebral column the costal elements remain undeveloped and fuse with the vertebrae (Williams & Warwick, 1980).

A typical cervical vertebra has a small body which displays prominent upturned rims laterally and an overhanging lip in midline inferiorly. Its spinous process is bifid and the transverse processes are distinguished by foramina transversaria. The short articular processes form a bony column cut obliquely into segments such that the facets on the superior processes face upward and backward while those of the inferior processes face downward and forward (Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

The body of a typical thoracic vertebra is characterized by bilateral upper and lower demifacets for articulation with the heads of ribs. Costal facets are also found at the tips of the transverse processes. The spinous processes are long, slant backward and down, and overlap. The articular surfaces of the superior articular processes lie on an arc of a circle whose centre lies approximately at the centre of the vertebral body, and these face posteriorly, slightly laterally and upward. Those of the inferior articular processes face forward, slightly medially and downward (Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

Attached to the thoracic vertebrae are the expanded costal elements of the vertebral arch, the 12 pairs of costae, each costa consisting of a rib bone and costal cartilage (Williams & Warwick, 1980). The head of each rib is shaped like a blunt arrowhead and articulates via synovial joints with its numerically corresponding vertebral body and that of the vertebra immediately above. The apex of the head is bound to the disc between the two vertebrae by a transversely-placed intra-articular ligament. The tubercle of the rib articulates with the transverse process of its numerically corresponding vertebra and is strongly attached to it by costo-transverse ligaments. By means of the superior costotransverse ligament, the tubercle of the rib is bound to the transverse process of the vertebra next above. Anteriorly the costae articulate directly and indirectly with the sternum. These articulations are all synovial except for that of the first rib which is a synchondrosis. Therefore, in the thoracic region a semi-rigid cage of bone is formed by the vertebral column posteriorly, the ribs laterally and the sternum anteriorly (Hollinshead, 1976; Last, 1978).

In the lumbar region a typical vertebra has a stout, massive body which is slightly higher anteriorly than posteriorly and thus is wedge-shaped. The spinous process is short, thick and quadrangular, while the transverse processes are long and thin. The facets of the articular processes are oriented vertically; those of the superior processes face medially and backward while those of the inferior processes face laterally and forward (Williams & Warwick, 1980; Clemente, 1985).

The sacrum, composed of five fused vertebrae, is the foundation platform upon which is balanced the superincumbent spinal column and thus it bears the weight of the head, trunk and upper extremities (Last, 1978; Cailliet, 1981). It is triangular in outline, concave anteriorly, convex posteriorly with a large auricular surface on either side for articulation with the hip bones. Superiorly, the base of the sacrum presents all the features of a typical vertebra in a slightly modified form. The upper surface articulates with the fifth lumbar vertebra and slopes downward and forward. Any tendency for the lumbar vertebra to slide forward on the sacrum is prevented in part by the large, upward projecting superior articular processes of the sacrum. These are directed medially and backward to articulate with the inferior processes of the fifth lumbar vertebra (Hollinshead, 1976; Williams & Warwick, 1980; Clemente, 1985).

JOINTS OF THE VERTEBRAL COLUMN

The anterior elements of adjacent vertebrae are joined by the fibrocartilaginous intervertebral disc, therefore the resultant articulation is classified as a symphysis. The intervertebral disc has two component parts - the peripheral anulus fibrosis and the central nucleus pulposus. The anulus fibrosus consists of concentric lamellae of dense fibrous tissue and fibrocartilage, each lamella being oriented obliquely to the one adjacent to it. The lamellae are firmly attached to the vertebral bodies (Ghadially, 1978; Stockwell, 1979; Clemente, 1985). The nucleus pulposus is a strongly hydrophilic colloidal gel that is confined and held under pressure by the anulus fibrosus. It consists of approximately 88% water and is essentially incompressable although its fluid nature allows it to change shape easily (Last, 1978; Lindh, 1980; Cailliet, 1981; Clemente, 1985).

The anterior elements of vertebrae are also joined by ligaments. The anterior longitudinal ligament extends from occiput to sacrum on the anterior aspect of the vertebral body and is thicker and narrower in the thoracic region than in the cervical or lumbar regions. It is firmly attached to the anterior surface of the intervertebral discs and the margins of the vertebral bodies, but is loosely attached to the middle of the bodies. The posterior longitudinal ligament also extends the length of the vertebral column and is located on the posterior aspect of the vertebral bodies, inside the vertebral canal. The margins of this ligament appear serrated as its fibres extend laterally to bind closely to the intervertebral disc and adjacent edges of the vertebral bodies while the fibres spanning the middle of the vertebral body form a narrow band that is not attached to bone (Williams & Warwick, 1980; Clemente, 1985). These ligaments also reinforce the anulus fibrosus of the intervertebral disc and hold the disc under tension (Kazarian, 1975; Hollinshead, 1976; Stockwell, 1979).

The posterior elements of the vertebral column are united by synovial joints between the superior and inferior articular processes. These apophyseal joints have all the features of a typical synovial articulation. Ligaments join the vertebral arches, spinous and transverse processes. The laminae of adjacent vertebrae are united by the strong, elastic ligamentum flavum. The spinous processes are joined by the thin, weak interspinous ligament and by the more superficial and strong supraspinous ligament. Adjacent transverse processes are joined by the intertransverse ligaments which form rounded cords in the thoracic region, but are thin and membranous in the cervical and lumbar regions (Kapandji, 1974; Romanes, 1976; Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

The articulation of the fifth lumbar vertebra with the base of the sacrum is very similar to that of the joints between the lumbar vertebrae. However, the lumbosacral intervertebral disc is very thick and is more wedge-shaped than the lumbar discs in order to accommodate the 30° angle between the two bones. Stability is enhanced by the widely spaced sacral superior articular processes and by the strong iliolumbar ligaments (Romanes, 1976; Lindh, 1980; Williams & Warwick, 1980; Clemente, 1985).

At the bilateral synovial sacroiliac joints the sacrum is firmly wedged between the iliac bones and is held in position by the strong sacroiliac, sacrospinous and sacrotuberous ligaments. These structures prevent the sacrum from being displaced downward and rotated anteriorly as a result of the superimposed weight of the vertebral column. The sacroiliac joint transmits the weight of the head, trunk and upper limb to the lower extremities and prevents the direct transmission of ground reaction forces to the vertebral column (Kapandji, 1974; Romanes, 1976; Last, 1978).

INNERVATION OF THE VERTEBRAL COLUMN

According to Bogduk (1983), the anterior and posterior elements of the vertebral column have separate innervation. The anterior element is supplied by the meningeal branches of spinal nerves, also known as the sinuvertebral nerves. These branches arise from the anterior ramus of the spinal nerve immediately after it emerges from the intervertebral foramen, then re-enter the vertebral canal via the intervertebral foramen. They ascend and descend within the vertebral canal to supply afferent and sympathetic fibres to the anterior aspect of the dural sac, blood vessels, posterior longitudinal ligament and anuli fibrosi (Williams & Warwick, 1980; Clemente, 1985). The posterior element is supplied by afferent branches of dorsal rami of spinal nerves. These pass through the intertransverse spaces to supply adjacent apophyseal joints and ligamenta flava (Hollinshead, 1965; Bogduk, 1983; Korkala et al., 1985).

BLOOD SUPPLY OF THE VERTEBRAL COLUMN

The blood supply to the bones and joints of the vertebral column is derived regionally from spinal branches of the vertebral artery, the ascending cervical branch of the inferior thyroid artery, dorsal branches of the posterior intercostal arteries, dorsal branches of lumbar arteries, and the superior branch of the lateral sacral artery. In general, these spinal branches enter the vertebral canal via the intervertebral foramina and then divide into two branches. The first branch supplies the spinal cord and meninges; the second branch divides into ascending and descending branches which anastomose with corresponding branches above and below to form two lateral and one central arterial chains on the posterior surface of the vertebral bodies (Williams & Warwick, 1980; Clemente, 1985).

Venous drainage is via the intricate plexuses that run along the length of the vertebral column. In general, the venous network can be divided into internal and external plexuses which anastomose freely and ultimately drain into the intervertebral veins. The external vertebral venous plexus consists of an anterior and a posterior plexus located on the periphery of the vertebral column anteriorly and posteriorly, while the anterior and posterior parts of the internal venous plexus lie within the vertebral canal. Drainage from the interior of vertebral bodies is via the basivertebral veins which communicate freely with both internal and external plexuses. All of these vessels drain into the intervertebral veins which accompany the spinal nerves through the intervertebral foramina and, in turn, drain into vertebral, inferior thyroid, posterior intercostal, lumbar and internal iliac veins (Williams & Warwick, 1980; Clemente, 1985).

MOVEMENTS OF VERTEBRAL COLUMN

Two adjacent vertebrae and their adjoining soft tissue constitute a motion segment (White & Hirsch, 1971). Each motion segment has three degrees of freedom, the movements possible being flexion and extension about a coronal axis in a sagittal plane, abduction and adduction about a sagittal axis in a coronal plane, and axial rotation about a vertical axis in a transverse plane. In addition, translation can occur in all three planes resulting in antero-posterior, lateral and vertical translation (Farfan, 1973; Panjabi, 1977).

Pure motion in any of the three planes rarely, if ever, occurs because the orientation of the articular surfaces of the apophyseal joints does not coincide exactly with the descriptive planes of motion. Vertebral motion is therefore the result of coupling in which motion about one axis is associated with translation or rotation about another axis (Farfan, 1973; Weis, 1975; White & Panjabi, 1978).

Movements in all three planes are possible in the cervical region, however most of the movement occurs in the upper portion of the spine between the skull, the atlas and the axis vertebrae. In the lower portion of the cervical spine, the orientation of the articular facets allows free flexion and extension and substantial lateral flexion, but only slight axial rotation. In addition to facet orientation, lateral flexion is enhanced by the lateral convexity of the inferior surface of the vertebral bodies and the corresponding concavity of the superior surface. Free extension is possible because the cervical spinous processes are short. In addition, the relative thickness of the cervical intervertebral discs contributes to freedom of movement in this region (Hollinshead, 1976; Romanes, 1976; Williams & Warwick, 1980; Cailliet, 1981; Clemente, 1985).

Range of motion is least in the thoracic region. The orientation of the articular facets allows rotation and lateral flexion, but restricts flexion and extension. In addition, flexion is limited by the ligamentum flavum, and extension is limited by the overlapping spinous processes.

All movement is severely restricted by the ribs and sternum, and also by the thin intervertebral discs (Weis, 1975; Romanes, 1976; Last, 1978; Williams & Warwick, 1980, Clemente, 1985). Although intersegmental movement in the thoracic spine is limited, the cumulative motion for the entire region is substantial (White & Panjabi, 1978).

In the lumbar region, the orientation of the articular facets allows free flexion, extension and lateral flexion; rotation is restricted. The intervertebral discs in this region are large and contribute to the range of motion in sagittal and coronal planes. The short spinous processes present no impediment to extension (Romanes, 1976; Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

All motions are possible at the lumbosacral joint, however flexion and extension are most free and rotation is least free because of the orientation of the articular facets. The intervertebral disc between the fifth lumbar and first sacral vertebrae is the thickest in the vertebral column, contributing to the range of motion. Axial rotation is also limited by the strong iliolumbar ligament (Basmajian, 1980; Williams & Warwick, 1980).

Little movement is possible at the sacroiliac joints because of the shape and orientation of the articular surfaces. The disposition of the intrinsic and extrinsic ligaments restricts motion to slight rotation about a coronal axis (Basmajian, 1980; Williams & Warwick, 1980; Clemente, 1985).

DEEP MUSCLES OF THE BACK

The deep back muscles extend from sacrum to occiput and are variably arranged in two columns on either side of the vertebral spinous processes. The deep back muscles are further subdivided into superficial and deep groups (Basmajian, 1980).

Erector spinae, the superficial group, has an extensive origin from the spinous processes of the lower two thoracic and all the lumbar vertebrae, the median and lateral sacral crests, and the posterior aspect of the iliac crest. In the lower lumbar area it forms one muscle mass; in the upper lumbar area it divides into three bundles which ascend to insert into vertebrae, ribs and skull. Spinalis, the most medial bundle, extends between the upper lumbar and upper thoracic vertebral spines. Longissimus, extending from the lumbar area to the posterior aspect of the skull, constitutes the intermediate bundle. In the thoracic region it attaches to the tips of the transverse processes of thoracic vertebrae. The lateral bundle, iliocostalis, also extends from the lumbar region to the posterior aspect of the skull. In the thoracic region it attaches to the angles of the lower six or seven ribs, and extends in relays from the angles of these ribs to those of the upper six ribs (Hollinshead, 1976; Williams & Warwick, 1980; Clemente, 1985).

Transversospinalis constitutes the deep group and is made up of three muscle bundles which fill the gutter between spinous and transverse processes. The bundles run obliquely upward and medially from transverse to spinous processes and extend the length of the column from sacrum to occiput. The most superficial bundle, semispinalis, is not found below the thoracic region; it arises from the transverse processes of the sixth to the twelfth thoracic vertebrae and is inserted into the spinous processes of the upper four thoracic and lower two cervical vertebrae. The intermediate bundle, multifidus, is the largest and has an extensive origin from the dorsal aspect of the sacrum, the aponeurosis of the overlying erector spinae, the posterior superior iliac spine, and the lumbar and thoracic transverse processes. It attaches by way of fascicles to the spinous processes of vertebrae two to three levels above. The deepest bundle, rotatores, is only represented well in the thoracic region. Its fascicles run from the transverse processes to the spinous process of the vertebra next above (Hollinshead, 1976; Last, 1978; Basmajian, 1980; Williams & Warwick, 1980; Clemente, 1985).

In the lumbar and thoracic regions both groups of deep back muscles are innervated by dorsal rami of spinal nerves. Blood supply is from the posterior intercostal, subcostal and lumbar vessels (Basmajian, 1980; Williams & Warwick, 1980).

The specific actions and functions of the deep back muscles are not yet entirely clear. However, it is generally held that both groups of muscles control trunk flexion from the orthograde position. Erector spinae is considered to be the chief extensor of the spine, is a side flexor and also possibly assists in rotation. Transversospinalis is considered to be a rotator and extensor of the vertebral column (Williams & Warwick, 1980; Clemente, 1985).

ABDOMINAL MUSCLES

Four large, flat muscles constitute the abdominal wall - rectus abdominis, obliquus externus abdominis, obliquus internus abdominis, and transversus abdominis.

Rectus abdominis is a long strap muscle extending from the crest of the pubis to the front of the xiphoid process and adjacent cartilages of ribs five to seven. It is separated from its fellow by the fibrous linea alba and is enclosed in a sheath formed by the aponeurosis of the other three abdominal muscles (Williams & Warwick, 1980; Clemente, 1985).

The external and internal oblique muscles extend diagonally across the abdomen, the middle fibres of one muscle running at right angles to those of the other.

External oblique originates from the external surface of the lower eight ribs and radiates donwards and medially to attach to the anterior half of the iliac crest, the posterior border being free. The middle and upper fibres end in an aponeurosis that is attached to the pubic tubercle and the length of the linea alba. The aponeurosis of external oblique contributes to the anterior lamina of the sheath of rectus abdominis. Between the anterior superior iliac spine (ASIS) and the pubic tubercle the lower free border of the aponeurosis forms the inguinal ligament (Basmajian, 1980; Williams & Warwick, 1980; Clemente, 1985).

Internal oblique arises from the thoracolumbar fascia, the anterior two-thirds of the iliac crest and the lateral two-thirds of the inguinal ligament. Its fibres fan upward and medially, the posterior fibres are almost vertical and unite the iliac crest and the rib cage while the uppermost fibres form a short, free superomedial border. The fibres arising from the inguinal ligament arch downward and medially to attach to the medial part of the pecten pubis while the middle fibres end in an aponeurosis. Above the level of the ASIS the aponeurosis of internal oblique splits at the lateral border of the rectus abdominis to contribute to both anterior and posterior laminae of the rectus sheath. Below the level of the ASIS the aponeurosis passes anterior to rectus abdominis, contributing only to the anterior lamina of its sheath (Williams & Warwick, 1980; Clemente, 1985).

The transversus abdominis is a thin muscle that takes origin from the thoracolumbar fascia, the anterior two-thirds of the iliac crest and the lateral one-half of the inguinal ligament. Its fibres run horizontally and end in an aponeurosis which superiorly blends with the linea alba and inferiorly inserts into the pecten pubis with the inferior fibres of internal oblique. The aponeurosis of transversus abdominis contributes to the posterior lamina of the rectus sheath above the level of the ASIS, and to the anterior lamina of the sheath below the level of the ASIS (Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

The abdominal muscles are innervated by the ventral rami of the lower six thoracic spinal nerves. Internal oblique and transversus abdominis also receive innervation from the ilio-inguinal and iliohypogastric branches of the first lumbar spinal nerve. The blood supply to the lateral abdominal wall is from the musculophrenic, lumbar, iliolumbar, deep circumflex iliac and inferior epigastric vessels. The upper part of rectus abdominis is supplied by the superior epigastric vessels, the lower by the inferior epigastric (Romanes, 1976; Last, 1978; Williams & Warwick, 1980; Clemente, 1985).

Electromyographic studies of the abdominal musculature have been summarized in the major anatomy textbooks. The abdominal musculature supports the viscera during sitting and standing. Contraction of the abdominal muscles raises the intra-abdominal pressure, converting the trunk into a rigid pillar and thus protecting the lumbar spine during weight lifting. This raising of the intra-abdominal pressure is also required during forced expiration and expulsion of abdominal contents (Kapandji, 1974; Williams & Warwick, 1980). In movements of the trunk, head raising and trunk flexion from the supine position are brought about by contraction of rectus abdominis assisted by the obliques; all muscles are active during lateral flexion from this position. During extension of the trunk from the orthograde position all muscles contract to prevent loss of equilibrium. Trunk rotation is brought about by contraction of the contralateral external oblique and the ipsilateral internal oblique (Hollinshead, 1976; Romanes, 1976; Last, 1978; Basmajian, 1980; Williams & Warwick, 1980; Clemente, 1985).

THORACOLUMBAR FASCIA

On the posterior aspect of the trunk, a sheet of connective tissue extends from sacrum to cervical region. It encloses the deep muscles of the back and provides attachment for two abdominal muscles and a superficial back muscle.

The thoracolumbar fascia is thin in the thoracic region and is attached medially to the vertebral spinous processes and supraspinous ligaments, laterally to the angles of the ribs and fascia covering the intercostal muscles (Williams & Warwick, 1980; Clemente, 1985).

In the lumbar region the thoracolumbar fascia is thick and attaches inferiorly to the median and lateral sacral crests, iliolumbar ligaments and iliac crests. It is split into three layers. The posterior layer passes superficial to erector spinae, enclosing it, and attaches medially to the lumbar spinous processes and supraspinous ligament. The middle layer lies between erector spinae and quadratus lumborum and is attached medially to the tips of lumbar transverse processes. The anterior layer lies between quadratus lumborum and psoas major and is attached medially to the anterior surface of lumbar transverse processes (Romanes, 1976; Basmajian, 1980; Williams & Warwick, 1980; Clemente, 1985).

The posterior and middle layers fuse at the lateral margin of erector spinae and then fuse with the anterior layer at the lateral margin of quadratus lumborum to form the aponeurotic origin of transversus abdominis and internal oblique. The posterior layer of the thoracolumbar fascia blends with the aponeurosis of latissimus dorsi and provides this muscle with attachment to the lumbar and sacral spinous processes (Williams & Warwick, 1980; Clemente, 1985).

BIOMECHANICS OF SAGITTAL TRUNK MOTION

Although there is no universally accepted definition of the term, there is general agreement that "biomechanics is the study of the structure and function of biological systems by means of the methods of mechanics"(Hatze, 1974, p.189). According to LeVeau (1977), mechanics comprises the areas of statics and dynamics, with statics being the study of bodies at rest or in equilibrium, and dynamics being the study of bodies in motion. Dynamics in turn can be subdivided into kinematics and kinetics. Frankel and Nordin (1980) define kinematics as "the branch of mechanics that deals with the motion of a body without reference to force or mass" (p.294), and kinetics as "the branch of mechanics that deals with the motion of a body under the action of given forces" (p.294). Therefore, a complete biomechanical description of trunk motion in any plane includes both kinematics and kinetics. A comprehensive review of trunk biomechanics is outside the scope of this thesis, however an overview of trunk kinematics and kinetics and a brief discussion of the theories that have been developed to explain experimental findings is necessary in order to understand the possible function of the trunk and its components during locomotion.

According to Yettram and Jackman (1980), the vertebral column can be considered, in an engineering sense, as both a mechanism and a structure. As a mechanism, it is a device for transmitting movement and, as a structure, it is a device for transferring force. Therefore both of these functions must be taken into consideration when discussing vertebral biomechanics.

Forward inclination of the trunk is a component of many daily activities and, during forward inclination, the upper part of the body must be both supported and balanced. Support requires transmission of internal and external forces, while balancing requires the generation of an adequate extensor force to keep the trunk from falling forward (Soderberg, 1986).

ANTHROPOMETRY

The trunk consists of pelvic, abdominal and thoracic segments (Winter, 1979). In practice however, all segments of the upper body are commonly used in biomechanical analyses and are termed Head, Arms and Trunk (H.A.T.) (Winter, 1979). Experiments conducted by Asmussen and Klausen (1962) showed that the line of gravity of the H.A.T. passed, on average, anterior to the thoracic curve and one centimeter anterior to the centre of the fourth lumbar vertebral body. Therefore, gravity will tend to increase the anterior curvature of the thoracic curve and decrease the posterior curvature of the lumbar curve. According to LeVeau (1977), the centre of mass of the total body is located anterior to the second sacral body, while that of the trunk lies approximately anterior to the eleventh thoracic vertebral body. To calculate the position of the centre of mass of the H.A.T. and of the trunk, Winter (1979) gave the figures of 62.6% and 50% respectively, of the length of the segment from C_7 to the greater trochanter, measured from the greater trochanter. The mass of the H.A.T. and of the trunk can be calculated as 67.8% and 49.7% respectively, of total body mass (Winter, 1979). Reid (1984), using computed tomography, calculated that the mean centre of mass of the trunk was located at 49.35% of the total segment length of the trunk measured from the greater trochanter to the suprasternal notch, while the mean segmental mass of the trunk was 52.58% of the total body mass.

RANGE OF MOTION

A variety of methods has been used to measure sagittal range of motion in the vertebral column. Twomey and Taylor

(1983), experimenting with postmortem material, found that the range of lumbosacral flexion was approximately 30°. Pearcy et al. (1984) used roentgenography to measure sagittal plane movement of the lumbar spine _in vivo_ and found the total range of lumbosacral motion to be 70°. In addition, they found that forward displacement or translation occurred only in the upper lumbar levels. Non-invasive measurement _in vivo_ is difficult because of the multisegmented nature of the vertebral column, however Loebl (1967) and recently Mayer et al. (1984) used an inclinometer while Anderson and Sweetman (1975) used a hydrogoniometer, to measure sagittal range. Based on reports in the literature, the American Academy of Orthopaedic Surgeons (1965), Farfan (1973), and White and Panjabi (1978) have compiled composite values for sagittal plane motion.

A comparison of published values for sagittal trunk motion is presented in Table 1. It can be seen that the compiled values for lumbosacral range are in close agreement with the recent _in vivo_ measurements of Pearcy et al.(1984). It can also be seen from the data of Anderson and Sweetman (1975) and of Pearcy et al. (1984), that generally there is a greater range of flexion than of extension in the $T_7 - S_1$ vertebral segment.

Functionally, the range of sagittal plane motion is not only the result of motion at individual thoracolumbar

Table 1. EXPERIMENTAL VALUES FOR RANGE OF TRUNK MOTION IN SAGITTAL PLANE MEASURED IN DEGREES

	STUDY 1			STUDY 2			STUDY 3			STUDY 4			STUDY 5			STUDY 6			STUDY 7		
	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.	Flex.	Ext.	Tot.
C_7-T_1															9						
T_1-T_2															4						
T_2-T_3															4						
T_3-T_4															4						
T_4-T_5															4						
T_5-T_6															4						
T_6-T_7						32									5						
T_7-T_8															6						
T_8-T_9	85	30													6						
T_9-T_{10}															6						
$T_{10}-T_{11}$															9						
$T_{11}-T_{12}$										39	18				12						
$T_{12}-L_1$						20									12						
L_1-L_2							3	3	7						12				8	5	13
L_2-L_3							3	3	7						14	30			10	3	14
L_3-L_4						65	9	9	18						15				12	1	13
L_4-L_5							12	10	22						17				13	2	16
L_5-S_1							12	6	18						20				9	5	14

1 = American Academy of Orthopaedic Surgeons, 1965 *
2 = Loebl, 1967 **
3 = Farfan, 1973 (rounded off) *
4 = Anderson & Sweetman, 1975 (rounded off) **
5 = White & Panjabi, 1978 *
6 = Twomey & Taylor, 1983 (cadaver material) **
7 = Pearcy et al., 1984 (rounded off) **
* = compiled values
** = experimental values

intervertebral joints, but also includes movement of the pelvis about the hip joints (Davis et al., 1965; Farfan, 1975). During anterior trunk motion or flexion, lumbar intervertebral joint motion accounts for 60° of the movement with a further 25° taking place at the hip joints (Mayer et al., 1984; Patwardhan et al., 1985). During posterior trunk motion or extension from the fully flexed position, initially the pelvis rotates posteriorly at the hip joints, followed by extension of the lumbar spine (Floyd & Silver, 1951; Farfan, 1975; Ortengren & Andersson, 1977; Kippers & Parker, 1984).

FORCES ACTING ON THE TRUNK

The trunk is subjected to forces that result from body weight, muscle activity, the passive elastic components of muscles and ligaments, and externally applied loads (Lindh, 1980). Soderberg (1986) identified five types of forces: compression, tension, shear, bending and torsion. The first four forces occur during sagittal motion of the trunk.

COMPRESSION

This occurs when equal and opposite loads are applied toward the surface of a structure, resulting in shortening and widening of the structure (Frankel & Nordin, 1980; Rodgers & Cavanagh, 1984). Both <u>in vitro</u> and <u>in vivo</u>

experiments have been conducted to determine the compression forces at the intervertebral joint.

Hirsch (1955) subjected the second and fourth lumbar intervertebral discs of postmortem material to compression and found that this caused the disc to bulge circumferentially. He applied both sudden and sustained loads to the discs and concluded that in sudden loading the disc acts as a shock absorber. Bartelink (1957), also using postmortem material, found that a mean compressive force of the equivalent of 3200 newtons (N) resulted in failure (Van Nostrand's Scientific Encyclopedia, p.665) of the intervertebral disc. Adams et al. (1980), in an effort to simulate physiological conditions, applied off-centre compressive loads to a lumbar intervertebral joint and observed that these loads resulted in tension in the posterior ligaments which, in turn, gave rise to a high compressive force within the intervertebral disc. Nachemson and Morris (1964) used a pressure transducer to measure indirectly compression loads on lumbar intervertebral discs _in vivo_. They observed that the lower lumbar discs have to support total loads of 880 to 1180 N in relaxed upright standing. Nachemson (1981) later reported revised values of 500 N for standing at ease, 600 N for 20° of trunk flexion and 1000 N for 40° of trunk flexion. Schultz, Andersson, Ortengren, Haderspeck & Nachemson (1982) measuring the L_3

intradiscal pressure *in vivo*, derived values of 440 N for standing at ease and 1450 N for 30° trunk flexion.

Lin et al. (1978) investigated the role of the posterior vertebral elements in both central and off-centre compression of the lumbar spine. Using postmortem material, they observed that in central compression (which rarely occurs under physiological conditions) and in anterior compression, the majority of the load was borne by the vertebral bodies and intervening discs. However, posterior compression resulted in part of the load being dispersed through the articular processes. They concluded that in hyperextension of the lumbar spine, the posterior elements of the intervertebral joint increase the load-bearing capacity of the lumbar spine. They also found that compressive loads of 1560 to 5800 N resulted in failure of the specimens and that this failure occurred in the vertebral body, not the anulus fibrosus.

Lorenz et al. (1983) studied the role of the lumbar articular facets in vertebral compression. They found that in cadaver material the average peak facet pressures were higher in posterior loading of the vertebral body than in central loading. Also, while peak facet pressures were high, the loads borne by the facets were not large and the facet load decreased with increasing central loads. They noted that failure of the vertebral body occurred with central loads of 1766 to 1864 N.

Researchers have also used equations in a mathematical model to predict the compression force on the intervertebral joint. Bartelink (1957) calculated that the compression on the fifth lumbar intervertebral disc would be 8900 N when a 91 kg load was lifted. Morris et al. (1961) derived values of 9225 N for the same load. Gracovetsky et al. (1981) calculated that combined body weight and psoas muscle activity would produce a compression force of 784 N at the lumbosacral intervertebral joint during relaxed upright standing with no external load.

TENSION

This occurs when equal and opposite loads are applied away from the surface of a structure, resulting in a lengthening and narrowing of the structure (Frankel & Nordin, 1980; Rodgers & Canvanagh, 1984). During lumbar flexion the centre of rotation lies within the intervertebral disc (Adams et al., 1980), therefore it can be expected that tension forces will be applied to the posterior structures of the intervertebral joint. Because measurement of tension *in vivo* would have to be invasive and would be very difficult to carry out, the majority of studies of tension force at the intervertebral joint have been performed on postmortem material. Mathematical models have been developed to predict tension values *in vivo* based on *in vitro* observations.

In studies of the intervertebral disc, Hirsch (1955) observed that sudden loading caused the disc at first to compress and then, fractions of a second later, to rebound. This rebound resulted in tension forces being applied to the disc. Nachemson and Morris (1964), measuring intradiscal pressure in vivo, calculated that, in normal discs, tensile forces of "sixty to eighty kilograms per square centimeter" (p.1091) existed in the posterior part of the anulus fibrosus. Galante (1967) experimented with sections of anulus fibrosus cut from postmortem lumbar discs and found that during tensile loading the anulus extended freely along the vertical axis. He observed that the tensile strength of the tissue of the anulus fibrosus was similar to that of tendon. Markolf (1972) studied the load-deformation behavior of intervertebral discs from thoracic and lumbar cadaver spines and found that the discs exhibited less stiffness when tension was applied to them than when compression was applied.

The behavior of the posterior ligamentous system has also been studied. Tkaczuk (1968) loaded the anterior and posterior longitudinal ligaments of isolated lumbar spines until failure. He observed that the average failure loads for these ligaments was the equivalent of 21 N for the anterior and 19 N for the posterior. He also observed that both longitudinal ligaments were prestressed by the pressure of the intervertebral disc. From his observations in vivo, Farfan (1975) concluded that the entire posterior ligamentous system was not subject to tension forces until approximately 60° of lumbar flexion had occurred. Adams et al. (1980) measured the resistance provided by the posterior ligamentous system in full and half lumbar flexion. They concluded that in full flexion the capsular ligaments accounted for 39% of the resistance, the intervertebral disc for 29%, the supraspinous and interspinous ligaments for 19%, and the ligamentum flavum for 13%. In half flexion, the intervertebral disc accounted for 38% of the resistance, the ligamentum flavum for 28%, the capsular ligments for 25%, and the supraspinous and interspinous ligaments for 8%. Panjabi et al. (1982) studied autopsy specimens of the L_{3-4} and L_{4-5} intervertebral joints. They found that in flexion the highest ligamentous tensions occurred in the supraspinous and interspinous ligaments, but also found differences between predicted and observed strain values for the supraspinous ligament, possibly due to deformation of the neural arch. Chazal et al. (1985) found that the intertransverse and posterior longitudinal ligaments and the ligamentum flavum were most resistant to tension during physiological ranges of lumbar motion in postmortem material.

Mathematical models for tensile forces applied to the vertebral column were developed by Gracovetsky et al. (1977;

1981). Based on *in vivo* observation, they postulated that the posterior ligamentous system is slack from zero to 40° of trunk flexion, but is taut from 40° to full flexion. They calculated that ligament tension during trunk flexion could reach the equivalent of 8000 N. Furthermore, because ligamentous tissue is passive, any tension that is applied to a ligament depends entirely upon external loads imposed upon it. They also considered the role of the thoracolumbar fascia. Experiments had shown that when longitudinal tension was applied to the thoracolumbar fascia it stretched longitudinally by 30% of its resting length but was also narrowed by 30% of its width. They concluded that a tension force applied to the lateral margins of the thoracolumbar fascia to prevent it from narrowing would result in an increase in the longitudinal tension in the structure.

SHEAR

This occurs when a load is applied to the surface of a structure resulting in angular deformation of the structure (Frankel & Nordin, 1980). Antero-posterior translation occurs between adjacent vertebrae at all levels of the column (Panjabi, 1977), subjecting the intervertebral disc, apophyseal joints and the ligaments to shear forces. The magnitude of the shear forces will increase during flexion of the trunk as gravity will tend to increase the amount of translation occurring at each joint (Soderberg, 1986).

Galante (1967) found that sections of anulus fibrosus cut from postmortem lumbar discs provided high resistance to shear forces. Based on in vivo observations, Farfan (1975) calculated that contraction of the extensor musculature of the back gave rise to shear forces at the intervertebral joints. He proposed that the shear force could be resisted by the apophyseal joints but, because of the limited strength of the neural arch, shear forces in the opposite direction would be required to protect both articular facets and intervertebral discs from damage. The role of the posterior ligaments and the apophyseal joints in resisting shear forces was investigated by Lin et al. (1978) and they concluded that these posterior structures prevent the vertebrae from slipping forward on one another. This was corroborated by Twomey and Taylor (1983) who sectioned the pedicles of postmortem lumbar spines and observed that this resulted in anterior translation of the vertebral body.

Gracovetsky et al. (1981) noted that the fibres of the midline posterior ligaments (supraspinous and interspinous) are angled and proposed that because of this angulation, these ligaments could produce a shear force in the opposite direction to that produced by anterior loads applied to the vertebral column.

BENDING

This occurs when a load is applied to a structure in such a way that the structure bends about an axis. The structure will therefore be subjected to a combination of compression and tension forces on opposite surfaces (Frankel & Nordin, 1980; Rodgers & Cavanagh, 1984).

During sagittal plane motion of the lumbar spine the axis of motion is considered to lie within the intervertebral disc, therefore the anterior aspect of the intervertebral joint will be subjected mainly to compression forces while the posterior aspect will be subjected to tension forces (Adams et al., 1980).

The magnitude of the bending force or moment is dependent not only on the applied force but also on the perpendicular distance between the point of application of the force and the axis of motion (Rodgers & Cavanagh, 1984). Lindh (1980) calculated that in relaxed standing, the perpendicular distance between the centre of the L_{4-5} intervertebral disc and the line of gravity is two centimetres. Assuming a body weight of 660 N (67 kg) and using Winter's (1979) value of .678 of Total Body Weight for the H.A.T., the forward bending moment in relaxed standing is 9 newton metres (Nm). However, when the trunk is inclined forward, the line of gravity is displaced anteriorly and this will increase the perpendicular distance or lever arm between the line of gravity and the axis of motion. At approximately 30° of trunk flexion the line of gravity lies roughly 25 cm from the axis of motion giving rise to a bending moment of 112 Nm.

INVESTIGATIONS OF TRUNK FUNCTION

Many investigations have been carried out to determine the function of erector spinae and rectus abdominis during sagittal plane motion.

Electromyography was used by Floyd and Silver (1951) to examine the activity of erector spinae during trunk flexion. They found that upright standing resulted in slight myoelectric activity, and that increasing angles of forward flexion gave rise to increasing levels of myoelectric activity. However, as the trunk approached full forward flexion, activity in erector spinae ceased and did not begin again until after trunk extension from maximum forward flexion was well underway. In further experiments, Floyd and Silver (1955) found that the position at which erector spinae became silent during trunk flexion was approximately the same as the position at which muscle activity reappeared during trunk extension from the fully flexed position. They also noted that trunk flexion was a combination of flexion of the vertebral column and flexion at the hip joint. They concluded that the erector spinae contracted to control trunk motion but that the weight of the trunk in full flexion was borne by the ligaments of the vertebral column. Kippers and Parker (1984) used photography and EMG to determine the positions of electrical silence and onset during trunk flexion and return to upright standing. They found that the onset and cessation of electrical silence occurred at approximately the same position: 90% of vertebral flexion and 60% of hip flexion.

Rectus abdominis was investigated by Floyd and Silver (1950) who found that trunk extension from the upright position elicited myoelectric activity whereas trunk flexion from the same position resulted in electrical silence. Similar results were obtained by Partridge and Walters (1959) and Flint and Gudgell (1965). The latter investigators concluded that rectus abdominis was active to position the trunk and balance the extensor moment produced by trunk extension.

The simultaneous myoelectric activity of erector spinae and rectus abdominis has also been investigated. Asmussen and Klausen (1962) found that erector spinae was active and rectus abdominis was inactive in quiet standing, concluding that the erect position of the vertebral column was maintained by erector spinae acting as an antagonist to the force of gravity. They also observed that these muscles were alternatively active during trunk sway with erector spinae active during forward sway and rectus abdominis during backward sway. They concluded that the muscles functioned like guy wires to maintain trunk stability during movement. Ekholm et al. (1982) studied the electrical activity of trunk muscles during various lifting tasks and found that rectus abdominis was inactive while erector spinae responded to increasing loads by becoming active earlier during trunk extension from full flexion. Schultz, Andersson, Ortengren, Bjork and Nordin (1982) applied resistance to trunk motion and noted that erector spinae activity closely reflected the magnitude of the net flexor moment while rectus abdominis activity reflected the net extensor moment.

It is well known that heavy lifting activities are invariably accompanied by a Valsalva maneuver in which the thoracic and abdominal cavities are pressurized by voluntary muscle contraction (Kapandji, 1974). Electromyographic studies of the Valsalva maneuver by itself have shown that erector spinae, rectus abdominis and the lateral abdominal oblique muscles are all active (Floyd & Silver, 1950; Campbell, 1952; Floyd & Silver, 1955; Ono, 1958; Bearn, 1961; Hatami, 1961a,b; de Sousa & Furlani, 1974; Strohl et al., 1981). In addition, Nachemson and Morris (1964) observed that intra-abdominal pressure increased during an isolated Valsalva maneuver.

The intra-abdominal pressure (IAP) generated during actual and simulated heavy lifting activities has also been studied. Davis and Troup (1964) found that the IAP increased during lifting while Grew (1980) observed that the IAP increased when flexor and extensor moments were each applied to the trunk. Kumar and Davis (1983) reported a close relationship between increased EMG and IAP in progressive loading of the vertebral column. Mairiaux et al. (1984) measured IAP during lifting in a fixed upright position and found a linear relationship between IAP and the moment of force at L_{4-5}. Hemborg et al. (1985) measured IAP and EMG of trunk muscles to determine which muscles contributed most to the rise in abdominal pressure. They concluded that the diaphragm was the most important muscle and attributed some importance to transversus abdominis, although acknowledging that little is known about this muscle.

Simultaneous measurement of myoelectric activity, intra-discal pressure (IDP) and IAP during lifting has also been done. Andersson et al. (1976) measured these parameters during lifting of a 100 N load with and without performing a Valsalva maneuver and found that disc pressure did not change significantly when the Valsalva maneuver was performed. The same techniques were used by Ortengren et al. (1981) who reported a linear relationship between load, EMG, IAP and

IDP. However, Schultz, Andersson, Ortengren, Haderspeck et al. (1982) reported low correlations between IDP and IAP during lifting and concluded that the level of IAP was not a good indicator of trunk loads.

DEVELOPMENT OF A THEORY OF TRUNK FUNCTION

The manner in which forces are balanced and transmitted by the trunk is not yet clear, therefore a number of theories of trunk function have been proposed. Any theory of trunk function must incorporate, reconcile and explain the data from the various <u>in vitro</u> and <u>in vivo</u> experiments that have been conducted on the bones, joints and muscles of the vertebral column.

A simple theory of trunk function in the sagittal plane is based upon the concept of moments (Tichauer, 1971). The centre of the lowest lumbar disc is considered to be a fulcrum from which extend a long anterior and a short posterior lever arm. The anterior arm lies between the disc centre and the anterior body wall while the posterior arm extends from the disc centre to the posterior limit of the erector spinae muscle. Anterior flexor moments are generated by the weight of the upper body and are balanced by posterior extensor moments generated by the extensor muscles of the back. Because the length of the anterior lever arm is approximately four times longer than the length of the posterior lever arm, the extensor force must be about four times the magnitude of the flexor force under conditions of equilibrium.

However, Bartelink (1957), using this theory, calculated that the moment produced by holding a 45 kg weight would generate a compression force at the fulcrum sufficient to cause failure of the disc or of the vertebral body, and recognized that some mechanism must exist to decrease the load on the disc. He proposed that this mechanism consisted of pressurization of the abdominal cavity by contraction of the transversus abdominis muscle during a Valsalva maneuver. An increase in pressure would convert the abdominal cavity into a fluid ball that would resist deformation and would transmit some of the load of the upper body to the pelvis. An increase in pressure would also give rise to a distending force between the thoracic and pelvic diaphragms which would produce an anterior extensor moment and would therefore decrease the net flexor moment and the compressive load on the disc. Morris et al. (1961) found that results of their experiments were congruent with Bartelink's proposal and used it to calculate the actual loads on the spine.

Bearn (1961) had another explanation for the increase in IAP seen during heavy lifting. He stated that contraction of the abdominal muscles would tend to flex the trunk and this would require synergic contraction of erector spinae to counteract the unwanted motion. The contraction of erector spinae would increase the magnitude of the posterior extensor moment which, in turn, would increase the compressive load on the disc. He observed that latissimus dorsi muscle was active during heavy lifting and proposed that the increase in IAP was required to prevent chest compression due to the strong contraction of latissimus dorsi. Bearn (1961) also commented on the finding of electrical silence in erector spinae during full trunk flexion. He stated that in this postion a flexor moment would be counterbalanced only by the tension in the interspinous ligaments and that any additional load might create a moment of sufficient magnitude to rupture the ligaments.

Farfan (1975) proposed a mechanism that accounted for the activity of the abdominal muscles, the electrical silence in erector spinae after 60° of trunk flexion, the fact that loads are borne entirely by ligaments, and the observed motion of the pelvis. He had observed that during trunk flexion from upright the extensor muscles worked alone to lower the trunk to approximately 60°, at which point tension developed in the previously slack posterior ligaments. Also, during trunk extension from full flexion, the lumbar spine remained flexed until the 60° point, thus maintaining tension on both posterior ligaments and muscles. Farfan proposed that there was a limit to the amount of extensor moment that the back muscles could generate; past that limit, ligament tension was required to support heavy loads while the turning moments would be supplied by the large glutei and hamstring muscles. He concluded that flexion was the optimum postion of the lumbar spine during lifting and that this postion was produced by contraction of the abdominal muscles, flexors of the trunk, during a Valsalva maneuver. He also stated that back muscle contraction and forward trunk inclination generate a large shear force at the intervertebral joint and proposed that contraction of the internal oblique muscle would result in a shear force of the opposite direction. Thus contraction of the abdominal muscles would serve to maintain the lumbar spine in the optimum position and also to decrease the shear force at the intervertebral joint.

Gracovetsky et al. (1977) agreed with Farfan that, without the posterior ligaments, the back muscles would be unable to sustain a heavy load. They noted that the supraspinous ligament and the thoracolumbar fascia both lie posterior to the deep back muscles and proposed that these two structures were the most effective elements to balance flexor moments at the intervertebral joints because they have a longer posterior lever arm than any of the back muscles. They also proposed that contraction of the underlying erector spinae would create hydraulic pressure that would push the thoracolumbar fascia posteriorly, increase tension in the midline ligaments and flatten the lumbar spine. Midline ligament tension would also be generated by an increase in IAP, caused by contraction of abdominal muscles, which would tend to push the thoracolumbar fascia posteriorly.

Building on this modified theory, Farfan (1978) proposed that stretching of the supraspinous ligament would cause it to straighten out between the third lumbar and second sacral spinous porcesses. Because the ligament is attached to the spinous processes of L_4 and L_5 and these lie within the lumbar lordosis, straightening of the supraspinous ligament would cause the fourth and fifth lumbar vertebrae to be pulled backwards. This would create a backward shear force which would tend to cancel the anterior shear produced by the weight of the upper body and any external load. Therefore the posterior ligaments serve not only to balance flexor moments, but also to neutralize anterior shear.

Gracovetsky et al. (1981) further modified the theory. They argued that ligaments, because they are passive, must be under active control of muscle at all times in order to modify their tension. They proposed that the longitudinal tension in the thoracolumbar fascia could be modified by contraction of the abdominal muscles, specifically internal oblique and transversus abdominis. These muscles are attached to the lateral margins of the thoracolumbar fascia and their contraction during trunk flexion would cause an antero-lateral force to be applied to the fascia, preventing it from narrowing and therefore increasing longitudinal tension in the structure. The correct angle of pull for the abdominal muscles is dependent upon the shape of the abdominal cavity which is maintained by the IAP generated by the contraction of the external oblique muscle. Therefore, the abdominal muscles, acting indirectly through the thoracolumbar fascia, generate an extensor moment to counterbalance the net flexor moment.

Bogduk and MacIntosh (1984) proposed further refinements to the theory based on their observations of the structure of the thoracolumbar fascia. They noted that the posterior layer of the fascia is split into two laminae, superficial and deep. The fibres of the superficial lamina are disposed in a caudo-medial direction while those of the deep lamina run in a caudo-lateral direction, giving the posterior layer a lattice-work appearance. The superficial and deep laminae fuse with the middle layer of the thoracolumbar fascia at the lateral border of erector spinae to form a raphe and both the internal oblique abdominal and the transversus abdominis have partial attachment to this raphe.

The superficial lamina provides the latissimus dorsi muscle with attachment to the lumbar spinous processes. Because of the orientation of the fibres of the lamina, contraction of latissimus dorsi will result in upward and lateral forces being exerted on the lumbar spinous processes. In addition, the superficial lamina acts as a retinaculum for the lower back muscles. The deep lamina also acts as a retinaculum for the lower back muscles, and in addition anchors the spinous processes of L_4 and L_5 to the ilia.

Bogduk and MacIntosh (1984) agreed with earlier proposals that contraction of back muscles results in expansion of the muscle mass which is resisted by the posterior layer of the thoracolumbar fascia, causing an increase in longitudinal tension. However, they postulated that the orientation of fibres in superficial and deep laminae also contributes to the generation of longitudinal tension. These fibres form a series of overlapping triangles with apices at the raphe and bases spanning two vertebral levels in midline. Lateral tension applied at the raphe will be spread over the area of a triangle and will generate reciprocal upward and downward forces at the basal corners of the triangle. These forces will tend to prevent separation of the lumbar spinous processes. Because of the attachment of internal oblique and transversus abdominis to the raphe, contraction of these muscles will generate lateral tension and therefore will generate an extensor moment.

It can be seen that the current theory of trunk function is far more complex than the initial concept of a simple balance of moments about the intervertebral disc. It takes into account the experimental findings about back and abdominal muscle activity, intra-discal pressure, intra-abdominal pressure and ligament strength, however further refinements will doubtless be made as more data are generated.

GAIT

Locomotion is defined as the act of movement from one geographical location to another (Inman et al., 1981). The normal adult commonly engages in bipedal locomotion in the form of walking or running. Walking itself is the process by which the erect body moves forward by continually falling from one limb to another and has been described as a series of prevented catastrophes (Rancho los Amigos, 1978; Soderberg, 1986).

Walking or gait is a repetitive process involving cyclical displacement of the body segments and demanding a co-ordinated response of trunk and limb muscles to gravitational, inertial and ground reaction forces (Gray & Basmajian, 1968; Lamoreux, 1971; Cappozzo et al., 1976). Murray et al. (1964) proposed that the stereotyped nature of walking represented the most energy efficient pattern of movement and further stated that to achieve this degree of efficiency requires highly developed sensory motor control. According to Conrad et al. (1983), the individual must be capable of simultaneously executing the relatively stereotyped motor act of walking, of adapting movement by using afferent input, motor planning and efferent output to meet changing external demands, and of maintaining constant balance by stabilizing the centre of gravity within narrow limits.

Although walking is a comparatively stereotyped process, differences exist between individuals and even within the same individual (Lamoreux, 1971; Inman et al., 1981). However, the magnitude of the inter- and intra-subject variation is considered to be much smaller than that seen in most other patterns of voluntary movement and certain characteristics of walking are universal (Murray et al., 1964; Lamoreux, 1971).

DETERMINANTS OF GAIT

During walking the moving body is supported alternately by the lower limbs in such a way that the contralateral limb swings forward as the body passes over the supporting limb (Inman et al., 1981). According to Perry (1985), the body is divided into a passenger and a locomotor unit during walking. The passenger unit consists of the head, thoracic, upper limb and pelvic segments; these do not contribute directly to the act of walking, but rather are carried relatively passively. The locomotor unit consists of the pelvic and lower limb segments which are the effectors of forward motion. The pelvis is therefore a transition segment, belonging to both the passenger and the locomotor units and joined to each at the highly mobile lumbar and hip joints.

Perry (1985) stated that walking has three components: progression, standing stability and energy conservation. Progression is achieved by the forward fall of the body weight plus the momentum of the swinging limb. Standing stability is maintained by muscle contraction in the trunk segment to ensure vertical alignment, and in the lower limb segments to provide a firm yet mobile column of support. Energy conservation is achieved by substituting momentum for muscle contraction whenever possible and by minimizing the displacement of the centre of gravity of the body from the line of progression.

The centre of gravity of the body is located within the pelvis (Inman et al., 1981). During walking the body moves in all three planes, therefore the centre of gravity will be displaced vertically, horizontally and in the plane of progression. The body is seen to oscillate up and down and to undulate from side to side as it moves through space. These displacements require the expenditure of energy and therefore, because energy conservation is a component of gait, the pattern of joint motion and muscle activity seen during walking is designed to minimize the magnitude of the displacement of the centre of gravity (Saunders et al., 1953; MacConaill & Basmajian, 1977).

Saunders et al. (1953) identified six factors that contribute to decreasing the excursion of the centre of gravity:

1. Pelvic rotation - the pelvis rotates about a vertical axis passing through the hip joint. The range of rotation is approximately 4° to each of right and left sides and the net effect of this movement is to decrease the magnitude of the vertical displacement of the centre of gravity.

2. Pelvic tilt - the pelvis tilts downward approximately 5° to the side of the unsupported limb as the body moves over the supporting limb. The net effect is also to decrease the magnitude of the vertical displacement of the centre of gravity.

3. Knee flexion - the knee joint flexes approximately 15° as the body is carried onto and over the supporting limb. This also decreases the magnitude of the vertical displacement of the centre of gravity.

4. & 5. - Foot and knee mechanisms - motion of the foot about the ankle results in an almost horizontal forward displacement of the knee of the supporting limb. This in turn smooths the path of the hip and pelvis and decreases the vertical displacement of the centre of gravity.

6. Lateral displacement of the pelvis - by shifting the pelvis horizontally, the centre of gravity is displaced laterally over the supporting limb during walking. A lateral shift of only four to five cm is required because the femur and tibia are angled in such a way that the femur is positioned in relative adduction. This angulation allows the feet to be maintained close to the line of progression of the centre of gravity and reduces the distance that the centre of gravity must be shifted in order to position it over the supporting limb.

The need for energy conservation also determines the speed at which walking occurs. Bobbert (1960) defined the optimum walking speed as that at which the energy expended per meter is minimal. It has been found that there is a curvilinear increase in energy expenditure as speed increases (Inman et al., 1981). The optimal speed for level walking varies between individuals, but has been calculated to range from approximately 55 to 80 meters per minute (m/min) for both men and women (Coates & Meade, 1961; Dean, 1965; Milner et al., 1971; Fitch et al., 1974).

KINEMATICS OF GAIT

TEMPORAL EVENTS

Murray et al.(1964) defined the walking or gait cycle as the interval between successive ground contacts of the same foot. They divided the gait cycle into two major phases: Stance and Swing. Stance is that period in which the foot is in contact with the ground, and Swing is that period in which the foot is off the ground and is moving forward. During walking the supporting foot remains in contact with the ground until the advancing foot makes ground contact and the period in which both feet are in contact with the ground is called the Double Support phase. During the gait cycle there are two periods of single limb support, one for each lower limb, and two periods of double limb support.

Each of the Stance and Swing phases can be considered to be composed of different events (Rancho los Amigos, 1978). The events occurring during Stance are:

1. Initial Contact - This represents the moment that the foot makes contact with the ground. In normal individuals, the first area of contact is the heel.
2. Loading Response - The foot is lowered to make full contact with the ground and the body weight begins to advance onto the foot.
3. Mid Stance - With the foot flat on the ground, the body is positioned over the supporting limb.
4. Terminal Stance - The body has moved ahead of the supporting limb and now falls towards the contralateral limb.
5. Pre Swing - The limb is prepared for swing by the shifting of the body weight onto the contralateral limb.

The events during Swing are:

1. Initial Swing - The limb is lifted from the ground and begins to advance.
2. Mid Swing - The limb is moved forward with the tibia perpendicular to the ground.

3. Terminal Swing - The limb advances to its furthest anterior excursion.

It can be seen that ipsilateral Mid Stance and contralateral Initial, Mid and Terminal Swing occur during the first single support period; while ipsilateral Initial, Mid and Terminal Swing and contralateral Mid Stance occur during the second single support period. Ipsilateral Initial Contact and Loading Response occur during the first double support period, while ipsilateral Terminal Stance and Pre Swing occur during the second double support period.

According to Inman et al.(1981), Stance phase begins with Heel Contact and ends with Toe Off; Swing phase begins with Toe Off and ends with Heel Contact. However, the first and last areas of contact are not necessarily the heel or the toe, therefore the more inclusive terms of Initial Contact and End of Weight Bearing are preferable (Quanbury, 1985).

If the gait cycle of one limb is taken to be 100%, the Stance, Swing and Double Support phases of both limbs can be expressed as percentages of this total gait cycle. It has been found that, on average, Stance constitutes 60% and Swing constitutes 40% of the total gait cycle in free-speed walking (Murray et al., 1964, 1966, 1969; Dubo et al., 1976; Cappozzo, 1983; Thurston & Harris, 1983; Kirtley et al., 1985). On average, the two Double Support phases constitute

10% each of the total gait cycle (Murray et al., 1964, 1966; Thurston & Harris, 1983; Kirtley et al, 1985). Murray et al. (1964) found no significant difference between the duration of Stance and Swing phases of right and left lower limbs within and between trials for the same subject. Therefore, if the right gait cycle is considered representative of a total gait cycle, then Initial Contact of the left foot occurs at approximately 50% and End of Weight Bearing occurs at approximately 10% of the total gait cycle (Thurston & Harris, 1983).

Besides considering the phases of gait as percentages of a total gait cycle, the time interval or duration of each phase can be measured (Murray et al., 1966). Investigators have found that the duration of the gait cycle in free-speed level walking ranged from 1.02 to 1.16 seconds (Murray et al., 1964, 1966, 1969; Waters & Morris, 1970; Cappozzo, 1983; Thorstensson et al., 1984). It has been found that the duration of the Stance phase decreased as walking speed increased (Murray et al., 1964, 1966; Kirtley et al., 1985; Nilsson et al., 1985). However, the duration of the Swing phase tended to remain constant regardless of changes in walking speed and therefore accounted for a greater percentage of the total gait cycle as walking speed increased (Murray et al., 1966). The comparative immunity of the Swing phase duration to changes in walking speed is attributed to the fact that Swing is essentialy a passive process in which the advancing limb acts like a pendulum (Kirtley et al., 1985).

The number of steps (successive ground contacts of alternate feet) taken within a one minute interval can also be counted to give the cadence of a particular gait (Drillis, 1958). In the investigations of Grieve and Gear (1966) the cadence of adults during level walking was found to range from 61 to 168 steps per minute, while Thurston and Harris (1983) found a range of 88 to 126 steps per minute. Other investigators have reported narrower ranges of 104 to 113 steps per minute for adults (Drillis, 1958; Murray et al., 1969; Waters et al., 1973; Winter, 1983; Kirtley et al., 1985). Cadence has been shown to be reciprocally related to the duration of the gait cycle and directly related to stride length and walking speed (Murray, 1967; Murray et al., 1969; Winter, 1983; Kirtley et al., 1985).

According to Winter et al. (1976), the natural cadence of an individual is close to that required for minimum energy expenditure and therefore represents the most efficient means of locomotion. Dean (1965) found that cadence was influenced by footwear and ground surface, but not necessarily by leg length. However, the results of Murray et al.(1966)

indicated that leg length may influence cadence as tall subjects could not normally attain as rapid a cadence as short subjects. Thurston and Harris (1983) found that cadence decreased with advancing age.

DIMENSIONS

The linear distance between successive ground contacts of the same foot and alternate feet can be measured to give stride length and step length, respectively (Murray et al., 1964). While the average stride length of men during level walking was found to be 156.5 cm (Murray et al., 1964), stride length has also been expressed as approximately 89% of the total body height at a freely chosen speed of walking (Murray, 1967; Murray et al., 1969). Because stride length is related to height, it follows that it will be greatest in tall subjects and least in short subjects. However, a recent study by Kirtley et al. (1985) showed that there was no significant correlation between stride length and height.

An increase in the speed of walking can be achieved not only by increasing cadence, but also by increasing step length (Murray et al., 1966; Smidt, 1971; Nilsson et al., 1985). Murray (1967) found that stride length decreased with advancing age, resulting in a decrease in walking speed. Kirtley et al. (1985) found a moderate correlation between stride length and age. It has also been found that men over age 65 increased the width of their stride by lateral rotation or out-toeing of the lower limb, possibly to increase the base of support (Murray et al., 1964).

DISPLACEMENT OF UPPER BODY SEGMENTS

Displacement is the change in position of a body and can be either linear or angular. In linear displacement every point within a body segment is displaced along parallel lines while in angular diisplacement the body segment rotates about an axis and therefore travels in an arc. Linear displacement is measured in millimeters or centimeters, angular displacement is measured in degrees or radians (Kelley, 1971; LeVeau, 1977; Gowitzke & Milner, 1980; Rodgers & Cavanagh, 1984).

<u>Linear Displacement</u>

Displacement can occur in vertical, lateral and antero-posterior directions.

Vertical Direction

Using photographic techniques, Saunders et al. (1953) observed that the centre of gravity was displaced up and down during the gait cycle. Subsequent investigators quantified the displacement of the head, trunk and pelvis and found that the mean vertical excursion of all segments was approximately four centimeters. The maximum upward displacement occurred during Mid Stance at 30% and 80% of the gait cycle when the trunk was centred over the single supporting limb. The maximum downward displacement occurred during Double Support at 5% and 55% of the gait cycle when one limb was positioned obliquely forward and the other obliquely backward. They also found that the patterns of vertical oscillation were similar, independent of height or age, however the amplitude of the total vertical excursion decreased with age (Murray et al., 1964, 1966, 1969; Waters & Morris, 1972; Waters et al., 1973; Cappozzo, 1981; Inman et al., 1981; Thurston & Harris, 1983; Thorstensson et al., 1984).

Lateral Direction

Investigators found that the mean total lateral displacement of the upper body segments was approximately five centimeters and the maximum excursion to each side occurred during Single Support at 28% and 75% of the gait cycle. The upper body shifted laterally over the supporting limb during Stance while the trunk shifted back towards the line of progression during Double Support, placing the head in a more central position (Murray et al., 1964, 1966; Waters et al., 1973; Inman et al., 1981; Thurston & Harris, 1983; Thorstensson et al., 1984). Murray et al. (1969) found that the total amount of lateral displacement was greater for older men. Cappozzo (1981) found a high degree of variability both between and within subjects for the pattern of lateral displacement of the head and shoulder girdle, however he observed a more regular pattern of pelvic motion within subjects.

Antero-posterior Direction

Although the body as a whole is carried forward through space during one gait cycle, the upper body segments are not displaced forward at a uniform rate, but rather oscillate forward and backward during the gait cycle. Waters et al. (1973) calculated the magnitude of this oscillation to be approximately .5 cm at the head, 1.8 cm at the T_{10} level and 2.6 cm at the S_2 level. Cappozzo (1981) confirmed that the pelvic segment had a larger displacement than did the thoracic and head segments, and found this to be true at all speeds of progression. It has been found that the body segments reach their maximum forward postion in the oscillation early in Stance phase at approximately 15% and 65% of the gait cycle while the maximum backward position is reached late in Stance phase at approximately 45% and 95% of the gait cycle (Inman et al., 1981; Thorstensson et al., 1984).

Velocity and acceleration during antero-posterior displacement have also been calculated. Inman et al. (1981) reported that the trunk exceeded its average velocity over the period of the gait cycle during the Double Support phases while it moved at less than its average velocity during the remainder of the gait cycle. Murray et al. (1964, 1966) found that the acceleration of the upper body segments was highest in Double Support phase just after Initial Contact at which time the trunk was descending to a lower, more anterior position. Likewise, acceleration was lowest early in Stance when the trunk was moving to its highest and most lateral position. These findings were supported by Waters et al. (1973) who concluded that acceleration of the upper body segments was related to the push off that occurs at the end of weight bearing, the deceleration of the Swing limb and the downward motion of the H.A.T. They related relative deceleration of the upper body segments to the backward thrust that occurs in late Stance, the forward acceleration of the Swing limb and the upward motion of the H.A.T. Cappozzo et al. (1978) found that the two points of peak acceleration of the pelvis occurred just before Initial Contact while the two points of peak deceleration occurred just after End of Weight Bearing.

Angular Displacement

Angular displacement or rotation can occur about coronal, sagittal and vertical axes in sagittal, coronal and transverse planes, respectively. Angular displacement can be measured using either a relative spatial reference system or an absolute spatial reference system. In the former, the position of a segment is measured with respect to the position of another body segment, e.g. the position of the thigh relative to the pelvis. In the latter system, the position of a segment is measured with respect to some external reference point, e.g. the position of the trunk relative to a line perpendicular to the ground (Winter, 1979). Because investigators have used two different measuring systems to quantify angular displacement of upper body segments, it is difficult to compare data from one study to another.

Coronal Axis/Sagittal Plane

Displacements about a coronal axis in a sagittal plane are commonly referred to as flexion and extension.

Using an absolute spatial reference system, Cappozzo et al. (1978) found that angular displacement of the trunk ranged from 2° at low walking speeds to 5° at high walking speeds. Thorstensson et al.(1982, 1984) found the net angular displacement of the trunk ranged from 1.5° to 10°. Relating angular displacement to the centre of gravity, Carlson and Thorstensson (1981) found that forward trunk inclination began during the initial part of Double Support and reached its maximum at the end of Double Support. Thorstensson et al. (1982) found that the trunk was rotated in a posterior direction at Initial Contact and that this continued throughout the Double Support phase. However, in later work Thorstensson et al. (1984) found that anterior trunk inclination was close to or had reached its maximum at Initial Contact while maximum posterior inclination occurred during Swing phase.

Thurston and Harris (1983) measured the angular displacement of the T_{12} to L_4 spinal segment relative to the pelvis. They found that peak anterior inclination of the segment occurred at approximately 10% and 60% of the gait cycle at End of Weight Bearing, while peak posterior inclination occurred at approximately 40% and 80% of the gait cycle in Mid Swing. In general, they observed that motion of the spinal segment was 90° out of phase with the pelvic segment although there was a great deal of between subject variability. They concluded that the relationship between sagittal motion of the pelvis and spine was still unclear.

Murray et al. (1964, 1966) used an absolute spatial reference system to measure motion of the pelvis and found a mean total excursion of 3° during level walking. Maximum anterior inclination of the pelvis occurred just before Initial Contact as the trunk inclined forward toward the next area of ground contact. Maximum posterior inclination was found to occur early in Mid Stance as the trunk moved into an erect position over the supporting limb. Lamoreux (1971) found an average of 3° each of anterior and posterior pelvic inclination for a total of 6°. Thurston and Harris (1983) measured pelvic motion relative to the ground and found that maximum anterior inclination occurred at 40% and 82% of the gait cycle (ipsilateral Swing, contralateral Stance), while posterior inclination occurred at 16% and 62% (ipsilateral Stance, contralateral Swing). They concluded that the inertia of the acceleration and deceleration of the swinging limb exerted a major influence on sagittal pelvic motion.

At the hip, a relative spatial reference measurement system has most commonly been used, the angle measured being that between the pelvic segment and the thigh segment. Murray et al. (1964), using a photographic technique, found values of 30° flexion and 10° extension at the hip during level walking. They observed that maximum hip flexion occurred at Initial Contact and that the hip extended as the trunk moved over the supporting limb. They noted that the termination of ipsilateral hip extension at approximately 50% of the gait cycle coincided with contralateral Initial

Contact. Johnston and Smidt (1969) used an electrogoniometer to measure hip angle during walking and recorded a mean of 37° flexion and 15° extension. They observed that extension began just prior to Initial Contact and then gradually increased until the end of Stance; flexion began just before End of Weight Bearing and continued until just prior to Initial Contact. Inman et al (1981) stated that maximum hip flexion occurred at 85% of the gait cycle while Nilsson et al. (1985) placed the onset of flexion during Double Support at 52% to 55% of the gait cycle. Both Smidt (1971) and Nilsson et al. (1985) found that increased walking velocity led to an increase in the net amplitude of hip motion.

Sagittal Axis/Coronal Plane

Displacements about a sagittal axis in a coronal plane are commonly referred to as abduction and adduction or lateral flexion.

Thurston et al. (1981) used an absolute spatial reference system to measure angular trunk displacement in the coronal plane. They calculated a displacement of approximately 7° from midline to each of the right and left sides. The maximum displacement occurred during Stance Phase, ipsilaterally at approximately 15% and contralaterally at approximately 70% of the gait cycle as the trunk flexed laterally to the side of the supporting limb. Similar results were obtained by Thorstensson et al. (1982) and Thurston and Harris (1983). Thorstensson et al. (1984) found that the trunk tilted to the ipsilateral side at Initial Contact and reached its maximum lateral flexion early in Double Support.

Inman et al. (1981) reported that the pelvis was level at Initial Contact, dropped to the side of the limb approaching End of Weight Bearing during Double Support, and then, shortly after End of Weight Bearing, rotated back to level position which was maintained until the next Initial Contact. Thurston et al. (1981) and Thurston and Harris (1983), using an absolute spatial reference system, found that a mean maximum pelvic angular displacement of 8° occurred during Swing, contralaterally at approximately 18% and ipsilaterally at approximately 67% of the gait cycle.

The angular displacement of the hip during level walking was measured with an electrogoniometer by Johnston and Smidt (1969). They found a total excursion of 12° of which 7° was abduction and 5° was adduction. Maximum abduction occurred during Swing and maximum adduction occurred at Mid Stance. Inman et al. (1981) reported similar values with maximum abduction occurring at 70% and maximum adduction at 20% of the gait cycle.

Vertical Axis/Transverse Plane

Displacement about a vertical axis in a transverse plane is commonly referred to as axial rotation.

Murray et al. (1964, 1966) used overhead photography to measure trunk rotation relative to reference markings on the ground. They found a high degree of variation between subjects but, on average, maximum rotation of the thorax was approximately 7° and occurred at Initial Contact with the thorax rotating to the side of the advancing foot. Similar results were obtained by Chapman and Kurokawa (1969) who used a mechanical external reference system. Gregersen and Lucas (1967) used transducers attached to Steinmann pins inserted into vertebral spinous processes to measure the relative rotation between vertebrae and sacrum during level walking. They found that maximum rotation of approximately 6° and 8° occurred at the T_1 and L_5 levels, respectively with no displacement observed at the T_7 level. They concluded that T_7 was an area of transition between vertebral rotation in the direction of the shoulder girdle superiorly and in the direction of the pelvic girdle inferiorly. Lumsden and Morris (1968) used a similar technique to measure the relative rotation at the lumbosacral joint and found a mean displacement of 1.5° which increased as walking speed increased. Thurston et al. (1981) also measured rotation of the trunk relative to the pelvis and found a maximum displacement in the lumbar spine of 8° which occurred at Initial Contact, resulting in rotation of the upper trunk towards the side of the advancing limb.

Rotation of the pelvis with respect to an external reference was measured by Murray et al. (1964, 1966) who found a great deal of variation between subjects. They calculated a maximum excursion of approximately 10° which occurred at Initial Contact, resulting in rotation of the pelvis away from the side of the advancing limb. Chapman and Kurokawa (1969) measured a similar amount of rotation and also noted that the point of reversal of rotation occurred at End of Weight Bearing. These findings were supported by the data of Sutherland and Hagy (1972), Thurston et al. (1981) and Thurston and Harris (1983).

Thoracic and pelvic rotation are 90° out of phase, i.e. at Initial Contact the thorax is maximally rotated to the ipsilateral side while the pelvis is maximally rotated to the contralateral side. Murray et al. (1964) proposed that this reciprocal motion has a damping effect which contributes to smoothness of forward translation of the body and, according to Thurston and Harris (1983), this reciprocal motion between thorax and pelvis maintains the orientation of the head in the direction of the plane of progression.

Hip motion relative to the pelvis was measured by Johnston and Smidt (1969) using an electrogoniometer. They found a total range of motion of 13° of which 9° was external rotation and 4° was internal rotation. Maximum external rotation occurred during Swing while maximum internal rotation occurred at Mid Stance.

TRUNK MUSCLE ACTIVITY DURING LEVEL LOCOMOTION

Waters and Morris (1972) stated that, compared to the lower limb, little attention had been paid to muscles of the trunk during ambulation. While research on trunk muscle activity during locomotion has been published in the intervening years, the bulk of gait research continues to be focused on the lower limb.

The body responds to external forces by contraction of muscles and electromyography (EMG) can be used as an indirect indicator of muscle activity (Asmussen & Klausen, 1962; Grieve, 1976). Tichauer (1971) stated that the surface EMG signal from the erector spinae muscle is a reliable indicator of the moment exerted on the lumbar spine. Although EMG has been used to study trunk muscle activity during sitting, standing and lifting, comparatively few studies have been carried out during walking.

Phasic Activity of Muscles

The relationship of onset, peak and cessation of myoelectric activity to the gait cycle can be measured.

Battye and Joseph (1966) monitored the electrical activity of the right erector spinae during level walking and found two distinct periods of activity coincident with the Double Support phases of the right gait cycle. Examination of their data shows that muscle activity in the first Double Support phase began at 96% of the previous gait cycle and continued until 9% of the gait cycle; activity in the second Double Support phase extended from 43% to 57% of the gait cycle. Letts et al. (1978) recorded activity from right and left paraspinal muscles and found biphasic and some triphasic activity within the gait cycle. They found that the phases of activity coincided with the Double Support periods and that the duration of the activity of each muscle group was symmetrical. Carlson and Thorstensson (1981) recorded activity from lumbar back muscles bilaterally and also found two main periods of activity per gait cycle at Initial Contact. In addition, they monitored muscle activity when the trunk was positioned at different inclinations and found that the period of activity of the lumbar muscles became more synchronous when the trunk was in flexion. Thorstensson et al. (1982) measured electrical activity of right and left longissimus and multifidus and also found a biphasic pattern that was similar on each side. A burst of myoelectric activity was seen bilaterally just before Initial Contact of both ipsilateral and contralateral feet with the ipsilateral burst consistently preceding the contralateral burst by 5% to 10% of the gait cycle. They noted that the two periods of electrical activity occurred during or at the end of an angular displacement directed backwards in the sagittal plane. These also coincided with activation of hip and leg extensor muscles. In addition, they observed bilateral symmetrical activity when sagittal plane movements predominated and asymmetrical activity when coronal plane movements predominated. On the basis of a biomechanical model, Cappozzo (1983) predicted that trunk extensor muscles would exert peak force just before or just after Initial Contact.

Rectus abdominis was investigated in 10 subjects by Sheffield (1962) who found no activity in the muscle during level walking at a cadence of 60 steps per minute. Using a biomechanical model, Cappozzo (1983) predicted that the mechanical activity of trunk flexors (abdominal muscles) would be small or absent.

The activity of erector spinae and rectus abdominis was measured simultaneously by Waters and Morris (1972). They observed peaks of electrical activity in iliocostalis and longissimus at contralateral Initial Contact and, less frequently, at ipsilateral Initial Contact. They found rectus abdominis to be active in five out of ten subjects and this activity occurred simultaneously with that of the back muscles.

Amount of Electrical Activity

The amount of myoelectric activity generated during the course of one gait cycle can also be measured. Guth et al. (1979) recorded the electrical activity of erector spinae in 20 normal subjects during level walking. They found activity in the lumbar and thoracic parts of the muscle to be approximately 6% and 5% respectively, of a reference contraction. They also observed significantly greater activity on the right side compared to the left. Thorstensson et al. (1982) found a difference in magnitude of the two bursts of activity of erector spinae during level walking. On the ipsilateral side they found the second burst, which occurred at contralateral Initial Contact, to be larger than the first burst which occurred at ipsilateral Initial Contact. Similarly, on the contralateral side they found the first burst, which occurred at ipsilateral Initial Contact, to be larger than the second burst which occurred at contralateral Initial Contact. As walking speed increased they observed that the total amount of electrical activity also increased. Waters and Morris (1972) studied the deep back muscles and rectus abdominis simultaneously and found that erect standing elicited activity in the back muscles, but none in rectus abdominis. During level walking they also observed that the magnitude of electrical activity of the ipsilateral deep back muscles was greater at contralateral Initial Contact than at ipsilateral Initial Contact and was not always present at ipsilateral Initial Contact.

Proposed Functions of the Trunk Muscles

As stated in previous sections, the upper body segments are displaced linearly and angularly in all three planes during locomotion. Therefore, trunk muscle activity can be related to motion in any or all planes.

In the sagittal plane, Battye and Joseph (1966) proposed that the back muscles contracted simultaneously to prevent the body from falling forwards. Waters and Morris (1972) also proposed this function for the back muscles. They noted that a ground reaction force greater than body weight is applied to the foot shortly after Initial Contact. This force is transmitted from the foot to the lower limb segments, pelvis and trunk and, because the force passes through the centre of gravity which lies anterior to the vertebral column, it tends to flex the trunk. This trunk flexion could be opposed by the contraction of erector spinae that is observed at Initial Contact. This proposal was supported by Carlson and Thorstensson (1981) and by Thorstensson et al. (1982).

In the coronal plane, Battye and Joseph (1966) proposed that contraction of the back muscles could prevent trunk side flexion. Waters and Morris (1972) noted that the trunk is displaced laterally during walking in order to balance over the supporting limb, and they proposed that the back muscles contracted to stabilize the trunk in the coronal plane. Thorstensson et al. (1982) observed that back muscles were active during trunk side flexion and concluded that the muscles acted to restrict motion in the coronal plane. From a biomechanical model, Cappozzo (1983) calculated that a moment would be exerted about the sagittal axis by contraction of the trunk side flexor muscles and that the magnitude of this moment would be half of that exerted by trunk extensor muscles about a coronal axis.

In the transverse plane, Battye and Joseph (1966) proposed that bilateral contraction of back muscles would prevent unwanted trunk rotation. On the other hand, Waters and Morris (1972) proposed that activity of the back muscles at Initial Contact exerted a force that contributed to the normal thoracic and pelvic rotation seen during level walking. Gracovetsky (1985) proposed that asymmetrical activity of paraxial muscles during spinal flexion in the sagittal plane will result in the generation of an axial torque, because of coupling. He postulated that lateral flexion of the trunk is sufficient to induce counter rotation in the transverse plane of the hip and shoulder and suggested that the pelvis is driven by the spine, not by the lower limb. He also proposed that the ipsilateral erector spinae, psoas, latissimus dorsi and trapezius are responsible for lateral trunk flexion and therefore cause pelvic and shoulder rotation.

Trunk muscles may also act as stabilizers. Waters and Morris (1972) proposed this function for rectus abdominis which they thought might exert a stabilizing force on the trunk. Letts et al. (1978) stated that the role of the paraspinal muscles in gait was poorly understood; from their results they concluded that reciprocal paraspinal activity could have a major role in stabilization of the vertebral column during locomotion. Carlson and Thorstensson (1981) contended that the primary function of the back muscles during locomotion is to control the stiffness of the trunk, thus restricting motion, and to maintain the trunk in the upright position.

CONTROL OF LOCOMOTION

Walking is a synchronized, complex pattern of movement requiring close co-ordination between the musculoskeletal apparatus and the nervous system (Lamoreux, 1971). During locomotion the body must be propelled smoothly forward in such a way that a minimum of effort is expended, and excessive mechanical stimulation of the sensory organs of eyes and labyrinth is avoided (Cappozzo, 1981).

Investigations of the control of locomotion have been carried out in fish, amphibians, arthropods and mammals (Pearson, 1976; Grillner, 1981). Experiments with cockroaches have shown that there is a group of interconnected neurons in the spinal cord that appear to generate a reciprocal rhythm in the hind leg. In the spinal cord of the cat there is a central pattern generator for each leg that is responsible for activating the limb muscles in the correct sequence for locomotion to occur. Removal of afferent input to the cord by transecting the dorsal roots does not change the temporal sequence of muscle activation, but that activation does become more susceptible to disruption. Therefore, stereotyped locomotor movements in both cockroach and cat appear to be generated by groups of neurons, called the central pattern generators, located in the lower levels of the spinal cord (Pearson, 1976; Grillner, 1981).

From the results of studies on cats, it would appear that initiation and maintenance of walking, plus modification of walking patterns in response to changes in terrain, are governed by higher centres. Locomotor centres have been found at different levels of the brain stem in the subthalamic locomotor region, the mesencephalic locomotor region and the pontine locomotor region. It is thought that these higher centres command and control the stereotyped locomotor patterns at spinal cord level by way of the rubrospinal, lateral vestibulospinal, reticulospinal and coerulospinal tracts (Grillner, 1981).

In addition to influences from higher centres, the neurons of the central pattern generators of the intact animal also receive sensory input from the limbs. This peripheral input regulates the duration and amplitude of the different phases of the gait cycle. During the Stance phase, a load is exerted on the extensors of the hip. At the end of the Stance phase the load decreases and this is thought to promote the initiation of the hip flexion that is required for Swing phase. It has also been found that flexion is initiated when the hip joint is placed in the position in which flexion would normally begin, and extension is initiated when the joint is placed in the position in which extension would normally begin. Finally, movement in the direction of flexion reinforces flexor muscle activity and, if movement in the direction of extension occurs, flexor muscle activity decreases (Grillner, 1981).

Co-ordination of locomotor activity is effected by the cerebellum which receives input from the periphery, the central pattern generators of the spinal cord and higher brain centres. Output from the cerebellum can, in turn, indirectly modify output from the central pattern generators via descending pathways from the reticular formation, vestibular nuclei and red nucleus. Thus the basic motor act of walking appears to be a stereotyped pattern, however precision of performance depends on sensorimotor integration of visual, vestibular and somatosensory elements (Grillner, 1981; Conrad et al., 1983).

Changes in walking surface or terrain require modification of locomotor patterns. According to Conrad et al. (1983), high external demands require a wide spectrum of sensory clues and the individual may respond to complex locomotor demands by using a protective gait mechanism. This is executed automatically and consists of a walking pattern in which bipedal contact is prolonged, the surface area of monopedal contact is increased ("flat foot") and speed is decreased, resulting in increased stability. This protective gait mechanism can be seen when an experimental animal is made to walk up or downhill. During uphill walking, the hind limbs extend further posteriorly and the forelimbs are placed differently. The reverse occurs during downhill walking. The purpose of these adjustments is to enhance balance by maintaining the centre of gravity between the moving points of support (Grillner, 1981).

It can be seen that locomotion is a complex motor activity with a basic rhythmic pattern which can be carried out automatically, but which can be modified, corrected and adapted in response to changes in the environment and in the volition of the animal or person (Grillner, 1981).

STAIR CLIMBING AND DESCENT

Andriacchi et al. (1980) stated that stair climbing and descending are common activities of daily living. However, the kinematics and kinetics of these activities are different from level walking and there is the potential for significant variations in the way different people climb stairs.

Stair climbing consists of elevation of the body on to the next highest step while stair descent consists of lowering the body to the step below. The dominant motions in these two activities take place in the sagittal plane (Joseph & Watson, 1967; Townsend & Tsai, 1976; Andriacchi et al., 1980).

Joseph and Watson (1967) found that both stair climb and stair descent had Stance, Swing and Double Support phases. In stair climb, Stance phase constituted approximately 71%, Swing phase approximately 29% and Double Support phases 29% of the gait cycle; in stair descent, Stance phase constituted approximately 63%, Swing phase 37% and Double Support phases 20% of the gait cycle. Corlett et al. (1972) investigated both stair and ramp walking and observed that the cycle duration in stair descent was less than that in ramp descent when the stair riser was 15 cm high and the slope of both stairs and ramp was 20°.

Displacements of lower limb segments during stair climb and descent have been measured. Townsend, Lainhart, et al. (1978) observed that foot clearance in stair climb was accomplished by hip and knee flexion. In stair climb Andriacchi et al. (1980) found that the maximum value of hip flexion in Stance was 33.8° and in Swing was 40.8°. At Initial Contact the hip was flexed, as the limb moved into Stance the hip moved into extension and this continued until End of Weight Bearing. From End of Weight Bearing to Mid Swing the hip moved into flexion, while from Mid Swing to Initial Contact the hip moved from maximum flexion into extension. In stair descent they found that the maximum value of hip flexion in Stance was 13.4° and in Swing was 23.0°. At Initial Contact the hip was slightly flexed, at End of Weight Bearing it was maximally flexed, while during Swing the amount of hip flexion decreased. They also found no significant difference in the amount of hip flexion in Swing phase between stair climb and stair descent.

The majority of investigators of stair walking who used electromyography as a tool have studied lower limb muscles. Townsend, Lainhart, et al. (1978) related rectus femoris and hamstring activity to trunk stabilization in both stair climb and descent. During these activities they found that rectus femoris and hamstrings were simultaneously active. They proposed that during Stance, rectus femoris was active to pull the trunk forward over the supporting foot while hamstrings were active to provide support. During stair climb hamstring activity was higher in subjects with greater forward trunk inclination, especially during the second Double Support phase. During stair descent hamstring activity was greatest during late Stance. Overall, they found that the patterns of lower limb muscles were more variable during stair climb than during stair descent. Andriacchi et al. (1980) stated that there were differences between the patterns of electrical activity of lower limb muscles in stair walking compared to level walking. These differences were evident in those muscles responsible for vertical displacement of the body ie. hip and knee extensors and ankle plantarflexors in stair climb, knee extensors and plantar flexors during stair descent. Lyons et al. (1983) studied the hip extensors and found that less myoelectric activity was generated in stair descent than in stair climb or level walking.

Back muscle activity during stair climb was studied by Joseph and Watson (1967) They found that myoelectric activity occurred at both ipsilateral and contralateral Initial Contact. From their description, it appears that the maximum peak of activity occurred during contralateral

Initial Contact. They stated that both erector spinae muscles contracted simultaneously in early Stance during the time that the trunk inclined forward and the body was displaced vertically. They proposed that this anterior trunk inclination assisted the forward displacement of the body and that bilateral erector spinae contraction controlled the amount of forward inclination of the trunk. They also observed that the contralateral erector spinae showed marked activity during the latter part of Stance but stated that they could not offer an explanation for this. During stair descent, biphasic erector spinae activity was also seen and occurred in the Double Support phase. It was proposed that the bilateral contraction of erector spinae prevented trunk flexion. Large variations in muscle activity were seen within subjects between trials during different sessions and this was attributed to the fact that individuals may walk up and down stairs with varying degrees of trunk flexion, thus giving rise to variable electromyograms.

Studies of energy expenditure during stair walking have also been carried out and it was shown by Corlett et al. (1972) that stair climbing was more efficient in terms of physiological cost than ramp climbing and that steps with high risers were more energy efficient than those with low risers. Fitch et al. (1974) observed that the pattern of energy expenditure was different in stair walking from that in ramp walking or level walking. During the latter two types of locomotion, subjects generally chose a speed of ambulation at which total energy cost per unit distance was minimal, however the speed chosen during stair climbing was seldom the most economical in terms of total energy expended. They also noted that, at the usual rate of climbing, the energy cost fell as stairs became steeper, but that the rate of energy expenditure was greater for stairs with high as opposed to low risers.

RAMP CLIMBING AND DESCENT

Few kinematic studies have been done of ramp walking, also referred to as grade, slope or incline walking. Dean (1965) considered ramp walking to be more complex than level walking.

Drillis (1958) observed that individuals adapted to insecure situations such as walking in the dark, on ice or ascending and descending slopes by taking shorter steps. However, Bobbert (1960), in experiments using a treadmill, found that stride length remained essentially the same when the gradient was changed, but that cadence decreased when the slope exceeded 8°. Dean (1965) found that cadence was unaffected for slopes up to 7°. Gray and Basmajian (1968) observed that walking speed increased during ramp descent and attributed this to acceleration due to gravity. Waters and

Morris (1970) found that the mean cycle duration for walking up a 5° incline was 1.2 seconds.

With respect to displacement of the body, Dean (1965) stated that, given a sufficiently steep grade, the vertical motion of the trunk would no longer be oscillatory, but rather would be a smooth rise or fall. Gray and Basmajian (1968) studied the displacement of the leg and foot during ramp walking and observed that in climbing the foot tended to be placed flat on the surface of the ramp rather than heel first which is the usual point of initial contact during level walking. They proposed that the slope of the ramp acted like an obstacle to the foot which could be cleared by increasing hip or knee flexion. They also observed increased eversion of the foot during ramp climbing and proposed that this modification possibly resulted in smoother, more efficient gait. Tokuhiro et al. (1985) observed that hip flexion was less in ramp descent than in ramp climbing.

Muscle activity of the leg during ramp walking was studied by Gray and Basmajian (1968) who found that all muscles except tibialis anterior showed lower mean activity levels in ramp descent compared to level walking. They explained this finding on the basis of gravity assisting the descent of the body, therefore less muscle activity would be required. However, they also recognized that walking speed increased during ramp descent because of the effect of gravity and that this might require increased muscle activity to control the descent of the body. A similar proposal was put forth by Pimental et al. (1982) who stated that concentric muscle contractions would also occur during ramp descent because the body is raised during the stepping cycle. Waters and Morris (1970) studied trunk muscle activity during ramp walking. They used a treadmill set at a 5° upward incline and found that the myoelectric activity of erector spinae and rectus abdominis was similar to that seen in level walking.

The efficiency of ramp walking has also been studied. Dean (1965) stated that the energy expenditure would increase in ramp climbing above a 20% grade, presumably reflecting the increased modification of limb activity but also due to the increasing tendency to incline the trunk forward. He calculated that the energy expenditure would be increased by approximately 21% when the trunk was inclined 45° anteriorly. Corlett et al. (1972) measured cardiac cost and oxygen consumption in ramp and stair climbing and found that these were slightly higher for ramps compared to stairs of the same slope. They also correlated body weight, leg length and oxygen consumption in ramp climbing and found the strongest correlation with energy cost to be leg length. Fitch et al. (1974) observed that subjects will tend to choose an optimum speed for minimum energy expenditure when climbing a ramp. They stated that energy cost per vertical meter decreased as slope increased until the slope exceeded 17°, after which energy cost increased. Also, ramps demanded a higher expenditure of energy per meter rise than stairs. However, when energy cost of both horizontal and vertical movement was taken into consideration, a ramp with a slope of less than 8° would be more economical than any stairway likely to be encountered in the course of normal activity. They recommended that ramps be used for long horizontal or vertical distances while stairs be used for short horizontal or vertical distances. During ramp descent, Pimental et al. (1982) observed that walking became more efficient as grade increased and attributed this to the effect of gravity which helps to move the load in the required direction, therefore the body muscles would be required to use fewer fibres during contraction. However, they recognized that to maintain a fixed rate of descent might require an increase in muscle contraction and therefore an increase in oxygen consumption as the pull of gravity would have to be resisited and controlled.

TECHNIQUES FOR MOTION ANALYSIS

In the last 200 years many different techniques have been used in an effort to investigate movement of the human body. Luigi Galvani, in the late 18th century, observed that electrical stimulation of the muscles of a frog resulted in contraction. Some seventy-five years later, Duchenne applied Galvani's discovery to human muscle and observed the action produced on the joint or joints that the stimulated muscle crossed. These experiments by Duchenne can be considered the beginning of the scientific study of human movement (Basmajian & DeLuca, 1985). At the end of the 19th century, the new technology of photography was used by Marey in France and Muybridge in the United States to record human motion (Winter, 1979). At the beginning of the 20th century, visual inspection and palpation was used systematically by Beevor (1903) to study the actions of muscles during specific movements. Electromyography, a technique based on Galvani's discovery that a contracting muscle generates electricity, was first used scientifically as a tool to investigate muscle function by Inman et al. (1944) in their study of the function of the muscles of the shoulder (Soderberg, 1986).

Over the last forty years, technical advances in electronics and instrumentation have resulted in an increase in the number of reliable and valid techniques for analyzing motion. Kinematic analysis can be done using direct measurement techniques such as electrogoniometry, and by using indirect measurement techniques such as photography and video recording. Kinetic analysis can be done using force plates and electromechanical dynamometers. The technique of electromyography can be used in both kinematic and kinetic studies to provide indirect information about the pattern and intensity of muscle contraction (Winter, 1979; Soderberg, 1986).

Because a kinematic approach was used in this investigation, a description of the techniques commonly used in kinematic studies will be given.

DIRECT MEASUREMENT TECHNIQUES

ELECTROGONIOMETRY

A goniometer is a device used to measure the angle produced between two bony segments when motion occurs in a particular plane. For clinical purposes, a simple device is used to measure joint movement; it consists of a protractor to which has been attached two arms, one static and one movable. The static arm extends from the axis of the protractor while the movable arm is riveted to the axis of the protractor and is free to move through 360°. The axis of the goniometer is placed over the axis of joint motion and a reading of the joint angle can be taken (Esch & Lepley, 1971; Scott & Trombly, 1983).

In an electrogoniometer a potentiometer is substituted for the protractor. There is a linear relationship between the shaft angle of the potentiometer and its electrical resistance. Therefore, if the device is mounted such that joint motion causes movement of the shaft of the potentiometer, a voltage applied across the potentiometer will be varied. Changes in joint angle will result in changes in the output voltage of the circuit containing the potentiometer. The output voltage can be relayed to a variety of output display and recording devices where the changes in voltage can be recorded in degrees (Liberson, 1965; Peat et al., 1976; Brown et al., 1979).

The advantages of electrogonimeters are that they are relatively inexpensive and that the output is available immediately, is easy to interpret and can be converted to a digital signal for storage on computer disc (Sutherland & Hagy, 1972; Winter, 1979; Gowitzke & Milner, 1980). The disadvantages are that electrogoniometers give relative, not absolute data on joint motion, measurement of multiaxial joints requires very complex instruments, they may require an inordinate amount of time to fit and align and, because they are attached directly to the body, they may interfere with normal motion (Sutherland & Hagy, 1972; Winter, 1979; Yack, 1984). In addition, errors may occur because of slippage of the electrogoniometer on the underlying skin, and because the linkage of the system is located outside the joint while motion is actually occurring inside the joint (Johnston & Smidt, 1969; Smidt, 1971; James & Orr, 1982). According to

Yack (1984), when using a triaxial electrogoniometer, errors in transverse and coronal plane measurements are likely to occur as displacement in the sagittal plane increases.

INDIRECT MEASUREMENT TECHNIQUES

PHOTOGRAPHY

Two photographic techniques have been used to record body motion - multiple exposure still photography and cinematography. In both these techniques the subject is free to move unhindered by any external measurement instruments (Smith, 1975).

Multiple Exposure Still Photography

In this technique a still camera is used to take multiple images of a particular motion sequence on a single photograph. This is done by keeping the camera shutter open for the duration of the motion sequence. Over this period of time the single frame of the film is exposed many times by an interrupted source of illumination with the result that multiple images are produced because the body is in a different position at each exposure (Grieve & Gear, 1966; Winter, 1979).

A 35 mm still camera or a Polaroid camera positioned at right angles to the plane of progression can be used to record motion in one plane (Grieve & Gear, 1966; Smith, 1975;

Gowitzke & Milner, 1980). To record motion in all three planes simultaneously, Cappozzo et al. (1978) used four 35 mm cameras placed in stereoscopic pairs with convergent optical axes.

A variety of methods can be used to illuminate the subject. The interrupted light method is carried out in a darkened room using a strobe light mounted directly below the camera. Each flash of the strobe light will illuminate the subject for several milliseconds and the flash rate can be set to repeat at regular intervals. With each flash, an image will be recorded on the film. Flash rates of 20 to 30 per second can be used and knowledge of the rate allows the time dimension to be recorded on the film. Dots or strips of reflective tape are placed on anatomical markers or along the long axes of limb segments of the subject to provide the image which will consist of a series of dots or bars on the single photograph. The subject should be photographed against a matte black backdrop which has scaling markers at the bottom of the field of view (Murray et al., 1964, 1969; Smith, 1975; Winter, 1979; Gowitzke & Milner, 1980).

A second method of illumination is the rotating slit shutter. In this technique a rotating disc is placed immediately in front of the camera lens. The disc has a slit cut in it which serves as a shutter opening, and it is driven at a constant speed which corresponds to the flash rate of the strobe method. Small light bulbs placed at joint centres or reflective markers placed along the axis of body segments are used and the subject is also photographed against a black backdrop (Eberhart & Inman, 1951; Smith, 1975).

A third method of illumination is that used by Cappozzo et al. (1978) who attached light emitting diodes (LEDs) to anatomical landmarks and then photographed the moving subject. The LEDs were activated for a period of three milliseconds and flashed at the rate of 30 per second. Multiple images of the LEDs were recorded on film.

In each of these techniques the end result is a photograph of a "stick" diagram or of a series of dots in which the position of the limb segments or of the body markers is recorded at equal intervals of time. Joint angles can be measured directly from the photograph, and X and Y co-ordinates of the illuminated points can be determined. If the strobe or disc rotation rate is too high, the photograph will contain multiple images that are so close together as to be undecipherable. Too low a rate will result in a photograph with only two or three recorded images for the entire motion and which therefore lacks sufficient detail (Smith, 1975; Winter, 1979; Gowitzke & Milner, 1980).

The advantages of multiple exposure still photography are that it is inexpensive and, if a Polariod camera is used, results can be available immediately. The disadvantages are that the multiple images on one photograph tend to limit measurement to range of motion, the flashing light in the strobe method may be distracting for the subject, and the LED method must be carried out in a darkened room and this may interfere with normal gait (Winter, 1979).

Cinematography

In this technique, a cine camera is used to take multiple photographs of a particular motion sequence. As with the multiple exposure technique, successive images are obtained of the displacement of body segments and measurement of linear and angular displacements can be made directly from each image (Smith, 1975; Yack, 1984).

According to Winter (1979) 16 mm cine cameras are most frequently used in motion studies. Cameras may be spring driven or may be powered by batteries or an ac source. The rate at which the film passes through the camera should be constant and should be low enough to record the entire event under investigation, but not so high as to waste film and require extra lighting. Frame rates of 48 per second or higher have been used in gait studies by Eberhart and Inman (1951), Liberson (1965) and Brandell (1977). Smith (1975) recommended that frame rates between 32 and 100 per second be used in motion analysis studies because higher rates were wasteful of film. However, cine recording is essentially a means of sampling human motion, therefore, according to Winter (1979), the Sampling Theorem can be applied to determine the optimum film rate at which sufficient images can be recorded in order to result in a valid kinematic analysis. The Sampling Theorem states that "the process signal must be sampled at a frequency at least twice as high as the highest frequency present in the signal itself" (Winter, 1979, p.28) In practice however, the sampling rate is usually four to five times the highest sample frequency. Because human movement has been found to contain frequencies of up to five to six Hertz (Hz), the frame rate of the cine camera should be 24 to 30 frames per second (Winter, 1979). Winter (1982) carried out a biomechanical analysis in which a comparison was made between the results when the the frame rate was 50 per second to those when the frame rate was 25 per second. He found the differences to be negligible and therefore recommended that a standard cine camera at a frame rate of 24 per second be used for normal and pathological gait studies. The advantages of using a standard camera are that it is less expensive than a high speed camera and the lower frame rate results in less film being used.

One or more cine cameras can be used to film movement. To acquire three-dimensional data, Sutherland and Hagy (1972) used three cameras placed orthogonally to the three planes. Timing devices on the cameras marked all three films simultaneously to ensure that the recordings were synchronous. According to Gowitzke and Milner (1980), three-dimensional information can be obtained using two cine cameras, provided that they are placed in such a way that marked body segments are in the field of view of each camera at all times.

Film can be synchronized with events in the gait cycle by using timing devices inside the camera or in the field of view (Sutherland & Hagy, 1972; Smith, 1975). Synchronization of film with electromyography has also been done. Gray and Basmajian (1968) recorded the EMG signal on the magnetic tape edge of 8 mm film. Sutherland and Hagy (1972) used an adapted camera to film simultaneously the subject walking and the EMG tracing on an oscilloscope. Brandell (1977) synchronized film and EMG by using simultaneous spike artifacts on the EMG recording and a light blink in the field of view of the cine camera.

As with multiple exposure still photography, markers must be placed on the body in order for consistent measurements to be taken from the cine film. It is recommended that black disc markers approximately 2.5 cm in diameter be placed over specific bony landmarks close to the joints or on joint lines. The centre of gravity of each body segment may also be marked, but the disc may shift due to muscle contraction beneath the skin (Sutherland & Hagy, 1972; Smith, 1975; Gowitzke & Milner, 1980).

A reference system is also required in order to relate displacement of body markers to the space through which the body moves. A distance scale and reference axes can be placed on a black cloth backdrop against which the subject is photographed. In this way, measurements of distance on the image can be converted to real distance by applying the appropriate conversion factor (Smith, 1975; Gowitzke & Milner, 1980; Yack, 1984).

After the film is developed, it is viewed by means of a special projector that can advance the film one frame at a time. The image of each frame is displayed on a screen that is set at a right angle to the axis of the lens of the projector. Direct measurement of displacements can be done on the projected image by using a ruler, protractor and dividers, however this is impractical for long runs of film as data extraction is extremely time consuming. A digitizing tablet or electronic grid interfaced to a computer can be used to decrease the amount of time spent in data extraction. The image is projected onto the surface of the digitizing tablet and the body markers in the image are touched with a cursor or special electronic pen. This causes a signal to be sent to the computer and this signal can be stored as an X and a Y co-ordinate. A computer program can be written to label the required number of body markers and it is recommended that a pattern recognition feature be written into the program to determine whether the markers have been digitized in the correct sequence. It is recommended that any computer program designed to store digitized co-ordinates, apply correction factors and manipulate data be tested carefully to ensure that it is reliable and that the data generated are valid (Sutherland & Hagy, 1972; Smith, 1975; Winter, 1979; Gowitzke & Milner, 1980).

A number of sources of error can occur with the cinematographic technique (Cappozzo, 1981). These include lack of definition of the film image, shifting of markers, optical distortion, errors in data extraction, and loss of information due to smoothing of the data (Smith, 1975). If the developed film is grainy and the image is lacking in definition, it may be difficult to detect the disc markers on the projected image, leading to inaccurate location of reference points. Shifting of markers may occur because of muscle contraction beneath the skin or because of stretching of the skin during movement. According to Cappozzo (1981) this movement of markers is unavoidable but can be minimized by careful choice of marker locations. Optical distortion may result because motion will not always occur only in a plane perpendicular to the camera axis and movement toward or away from the camera will result in apparent changes in body dimensions. This "parallax error" can be minimized in three ways: first, by placing the camera as far away from the subject as possible (approximately 4 to 12 m) as this will reduce the size of the angle between the camera and a body marker; second, by keeping the axis of the camera at a right angle to the subject throughout the movement; and third, by incorporating a mathematical correction for the error into the computational formulae used to calculate displacements. Errors in data extraction can occur during the digitization process and these result from malalignment of the cursor or special pen on the image of the anatomical marker. Smoothing of the data can result in loss of genuine information in addition to the removal of extraneous information. This can be minimized by selecting an appropriate smoothing technique (Eberhart & Inman, 1951; Gray & Basmajian, 1968; Sutherland & Hagy, 1972; Smith, 1975; Winter, 1979; Gowitzke & Milner, 1980; Cappozzo, 1981; Winter, 1983).

The advantages of cinematography are that it offers no restriction to movement, the process of filming is not time consuming, it provides absolute as opposed to relative co-ordinates, and it produces a permanent visual record of the motion. The disadvantages are that equipment, purchase and processing of film are expensive, and the amount of time required to extract the data from each frame of film can be lengthy (Winter, 1979).

VIDEO RECORDING

In this technique a television (TV) camera and video recording system are used to capture changes in displacement of body segments. The recording can be played back in slow motion or frame-by-frame and it can be converted into a digital format by TV-computer interface (Winter et al., 1972a). The field rate of TV cameras in North America is 60 Hz, therefore the sampling rate can be considered high enough to be appropriate for normal and pathological gait studies (Winter, 1982).

The TV camera should be aligned in such a way that the axis of the camera lens is perpendicular to the plane of progression. The camera and a monitor can be mounted on a cart approximately 3 m from the walkway. The mobile cart allows the subject to be tracked during motion while the monitor allows focus and lighting to be adjusted easily (Winter, 1979; Gowitzke & Milner, 1980).

As with cinematography, video recording can be synchronized with events in the gait cycle and with the electromyogram (Letts et al., 1975; Dubo et al., 1976; Pare et al., 1981).

Reflective body and background reference markers are also used in this technique. The subject moves against a black matte backdrop and the lighting is adjusted to provide a very high contrast image. Thus the image recorded is of white dots against a black background and this enables a one-bit conversion of the data for computer analysis (1 for white, inside a marker; 0 for black, outside a marker). A computer program can then be used to calculate the centre of the markers, apply correction factors for optical distortion, and determine the absolute co-ordinates (Winter et al., 1972a). In addition, linear and angular displacements can be measured directly from the image on the monitor screen (Yack, 1984).

The advantages of video recording are that the cost of equipment is not high, the technique offers no impediments to movement, minimal time is required to attach body markers, absolute as opposed to relative co-ordinates are produced, and a record of movement that can be replayed instantly and repeatedly is produced. The disadvantages are that extra lighting is required and that the maximum field rate of 60Hz may not be sufficiently high to capture very rapid motion (Winter, 1979, 1982).

OPTOELECTRONIC TECHNIQUES

In recent years the use of TV cameras interfaced to computers has given rise to two automated motion analysis systems - Selspot and Vicon. The principle behind both systems is automatic location and conversion to x and y co-ordinates of anatomical markers (Winter, 1979).

In the Selspot system infrared LEDs are used as anatomical markers and are flashed sequentially in a particular order. The LEDs are connected to a power supply and switching circuit by individual cables. The TV camera uses a standard lens to focus the LED flash onto a special semi-conductor diode surface in the focal plane of the camera. The location of the image of the LED flash gives rise to two signals, one for the x co-ordinate, the other for the y co-ordinate. Because each LED flashes sequentially in a predetermined order, a series of x and y co-ordinate signals are generated for each anatomical marker and can be fed to a tape recorder or directly to a high speed computer for storage (Winter, 1979; Gowitzke & Milner, 1980; Nilsson et al., 1985).

The Vicon system consists of two to four TV cameras which contain light sensitive diodes and which are interfaced to a digital computer. Special electronic circuits scan the x and y connections to each diode. Strobe lights are mounted near the lenses of the cameras and their flash rates are synchronized with the field-scan of the camera. The cameras are on line with the computer which samples at 20 millisecond intervals.

Reflective markers are attached to anatomical landmarks. Bright spots reflected from the markers will illuminate the diodes and interface sensitivity is set to trigger when a bright spot is scanned. In this way the position of each anatomical marker is stored as an x - y co-ordinate pair determined by the number of the TV raster line on which the spot was encountered and the distance of the spot along that line (Winter, 1979; Thurston et al., 1981; Thurston & Harris, 1983; Kirtley et al., 1985).

In both systems a computer program identifies and labels each marker, smooths the signal and performs data transformations (Andriacchi et al., 1979; Winter, 1979).

The advantages of optoelectronic systems are that three dimensional data can be generated, data extraction is done quickly and random errors are minimized because all calculations are done by the computer (Thurston, 1985). The major disadvantages are that the systems are complex and expensive. In addition, in the Selspot system ambient infrafed radiation, variations in the intensity of the LEDs and electrical disturbances can all distort the stored signal (Andriacchi et al., 1979). In the Vicon system the markers are not always in view or do not always trigger a signal, therefore an interpolation algorithm must be used to provide the missing data and this can be a source of error (Kirtley et al., 1985). Yack (1984) recommended that optoelectronic systems be thoroughly tested to ensure that the data they generate are reliable and valid.

DEVICES TO RECORD TEMPORAL EVENTS

In studies of human locomotion foot timing information is considered to be the essential synchronizing signal for correlation of all other data (Winter et al., 1972b). Therefore, in a kinematic analysis, some method of demarcating and recording the major events of the gait cycle is required in order to calculate accurately cadence, length of cycle, and duration of Stance and Swing phases. It is also required in order to correlate accurately the position of the foot with EMG signals and to determine accurately the sequence of the points of contact of the foot during normal and pathological locomotion (Eberhart & Inman, 1951; Gray & Basmajian, 1968; Waters & Morris, 1972; Winter et al., 1972b; Fitch et al., 1974; Cappozzo et al., 1976; Townsend, Lainhart et al., 1978; Townsend, Shiavi et al., 1978; Gowitzke & Milner, 1980; Bogardh & Richards, 1981).

Foot switches are the devices most commonly used to demarcate the temporal events in the gait cycle. A review of 27 gait studies revealed that foot switches of various types were used in 21 or 78% of the investigations. Other methods used to indicate temporal events included use of a special walkway, photography, and use of a force plate (Milner et al., 1971; Winter et al., 1972b; Cappozzo, 1981; Mizrahi et al., 1982).

In its simplest form a foot switch consists of a switch connected in series in an electrical circuit powered by a dc source. The switch is placed on the plantar surface of the foot and is designed to close when the foot is in contact with the floor; a device to record output voltage can be incorporated into the circuit (Winter et al., 1972b; Tata, 1980). Foot switches can also be connected in parallel with different resistors in a circuit. In this way, closure of each individual foot switch will result in a voltage output that will be different from that of the other foot switches. It then becomes possible to identify which points of the foot are in contact with the ground at specific times in the gait cycle (Tata, 1980).

Foot switches can be built into special shoes, attached to the sole of the shoe or attached to special inserts that fit into the shoe. In all cases, the foot switches should not noticeably affect normal gait, should be capable of giving reliable signals over long periods of use, and should be capable of being used in all normal walking areas (Winter et al., 1972b).

Lamoreux (1971) conducted a gait study in which he used a foot switch built into the heel of the shoe. Winter et al. (1972b) designed a special shoe with five recessed microswitches - two switches were placed in each of the heel and ball areas of the sole and one was placed in the toe area. These adapted shoes were used in gait studies by Letts et al. (1975, 1978) and Dubo et al. (1976).

Waters et al. (1973) taped foot switches to the heel area of the subject's shoes and Dietz et al. (1979), in a study of running, taped foot switches to the ball area of the subject's shoes. Two foot switches attached to the heel and toe areas of shoes were used by Eberhart and Inman (1951), Fitch et al. (1974) and Tokuhiro et al. (1985). Three point attachment at heel, ball and toe areas was used by Sheffield (1962) and by Gray and Basmajian (1968).

Multiple foot switches incorporated into shoe insoles were used by Townsend, Lainhart et al. (1978) and by Townsend, Shiavi et al. (1978). These switches controlled the voltage output level of two amplifiers. Tata (1980) also used foot switches attached to shoe insoles and placed the switches at the heel, ball and toe areas.

Special walkways have also been used as a means of delimiting temporal events. Milner et al. (1971) used an aluminum walkway and fitted their subjects with shoes which had wire mesh electrodes attached to the heel and toe areas. When an electrode contacted the walkway, a voltage unique to that electrode was produced and recorded. Cappozzo (1981) used a level walkway which had 5 mm wide adhesive metal strips placed 1 mm apart attached to it. The sole of the subject's shoe was also covered with the metal strips and, when the shoe contacted the walkway, a signal was generated and recorded. A similar system was used by Mizrahi et al. (1982).

The position of the foot in space can also be determined using imaging techniques such as still and cinephotography, video recording and optoelectronic systems. Murray et al. (1964, 1966, 1969) determined the beginning and end points of foot contact from multiple image photographs while Battye and Joseph (1966) extracted these data from cinefilm. However, according to Winter et al. (1972b), this approach is limited by the size and definition of the image and by the sampling rate of the camera. Thorstensson et al. (1982), using the Selspot system, defined the different phases of the gait cycle by relating them to the recorded angular displacement of the ipsilateral knee joint.

A force plate can be used to determine the point of foot contact and the end of forward progression in the Stance phase. However, the use of this device is restricted to a walkway and only one stride can be analyzed (Winter et al, 1972b).

ELECTROMYOGRAPHY

THE MOTOR UNIT ACTION POTENTIAL

The structural unit of skeletal muscle is the individual muscle cell or fibre while the functional unit is the motor unit. The motor unit consists of the synaptic junction at an alpha motor neuron in the ventral horn of the spinal cord, the alpha motor neuron plus its axon, the neuromuscular junctions, and all the muscle fibres innervated by that particular alpha motor neuron (Winter, 1979; Gordon, 1982; Basmajian & DeLuca, 1985).

Muscle fibres from adjacent motor units interdigitate. The territory of an individual motor unit is approximately one third of the cross-sectional area of the muscle in which it is located (Basmajian & DeLuca, 1985).

Motor units within a muscle vary in size depending upon the function of the particular muscle. Small muscles responsible for fine movements have few muscle fibres per axon and are said to have a low innervation ratio, while muscles responsible for gross movement have many muscle fibres per axon and are said to have a high innervation ratio. The size of the motor unit corresponds to the size of its alpha motor neuron, with small motor neurons innervating fewer muscle fibres than larger motor neurons (Gordon, 1982; Basmajian & DeLuca, 1985).

Excitation of the alpha motor neuron leads to depolarization of the sarcolemma of each individual muscle fibre within the motor unit. This results in a mechanical twitch. The multiple individual fibre depolarizations or action potentials are collectively known as the motor unit action potential. Because all the muscle fibres in the motor unit do not contract simultaneously, the electrical potential generated by the motor unit action potential has a median duration of 9 milliseconds (msec) (Basmajian, 1978).

The total amplitude of the motor unit action potential is measured in micro or millivolts. This amplitude is a function of the number of individual muscle fibre action potentials generated and also of the distance of the motor unit from the electrodes used to record it. Because the motor unit action potential is dissipated into the surrounding tissues, the closer the placement of the electrodes with respect to the contracting motor unit, the higher will be the potential recorded. Therefore the amplitude of the motor unit action potential depends not only upon the size of the motor unit that generates it, but also upon the distance of the motor unit from the recording electrode (Fleck, 1962; Basmajian, 1978; Winter, 1979; Basmajian & DeLuca, 1985).

An increase in muscle tension can be accomplished first by increasing the frequency at which the motor unit is stimulated and second, by the recruitment of additional motor units. Each motor unit has a limit to its "firing rate" or frequency of discharge. The normal upper limit of activation is considered by Basmajian (1978) to be 50 Hz while Hayes et al. (undated) stated that some motor units may discharge at rates as high as 100 Hz. Once the maximum frequency of firing of a motor unit is reached, another motor unit is recruited and it likewise responds to increased tension levels by increasing the firing frequency to its limit at which time a third motor unit is recruited. During contraction the smallest motor units are recruited first with larger motor units being added as the requirements for tension increase (Winter, 1979; Gordon, 1982; Basmajian & DeLuca, 1985).

THE MYOELECTRIC SIGNAL

Electrodes placed inside or over a muscle will record the sum of the electrical activity produced by the depolarization and repolarization waves of all contracting muscle fibres in the immediate vicinity of the electrodes (Eberhart & Inman, 1951; Grieve, 1976; Winter, 1979; Gans & Gorniak, 1980; Soderberg & Cook, 1984). Therefore the myoelectric interference pattern will consist of positive and negative spikes of varying amplitudes representing the superimposed activity of all motor units in the area (Winter, 1979). The amplitude of the myoelectric signal is affected by the thickness of the soft tissue interposed between muscle and electrode, the orientation of the electrode with respect to the direction of muscle fibres, and the placement of the electrode relative to the motor end plate regions of the muscle (Andersson & Ortengren, 1974).

The spectrum of the electromyograph (EMG) ranges from 5 to 2000 Hz with most of the signal concentrated in the band between 20 and 200 Hz. Sources other than contracting muscle can generate frequencies that lie within the EMG spectrum and are therefore capable of distorting the EMG signal. The most common distortions or artifacts arise from the contraction of heart muscle (EKG signal) which has frequencies up to 100 Hz, the motion of electrodes or cables which generates frequencies in the 10 Hz range, and 60 Hz interference from power supply mains or nearby electrical equipment (O'Connell & Gardner, 1963; Grossman & Weiner, 1966; McLeod, 1973; Winter, 1979).

As the level of contraction increases, progressively more motor units are recruited and these fire more frequently giving rise to an increasingly complex EMG pattern. However, there is no definite relationship between the amplitude of the motor unit action potentials and the tension generated by the muscle (Eberhart & Inman, 1951). The length of the muscle during contraction can alter the relationship of the EMG to tension and the EMG actually decreases when tension increases at longer muscle lengths. This may be due to the fact that the muscle bulk under the electrodes decreases as the muscle elongates, resulting in fewer motor units located in the immediate vicinity of the electrodes. It may also be due to the fact that tension is greater in elongated muscle. This tension may stimulate the muscle spindle and this will cause inhibition of the alpha motor neuron which, in turn, will result in a decrease in the number of generated motor unit action potentials and a smaller myoelectric signal (Soderberg & Cook, 1984).

It has been found that the EMG generated during a concentric contraction is greater than that generated during an eccentric contraction for similar levels of tension. In eccentric contractions the elastic elements of the muscle may be used to generate tension, therefore less activity will be required from the contractile elements, resulting in a smaller myoelectric signal (Soderberg & Cook, 1984).

While a pure linear relationship between muscle tension and the various parameters of the myoelectric signal has not been found, there is consensus that a functional relationship does exist (Basmajian et al., 1975; Ortengren & Andersson, 1977; Magora & Gonen, 1978; Winter, 1979). According to Rosenfalck (1960) the purpose of measuring the degree of electrical activity of a muscle is to obtain some measure of the mechanical activity of that muscle, information that is difficult to obtain in any other way.

SIGNAL ACQUISITION

The electrical signal from contracting muscle must be received, amplified, processed and recorded before it can be evaluated either qualitatively or quantitatively.

Electrodes

The EMG signal is received by electrodes which must transmit an undistorted signal to the amplifier (Winter, 1979). According to Soderberg and Cook (1984), the function of the electrode is to convert the ionic current of the motor unit action potential into an electronic current which can be displayed, recorded and stored.

Bipolar electrodes, which allow the voltage difference between electrodes to be amplified and processed, are used to record the EMG signal (Basmajian et al., 1975). Two basic types of bipolar electrodes are used in kinesiological studies - fine wire indwelling and surface electrodes. While the use of one type over the other is controversial, the choice depends upon the experimental problem and an understanding of the advantages and limitations of each type.

At the present time indwelling electrodes most frequently consist of two fine wires which are inserted into the muscle by a fine bore hypodermic needle. According to Basmajian & DeLuca (1985), this type of electrode is easily implanted and withdrawn, is relatively painless, gives clear signals and is broad in its pick up. However, Ortengren and Andersson (1977) and Soderberg and Cook (1984) stated that indwelling electrodes have a limited pick up area and therefore do not always give a representation of activity in the whole muscle. Indwelling electrodes are subject to dislocation, kinking and displacement during muscle contraction (Jonsson, 1968; Jonsson & Komi, 1973; Ortengren & Andersson, 1977). In addition, their reliability in day to day application is not as good as that of surface electrodes and therefore they are considered inappropriate for long term studies (Komi & Buskirk, 1970). However, these are the electrodes of choice when precision of measurement is required or when deep or closely placed muscles are to be studied (O'Connell & Gardner, 1963; Basmajian, 1973; Letts et al., 1978; Perry et al., 1981; Basmajian & DeLuca, 1985).

Surface electrodes usually consist of recessed discs of a metal and one of its salts, most commonly silver-silver chloride. They pick up electrical activity from a relatively large volume of underlying muscle, therefore the signal they transmit is considered representative of the whole muscle (Close et al. 1960; Jonsson, 1968; Ortengren & Andersson, 1977). Komi and Buskirk (1970) found that their reliability in day to day applications was good. Surface electrodes are easy to apply and cause no discomfort to the subject as they are attached by adhesive tape or by double-sided adhesive cuffs (Ortengren & Andersson, 1977; Basmajian & DeLuca, 1985). The chief disadvantage of surface electrodes is that their use should be restricted to large, superficial muscles. Signals from superficial muscles will mask the attenuated signal from deeper muscles, therefore the signals from deeper muscles will be distorted. In the case of small muscles, surface electrodes may pick up the signal from muscles adjacent to the ones being studied, again resulting in a distorted signal (Ortengren & Andersson, 1977; Perry et al., 1981; Basmajian & DeLuca, 1985). Consequently, surface electrodes should only be used when simultaneous activity or interplay of activity are being studied in fairly large, superficial groups of muscles, where global pick up is desired, or when overlap in muscle function is inconsequential (Perry et al., 1981; Basmajian & DeLuca, 1985).

Because there is no prescribed method for placement of electrodes, the location of surface electrodes must be done with care (Soderberg & Cook, 1984). Zuniga et al. (1970) found that the average amplitude of myoelectric potentials recorded by surface electrodes was greatest in the middle of the muscle belly and decreased when the position of the electrodes was moved toward the ends or sides of the muscle. In addition, they found that surface electrodes could detect myoelectric activity from quite distant muscles. Perry et al. (1981) stated that localization of the myoelectric signal decreased as spacing between electrodes increased, leading to contamination of the signal by other muscles. This type of signal contamination is referred to as cross-talk (Basmajian & DeLuca, 1985). Localization of the signal from a specific muscle may also be affected by the fact that the skin to which the electrodes are attached moves with respect to the underlying muscle, consequently the signal may not be derived consistently from the same motor units (Marras et al., 1984). Therefore, to ensure optimum localization of the myoelectric signal, surface electrodes should be placed close together over the centre of the belly of the muscle parallel to the direction of the muscle fibres (Waterland & Shambes, 1969; Zuniga et al., 1970; Perry et al., 1981; Soderberg & Cook, 1984; Basmajian & DeLuca, 1985).

The myoelectric signal can be affected by any motion between surface electrodes and the skin and also by the impedance of the tissues that lie between electrodes and muscle. Contact can be improved by securing the electrodes to the skin with adhesive collars and tape. Conductivity can be increased by interposing a saline paste between the skin and the active surface of the electrode, and by reducing skin impedance by carefully cleaning the skin with alcohol to remove protective oils and the dead layer of cells (Basmajian & DeLuca, 1985).

Amplification

The amplitude of the myoelectric signal is low; it ranges from approximately 100 microvolts (μV) to 5 millivolts (mV). Amplification of the EMG signal is therefore required and the amplifier used to do this must be capable of augmenting the signal without distortion. The amplifier should have a range of gains (ratio of output voltage to input voltage) from 100 to 10,000 (Winter, 1979). Also, to avoid attenuation of the signal due to voltage drop across resistances, it is recommended that the input impedance of the amplifier be at least 100 times greater than the electrode/skin impedance, with one megohm usually adequate for surface electrodes (Winter, 1979). The amplifier must be capable of amplifying, without attenuation, all frequencies present in the EMG signal; these range from approximately 5 to 2000 Hz. For surface electrodes it is recommended that the frequency response of the amplifier be sufficient to accommodate a range of 10 to 1000 Hz; for indwelling electrodes the frequency response should accommodate a range from 20 to 2000 Hz (Winter, 1979).

Because most of the EMG signal is concentrated in the band between 20 and 200 Hz, filters may be used to remove some of the frequencies generated by amplifier noise, tissue noise and movement artifact (Winter, 1979; Basmajian & DeLuca, 1985). Movement artifact due to motion of electrodes and/or cables can also be reduced by using small preamplifiers located close to the electrode site (Milner et al., 1971; McLeod, 1973; Winter, 1979). External interference from machinery and power sources may distort the myoelectric signal, therefore a differential amplifier is used which will subtract the signal received by one of the bipolar electrodes from that received by the other electrode. As a result, if a signal is received that is common to both electrodes (common mode signal), the output from the amplifier will be zero and the interference will be eliminated. However, perfect subtraction never occurs and the ability of the amplifier to suppress the common mode signal is referred to as the common mode rejection ratio (CMRR), usually expressed in decibels (dB). It is recommended that a CMRR of 80 dB or higher be used, that is, all but one ten thousandth of the common mode signal is rejected (Winter, 1979).

Processing

The amplified raw EMG signal is difficult to quantify and cannot be faithfully reproduced by, for example, pen recorders. In this type of recording device, the inertia of the pens prevents response to the very rapid changes in the EMG signal beyond 60 Hz. Therefore, some method of processing the raw EMG signal is required. Three types of processing are commonly used: half or full wave rectification, full wave rectification plus lowpass filtering (linear envelope detector), and integration (Winter, 1979).

In full wave rectification the signal has a positive polarity, does not cross the baseline, and fluctuates with the strength of the muscle contraction. The amplitude of the spikes can be measured in millivolts or phasic muscle activity can be evaluated, however the chief use of full wave rectification is as an input to other processing techniques (Hayes et al., undated; Winter, 1979).

The full wave rectified signal can be lowpass filtered to remove all the rapidly changing components of the wave form, leaving an indication of the intensity of the activity at a specified instant in time, or a moving average. The moving average follows the trend of the electromyogram and closely resembles the shape of the raw curve. The wave form thus generated is called a linear envelope and is measured in millivolts. It is far easier to quantify than the raw or full wave rectifed signal (Winter, 1979).

Integration processing techniques use the signal parameters of amplitude and time, therefore measurements are expressed in millivolt seconds (mV.s). Three methods of integration are in common use: integration that starts at a preset time and continues during the total time of the muscle contraction; the integrated signal that can be reset to zero at regular intervals of time, usually 40 to 200 msec, then repeated; or the integrated signal that can be reset to zero when it reaches a specified voltage level then repeated (Winter, 1979).

According to Siegler et al. (1985), despite great efforts in recent years to develop new EMG processing techniques, there is no general agreement as to which technique is best, and researchers select a technique on the basis of subjective preference and convenience. The selection of the proper processing technique depends upon the problem being investigated. The raw EMG signal may be most appropriate when small magnitudes of muscle activity are being studied (Kelley, 1971). If measurement of the level of muscle activity is required, integration may be a useful processing technique. Because there is a time lag between the raw and filtered signal in the linear envelope, this technique should be used with caution if precise measurement of the timing of the EMG signal is required. This time lag, caused by the lowpass filter, must be taken into account if the analog EMG signal is to be converted to digital format for computer storage and if the digitized data are to be smoothed again with software filters (Winter, 1979; Halbertsma & DeBoer, 1981). However, because the linear envelope closely follows the rising and falling of muscle tension in a contraction, linear envelope processing appears to be the best way to produce an analog representation of the EMG signal that can be shared between laboratories or used in clinical settings (Winter, 1984).

Recording

The EMG signal must be recorded in order for it to be studied and measured. A wide variety of recording techniques is available and selection of the most appropriate one depends upon the nature of the experimental problem. The raw EMG signal contains high frequency components, therefore it is best recorded on a cathode ray oscilloscope and later photographed. The raw signal can also be recorded by a light beam or an ink jet recorder with a high frequency response, or on magnetic tape (Kelley, 1971; McLeod, 1973). For processed signals, pen writers can be used. These are inexpensive and easily serviced, their chief limitation being a relatively low frequency response (Grossman & Weiner, 1966; McLeod, 1973; Grieve, 1975; Basmajian & DeLuca, 1985). EMG signals that have been recorded on magnetic tape can be converted to digital format for computer processing and analysis and, with the advent of microcomputers, it is now possible to record the EMG signal directly onto floppy discs (Basmajian & DeLuca, 1985).

Because computers must receive signals in numerical form, the analog EMG signal must be converted to a digital format. This is done by feeding the signal to the input terminals of an analog-to-digital convertor. The signal is sampled at a rate controlled by the computer and is changed into a series of short duration pulses of the same amplitude as the original analog signal at the time it was sampled. The sampled pulse is converted to binary format and the signal is then represented by a number of bits or "words" which can be stored in the computer memory. The sampling rate of the computer should be governed by the Sampling Theorem, i.e. the sampling rate should be at least twice the value of the highest frequency component in the EMG signal (McLeod, 1973; Winter, 1979; Basmajian & DeLuca, 1985).

SIGNAL EVALUATION

Interpretation of EMG data is complicated because the recorded signal reflects only the activity of the contractile elements of the muscle and does not represent the role of the elastic elements which are in series and in parallel with the contractile elements. In addition, it must be remembered that the EMG signal is a manifestation of the electrical events occurring in the sarcolemma of contracting muscle fibres and is not a representation of movement of cross-bridges in the sarcomere (Gordon, 1982; Soderberg & Cook, 1984).

According to Basmajian (1978), the evaluation of recordings is the most abused element of electromyography. To date there is no universally approved method of analysis and quantification of the EMG signal and, as with all other aspects of electromyography, the evaluation method depends upon the the experimental problem. Analysis can be subjective or objective, although both methods should be used for most EMG recordings (Kelley, 1971).

Subjective analysis involves careful visual inspection of the EMG recording and a system of classification is used that assigns symbols and descriptors to represent different levels of magnitude of electrical activity. The investigator who uses this approach should be well trained and have a great deal of experience (Kelley, 1971). Hirose et al. (1974) and Ortengren and Andersson (1977) stated that it is difficult to distinguish between levels of activity or to compare levels of high activity when using a subjective analysis method. Magora and Gonen (1978) pointed out that visual analysis of the EMG recording is time consuming and often inaccurate.

Objective analysis involves a variety of techniques: measurement of temporal or positional events, amplitude, frequency, or of the amount of electrical activity generated over a defined period of time (Willison, 1963; Kelley, 1971; Hirose et al., 1974; Basmajian et al., 1975; Ortengren &

Andersson, 1977; Basmajian & DeLuca, 1985). Events such as onset, peak and cessation of muscle activity can be quantified by comparing them to an external standard or event such as a tracing from an electrogoniometer or from footswitches, a synchronizing light pulse on cinefilm, or event markers on a pen recorder (O'Connell & Gardner, 1963; Gray & Basmajian, 1968; Sutherland & Hagy, 1972; Peat et al., 1976; Brandell, 1977; Tata, 1980; Quanbury, 1981; Nilsson et al., 1985).

Measurements of amplitude, frequency, or total amount of electrical activity cannot be compared directly between muscles of the same subject or between subjects because of the uncontrolled variation in the size of the motor unit pools being sampled (Grossman & Weiner, 1966; Lyons et al., 1983; Soderberg & Cook, 1984). Between muscle and between subject comparisons can only be made if some standard is set for each muscle against which all subsequent activity of that muscle can be measured (Basmajian, 1978; Perry et al., 1981). One method of establishing a standard is to record a maximum voluntary isometric contraction from each muscle under investigation. All subsequent activity of the muscle is compared to this maximum voluntary isometric contraction and is expressed as a percentage of it. This normalization process allows comparison of electrical activity between muscles in the same or in a different subject (Basmajian et al., 1975; Letts et al., 1978; Perry et al., 1981; Lyons et al., 1983; Soderberg & Cook, 1984; Miller, 1985).

Although comparison to the maximum voluntary isometric contraction is commonly used as a normalization procedure, results of a study by Yang and Winter (1983) indicated that it may not be the most reliable method. The assumption made when using a maximum volutary isometric contraction is that all the motor units in the recording area of the electrode are firing at their maximum rate. However, because of synergic contraction of muscle groups, it may be that no one muscle contracts at its maximum even during maximum voluntary effort.

Another method of establishing a standard for muscle contraction is to measure the electrical activity of a muscle when the joint it crosses is in a defined position and the muscle is required to work against a known external force. This method was used by Andersson and Ortengren (1984) to establish a baseline standard for contraction of erector spinae muscle. Subjects were required to stand in 30° trunk flexion (measured with a hydrogoniometer at the T_{10} level) and hold a 10 kg weight in each hand for a duration of 30 seconds during which time electrical activity was recorded. This activity then served as the standard against which subsequent contractions were measured.

Manual measurement of the various parameters of the EMG signal is time consuming and may be inaccurate. Therefore researchers are now using computers to analyze the signal. The advantages of using a computer are that it is accurate, uniform, efficient and not arbitrary. In addition, because of its speed, large amounts of EMG data can be sampled by the computer, possibly giving a better representation of muscle activity than could be obtained from fewer data (Hirose et al., 1974; Basmajian et al., 1975; Grieve, 1975; Halbertsma & deBoer, 1981).

CHAPTER III

MATERIALS AND METHODS

INTRODUCTION

The general purpose of this study was to investigate trunk muscle activity in the types of locomotion commonly found in the workplace; that is, level walking, stair climbing, stair descent, ramp climbing and ramp descent. The specific aims of the investigation were:

For each of Level Walking, Stair Climbing, Stair Descent, Ramp Climbing and Ramp Descent –

1. to measure and compare trunk motion in the sagittal plane.

2. to measure and compare temporal events in the gait cycle.

3. to measure and compare the myoelectric activity of the trunk muscles considered to be primarily responsible for motion in the sagittal plane, namely erector spinae and rectus abdominis.

4. to determine and compare the relationships between onset, peak and cessation of myoelectric activity to trunk motion in the sagittal plane and temporal events in the gait cycle.

The equipment and facilities of the Locomotion Laboratory, Rehabilitation Centre for Children, Winnipeg, Manitoba were used in this investigation.

SUBJECTS

One of the aims of the study was to acquire normal data for later comparison to data from subjects with low back injury sustained in the workplace. In 1985, 79.5% of claimants to the Workers Compensation Board of Manitoba were male and 54.5% of claimants were between the ages of 20 and 34 (Workers Compensation Board of Manitoba, 1985). Because the majority of injured workers in Manitoba are male, only male subjects were selected for this study.

Subjects were chosen on the basis of their availability (Currier, 1984). They were required to be between the ages of 18 and 35, in good health, and of normal weight. Exclusion criteria were a history of spinal pathology and active lower limb pathology as these might have affected the subject's gait.

The Faculty Committee on the Use of Human Subjects in Research approved this study, all subjects were volunteers and received no remuneration. Each subject was required to sign a consent form before participating in the study (Fig. 1). Twenty subjects were selected for the study. Complete data were collected from 18 subjects and their descriptive data are presented in Table 2.

FIGURE 1.

HE UNIVERSITY OF MANITOBA

FACULTIES OF MEDICINE AND DENTISTRY
Department of Anatomy

730 William Avenue
Winnipeg, Manitoba
Canada R3E 0W3

CONSENT FORM

I hereby consent to act as a subject in a research project on the role of the trunk muscles during level walking, stair climbing and descent, and ramp climbing and descent. The procedure, which includes placement of surface electrodes on back and abdomen, attachment of foot switches to shoes and photographic recording, has been explained to me fully.

Data from cinefilm, foot switches and surface electrodes will be collected and stored by computer for analysis. Photographs will also be taken and these, with the computer data, will be used for analytical purposes and may also serve as documentation in research papers and medical lectures.

My participation in this research project is voluntary and I reserve the right to withdraw immediately from the procedure whenever I wish.

Signature of subject:____________________________________

Date:____

TABLE 2. DESCRIPTIVE DATA OF SUBJECTS

NUMBER OF SUBJECTS: 18

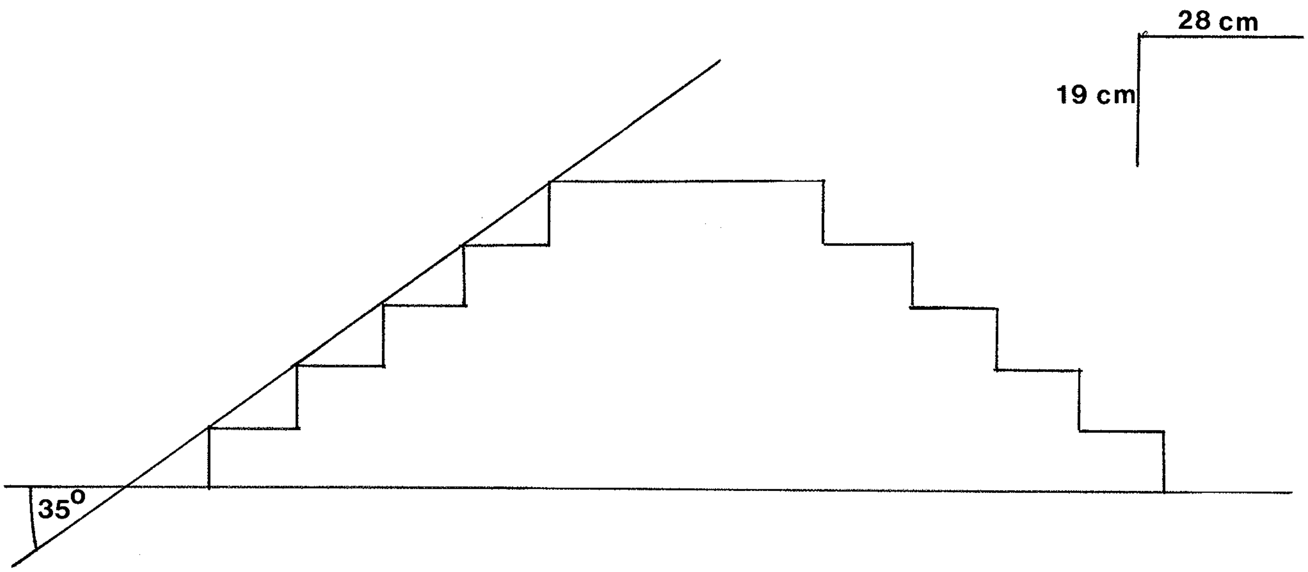
AGE:	MEAN	26.2	3.9 yrs
	RANGE	20.4 - 32.3 yrs	
HEIGHT:	MEAN	1.74	.06 m
	RANGE	1.62 - 1.83 m	
WEIGHT:	MEAN	69.9	7.3 kg
	RANGE	56.7 - 85.2 kg	

APPARATUS

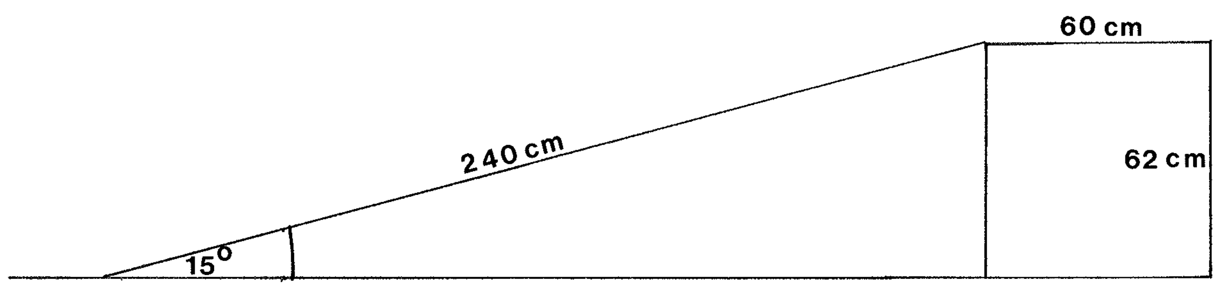
The existing walkway at the Locomotion Laboratory, Rehabilitation Centre for Children was used. It is 9.5 m long, 1.8 m wide and is elevated 0.4 m from the floor. The dimensions of this walkway are similar to those of walkways used by Cappozzo et al (1976), Mizrahi et al. (1982) and Winter (1982, 1983).

An existing set of portable stairs was modified for the study to add an extra step for a total of five steps. The dimensions of the stairs met the recommended standards of the Workplace Safety and Health Division of the Province of Manitoba (Krywulak, 1983). These standards recommend that the slope for stairs be between 30° and 35°. For stairs with a slope of 35°, the recommended height of the riser is 19 cm, the recommended length of the tread and nosing is 26 cm. Fitch et al. (1974), in their investigation of various combinations of stair dimensions, recommended a riser height of 10 to 18 cm and a tread length of 28 to 37 cm. The dimensions of the stairs used in this study were: slope 35°, riser 19 cm, tread and nosing 28 cm (Fig. 2). These dimensions are similar to those of stairs used by Bruce et al. (1967), Joseph and Watson (1967), Townsend, Lainhart et al. (1978), Townsend, Shiavi et al. (1978), Andriacchi et al. (1980) and Lyons et al. (1983).

FIGURE 2. DIAGRAM OF STAIRS AND RAMP USED IN THE STUDY.



DIMENSIONS OF STAIRS



DIMENSIONS OF RAMP

Figure 2.

An adjustable portable ramp was designed and constructed to meet the recommended standards of the Workplace Safety and Health Division of the Province of Manitoba which state that the maximum preferred slope for ramps and inclines is 15° (Krywulak, 1983). Fitch et al. (1974), in their investigation of ramp dimensions found that the total energy cost per vertical meter decreased as the slope increased until a slope of 17° was reached, after which the energy cost rose. The dimensions of the ramp used in this investigation are presented in Fig. 2. Previous studies of slope walking on a treadmill were conducted by Waters and Morris (1970) and by Brandell (1977); in both studies the maximum slope of the treadmill was 10°. Corlett et al. (1972) constructed a ramp that could have the slope adjusted from 11° to 30°.

ELECTROMYOGRAPHY

CHOICE OF ELECTRODES

Forty-six studies of erector spinae were reviewed; only the most relevant will be cited. Surface electrodes were used in 30 studies, indwelling electrodes were used in 12 studies and both types of electrodes were used in four studies. Not all investigators gave reasons for their choice of one type of electrode over another. Ease of application, lack of discomfort for the subject and sampling from a large and therefore representative volume of muscle were reasons given for selection of surface electrodes in the studies of Eberhart and Inman (1951), Klausen (1965), Andersson and Ortengren (1974), and Kippers and Parker (1984). Golding (1952) used surface electrodes as, in his opinion, these gave findings similar to those from indwelling electrodes. Pauly (1966) and Pauly and Steele (1966) chose indwelling electrodes because they allow study of deep as well as superficial muscles and also allow study of individual muscles without interference from adjacent muscles. Waters and Morris (1972) chose indwelling electrodes because they considered surface electrodes to be too prone to motion artifact to be able to record accurately from a walking subject. Floyd and Silver (1951, 1955) used both surface and indwelling electrodes as they wished to record simultaneously from superficial and deep portions of the back muscles. Letts et al. (1978) used both indwelling and surface electrodes because of the possibility of signal contamination by the activity of multiple overlying paraspinal muscles.

Thirty-three studies of rectus abdominis were reviewed; again, only the most relevant will be cited. Surface electrodes were used in 22 studies, indwelling were used in 10 studies and both types of electrodes were used in one study. Again, reason for choice was not always given. Ease of application, lack of discomfort and the ability to sample from a large volume of muscle were among the reasons given for selection of surface electrodes by Campbell and Greene (1953), Ono (1958), Hatami (1961a), Klausen (1965) and Booth et al. (1980).

In the present study it was decided that surface rather than indwelling electrodes would be used because the two muscles under investigation are large and superficial and therefore it is not difficult to obtain uncontaminated signals from them. Also, because rectus abdominis is relatively thin, it might be difficult to judge the depth of penetration during insertion of indwelling electrodes. Too deep an insertion might result in accidental penetration of the parietal peritoneum.

POSITION OF ELECTRODES

To establish precisely where surface electrodes should be placed over erector spinae and rectus abdominis muscles, a review of major anatomy textbooks and of relevant literature was carried out.

ERECTOR SPINAE

Specific electrode placement in EMG studies of erector spinae muscle has varied. Researchers have placed electrodes in lumbar areas, thoracic areas and in both lumbar and thoracic areas simultaneously.

Erector spinae activity was sampled at 12 levels from C_4 to S_2 by Joseph and McColl (1961). Blackburn and Portney (1981) also sampled activity from cervical to sacral levels with surface electrodes placed at the levels of the spinous processes of C_5, T_6, L_3 and S_1 vertebrae. Activity from the level of the C_4, T_8 and L_4 spinous processes was sampled by Schultz, Andersson, Ortengren, Bjork and Nordin (1982) and by Schultz, Andersson, Ortengren, Haderspeck and Nachemson (1982). Activity of erector spinae from upper thoracic to lower lumbar levels was investigated by Andersson et al. (1976, 1977) and by Ortengren et al. (1981), while activity from lower thoracic to lower lumbar levels was investigated by Floyd and Silver (1951), Pauly (1966), Pauly and Steele (1966), Jonsson (1970), Letts et al. (1978), Soderberg and Barr (1983) and by Andersson and Ortengren (1984). Morris et al. (1962) and Waters and Morris (1970) sampled erector spinae activity in the lower thoracic region by inserting indwelling electrodes into the muscle over the posterior aspect of the ribs medial to their angle.

Of those researchers who confined their investigation of erector spinae to the lumbar area, muscle activity was simultaneously sampled at various levels by Golding (1952), Portnoy and Morin (1956), Morris et al. (1961), Battye and Joseph (1966), Joseph and Watson (1967), and Wolf and Basmajian (1978). Others have confined their investigation to one specific level in the lumbar area. For example, Chapman and Troup (1969) placed electrodes at the level of the second lumbar spinous process, while Floyd and Silver (1955), Berkson et al. (1977), Cooper (1982) and Mayer et al. (1985) placed electrodes at the level of the third lumbar spinous process. Bendix et al. (1984) placed electrodes at the level of L_4, while Ekholm et al. (1982) located their electrodes between the fourth and fifth lumbar levels. It was found that the majority of investigators of erector spinae activity placed surface electrodes at the level of the third lumbar spinous process.

Centering of electrodes has also varied. Morris et al. (1961) placed electrodes in the main muscle mass, Morris et al. (1962) inserted electrodes one quarter of the distance from the angle of the rib to the vertebral spinous process, while Battye and Joseph (1966) and Joseph and Watson (1967) placed electrodes along the lateral border of the muscle. Specific distances measured in centimeters from vertebral spinous processes were used in 15 studies and, in 12 of the 15, the distance used was 3 cm. Rather than use a direct measurement in centimeters from a vertebral spinous process, Pauly (1966), Pauly and Steele (1966) and Cooper (1982) placed electrodes halfway between the vertebral spinous process and the lateral border of erector spinae muscle, thus ensuring that the location of the electrodes was over the main muscle mass, but in a standardized location.

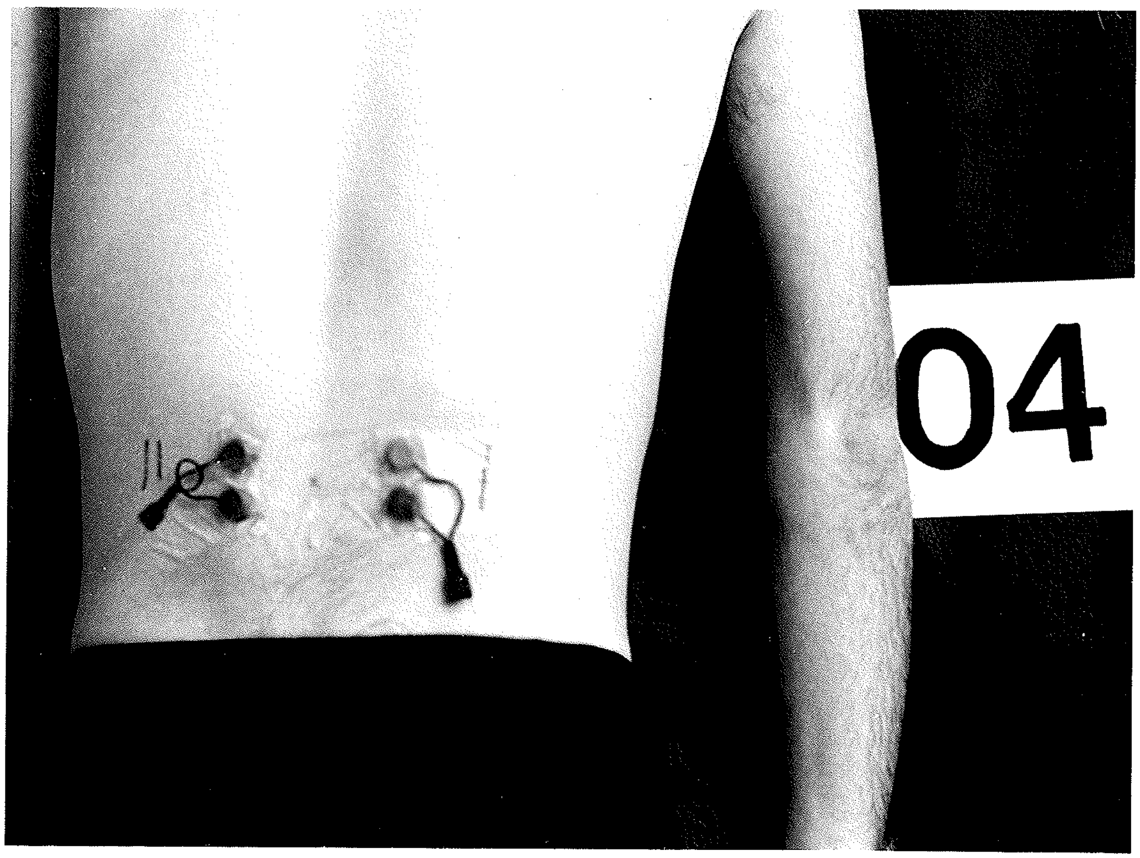
Floyd and Silver (1955), Portnoy and Morin (1956), Battye and Joseph (1966), and Cooper (1982) stated that a check was done to rule out interference from other muscles, however only Morris et al. (1962) and Cooper (1982) reported that cadaver specimens had been studied to verify electrode placement.

Following a review of major anatomy texts, reports in the literature and the results of a previous study (Cooper, 1982), it was decided that electrodes would be situated over erector spinae at the level of the third lumbar spinous process. At this level erector spinae forms a large, prominent, fleshy mass and is covered only by the posterior layer of the thoracolumbar fascia. The landmarks used for centering the electrodes over the muscle mass were the third lumbar spinous process and the lateral border of erector spinae. These points were marked with water soluble ink and the electrodes were placed halfway between the two points, parallel to the muscle fibres (Fig 3).

RECTUS ABDOMINIS

Electrode placement for rectus abdominis in EMG studies has varied, however 22 of 26 investigators who reported on electrode placement used the umbilicus as a landmark. Of

FIGURE 3. PLACEMENT OF ELECTRODES ON ERECTOR SPINAE.


04

these 16 used the umbilicus as the sole landmark while six used both the umbilicus and the xiphoid process as landmarks. Morris et al. (1961), Lipetz and Gutin (1970), and Booth et al. (1980) did not report specific locations for electrode placement. Mayhew et al. (1983) used the xiphoid process as the sole landmark while Cooper (1982) used the anterior superior iliac spine (ASIS) and the costal margins as bony landmarks.

Centering of electrodes has also varied and descriptions of centering were found in eight investigations. Waters and Morris (1970) placed indwelling electrodes one centimeter lateral to the linea alba. Electrodes were placed two centimeters from midline by DeSousa and Furlani (1974), Lansing and Meyerink (1981), Schultz, Andersson, Ortengren, Bjork and Nordin (1982), Schultz, Andersson, Ortengren, Haderspeck and Nachemson (1982), and by Mayhew et al. (1983); Bendix et al (1984) centered electrodes three centimeters from the midline. Cooper (1982) centered electrodes by placing them halfway between the linea alba and the lateral border of the rectus abdominis muscle.

Although the possibility of interference from adjacent muscles was raised by Floyd and Silver (1950), Klausen (1965) and Battye and Joseph (1966), only Waters and Morris (1970) stated that a cadaver study was conducted to confirm electrode placement. Cooper (1982) conducted an experiment to locate interference-free positions for surface electrodes placed over the abdominal musculature.

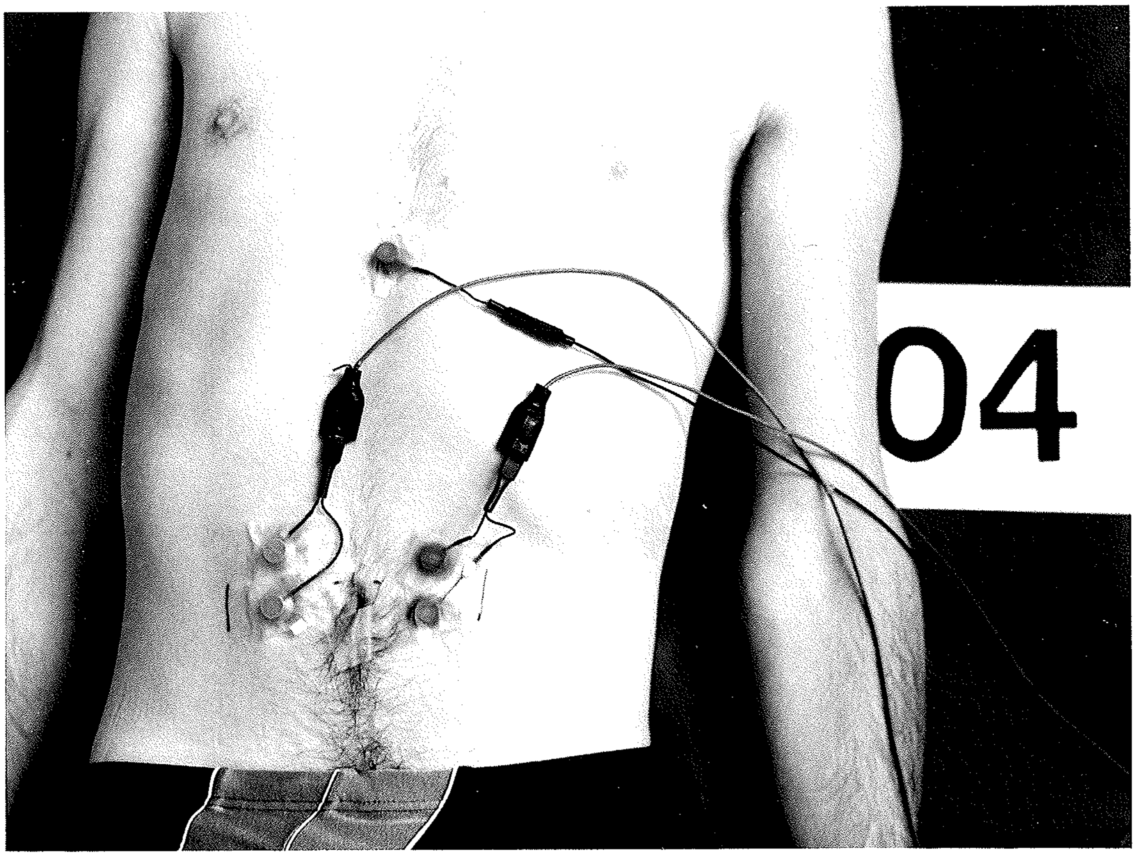
Placement of electrodes over rectus abdominis in the present investigation was based on the study by Cooper (1982) in which bony landmarks were used to standardize electrode location in all subjects. Using water soluble ink, a line was drawn from the xiphoid process to the symphysis pubis. At a point halfway along this line a horizontal line, perpendicular to the first, was drawn, extending to the lateral margin of the rectus abdominis muscle. Electrodes were placed parallel to muscle fibres, equidistant above and below the midpoint of this horizontal line (Fig. 4).

ELECTROMYOGRAPHIC EQUIPMENT

Paired Beckman silver-silver chloride surface electrodes were used. These had an overall diameter of 14 millimeters (mm), with an active surface of 5 mm^2. By using adhesive cuffs as a guide, spacing between the electrodes of each pair was kept constant at 10 mm. The hollow portions of the electrode were filled with electrode electrolyte gel and the electrodes were fastened to the skin by double sided adhesive cuffs and Millipore brand surgical tape.

In order to minimize motion artifacts, each pair of electrodes was attached at the electrode site to a preamplifier with an input impedance of 2.2 megohms. Wires

FIGURE 4. PLACEMENT OF ELECTRODES ON RECTUS ABDOMINIS.


04

from the preamplifiers were conveyed to a junction box which could be belted to the subject's waist. The junction box was connected to the amplifier by a single flexible multiconductor cable.

An eight-channel amplifier was used which was designed and constructed by the Biomedical Engineering Department, Rehabilitation Centre for Children. It has a bandwidth of 50 to 300 Hz with an input impedance of 2.2 megohms and a common mode rejection ratio of 90 dB. The EMG signal was full wave rectified and lowpass filtered to produce a linear envelope configuration. A first order lowpass filter was used with a time constant of 45 msec and a frequency cutoff of 35 Hz.

The analog signal from the muscles was recorded by a Gould Brush 8 channel ink chart recorder (Model 481). The recorder has a bandwidth frequency response of 50 divisions $\pm$ 1 division dc to 40 Hz, 10 divisions $\pm$ 1 division dc to 100 Hz. A rectilinear trace presentation was given with the sensitivity of the recorder adjustable from 1 mV/division to 500 V/division. The analog chart record was used as a backup for computer recording.

Analog to digital (A/D) conversion of the EMG signal was carried out on a Hewlett-Packard 6940B multiplexer with a 69336B High Speed Scanner Card and a 69920A Timer Pacer Card which allowed all EMG channels to be sampled almost instantaneously. The resolution of the A/D convertor is 12 bits with a dynamic range set at ± 10 V which gives a 5 mV resolution. The sampling rate of the multiplexer was 50 Hz per channel and the digital data were sent to a Hewlett-Packard 9826 microcomputer and stored on a floppy disc.

CINEMATOGRAPHIC EQUIPMENT

Cinefilm was used to record trunk position in the sagittal plane.

A 16 mm Teledyne DBM 54 motor driven camera fitted with an Angenieux f10 lens was used at a speed of 50 frames per second. The frame rate was checked with a synchronizing light during the course of the study. Kodak 16 mm 7240 double perforated color film with an ASA of 125 (tungsten lighting) was used.

For level walking the camera was mounted on a mobile cart located 4 m from one edge of the walkway. For stair and ramp walking the camera was mounted on a tripod kept at constant height, 4 m from the same edge of the stairs and ramp.

A black cloth backdrop was mounted immediately behind each apparatus. Against the backdrop and within the field of view of the camera were mounted a series of circular reflective reference markers, each 7 cm in diameter and spaced 24 cm apart measured from center to center.

In order to synchronize the film with the EMG signals, a portable signal light was attached to the walkway, stairs or ramp, within the field of view of the camera. The light consisted of a cluster of 4 LEDs powered by a 9 V battery, and was connected in parallel to the chart recorder and the multiplexer. When the light circuit was closed at the beginning of filming, an analog signal was traced on one channel of the chart recorder and the computer stored the digital data from muscles and footswitches only during the period in which the light circuit was closed.

After the exposed film was developed, it was projected using a L-W Photo Optical Data Analyzer. This device allowed the film to be projected one frame at a time onto the surface of a digitizing tablet (GT Corp.) which had a 50 cm by 50 cm active area and a resolution of .025 mm.

FOOTSWITCHES

The temporal events in the gait cycle were recorded by means of three footswitches (Tapeswitch Corp. of America) attached with adhesive tape to the sole of each of the subject's shoes. One footswitch was positioned parallel to the long axis of the foot at the region of the hallux, the second was aligned with the position of the metatarsal heads, and the third was placed perpendicular to the long axis of the foot at the calcaneus. The switches were connected to the power supply and resistor network in such a way that discrete voltage levels were obtained for any combination of switch closures. In this way various events in the Stance phase of gait could be identified. The footswitch signal was conveyed to the multichannel amplifier system and from there simultaneously to the chart recorder and multiplexer where it was A/D converted with the EMG signals and sent to the computer for storage.

EXPERIMENTAL PROTOCOL

The Pre-run Check List and the Experiment Routine used in this study are presented in Appendix I.

Each subject wore shorts or bathing suit and his own shoes (Joseph & Watson, 1967; Cappozzo et al., 1978; Townsend, Lainhart, et al., 1978; Cappozzo, 1981; Winter, 1983; Thorstensson et al., 1984; Nilsson et al., 1985). The footswitches were attached to the sole of each shoe with adhesive tape. Using the analog signal on the chart recorder, the footswitch output was tested both before and after the subject put on the instrumented shoes.

Bony landmarks were used to standardize electrode application. All electrode pairs were applied by the same person, using the same measuring tools. Skin resistance was reduced to practical levels by rubbing it vigorously with an alcohol swab (Quanbury, 1981). To avoid mistakes, the leads to the junction box were attached to the electrodes and footswitches in the same order each time. The junction box was attached comfortably to the subject's waist with an adjustable belt.

A maximum voluntary isometric contraction (MVC) was obtained from each muscle in order to establish a standard of muscle activity against which all subsequent contractions of that particular muscle could be measured. The MVC was done using the positions recommended by Daniels and Worthingham (1980) for testing normal strength of erector spinae and rectus abdominis muscles in order to ensure maximum motor unit recruitment from each muscle (Lyons et al., 1983). Three maximum effort trials, each of 5 sec duration, were obtained for each muscle and recorded on the chart recorder. The first two trials were done for practice and warmup and the third trial was stored by the computer (Bigland-Ritchie et al., 1983; Lyons et al., 1983).

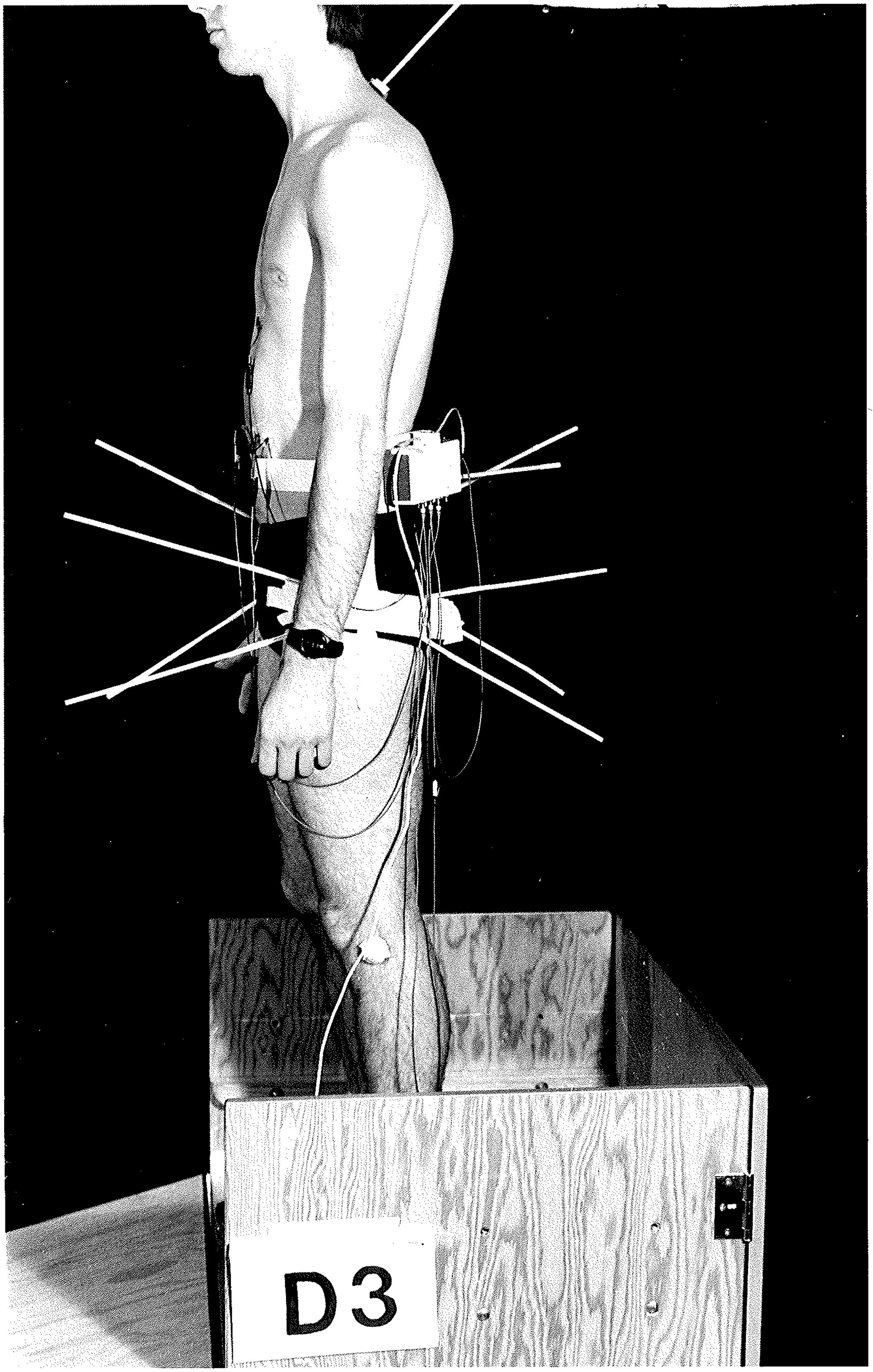
Surface targets placed over anatomical landmarks were used to demarcate specific body segments. Surface targets are widely used in motion studies although it is recognized that there will be relative movement of the skin and target over the underlying bony landmark. A study was done by Thorstensson et al. (1982) to determine the extent of the movement between target and anatomical landmark that occurred when the trunk was placed in different positions; it was found that the discrepancy never exceeded 2 mm. Other investigators have calculated that the error that occurs when using skin mounted targets during studies of normal walking is approximately 8% (Thurston et al., 1981; Thurston & Harris, 1983; Thurston, 1985). The alternative to surface targets is pins surgically inserted into bony landmarks (Eberhart & Inman, 1951), however the pain resulting from such a procedure might alter the normal gait pattern.

The targets used in this study were placed over the C_7 and L_5 spinous processes, the right and left greater trochanters and at the centre of the lateral aspects of the right and left knee joint lines (Eberhart & Inman, 1951; Thurston et al., 1981; Thorstensson et al., 1982; Thurston & Harris, 1983; Winter, 1983; Kirtley et al., 1985). The C_7 and L_5 targets delimited the trunk segment, the L_5 and greater trochanter targets delimited the pelvic segment, while the greater trochanter and knee joint line targets delimited the thigh segment. All targets were applied by the same person.

The C_7 and L_5 targets consisted of stiff white plastic rods, the tips of which were wrapped in reflective tape. The rods measured 14 cm in length and were mounted in foam rubber blocks which were taped securely to the skin (Thorstensson et al., 1982, 1984). For the greater trochanters an X-shaped target of 60 cm long reflective plastic rods was used. The point of intersection of the rods was placed over the upper portion of the greater trochanter and taped firmly in place (Fig. 5). This type of marker was necessary because a small marker on the greater trochanter might be obscured by the upper limb during walking (Kirtley et al., 1985). In addition, requiring the subject to hold the upper limb away from the greater trochanter in order to keep the marker in view might result in disruption of the normal walking pattern. The knee joint target consisted of a styrofoam semicircle covered with reflective plastic film and fastened to the skin with double-sided adhesive tape (Thurston et al., 1981; Thurston & Harris, 1983; Kirtley et al., 1985).

To avoid bias, the sequence of use of the walkway, ramp and stairs was randomly determined by coin toss (Corlett et al., 1972). Because the position of the camera was changed between the level walking trials and the ramp/stair trials, it was decided that the randomization should be done in two steps to ensure that the camera was moved only once during each experiment. In the first step the walkway was considered as one option and the stairs and ramp were jointly considered as the other option. Therefore the first coin toss determined which of the walkway or the stair/ramp combination would be used first. In the second step, the coin toss was used to determine which of stairs or ramp would

FIGURE 5. PLACEMENT OF SURFACE TARGETS TO DELIMIT BODY SEGMENTS.


D3

be used first. Given these conditions, the sequence of use of the apparatus shown in Fig. 6 was used. The stairs and ramp were kept in a constant position by aligning them with markers taped to the floor.

The different types of locomotion that resulted from use of the apparatus were designated Level Walking, Stair Climbing, Stair Descent, Ramp Climbing and Ramp Descent. Each subject was required to perform three trials of each type of locomotion and the middle stride of each trial was used for data analysis (Townsend, Lainhart et al., 1978; Lyons et al., 1983). To ensure that the walking patterns were as normal as possible in an experimental situation, subjects were instructed to walk on the apparatus at a comfortable speed (Battye & Joseph, 1966; Cappozzo et al., 1976; Dubo et al., 1976; Pare et al., 1981; Thurston et al., 1981; Thurston & Harris, 1983; Kirtley et al., 1985). To minimize the effects of warmup and fatigue, the subject was required to rest for five minutes between each apparatus (Andersson et al., 1976; Brandell, 1977).

All subjects were thanked for their participation in the study.

FIG. 6. SEQUENCE OF USE OF APPARATUS

SUBJECT NUMBER	APPARATUS
01	Walkway, Stairs, Ramp
02	Stairs, Ramp, Walkway
03	Walkway, Ramp, Stairs
04	Walkway, Ramp, Stairs
05	Walkway, Ramp, Stairs
06	Walkway, Stairs, Ramp
07	Ramp, Stairs, Walkway
08	Ramp, Stairs, Walkway
09	Walkway, Stairs, Ramp
10	Walkway, Stairs, Ramp
11	Ramp, Stairs, Walkway
12	Walkway, Stairs, Ramp
13	Ramp, Stairs, Walkway
14	Walkway, Stairs, Ramp
15	Ramp, Stairs, Walkway
16	Walkway, Stairs, Ramp
17	Stairs, Ramp, Walkway
18	Ramp, Stairs, Walkway

DATA COLLECTION AND EXTRACTION

Fig. 7 shows the arrangement of the instrumentation used in this study to collect the data on EMG, temporal events and body segment displacement.

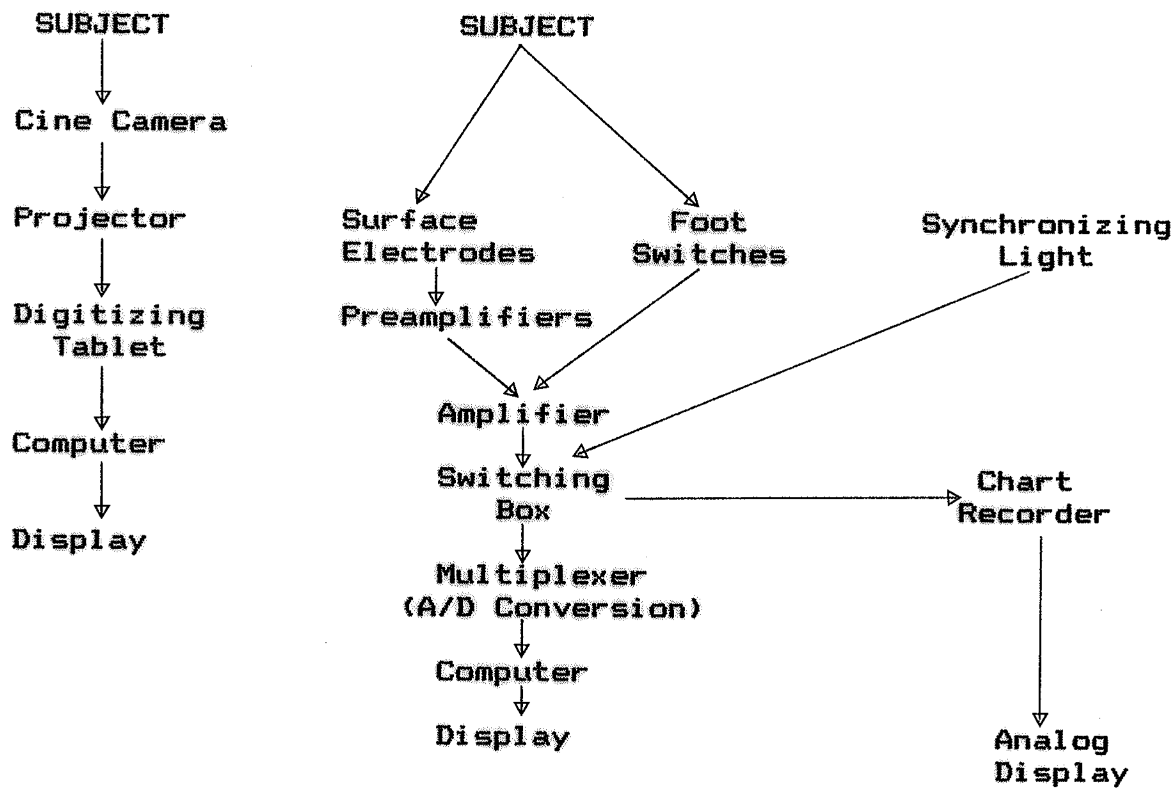
A computer program, LRS, was specifically written for this study. Values derived from the program were cross checked with analog data from the chart recorder and with values from manual measurement to ensure that reliable and valid data were generated by the program. A copy of the program was written onto 20 floppy discs, one for each subject. All the data files for that subject were stored on his own program disc.

EMG AND FOOTSWITCH DATA

The subprogram EMGIN was used to collect the reference contraction (MVC) data; this resulted in one file of raw EMG data for each muscle being stored on disc for a total of four reference contraction EMG data files.

EMGIN was also used to collect the EMG and footswitch data during the five types of locomotion (3 trials each). The computer sampled the EMG and footswitch signals only during the interval in which the synchronizing light was on, resulting in one file of raw data per trial per apparatus

FIG. 7. FLOW DIAGRAM OF DATA ACQUISITION



being stored on disc. A total of 15 files of raw EMG data per subject was generated by EMGIN.

The subprogram RFCCALC allowed an analog signal of each MVC to be displayed on the computer monitor. The beginning and end of a representative part of the MVC signal was demarcated by using a movable vertical line marker that was manually scrolled across the computer monitor. Each position of the vertical marker was indicated by a frame number on the monitor and the frame numbers of the beginning and end of the representative sample were recorded on paper to allow for cross checking at a later date, if necessary. The marked MVC signal was then stored to give one file of refined MVC data per muscle for a total of four refined MVC files per subject.

Subprogram EMGCALC allowed analog footswitch signals and analog EMG signals from each muscle during each trial to be displayed individually on the computer monitor. The temporal events of Initial Contact and End of Weight Bearing were marked on each footswitch tracing by using a movable vertical arrow and soft keys. The right gait cycle was used as the reference against which all other measurements were made. Once marked on the footswitch tracings, the points of Right Initial Contact and Right End of Weight Bearing also appeared on the EMG tracings for each muscle and on these tracings the additional points of onset and cessation of myoelectric activity could be marked using the movable vertical arrow and soft keys. In addition, up to three peaks of maximum amplitude of each myoelectric signal were marked and labelled with the movable arrow and soft keys. The position of the movable arrow was indicated by a frame number on the computer monitor and the frame numbers of all marked positions were recorded on paper.

Results from preliminary runs showed that EMG activity from rectus abdominis during locomotion was low compared to the activity recorded during a MVC. This resulted in a "noisy" signal being displayed on the computer monitor when the EMGCALC subprogram was run, making identification of onset, peak and cessation positions difficult. To remedy this the gain on the rectus abdominis channels of the amplifier was increaed from 2,500 during the MVC to 4,000 during the locomotion trials and a conversion formula was written into the EMGCALC subprogram to compensate for the increase in gain.

Once all the designated events were marked on the footswitch and EMG tracings, the length and cadence of a right gait cycle and the positional events of both right and left gait cycles were calculated by the subprogram and were displayed on the computer monitor. The amount of myoelectric activity that occurred during the right gait cycle was calculated and displayed as a percentage of a MVC; the points of onset, peaks (1,2 and 3) and cessation of myoelectric activity that occurred during the right gait cycle were calculated and displayed as percentages of the right gait cycle. These refined data could be stored, giving 15 files per subject (5 types of locomotion, 3 trials each). The refined data could also be printed. From this printout an average for each parameter for the three trials of each type of locomotion was calculated and recorded on a data sheet. In addition, the mean values were entered into a computer file using the STATS PLUS program (Human Systems Dynamics) for later statistical analysis.

BODY SEGMENT DISPLACEMENT DATA

The processed film of each subject was first projected from beginning to end onto the digitizing tablet in order to check the sequence of the types of locomotion and of the trials within each type. A frame of film from the middle of each trial was selected and the position of the C_7, L_5 and greater trochanter markers on the projected image was carefully traced onto a sheet of clear plastic film with an indelible pen. This plastic sheet served as a template for alignment of the electronic cursor of the digitizing tablet during that particular trial, ensuring consistency of digitization.

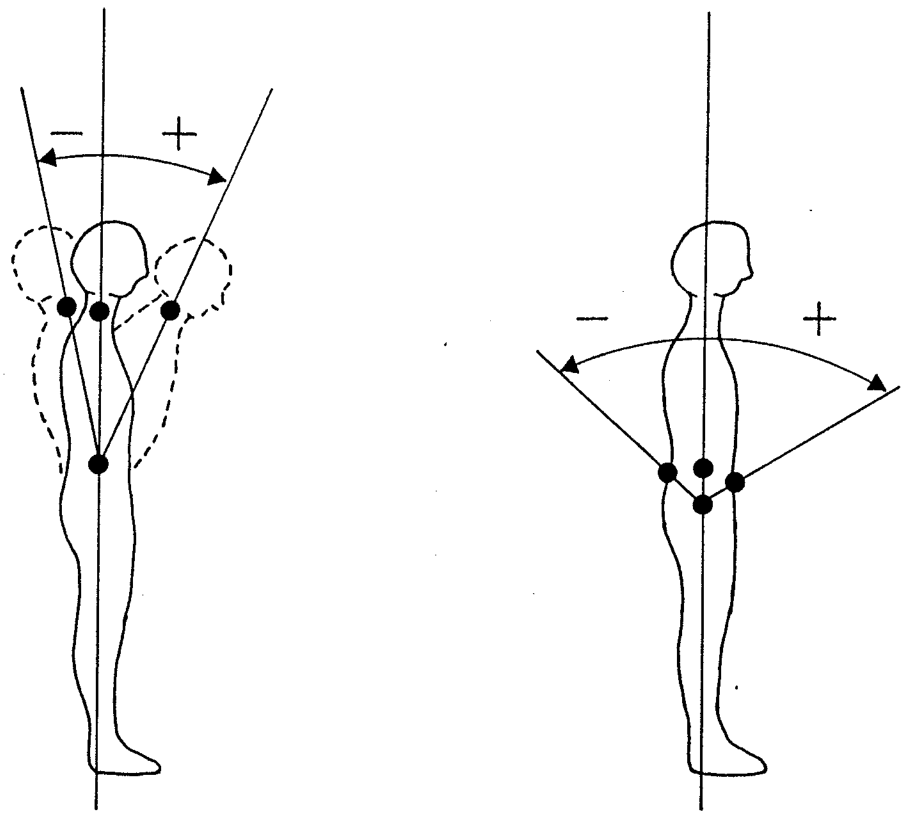
The DIGIN subprogram was used to extract and store the digitized X and Y co-ordinates from each body marker for each frame of film. A correction for the fact that the body markers were not in the same plane as the background markers was written into the subprogram and it was calculated that the maximum error that would result from measurements taken at the extremes of the field of view of the camera was 0.3°. The synchronizing light was used to identify the first frame of the trial to be digitized. The subprogram required two adjacent background markers to be digitized first, followed by the body markers in the sequence C_7, L_5, greater trochanter, knee. A sequence check was built into the subprogram; an audible prompt was given if any markers were digitized in an incorrect sequence. After all the frames in a particular trial were digitized, the co-ordinate data were stored and a printout of the raw X and Y co-ordinates was generated.

The DIGCALC subprogram was used to smooth and then convert the raw X and Y co-ordinates into angular displacement data for the defined body segments (Winter, 1979). Because of noise due to vibrations in the cinecamera, imperfect alignment of film in the camera, and human errors in the digitization process (Winter, 1979), the raw X and Y co-ordinates were filtered with a second order Butterworth lowpass digital filter with a cutoff frequency of 5 Hz. The data were filtered in the forward and backward time direction to remove any phase shift that the filter might introduce.

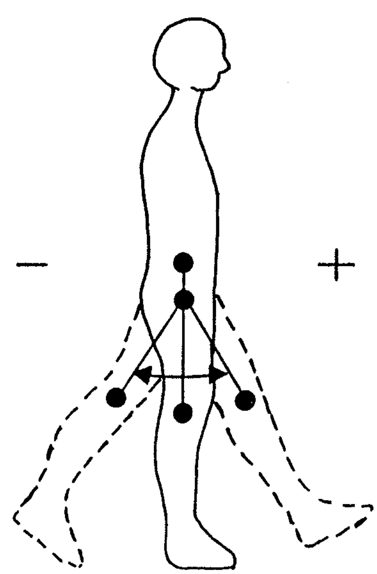
The filtered X and Y co-ordinates were used in the DIGCALC subprogram to calculate the position of the trunk and pelvic segments relative to the vertical (Thorstensson et al., 1982; Thurston & Harris, 1983; Bejjani et al., 1984; Thorstensson et al., 1984, 1985), and to calculate the angle between the pelvic and thigh segments, designated the hip angle (Fig. 8). These data could be stored, giving 15 files of filtered displacement data per subject (5 types of locomotion, 3 trials each). A printout could be generated of trunk inclination, pelvic inclination and hip angle values for each frame of the film that was digitized in a designated trial. In addition, plots of trunk inclination, pelvic inclination and hip angle curves during the course of the digitized trial could be generated on the computer monitor and a hard copy could be printed.

The temporal events occurring during one right gait cycle (Right Initial Contact, Left End of Weight Bearing, Left Initial Contact, Right End of Weight Bearing, Right Initial Contact) were marked on the printout of the angular displacement values and from this the segment values at 5% intervals of the right gait cycle (100%) were extracted. The values for each of the three trials of one type of locomotion were averaged and entered onto a data sheet. The mean values were used to plot graphic representations of the pattern of angular displacements that occurred during a specific type of

FIGURE 8. DEFINITION OF BODY SEGMENTS.



TRUNK AND PELVIC INCLINATION WITH RESPECT TO VERTICAL



HIP ANGLE – BETWEEN PELVIC AND THIGH SEGMENTS locomotion. The mean values were also entered into a computer file using the STATS PLUS program for later statistical analysis.

STATISTICAL ANALYSIS

At the conclusion of the data extraction process the Right Gait Cycle data available for statistical analysis for each of Level Walking, Stair Climbing, Stair Descent, Ramp Climbing and Ramp Descent were:

- mean duration of Right Gait Cycle, in seconds.
- mean cadence of Right Gait Cycle, in steps per minute.
- mean positions of onset, peaks (1,2 and 3) and cessation of myoelectric activity of the right erector spinae, left erector spinae, right rectus abdominis and left rectus abdominis muscles, expressed as percentages of the Right Gait Cycle.
- mean amount of myoelectric activity of the right erector spinae, left erector spinae, right rectus abdominis and left rectus abdominis muscles during a Right Gait Cycle, expressed as a percentage of a MVC.
- mean inclination of the trunk with respect to the vertical at 5% intervals of the Right Gait Cycle, measured in degrees.

– mean inclination of the pelvis with respect to the vertical at 5% intervals of the Right Gait Cycle, measured in degrees.

– mean hip angle at 5% intervals of the Right Gait Cycle, measured in degrees.

Statistical analysis of the data was done using the STATS PLUS and ANOVA][programs (Human Systems Dynamics, 1983) with an Apple IIe computer. In all instances the probability level was set at .05.

CHAPTER IV

RESULTS

As discussed in Materials and Methods, each subject performed five types of locomotion - Level Walking, Stair Climbing, Stair Descent, Ramp Climbing and Ramp Descent. Three trials of each type of locomotion were completed, with the middle stride period of each trial selected for analysis. The Right Gait Cycle (RGC) was used as the standard against which all other measurements were made (Winter, 1986). Because of the placement of the experimental apparatus, some measurements had to be made from the left side of the body. According to Winter et al. (1976), during a gait study the data for one lower limb, suitably displaced in time, can be used as representative data for the opposite limb. In the present study the mean percentages of Stance phase for Right and Left Gait Cycles of each type of locomotion were calculated from the temporal events data; these values are presented in Appendix II,1-5. The data for Right and Left Stance were then compared between the different types of locomotion by means of a repeated measures two-way (type x side) analysis of variance (Linton & Gallo, 1975; Sokal & Rohlf, 1981). No significant difference was found between Right and Left Stance in any type of locomotion. It was therefore concluded that the Right and Left Gait Cycles were essentially symmetrical and measurements from the left side of the body, appropriately displaced in time, could be considered to be representative of the right side of the body.

DURATION OF THE RIGHT GAIT CYCLE

For each trial of each type of locomotion, two successive Right Initial Contacts (RICs) were marked on the displayed Right Footswitch tracing and the EMGCALC subprogram generated the mean time interval between the RICs for the three trials. The mean data for each subject can be found in Appendix II,6. These data were subjected to a repeated measures analysis of variance followed by a post hoc Tukey's Test. The summary of these analyses is presented in Table 3. It can be seen from this table that the duration of the Right Gait Cycle in the two climbing types of locomotion was significantly longer than in the two descending types. There was no significant difference in the duration of the gait cycle between Stair Climbing and Ramp Climbing or between Stair Descent and Ramp Descent. The duration of the gait cycle in Level Walking was significantly shorter than in Ramp Climbing, but significantly longer than in the two descending types of locomotion.

CADENCE

In addition to the duration of the Right Gait Cycle, the EMGCALC subprogram generated the mean number of steps taken per minute over the three trials of each type of locomotion. The mean data for each subject can be found in Appendix II,7.

Table 3. Summary of Data for Mean Duration (secs) of RGC in All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	1.2	1.3	1.1	1.4	1.1
SD	0.1	0.2	0.2	0.2	0.1

Summary of Statistical Analyses of Comparison of Mean Duration of Right Gait Cycle between all Types of Locomotion

ANOVA $F(4,85) = 21.0$ $p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	.01	.01	.01
SC		–	.01	NS	.01
SD			–	.01	NS
RC				–	.01

KEY: L = Level SC = Stair Climb SD = Stair Descent
RC = Ramp Climb RD = Ramp Descent

These data were subjected to a repeated measures analysis of variance followed by a _post hoc_ Tukey's Test. The summary of these analyses is presented in Table 4. It can be seen from this table that the cadence during the two climbing types of locomotion was significantly lower than that during the two descending types. There was no significant difference in cadence between Stair and Ramp Descent, however the cadence in Ramp Climbing was significantly lower than that in Stair Climbing. The cadence in Level Walking was significantly higher than in Ramp Climbing, but significantly lower than in the two descending types of locomotion.

TEMPORAL EVENTS

For each of the three trials of each type of locomotion, the EMGCALC subprogram generated a value for the point of Right End of Weight Bearing (REWB), Left Initial Contact (LIC), and Left End of Weight Bearing (LEWB) and expressed these values as percentages of the Right Gait Cycle. The means for each temporal event in each type of locomotion were then calculated. Because the footswitch tracings were demarcated manually on the displayed tracings, some errors in marking may have occurred, therefore the mean data for temporal events were expressed to the nearest whole number (Davis & Foote, 1956).

Table 4. Summary of Data for Mean Cadence (number of steps per minute) of All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	97.9	97.8	116.0	90.2	112.6
SD	8.5	13.8	18.9	11.2	12.0

Summary of Statistical Analyses of Comparison of Mean Cadence during all Types of Locomotion

ANOVA $F\ (4,85) = 33.2$ $p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	-	NS	.01	.01	.01
SC		-	.01	.01	.01
SD			-	.01	NS
RC				-	.01

KEY: L = Level SC = Stair Climb SD = Stair Descent
RC = Ramp Climb RD = Ramp Descent

A graphic comparison of the temporal events in each type of locomotion is presented in Fig. 9; the raw data can be found in Appendix II,8-12. It can be seen that the values for LEWB ranged between 10% (Stair Descent) and 15% (Ramp Climbing), those for LIC ranged between 47% (Stair Descent) and 51% (Level Walking, Stair Climbing, Ramp Climbing), and those for REWB ranged between 61% (Stair Descent, Ramp Descent) and 64% (Stair Climbing, Ramp Climbing). Variation for each temporal event between the different types of locomotion was small, however a repeated measures analysis of variance revealed significant differences between locomotion types for LIC and LEWB, but no significant difference between types for REWB. A <u>post hoc</u> Tukey's Test on the LIC data (Table 5) showed that there was no significant difference between Stair and Ramp Climbing, between Stair and Ramp Descent, or between Level Walking, Ramp Descent and the two climbing types of locomotion. There was a significant difference only between the two climbing types and Stair Descent, and between Level Walking and Stair Descent. A <u>post hoc</u> Tukey's test on the values for LEWB (Table 6) showed that there was no siginificant difference between Stair or Ramp Climbing or between Stair and Ramp Descent; there was a significant difference between the two climbing and the two descending types.

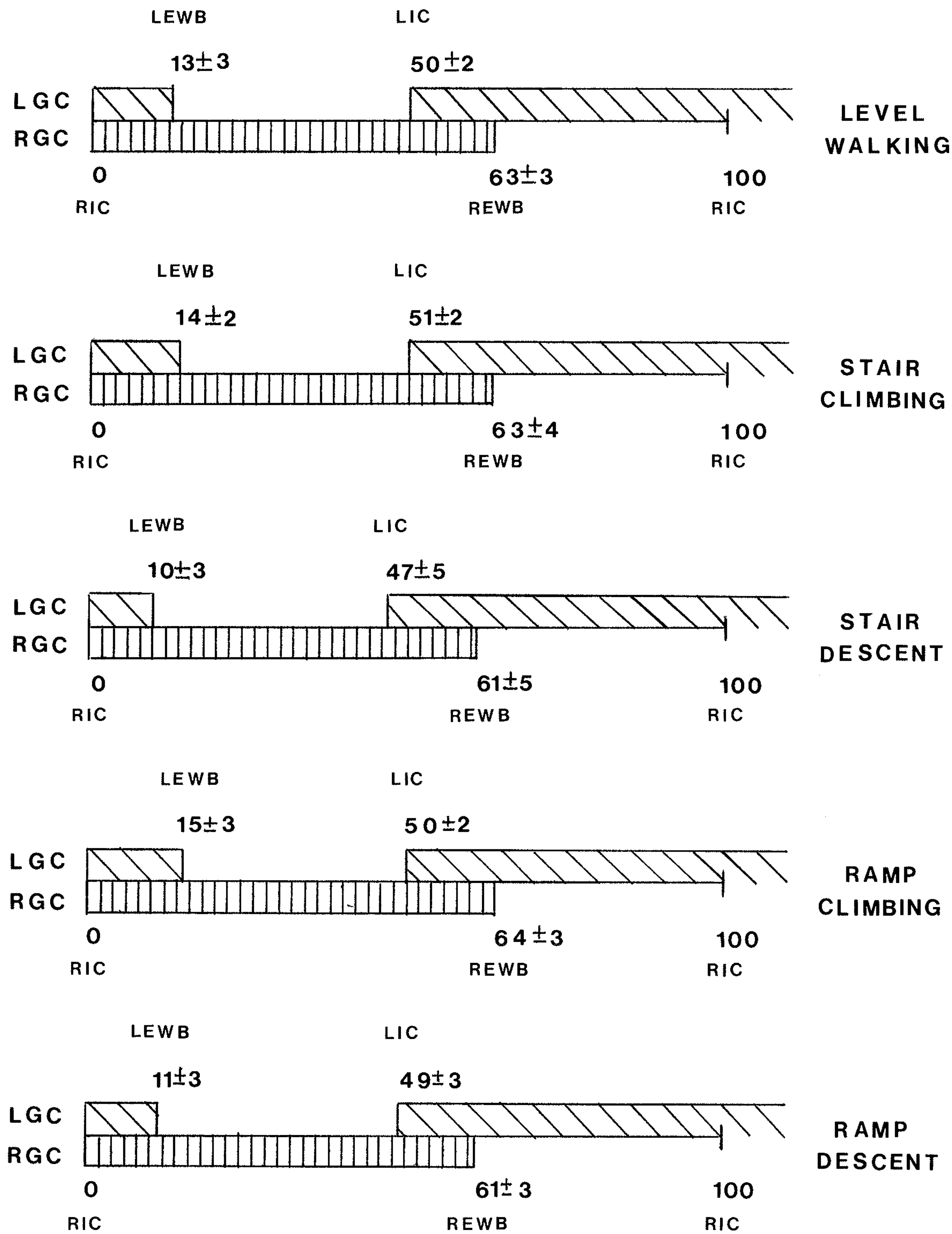


Fig. 9. COMPARISON OF TEMPORAL EVENTS DURING EACH LOCOMOTION TYPE (%of RGC)

Table 5. Summary of Data for Left Initial Contact (% of RGC) during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	51	51	47	51	49
SD	2	2	5	2	3

Summary of Statistical Analyses of Comparison of Left Initial Contact during All Types of Locomotion

ANOVA $F\ (4,85) = 5.3$ $\quad p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	.01	NS	NS
SC		–	.01	NS	NS
SD			–	.01	NS
RC				–	NS

KEY: L = Level SC = Stair Climb SD = Stair Descent
RC = Ramp Climb RD = Ramp Descent

Table 6. Summary of Data for Left End of Weight Bearing (% of RGC) during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	13	14	10	15	11
SD	3	2	2	3	3

Summary of Statistical Analyses of Comparison of Left End of Weight Bearing during All Types of Locomotion

ANOVA $F\ (4,85) = 13.0 \quad p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	.05	.05	NS
SC		–	.01	NS	.01
SD			–	.01	NS
RC				–	.01

KEY: L = Level SC = Stair Climb SD = Stair Descent
RC = Ramp Climb RD = Ramp Descent

DISPLACEMENT IN THE SAGITTAL PLANE

The positions of the trunk and pelvic segments relative to the vertical were measured at 5% intervals of the Right Gait Cycle (Thorstensson et al., 1982, 1984, 1985; Thurston & Harris, 1983; Bejjani et al., 1984). The angle between the pelvic segment and the thigh segment was measured and designated the hip angle. Because of possible errors inherent during photographic recording and digitization, trunk and pelvic segment inclination and hip angle data are expressed to the nearest degree.

The raw data for trunk and pelvic inclination are presented in Appendix II,13-22.1. These data are presented according to kinematic convention, that is, counterclockwise motion of the trunk and pelvic segments is defined as positive and clockwise motion is defined as negative (Winter, 1986). Therefore, because the right side of the body is being considered, anterior inclination of the trunk and pelvis is negative and posterior inclination is positive. However, for clarity of illustration, graphic representation of the trunk and pelvic segment displacement data will be presented in the format used by Thorstensson et al. (1982, 1984, 1985), Thurston and Harris (1983) and Bejjani et al. (1984): anterior inclination will be plotted in the positive direction of the y axis and posterior inclination will be plotted in the negative direction of the y axis (Fig. 8).

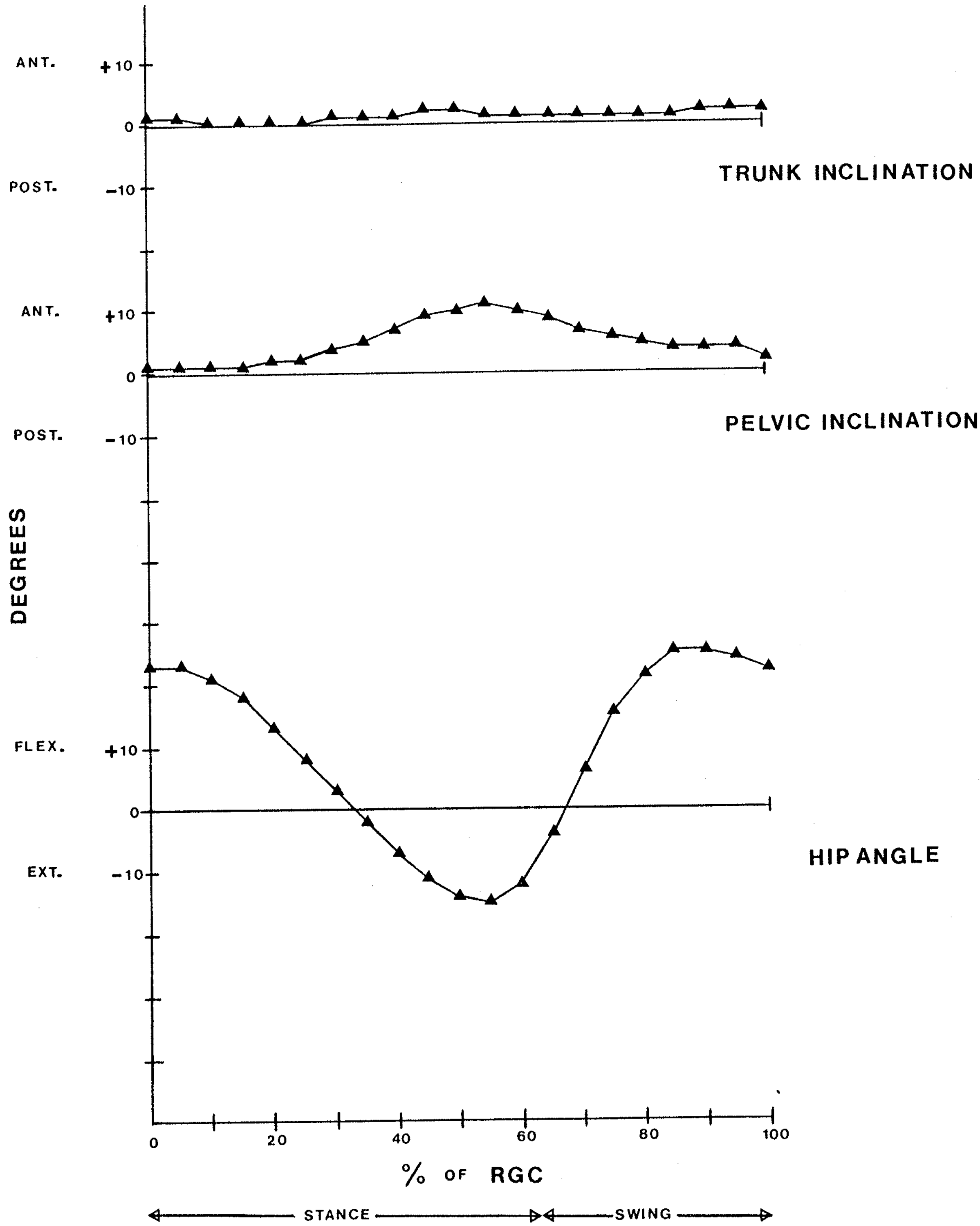
The hip angle was also measured at 5% intervals of the Right Gait Cycle and the raw data are presented in Appendix II,23-27.1, according to kinematic convention in which counterclockwise motion is defined as positive and clockwise motion is defined as negative. Because the right side of the body is being considered, hip flexion is positive and hip extension is negative (Winter, 1986). Graphic representation of hip angle data will be presented in the format used by Inman et al. (1981) and Perry (1985): hip flexion will be plotted in the positive direction of the y axis and hip extension will be plotted in the negative direction of the y axis.

LEVEL WALKING

A graphic representation of displacement of body segments in the sagittal plane during the Right Gait Cycle of Level Walking is presented in Fig. 10.

The trunk segment in eight subjects was held in anterior inclination throughout the gait cycle, in six subjects it was held in posterior inclination, and in four subjects the trunk moved into both anterior and posterior inclination. Overall, the trunk segment was held in slight anterior inclination throughout the majority of the gait cycle. When the values

FIGURE 10. BODY SEGMENT POSITIONS DURING LEVEL WALKING.



BODY SEGMENT POSITIONS DURING LEVEL WALKING for each individual are considered, the maximum anterior inclination observed was 8°, maximum posterior inclination was 5°. Over the 18 subjects, mean maximum anterior inclination of 2° occurred at the end of both Right and Left Swing; the neutral (0°) position occurred during the period from Left End of Weight Bearing to Right Mid Stance. The trunk excursion for each subject was calculated by subtracting the maximum posterior inclination from the maximum anterior inclination and a mean value for the 18 subjects was calculated. The mean trunk segment excursion in Level Walking was $3 \pm 1°$.

The pelvic segment was maintained in a position anterior to the vertical in 12 subjects and moved both anterior and posterior to the vertical in six subjects. Overall, the pelvic segment was held anterior to the vertical throughout the gait cycle. Individually, maximum anterior inclination observed was 18°, maximum posterior inclination was 9°. Over the 18 subjects, mean maximum anterior inclination of 11° occurred during the second Double Support phase and minimum anterior inclination of 1° occurred during the first Double Support phase. The mean pelvic segment excursion of all subjects was $11 \pm 4°$.

When hip angle is considered, all subjects moved from a position of hip flexion at the beginning of the gait cycle, into hip extension at approximately 50% of the gait cycle, and back into flexion at the end of the gait cycle. Individually, maximum hip flexion was 30°, maximum hip extension was 19°. Over the 18 subjects, mean maximum hip flexion of 25° occurred just before the end of Right Swing, mean maximum hip extension of 15° occurred during the second Double Support phase. The mean hip angle excursion was 40 $\pm$ 3°.

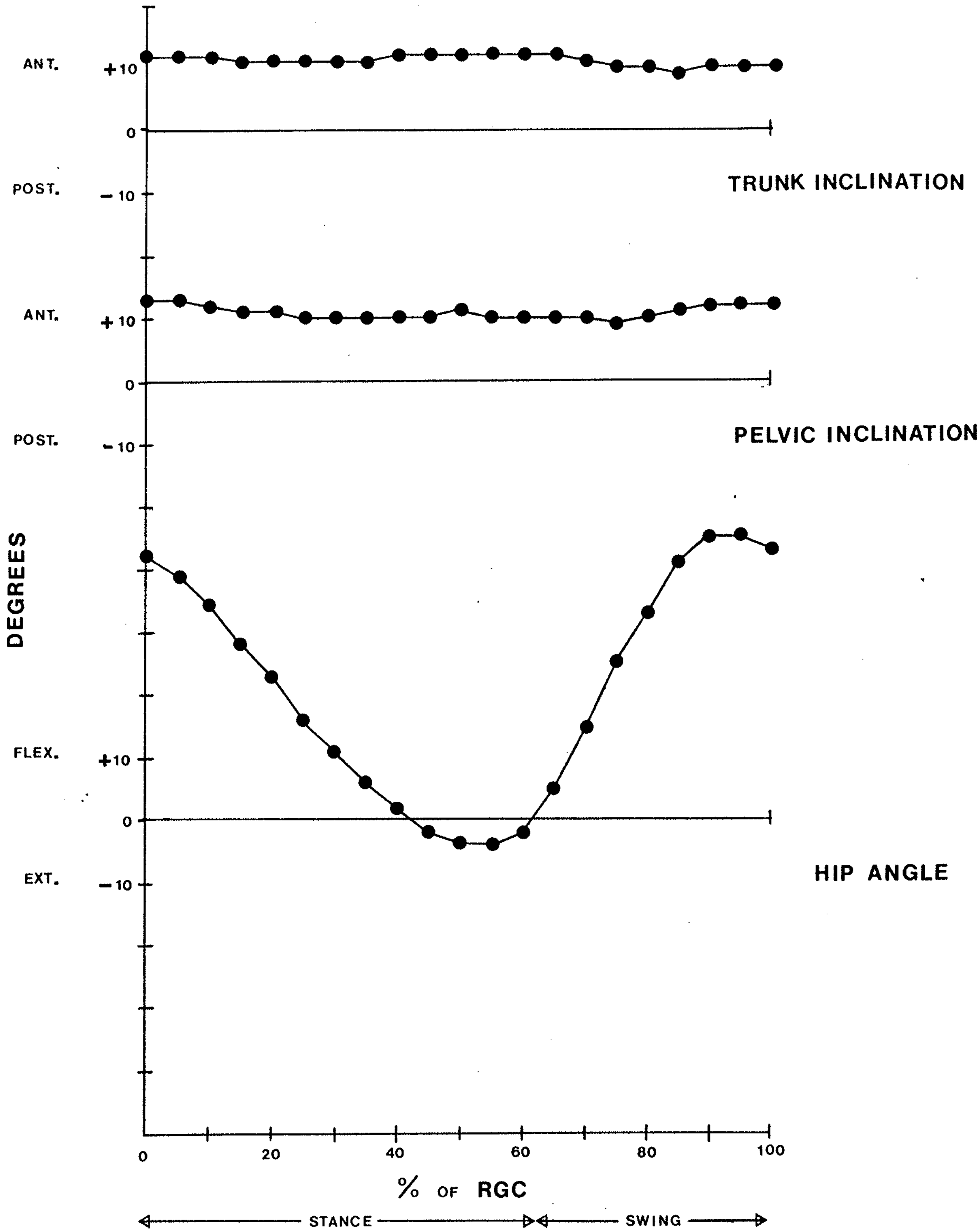
STAIR CLIMBING

A graphic representation of the displacement of body segments in the sagittal plane during Stair Climbing is presented in Fig. 11.

The trunk segment of all subjects was held in anterior inclination throughout the gait cycle. Individually, maximum anterior inclination was 23°, minimum anterior inclination was 1°. Over the 18 subjects, mean maximum anterior inclination of 12° occurred during the first Double support Phase; mean minimum anterior inclination of 9° occurred during Right Swing. The mean trunk segment excursion of the 18 subjects was 4 $\pm$ 1°.

The pelvic segment was held in anterior inclination throughout the gait cycle in 17 subjects; in one subject the pelvic segment moved into slight posterior inclination. Individually, maximum anterior inclination was 23°, maximum posterior inclination was 3°. Over the 18 subjects, mean

FIGURE 11. BODY SEGMENT POSITIONS DURING STAIR CLIMBING.



BODY SEGMENT POSITIONS DURING STAIR CLIMBING maximum anterior inclination of 13° occurred during the first Double Support phase; mean minimum anterior inclination of 9° occurred during Right Swing. The mean pelvic segment excursion of the 18 subjects was 7±3°.

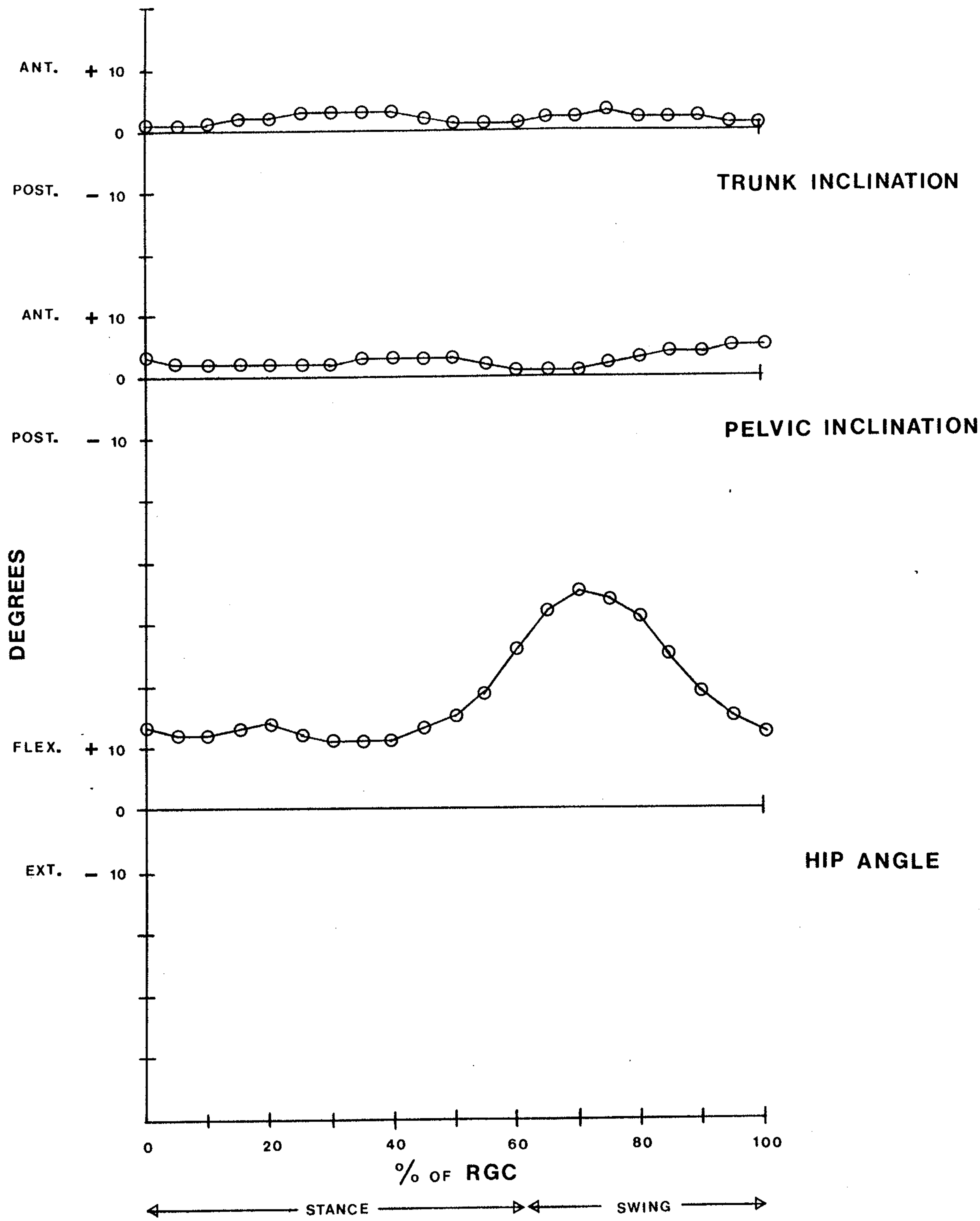
When hip angle is considered, 16 subjects moved from hip flexion at the beginning of the gait cycle, into hip extension at approximately 50% of the cycle, and then back into hip flexion at the end of the gait cycle. In one subject the hip moved from flexion to neutral (0°) in mid cycle and in one subject the hip remained in flexion throughout the gait cycle. Individually, maximum hip flexion was 53° and maximum hip extension was 10°. Over the 18 subjects mean maximum hip flexion of 45° occurred just before the end of Right Swing and mean maximum extension of 4° occurred during the second Double Support phase. The mean hip angle excursion of the 18 subjects was 51±4°.

STAIR DESCENT

A graphic representation of displacement of body segments in the sagittal plane during Stair Descent is presented in Fig. 12.

During Stair Descent the trunk segment was held in anterior inclination throughout the gait cycle in eight subjects, in posterior inclination throughout in three subjects and in seven subjects it moved into both anterior

FIGURE 12. BODY SEGMENT POSITIONS DURING STAIR DESCENT.



BODY SEGMENT POSITIONS DURING STAIR DESCENT and posterior inclination during the gait cycle. Overall, the trunk segment was held in slight anterior inclination throughout Stair Descent. Individually, maximum anterior inclination was 12°, maximum posterior inclination was 5°. Over the 18 subjects, mean maximum anterior inclination of 3° occurred during Right Mid Stance and Right Swing; mean minimum anterior inclination of 1° occurred during the first and second Double Support phases and at the end of Right Swing. The mean trunk segment excursion of the 18 subjects was 4±2°.

The pelvic segment was held in anterior inclination throughout Stair Descent in 10 subjects, in posterior inclination throughout in four subjects, and in four subjects it moved into both anterior and posterior inclination. Overall, the pelvic segment was held in slight anterior inclination throughout Stair Descent. Individually, maximum anterior inclination was 14°, maximum posterior inclination was 10°. Over the 18 subjects, mean maximum anterior inclination of 5° occurred at the end of Right Swing, and mean minimum anterior inclination of 1° occurred at the beginning of Right Swing. The mean pelvic segment excursion of the 18 subjects was 6±3°.

The hip angle of all subjects remained in flexion throughout Stair Descent with maximum hip flexion occurring at approximately 75% of the gait cycle. Individually, maximum hip flexion was 41°, minimum hip flexion was 4°. Over the 18 subjects mean maximum hip flexion of 35° occurred at the beginning of Right Swing, mean minimum hip flexion of 11° occurred during Right Mid Stance. The mean hip angle excursion of the 18 subjects was 26±3°.

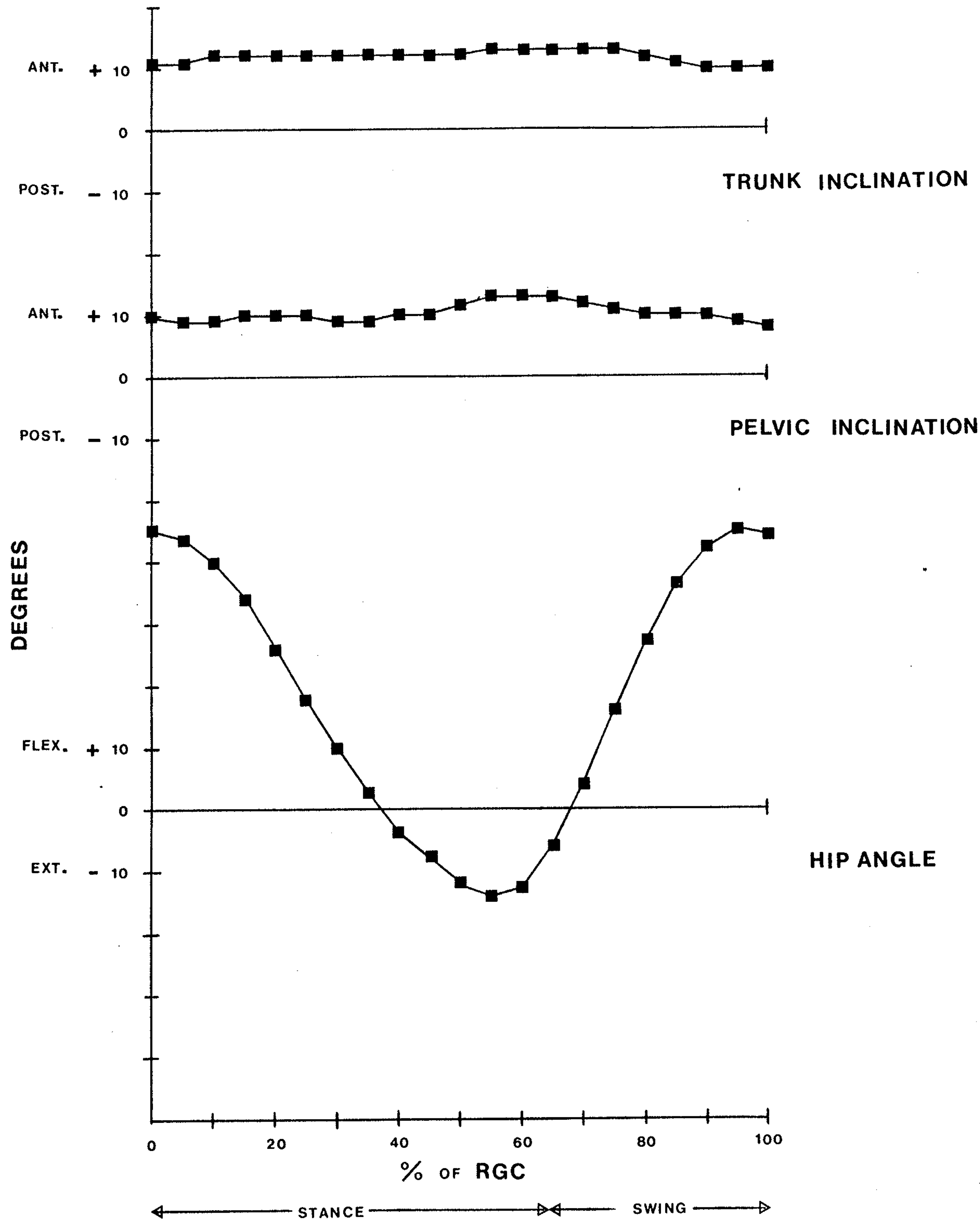
RAMP CLIMBING

A graphic representation of the displacement of body segments in the sagittal plane during Ramp Climbing is presented in Fig. 13.

During Ramp Climbing the trunk segment was held in anterior inclination throughout the gait cycle in all subjects. Individually, maximum anterior inclination was 24°, minimum anterior inclination was 2°. Over the 18 subjects, mean maximum anterior inclination of 13° occurred during the second Double Support phase into Right Swing; mean minimum anterior inclination of 10° occurred at the end of Right Swing. The mean trunk excursion of the 18 subjects was 5±2°.

The pelvic segment was held in anterior inclination throughout the gait cycle in 17 subjects; in one subject the pelvis moved into both anterior and posterior inclination. Individually, maximum anterior inclination was 21°, maximum posterior inclination was 1°. Over the 18 subjects, mean maximum anterior inclination of 13° occurred during the

FIGURE 13. BODY SEGMENT POSITIONS DURING RAMP CLIMBING.



BODY SEGMENT POSITIONS DURING RAMP CLIMBING second Double Support phase into Right Swing; mean minimum anterior inclination of 8° occurred at the end of Right Swing. The mean pelvic segment excursion of the 18 subjects was $9 \pm 3°$.

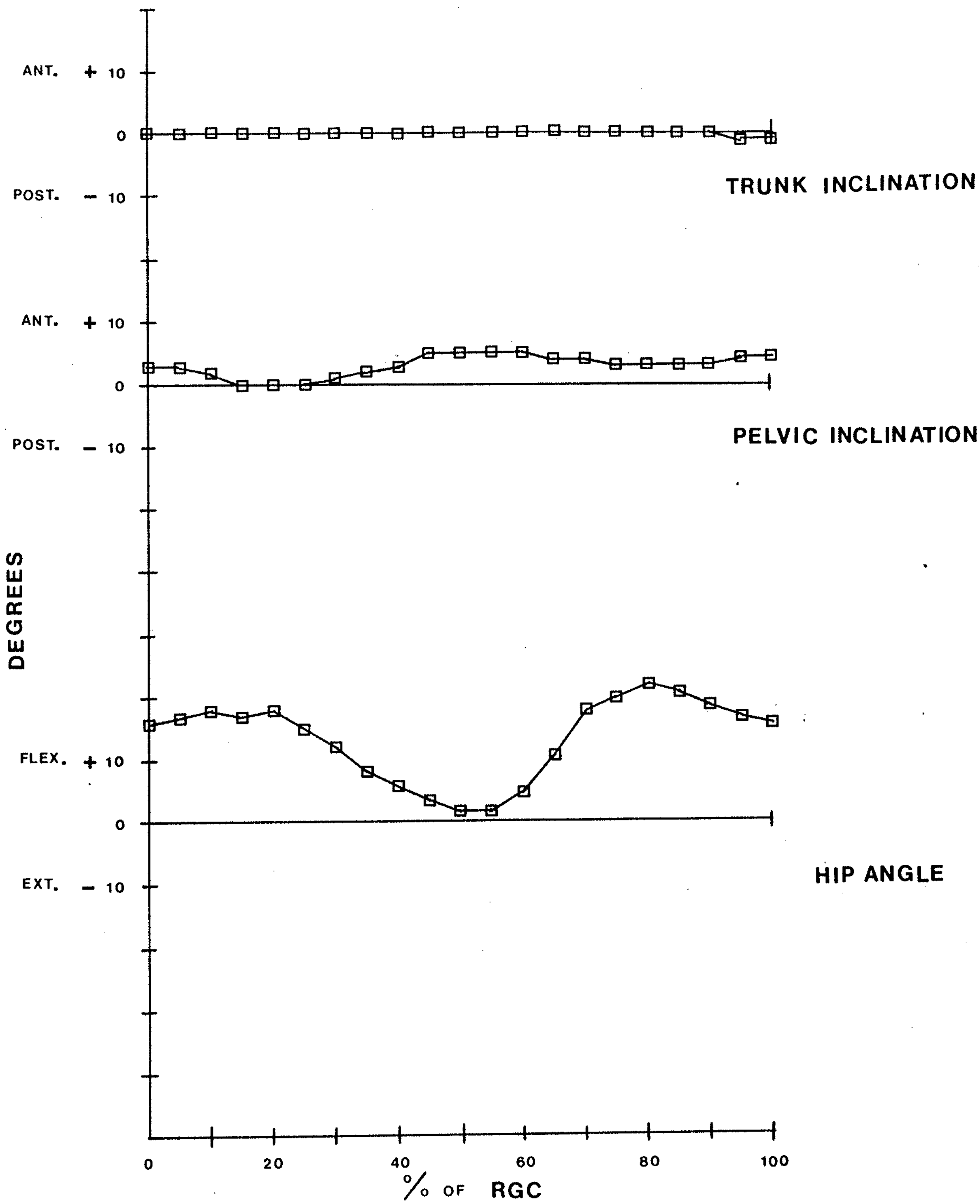
When hip angle is considered, all subjects moved from hip flexion at the beginning of the gait cycle into hip extension at approximately 50% of the gait cycle and back into hip flexion at the end of the cycle. Individually, maximum hip flexion was 54°, maximum hip extension was 23°. Over the 18 subjects mean maximum hip flexion of 45° occurred at Right Initial Contact, mean maximum hip extension of 14° occurred during the second Double Support phase. The mean hip angle exursion of the 18 subjects was $62 \pm 7°$.

RAMP DESCENT

A graphic representation of the displacement of body segments in the sagittal plane during Ramp Descent is presented in Fig. 14.

During Ramp Descent the trunk segment was held in posterior inclination throughout the gait cycle in nine subjects, in anterior inclination throughout in 5 subjects, and in four subjects the trunk segment moved into both anterior and posterior inclination. Overall, the trunk was held in the neutral (0°) position throughout the majority

FIGURE 14. BODY SEGMENT POSITIONS DURING RAMP DESCENT.



BODY SEGMENT POSITIONS DURING RAMP DESCENT of the gait cycle during Ramp Descent, moving into 1° of posterior inclination at the end of Right Swing. Individually, maximum anterior inclination was 8°, posterior inclination was 6°. The mean excursion of the trunk segment in the 18 subjects was $3\pm1°$.

The pelvic segment was held in anterior inclination throughout Ramp Descent in eight subjects, in posterior inclination throughout in two subjects, and in eight subjects it moved into both anterior and posterior inclination. Overall, the pelvic segment was held in anterior inclination during the majority of the gait cycle. Individually, maximum anterior inclination was 22°, maximum posterior inclination was 11°. Over the 18 subjects, mean maximum anterior inclination of 5° occurred during the second Double Support phase and the neutral (0°) position was reached during Right Stance. The mean excursion of the pelvic segment over the 18 subjects was $8\pm4°$.

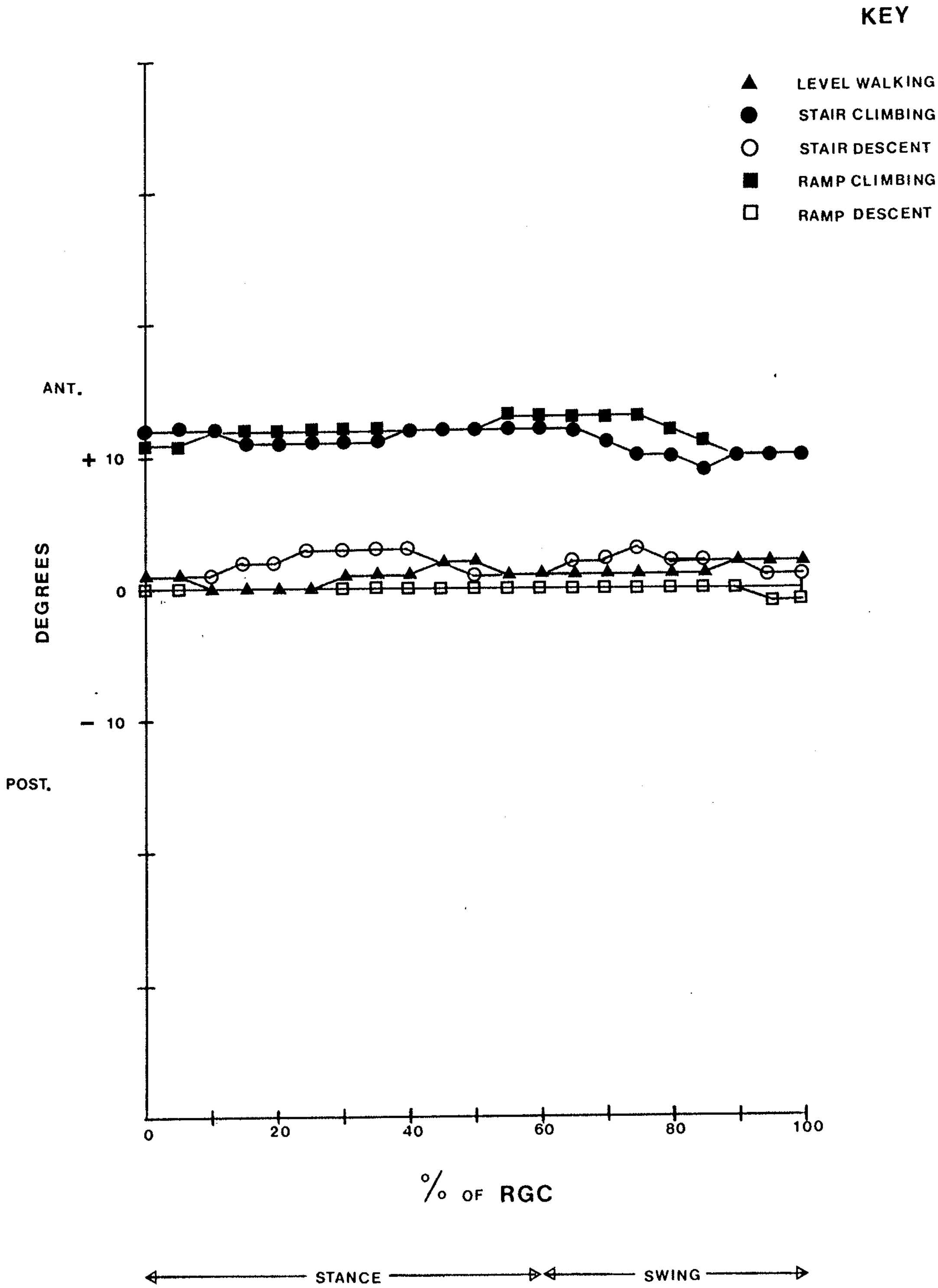
The hip angle of 12 subjects remained in flexion throughout Ramp Descent; in six subjects the hip went into slight extension at approximately 50% of the gait cycle. Overall, the hip remained in flexion throughout the gait cycle. Individually, maximum hip flexion was 29°, maximum hip extension was 12°. Over the 18 subjects, maximum hip flexion of 22° occurred during Right Mid Swing and minimum hip flexion of 2° occurred during the second Double Support phase. The mean excursion of the hip over the 18 subjects was $22 \pm 6°$.

COMPARISON OF DISPLACEMENTS IN DIFFERENT TYPES OF LOCOMOTION

The values for trunk segment inclination, pelvic segment inclination and hip angle were compared at 5% intervals from 0% to 100% of the Right Gait Cycle between each type of locomotion. In all instances the statistical test used was a repeated measures analysis of variance followed by a post hoc Tukey's Test.

A graphic comparison of the trunk inclination curves in the different types of locomotion is presented in Fig. 15. It can be seen that the Stair Climbing and Ramp Climbing curves are very similar to each other while the Level Walking, Stair Descent and Ramp Descent curves are all similar to each other. Results of the statistical analysis revealed that the Stair Climbing and Ramp Climbing curves were significantly different only between 70% to 80% of the gait cycle (Right Swing phase) when anterior inclination was greater during Ramp Climbing than during Stair Climbing. Comparison of the two descending types of locomotion revealed that these curves were significantly different only between 25% to 40% (Right Mid Stance) and between 80% to 90% (Right Terminal Swing) of the gait cycle when anterior inclination was greater during Stair Descent. Comparison of Level

FIGURE 15. COMPARISON OF TRUNK INCLINATION IN THE SAGITTAL PLANE.

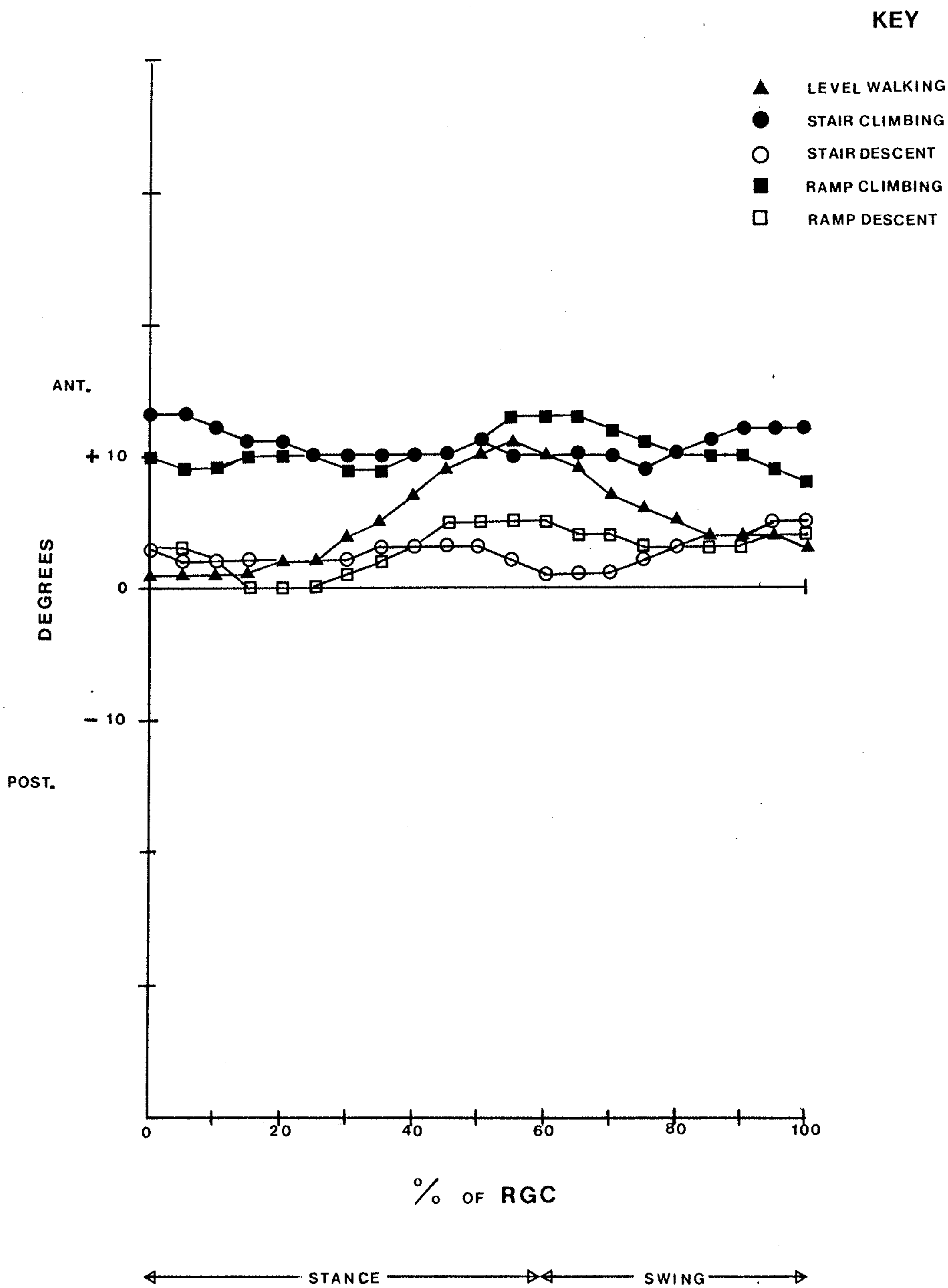


COMPARISON OF TRUNK INCLINATION IN THE SAGITTAL PLANE

Walking to Stair Descent showed that the curves were significantly different only between 15% to 35% (Right Stance) of the cycle when there was greater anterior inclination during Stair Descent. Comparison of Level Walking to Ramp Descent showed that there was a significant difference only between 80% to 100% (Right Terminal Swing) of the cycle when there was greater anterior inclination during Level Walking.

A graphic comparison of the pelvic inclination curves in the different types of locomotion is presented in Fig. 16. Again, it can be seen that the curves of the two climbing types of locomotion are similar as are the curves of the two descending types. Results of statistical tests showed that there was no significant difference between the curves of the two climbing types of locomotion or between the curves of the two descending types. Comparison of Level Walking to the two climbing locomotion types revealed that there was significantly less anterior pelvic inclination in Level Walking except during the second Double Support phase, between 40% to 74% for Stair Climbing and between 40% to 60% for Ramp Climbing, when there was no significant difference between the three curves. Comparison of Level Walking to the two descending types of locomotion showed that the curves were significantly different only during the second Double Support phase, between 50% to 75% in the case of Stair

FIGURE 16. COMPARISON OF PELVIC INCLINATION IN THE SAGITTAL PLANE.

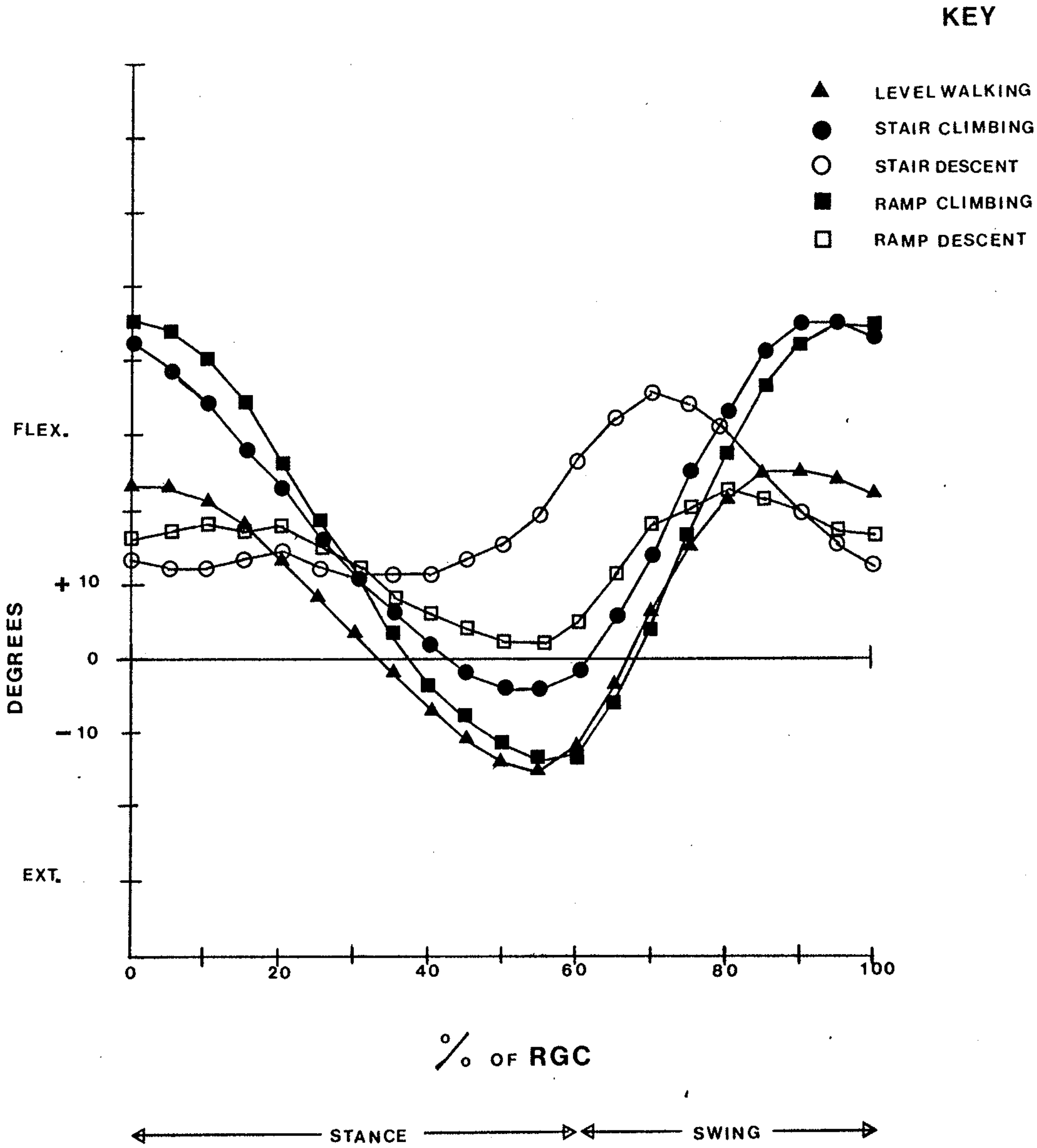


COMPARISON OF PELVIC INCLINATION IN THE SAGITTAL PLANE

Descent and 50% to 65% in the case of Ramp Climbing; at these times there was greater anterior inclination during Level Walking.

A graphic comparison of the hip angle curves in the different types of locomotion is presented in Fig 17. It can be seen that the form of the curve of the Level Walking, Stair Climbing, Ramp Climbing and Ramp Descent types of locomotion is basically similar, while that of Stair Descent is different. Statistical analysis showed that the curves of the two climbing types of locomotion were significantly different between 0% and 20% (Right Initial Contact into the first Double Support phase) when hip flexion was greater during Ramp Climbing, and between 40% to 90% (Terminal Stance into Right Swing) when there was greater hip extension in Ramp Climbing and greater hip flexion in Stair Climbing. Comparison of the curves in the two descending types of locomotion showed that there was significantly greater hip flexion in Stair Descent except between 25% to 35% (Right Mid Stance), and between 90% to 95% (Right Terminal Swing). When Level Walking was compared to Stair Climbing it was found that the curves were significantly different throughout the gait cycle, with greater hip flexion occurring during Stair Climbing and greater hip extension during Level Walking. Comparison of Level Walking to Ramp Climbing showed that there was significantly greater hip flexion during Ramp

FIGURE 17. COMPARISON OF THE HIP ANGLE IN THE SAGITTAL PLANE.



COMPARISON OF HIP ANGLE IN THE SAGITTAL PLANE

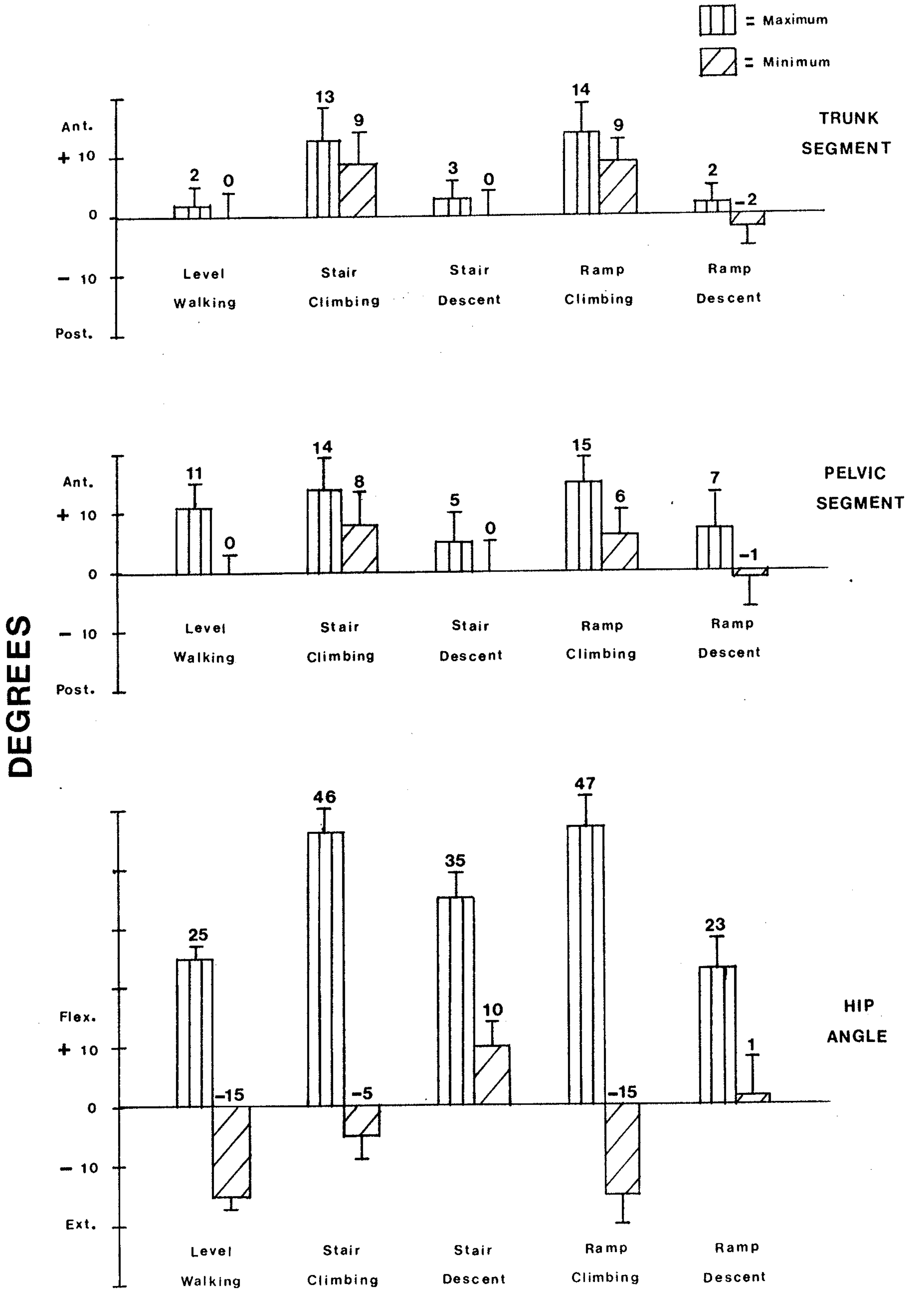
Climbing, but that there was no significant difference between the two types of locomotion during the period from 40% to 75% (Right Terminal Stance to Right Mid Swing). Comparison of Level Walking to the two descending types of locomotion showed that the curves were significantly different throughout the Right Gait Cycle; in the descending types of locomotion the hip did not move into extension past the neutral (0°) position.

COMPARISON OF MAXIMUM AND MINIMUM BODY SEGMENT DISPLACEMENTS

For each of the 18 subjects the values for the maximum anterior and posterior inclination of the trunk and pelvic segments during the Right Gait Cycle of each type of locomotion were identified, and the mean of each of the two extreme positions in the 18 subjects was calculated. Likewise, the values of maximum hip flexion and extension during the Right Gait Cycle of each type of locomotion were identified and the mean of each of the two positions in the 18 subjects was calculated. A graphic comparison of the mean extremes of body segment positions in the five different types of locomotion is presented in Fig. 18. A repeated measures analysis of variance and post hoc Tukey's Test was done for each set of values.

For the trunk segment it can be seen that maximum anterior inclination occurred during Stair and Ramp Climbing;

FIGURE 18. COMPARISON OF MAXIMUM AND MINIMUM BODY SEGMENT POSITIONS.



COMPARISON OF MAXIMUM AND MINIMUM BODY SEGMENT POSITIONS BETWEEN LOCOMOTION TYPES there was no significant difference between these values. There was no significant difference between maximum anterior inclination in Level Walking, Stair Descent and Ramp Descent. Maximum posterior inclination occurred during Ramp Descent, however this value was not significantly different from that during Level Walking or Stair Descent.

For the pelvic segment, maximum anterior inclination also occurred during Stair and Ramp Climbing; there was no significant difference between these values. There was also no significant difference between maximum anterior inclination of the two descending types of locomotion. Maximum anterior inclination during Level Walking was significantly greater than that during Stair Descent and significantly less than that during Ramp Climbing. Maximum posterior inclination occurred during Ramp Descent, however this was not significantly different from that during Level Walking or Stair Descent.

Maximum hip flexion occurred during the two climbing types of locomotion; there was no significant difference between these values. There was no significant difference between maximum hip flexion during Level Walking and Ramp Descent; maximum hip flexion in Stair Descent was significantly greater than that during Level Walking or Ramp Descent. The position of maximum hip extension occurred during Level Walking and Ramp Climbing and the values for these two types of locomotion were significantly greater than those of the other types of locomotion.

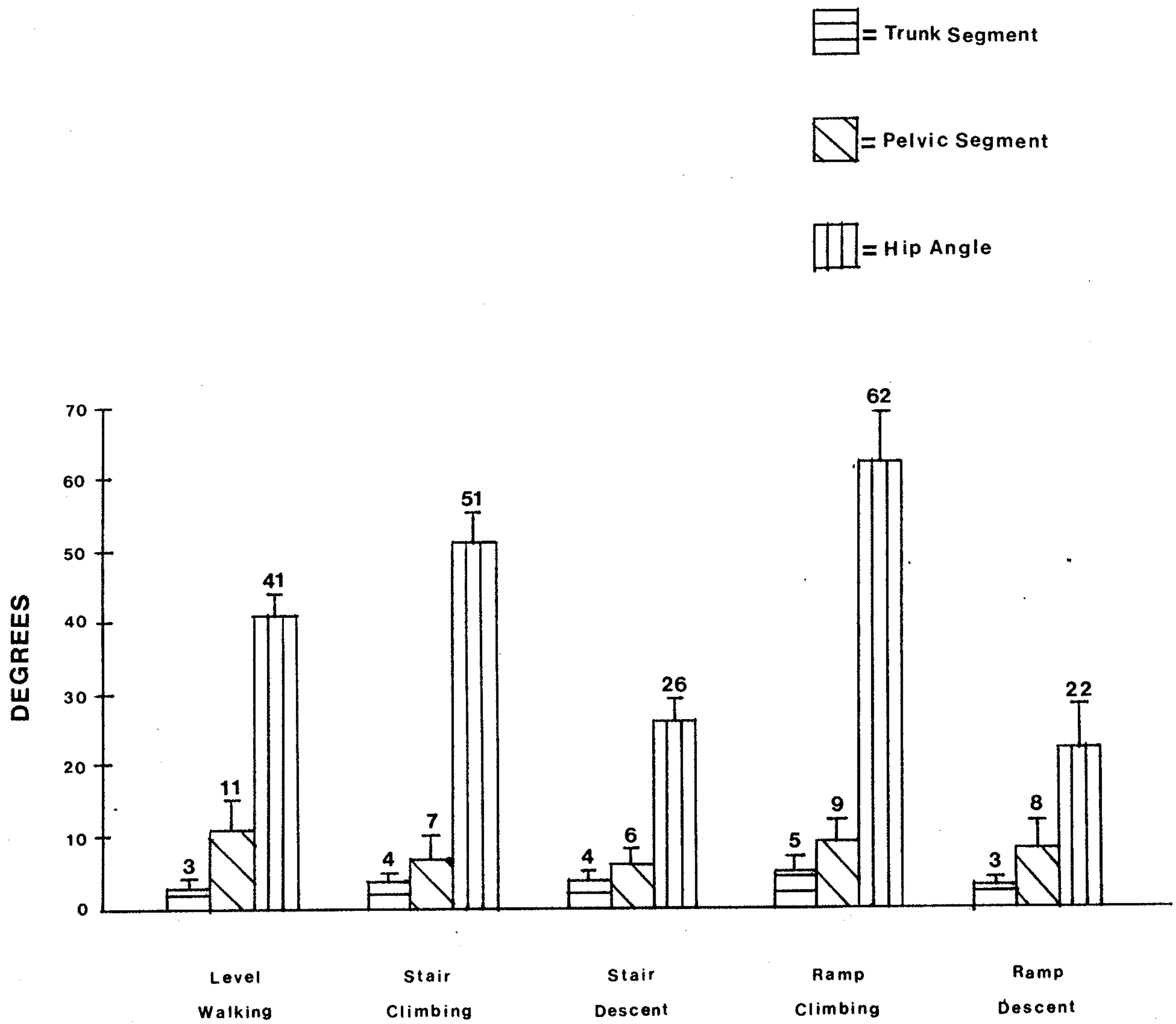
COMPARISON OF BODY SEGMENT EXCURSION

For each of the 18 subjects, the total excursion that occurred in the trunk and pelvic segments during a right gait cycle was calculated by subtracting the maximum anterior inclination value from the maximum posterior or minimum anterior inclination value. Likewise, values for hip excursion were calculated by subtracting the maximum flexion value from the maximum extension or minimum flexion value. The values for each segment were averaged and compared across the five types of locomotion by means of a repeated measures analysis of variance and <u>post hoc</u> Tukey's Test. A graphic comparison of the mean excursion of each of the body segments is presented in Fig.19.

For the trunk segment it can be seen that the total excursion was small in all types of locomotion, ranging from 3° to 5°. The greatest excursion occurred during Ramp Climbing, but this was not significantly greater than the excursion during Stair Climbing. There was no significant difference between the trunk segment excursion in Level Walking, Stair Descent and Ramp Descent.

The amount of excursion in the pelvic segment was greater than that of the trunk segment, ranging from a high

FIGURE 19. COMPARISON OF BODY SEGMENT EXCURSIONS BETWEEN LOCOMOTION TYPES.



COMPARISON OF BODY SEGMENT EXCURSION BETWEEN LOCOMOTION TYPES of 11° to a low of 6°. The greatest amount of excursion occurred during Level Walking, however this was not significantly different from that during Ramp Climbing. There was no significant difference between the amount of pelvic excursion that occurred during Ramp Climbing, Stair Climbing, Stair Descent or Ramp Descent.

At the hip the maximum and minimum excursion was 62° and 22°, respectively. Maximum hip excursion occurred during Ramp Climbing and this was significantly greater than during any other type of locomotion. The least amount of hip excursion occurred during the two descending types of locomotion; there was no significant difference between them.

AMOUNT OF TOTAL MYOELECTRIC ACTIVITY

Once the Right Gait Cycle had been marked on the Right Footswitch tracing, the EMGCALC subprogram measured the area under the curve of the linear envelope of a designated muscle, compared this value to the area of the linear envelope of the MVC of the muscle over a one second period, and generated a value that represented the percentage of activity produced by the muscle during the Right Gait Cycle. As it is impossible to determine if all motor units of a muscle are being recruited during the MVC, the value obtained from this contraction may be only an approximation of the maximum performance of the muscle.

Consequently, values calculated for myoelectric output and expressed as a percentage of a maximum voluntary contraction are an estimate of the activity of the muscle relative to its maximum possible performance, and are therefore expressed to the nearest whole number. The values are, however, useful in that some quantitative comparison can be made between muscles and a determination can be made of the amount of activity generated by a muscle during a sequence of movements (Basmajian, 1978; Letts et al., 1978; Perry et al., 1981).

The raw data for the mean amounts of total electrical activity of right erector spinae, left erector spinae, right rectus abdominis and left rectus abdominis in each of the five types of locomotion can be found in Appendix II,28-32. For each type of locomotion the mean total muscle activity was compared between muscles using analyses of variance and post hoc Tukey's Tests. A summary of the data and of the statistical tests is presented in Tables 7-11. It can be seen that for each type of locomotion there was no significant difference between the level of electrical activity of right and left erector spinae and between right and left rectus abdominis muscles. In all instances the two erector spinae muscles showed more activity than the two rectus abdominis muscles.

Because there was no significant difference between the activity of right and left sides of the two muscles, the mean

Table 7. Summary of Data for Mean Amount of Total Muscle Activity (% MVC) during Level Walking

	RES	LES	RRA	LRA
$\bar{X}$	12	13	3	4
SD	7	7	3	3

Summary of Statistical Analyses of Comparison of Mean Amount of Total Muscle Activity during Level Walking

ANOVA $F\ (3,51) = 28.3$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.01	.01
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 8. Summary of Data for Mean Amount of Total Muscle Activity (% MVC) during Stair Climbing

	RES	LES	RRA	LRA
$\bar{X}$	22	30	3	3
SD	12	18	2	2

Summary of Statistical Analyses of Comparison of Mean Amount of Total Muscle Activity during Stair Climbing

ANOVA $F\ (3,51) = 32.3 \quad p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.01	.01
RRA			–	NS

Key: RES = Right Erector Spinae RRA = Right Rectus Abdomins
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 9. Summary of Data for Mean Amount of Total Muscle Activity (% MVC) during Stair Descent

	RES	LES	RRA	LRA
$\bar{X}$	12	10	3	3
SD	7	5	3	3

Summary of Statistical Analyses of Comparison of Mean Amount of Total Muscle Activity during Stair Descent

ANOVA $F(3,68) = 14.5$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.01	.01
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 10. Summary of Data for Mean Amount of Total Muscle Activity (% MVC) during Ramp Climbing

	RES	LES	RRA	LRA
$\bar{X}$	22	30	3	3
SD	12	21	3	2

Summary of Statistical Analyses of Comparison of Mean Amount of Total Muscle Activity during Ramp Descent

ANOVA F (3,68) = 22.2 p < .001

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.01	.01
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 11. Summary of Data for Mean Amount of Total Muscle Activity (% MVC) during Ramp Descent

	RES	LES	RRA	LRA
$\bar{X}$	12	11	4	5
SD	9	7	3	5

Summary of Statistical Analyses of Comparison of Mean Amount of Total Muscle Activity during Ramp Descent

ANOVA $F\ (3,68) = 7.6$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.05	.05
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

of the activity of the erector spinae as a whole and of the rectus abdominis as a whole was calculated for each subject (II,33-34), and these values were used to compare the activity of each muscle between the different types of locomotion (Fig. 20). A repeated measures analysis of variance and post hoc Tukey's Test were done on each set of data and the summaries are presented in Tables 12 and 13.

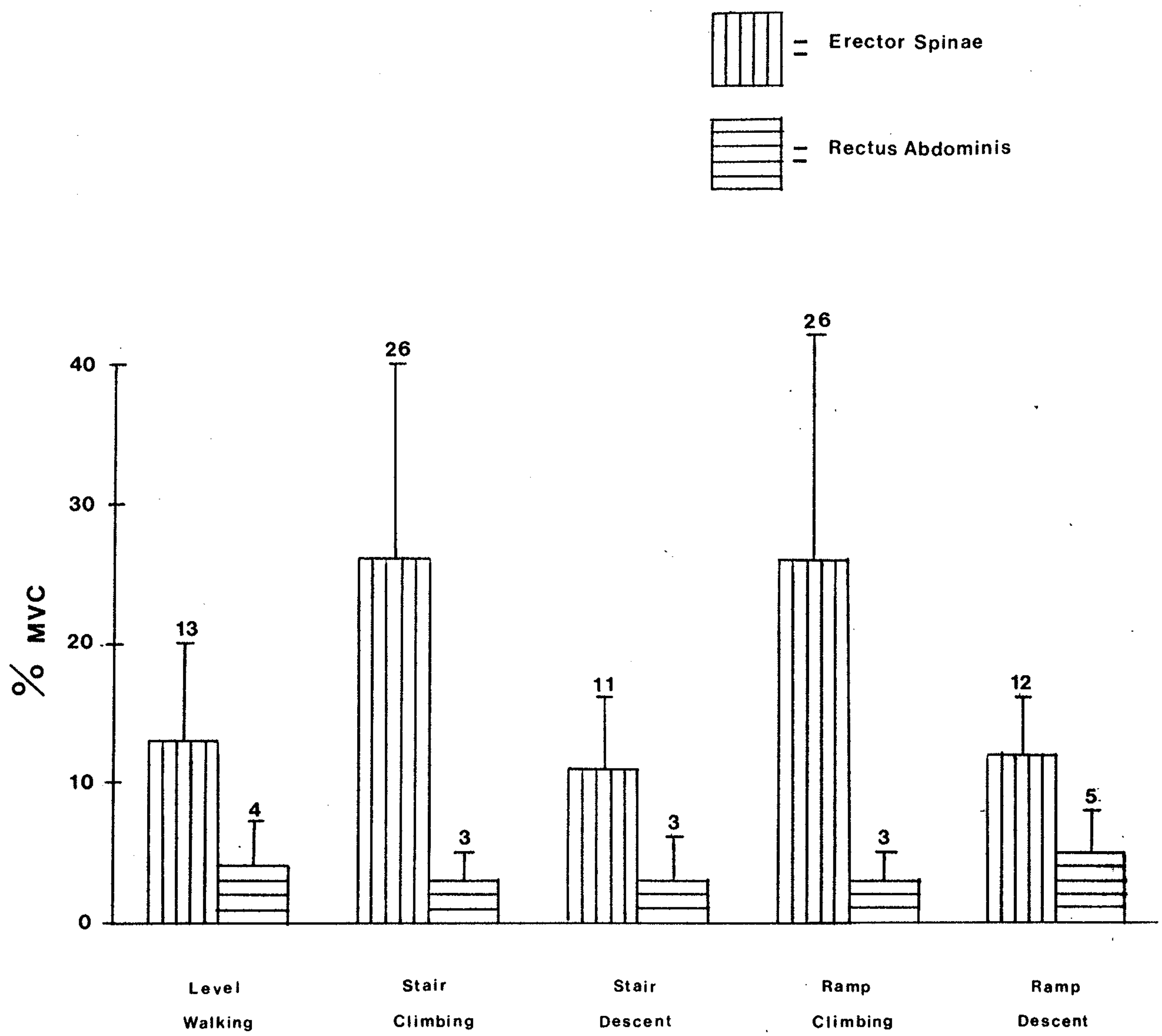
For erector spinae it can be seen from Fig. 20 and Table 12 that the greatest amount of electrical activity occurred during the two climbing types of locomotion, the least amount during Stair Descent. There was no significant difference between Stair Climbing and Ramp Climbing, or between Level Walking compared to Stair Descent and Ramp Descent.

For rectus abdominis, it can be seen from Fig. 20 and Table 13 that electrical activity was low in all types of locomotion with the greatest amount seen during Ramp Descent and the least amount seen during Stair Climbing, Ramp Climbing and Stair Descent. A significant difference in amount of total electrical activity was found only between the two climbing types of locomotion when compared to Ramp Descent.

PHASIC ACTIVITY OF TRUNK MUSCLES

The sequence of participation of each muscle during the Right Gait Cycle in each type of locomotion was investigated.

FIGURE 20. COMPARISON OF MEAN TOTAL MUSCLE ACTIVITY BETWEEN LOCOMOTION TYPES.



COMPARISON OF MEAN TOTAL MUSCLE ACTIVITY

BETWEEN LOCOMOTION TYPES

Table 12. Summary of Data for Mean Amount of Muscle Activity (% MVC) in Erector Spinae during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	13	26	11	26	12
SD	7	14	5	16	8

Summary of Statistical Analyses of Comparison of Mean Amount of Erector Spinae Activity during All Types of Locomtion

ANOVA $F\ (4,68) = 19.5$ $p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	.01	NS	.01	NS
SC		–	.01	NS	.01
SD			–	.01	NS
RC				–	.01

KEY: L = Level SC = Stair Climbing SD = Stair Descent
RC = Ramp Climbing RD = Ramp Descent

Table 13. Summary of Data for Mean Amount of Muscle Activity (% MVC) of Rectus Abdominis during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	4	3	3	3	5
SD	3	2	3	2	3

Summary of Statistical Analyses of Comparison of Mean Amount of Muscle Activity of Rectus Abdominis during All Types of Locomotion

ANOVA $F\ (4,68) = 2.7$ $p < .04$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	NS	NS	NS
SC		–	NS	NS	.05
SD			–	NS	NS
RC				–	.05

KEY: L = Level Walking SC = Stair Climbing SD = Stair Descent
RC = Ramp Climbing RD = Ramp Descent

As described in Materials and Methods, the onset, peak and cessation of myoelectric activity of each EMG tracing were selected for evaluation. These points were marked manually on the displayed EMG tracings and the EMGCALC subprogram generated a value for each parameter, expressed as a percentage of the RGC.

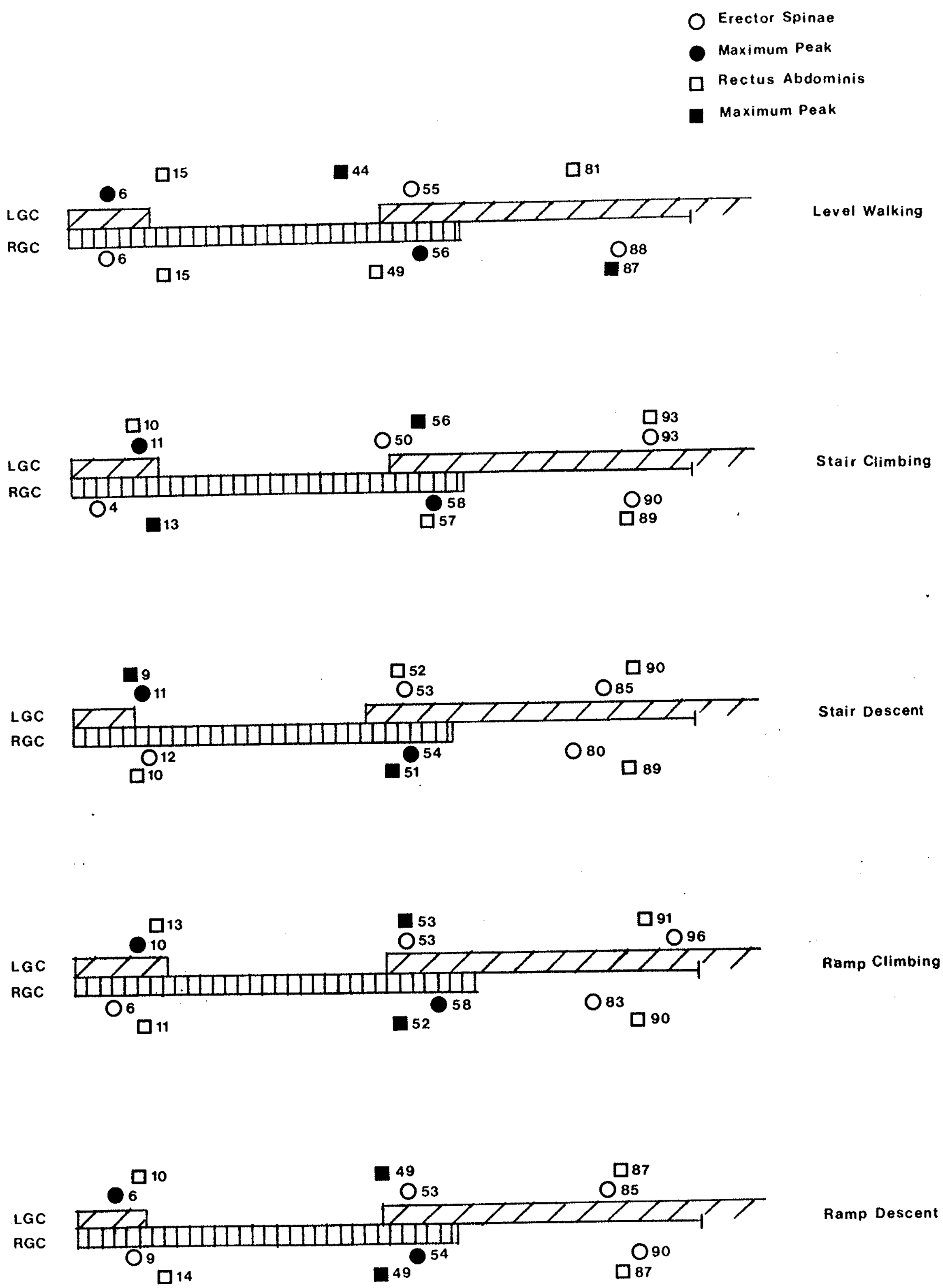
It was found that muscles were either active throughout the entire gait cycle or were inactive, therefore no onset or cessation points occurred. Three distinct peaks of myoelectric activity during the Right Gait Cycle could be identified for the majority of muscles.

PEAK 1 EMG ACTIVITY

The mean points of Peak 1 EMG activity of the right and left erector spinae and the right and left rectus abdominis during the five different types of locomotion are presented in Fig 21; the raw data can be found in Appendix II,35-39. An analysis of variance and <u>post hoc</u> Tukey's Test were used to compare the Peak 1 activity between all muscles in each type of locomotion.

During Level Walking (II,35) both erector spinae muscles showed a peak of activity in all subjects during the first Double Support phase. In the right and left rectus abdominis muscles a peak of activity was observed in 11 and six subjects, respectively; the peak in both abdominal muscles

FIGURE 21. COMPARISON OF PHASIC MUSCLE ACTIVITY BETWEEN LOCOMOTION TYPES.



COMPARISON OF PHASIC MUSCLE ACTIVITY

BETWEEN LOCOMOTION TYPES

(% RGC)

occurred at the end of the first Double Support phase, at Left End of Weight Bearing. The statistical analysis (Table 14) revealed that there was no significant difference between the Peak 1 points of the right and left erector spinae muscles or between the Peak 1 points of the right and left rectus abdominis muscles. Comparison of erector spinae to rectus abdominis showed that Peak 1 activity in the rectus abdominis (15%) occurred significantly later in the gait cycle than Peak 1 activity in erector spinae (6%).

During Stair Climbing (II,36) it can be seen that in all subjects both erector spinae muscles showed a peak of activity during the first Double Support phase of the gait cycle. A peak of activity in the right and left rectus abdominis was observed in 16 and 15 subjects, respectively and occurred toward the end of the first Double Support phase close to Left End of Weight Bearing. Results of the statistical analysis (Table 15) showed that Peak 1 in the right erector spinae occurred significantly earlier than that in the other muscles. There was no significant difference in the Peak 1 points between the three remaining muscles.

During Stair Descent (II,37), both erector spinae muscles showed a peak of activity in all subjects and this occurred at the end of the first Double Support phase at the point of Left End of Weight Bearing. Peak 1 activity of the right and left rectus abdominis muscles was seen in 12 and 15

Table 14. Summary of Data for Mean Point of Peak 1 EMG Activity (% of RGC) during Level Walking

	RES	LES	RRA	LRA
$\bar{X}$	6	6	15	15
SD	2	2	4	8

Summary of Statistical Analyses of Comparison of Peak 1 EMG Activity during Level Walking

ANOVA $F\ (3,49) = 21.7$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	.01	.01
LES		–	.01	.01
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 15. Summary of Data for Mean Point of Peak 1 EMG Activity during Stair Climbing (% of RGC)

	RES	LES	RRA	LRA
$\bar{X}$	5	11	10	13
SD	3	3	4	4

Summary of Statistical Analyses of Comparison of Peak 1 EMG Activity during Stair Climbing

ANOVA $F\ (3,63) = 15.7 \quad p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	.01	.01	.01
LES		–	NS	NS
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

subjects, respectively and, similarly to erector spinae, occurred at the end of the first Double Support phase. Statistical analysis showed that there was no significant difference between the Peak 1 events of all four muscles.

During Ramp Climbing (II,38) both erector spinae muscles showed a peak of activity in all subjects and this occurred during the first Double Support phase of the gait cycle. The right and left rectus abdominis muscles showed Peak 1 activity in 15 and 14 subjects, respectively and this peak also occurred during the first Double Support phase. Results of the statistical analysis (Table 16) showed that Peak 1 activity in the right erector spinae occurred significantly earlier in the gait cycle than that of the other muscles. There was no significant difference in the Peak 1 events between the left erector spinae and the two rectus abdominis muscles.

During Ramp Descent (II,39), both erector spinae muscles showed a peak of activity in all subjects and this occurred during the first Double Support phase. Peak 1 activity in the right and left rectus abdominis muscles was observed in 13 and 15 subjects, respectively and occurred at the end of the first Double Support phase, close to Left End of Weight Bearing. Statistical analysis (Table 17) showed that Peak 1 activity in the left erector spinae occurred significantly earlier in the gait cycle than that of the right rectus

<u>Table 16. Summary of Data for Mean Point of Peak 1 EMG Activity (% of RGC) during Ramp Climbing (% of RGC)</u>

	RES	LES	RRA	LRA
$\bar{X}$	5	10	11	13
SD	2	3	7	6

<u>Summary of Statistical Analyses of Comparison of Peak 1 EMG Activity during Ramp Climbing</u>

ANOVA (3,61) = 6.7 P < .001

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	.05	.01	.01
LES		–	NS	NS
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 17. Summary of Data for Mean Point of Peak 1 EMG Activity (% of RGC) during Ramp Descent

	RES	LES	RRA	LRA
$\bar{X}$	9	6	14	10
SD	5	5	8	6

Summary of Statistical Analyses of Comparison of Peak 1 EMG Activity during Ramp Descent

ANOVA $F\ (3,60) = 5.0$ $P < .004$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	NS	NS
LES		–	.01	NS
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

abdominis, otherwise there was no significant difference in Peak 1 events between the muscles.

The point of Peak 1 EMG activity of each muscle was also compared between the five types of locomotion (II,40-43). Statistical analyses using a repeated measures analysis of variance and a post hoc Tukey's Test showed that significant differences between the different types of locomotion occurred only for the erector spinae muscles. Tables 18 and 19 present the summary data for these muscles.

In the right erector spinae it can be seen that Peak 1 activity occurred earliest during Stair Climbing and Ramp Climbing and latest during Stair Descent. There was no significant difference between the Peak 1 points of the two climbing or of the two descending types of locomotion. Peak 1 activity in Level Walking occurred significantly earlier than in Stair Descent, but was not significantly different compared to the other types of locomotion. The Peak 1 points of the two climbing types of locomotion occurred significantly earlier than those of the two descending types of locomotion.

In the left erector spinae it can be seen that Peak 1 activity occurred earliest during Level Walking and Ramp Descent and latest during Stair Climbing and Stair Descent. There was no significant difference between the points of Peak 1 activity during Stair Climbing, Ramp Climbing and

Table 18. Summary of Data for Mean Point of Peak 1 EMG Activity (% of RGC) of Right Erector Spinae during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	6	5	12	5	9
SD	2	3	6	2	5

Summary of Statistical Analyses of Comparison of Peak 1 EMG Activity of Right Erector Spinae in All Types of Locomotion

ANOVA $F(4,85) = 12.5$ $P < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	.01	NS	NS
SC		–	.01	NS	.01
SD			–	.01	NS
RC				–	.05

KEY: L = Level SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 19. Summary of Data for Mean Point of Peak 1 EMG Activity (% of RGC) of Left Erector Spinae during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	6	11	11	10	6
SD	2	3	7	3	5

Summary of Statistical Analyses of Comparison of Peak 1 Left Erector Spinae Activity during All Types of Locomotion

ANOVA $F\ (4,85) = 6.6 \quad p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	.05	.01	.05	NS
SC		–	NS	NS	.01
SD			–	NS	.01
RC				–	.05

KEY: L = Level Walking SC = Stair Climbing SD = Stair Descent
RC = Ramp Climbing RC = Ramp Descent

Stair Descent. The Peak 1 points during Level Walking and Ramp Descent were found to be significantly different from those of Stair Climbing, Ramp Climbing and Stair Descent.

There was no significant difference between the Peak 1 points of the right or of the left rectus abdominis muscles during the five different types of locomotion, therefore no summary of the data or statistical analyses are given.

In summary, both erector spinae muscles of all subjects showed a peak of EMG activity during the early part of the Right Gait Cycle, during or close to the end of the first Double Support phase. This activity occurred at essentially the same time in the gait cycle except in Stair and Ramp Climbing when the peak activity of the left erector spinae occurred later in Double Support than that of the right erector spinae. A peak of activity early in the Right Gait Cycle was not always present in the rectus abdominis muscles. When present, the peak occurred toward the end of the first Double Support phase at essentially the same time, regardless of the type of locomotion.

PEAK 2 EMG ACTIVITY

The mean points of Peak 2 EMG activity of the four muscles during each of the five types of locomotion are presented in Fig. 21; the raw data can be found in Appendix II,44-48. Analyses of variance and <u>post hoc</u> Tukey's Tests were used to compare the Peak 2 activity between muscles in each type of locomotion.

During Level Walking (II,44), both erector spinae muscles in all subjects showed a second peak of EMG activity in the middle of the second Double Support phase. A second peak of activity was also seen in all subjects in the EMG tracings of both rectus abdominis muscles, and this occurred at the end of Left Swing. Statistical analysis (Table 20) showed that there was no significant difference between the Peak 2 points of right compared to the left erector spinae muscles or between right compared to the left rectus abdominis muscles. There was also no significant difference between the Peak 2 points in the right rectus abdominis and in the two erector spinae muscles. There was a significant difference between the Peak 2 points of the left rectus abdominis compared to the left and right erector spinae muscles.

During Stair Climbing (II,45), Peak 2 activity in both erector spinae muscles and in the left rectus abdominis muscle was present in 18 subjects; in the right rectus abdominis muscle it was present in 17 subjects. It can be seen that the mean Peak 2 activity in the right erector spinae and in both rectus abdominis muscles occurred in the middle of the second Double Support phase while that of the left erector spinae occurred at the end of Left Swing, just

Table 20. Summary of Data for Mean Point of Peak 2 EMG Activity during Level Walking (% of RGC)

	RES	LES	RRA	LRA
$\bar{X}$	56	55	49	44
SD	2	3	13	12

Summary of Statistical Analyses of Comparison of Peak 2 EMG Activity in Level Walking

ANOVA $F(3,68) = 7.3$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	NS	NS	.01
LES		–	NS	.01
RRA			–	NS

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spina LRA = Left Rectus Abdominis

before Left Initial Contact. Statistical analysis (Table 21) indicated that there was no significant difference between the Peak 2 points of the right and left rectus abdominis and the right erector spinae muscles. The Peak 2 points of these muscles occurred, however, significantly later in the gait cycle than that of the left erector spinae.

During Stair Descent (II,46) a second peak of activity was present in the erector spinae muscles of all subjects; in the rectus abdominis muscles it was present in 17 subjects. It can be seen that the Peak 2 activity in all four muscles occurred during the middle of the second Double Support phase. Statistical analysis showed that there was no significant difference between the Peak 2 points of the four muscles.

During Ramp Climbing (II,47) a second peak of EMG activity was present in each muscle of all subjects and occurred during the second Double Support phase. Statistical analysis revealed that there was no significant difference between the Peak 2 points of the four muscles.

During Ramp Descent (II,48) a second peak of activity was present in right and left erector spinae and left rectus abdominis muscles in all subjects; in the right rectus abdominis it was present in 17 subjects. In the erector spinae muscles Peak 2 occurred in the middle of the second Double Support phase; in the rectus abdominis muscles it

Table 21. Summary of Data for Mean Point of Peak 2 EMG Activity during Stair Climbing (% of RGC)

	RES	LES	RRA	LRA
$\bar{X}$	58	50	57	56
SD	2	6	6	9

Summary of Statistical Analyses of Comparison of Peak 2 EMG Activity during Stair Climbing

ANOVA $F\ (3,67) = 7.4$ $p < .001$

Results of post hoc Tukey's Test

	RES	LES	RRA	LRA
RES	–	.01	NS	NS
LES		–	.01	.01
RRA			–	NS

Key: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

occurred at the end of Left Swing. There was, however, no significant difference between the Peak 2 points of all muscles.

The point of Peak 2 EMG activity of each muscle was also compared between the five types of locomotion (II,49-52). Although differences between means were small, statistical analyses using a repeated measures analysis of variance and post hoc Tukey's Test showed that significant differences between the types of locomotion existed for the right erector spinae and left rectus abdominis muscles. Tables 22 and 23 present the summary data for these muscles. In the right erector spinae it can be seen that Peak 2 activity occurred significantly earlier in the two descending types of locomotion than in the two climbing types. There was no significant difference between Stair Climbing and Ramp Climbing or between Stair Descent and Ramp Descent.

For the left rectus abdominis Peak 2 activity occurred at essentially the same time in the Right Gait Cycle in all types of locomotion, however it occurred significantly earlier in Level Walking compared to Stair Climbing.

In summary, a second peak of EMG activity was seen in all four muscles in the majority of subjects. Both erector spinae muscles showed a peak of activity during the second Double Support phase of the Right Gait Cycle. This activity occurred at essentially the same time in the gait cycle

Table 22. Summary of Data for Mean Point of Peak 2 EMG Activity (% of RGC) of Right Erector Spinae during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	56	58	54	58	54
SD	2	2	8	3	3

Summary of Statistical Analyses of Comparison of Peak 2 Right Erector Spinae Activity during All Types of Locomotion

ANOVA $F(4,68) = 5.2$ $p < .001$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	NS	NS	NS
SC		–	.05	NS	.05
SD			–	.05	NS
RC				–	.05

KEY: L = Level Walking SC = Stair Climbing SD = Stair Descent
RC = Ramp Climbing RD = Ramp Descent

Table 23. Summary of Data for Mean Point of Peak 2 EMG Activity (% of RGC) of Left Rectus Abdominis during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	44	56	52	53	49
SD	12	9	9	10	9

Summary of Statistical Analyses of Comparison of Peak 2 Left Rectus Abdominis Activity during All Types of Locomotion

ANOVA $F_{(4,84)} = 4.1$ $p < .004$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	.01	NS	NS	NS
SC		–	NS	NS	NS
SD			–	NS	NS
RC				–	NS

KEY: L = Level Walking SC = Stair Climbing SD = Stair Descent
RC = Ramp Climbing RD = Ramp Descent

except in Stair Climbing in which the Peak 2 activity in the left erector spinae occurred at the beginning of the Double Support phase. In the rectus abdominis muscles the presence of a second peak of activity was more variable than in the erector spinae muscles. Peak 2 activity in the rectus abdominis muscles occurred at essentially the same time in the gait cycle in all types of locomotion; that is, during or close to the beginning of the second Double Support phase of the Right Gait Cycle.

PEAK 3 EMG ACTIVITY

The mean points of the third peak of EMG activity in all four muscles are presented in Fig. 21; the raw data are contained in Appendix II,53-57. Comparison of the Peak 3 events between muscles in each type of locomotion was done using analyses of variance and post hoc Tukey's Tests.

During Level Walking (II,53) a third peak of muscle activity during the Right Gait Cycle was rarely present in the erector spinae muscles; in the right muscle it was present in only four subjects and it was absent in the left muscle. In the right erector spinae Peak 3 activity occurred at the end of Right Swing. In the right and left rectus abdominis muscles a third peak of EMG activity was seen in 16 and 15 subjects, respectively and also occurred toward the end of Right Swing. Statistical analysis showed that there was no significant difference between the Peak 3 events of right erector spinae and the two rectus abdominis muscles.

During Stair Climbing (II,54) a third peak of muscle activity was variably present in the erector spinae muscles; in the right muscle it occurred in 13 subjects while in the left muscle it was present in only three subjects. Peak 3 activity was also variably present in the rectus abdominis muscles, occurring in 11 and eight subjects in right and left muscles respectively. Peak 3 activity in all muscles occurred at the end of Right Swing. Statistical analysis of the data showed that there was no significant difference between the Peak 3 points of the four muscles.

During Stair Descent (II,55) a third peak of EMG activity was not always seen. In the right and left erector spinae it was present in eight and three subjects, respectively; in the right and left rectus abdominis it was present in 16 and 10 subjects, respectively. When present, Peak 3 activity in all muscles took place at the end of Right Swing. Statistical analysis showed no significant difference in the points of Peak 3 activity between the four muscles.

During Ramp Climbing (II,56) a third peak of EMG activity was occasionally seen in the erector spinae muscles; for the right muscle it was present in four subjects while in the left muscle it was present in only one subject. A third peak of activity in the rectus abdominis muscles was seen in the majority of subjects, being present in the right muscle in 10 subjects and in the left muscle in nine subjects. When present, Peak 3 activity occurred during Right Swing. When the points of Peak 3 activity of the four muscles were compared, there was no significant difference between them.

During Ramp Descent (II,57) a third peak of activity was occasionally seen in the erector spinae muscles; it was present in the right muscle in eight subjects and in the left muscle in three subjects. A third peak of activity was seen in the rectus abdominis muscles of the majority of subjects, being present in the right and left muscles of 15 and 14 subjects, respectively. In all muscles Peak 3 occurred at the end of Right Swing and statistical analysis showed that there was no significant difference between muscles.

The mean point of Peak 3 EMG activity of each muscle was also compared between the five types of locomotion (II,58-61). Differences between the means were small; a repeated measures analysis of variance and *post hoc* Tukey's Test showed that a significant difference existed only for the left rectus abdominis muscle (Table 24). It can be seen that in this muscle Peak 3 occurred earliest during Level Walking and latest during Ramp Climbing. Only this comparison was statistically significant.

Table 24. Summary of Data for Mean Point of Peak 3 EMG Activity (% of RGC) of Left Rectus Abdominis during All Types of Locomotion

	L	SC	SD	RC	RD
$\bar{X}$	81	89	90	91	87
SD	10	5	4	7	9

Summary of Statistical Analyses of Comparison of Peak 3 Left Rectus Abdominis Activity during All Types of Locomotion

ANOVA $F(4,51) = 2.73 \quad p < .04$

Results of post hoc Tukey's Test

	L	SC	SD	RC	RD
L	–	NS	NS	.05	NS
SC		–	NS	NS	NS
SD			–	NS	NS
RC				–	NS

KEY: L = Level Walking SC = Stair Climbing SD = Stair Descent RC = Ramp Climbing RD = Ramp Descent

In summary, a third peak of EMG activity was seen more frequently in the rectus abdominis muscles than in the erector spinae muscles. In all muscles it was present less frequently than Peak 1 or Peak 2 activity. For those subbjects in which Peak 3 occurred, it was found at the end of Right Swing.

MAXIMUM PEAK EMG ACTIVITY

For every muscle in each type of locomotion the peak of EMG activity with the highest amplitude during the Right Gait Cycle was identified (Fig. 21). In the right erector spinae muscle the second peak of EMG activity, occurring during the second Double Support phase, showed maximum amplitude and this occurred consistently in all five types of locomotion. In the left erector spinae the first peak of EMG activity, occurring during the first Double Support phase, showed maximum amplitude and this occurred consistently in all five types of locomotion. The point of maximum amplitude was less consistent in the rectus abdominis muscles. In the right rectus abdominis the point of maximum amplitude during Level Walking occurred during Right Swing while in Stair Climbing it occurred during the first Double Support phase; in all other types of locomotion it occurred during the second Double Support phase. In the left rectus abdominis the point of maximum amplitude occurred during the second Double

Support phase in all types of locomotion except Stair Descent in which it occurred during the first Double Support phase.

CHAPTER V

DISCUSSION

CYCLE DURATION

In this study it was found that the duration of the Right Gait Cycle during free-speed level walking was 1.2 ± 0.1 sec. This value is similar to the range of 1.02 to 1.16 sec reported by Murray et al. (1964 ,1966 ,1969), Waters and Morris (1970), Cappozzo (1983), and Thorstensson et al. (1984). It was also found that the percentage of the gait cycle represented by the Stance phase decreased as the duration of the gait cycle decreased and this was consistent with the findings of Murray et al. (1964, 1966), Kirtley et al. (1985) and Nilsson et al. (1985). However, when the percentages of the gait cycle represented by Stance were compared between the different types of locomotion, the differences were found to be not significant.

Corlett et al. (1972) found that the duration of the gait cycle in Stair Descent was less than that in Ramp Descent for the same slope. In the present study no significant difference was found between the cycle duration of the two descending types of locomotion, however the slope of the stairs and the slope of the ramp were not equal. Waters and Morris (1970) found that the duration of the Right Gait Cycle during climbing a ramp with a 5° incline was 1.2 sec; in this study the duration was longer (1.4 sec), possibly because the slope was greater (15°).

The duration of the gait cycle was significantly greater during the two climbing types of locomotion compared to the two descending types. This can be explained by the fact that the direction in which the body moves during walking up a ramp or stairs is opposite to that of the force of gravity. Gravity therefore tends to decelerate the body and decrease the walking speed. The direction in which the body moves when walking down a ramp or stairs is the same as that of the force of gravity. Gravity therefore tends to accelerate the body and increase the walking speed (Gray & Basmajian, 1968). The duration of the gait cycle during Level Walking was not significantly different from that during Stair Climbing, but was significantly shorter than that during Ramp Climbing. This may be attributed to the fact that Stair Climbing is generally a more familiar activity than Ramp Climbing. The subjects might therefore have responded to the increased locomotor demands presented by the relatively unfamiliar ramp climbing by decreasing speed and prolonging bipedal contact. This would serve to enhance stability and maintain the centre of gravity between the moving points of support (Grillner, 1981; Conrad et al., 1983). In addition, during ramp climbing the foot must accommodate to a sloped surface as opposed to the flat surface of stairs which might result in decreased walking speed and increased duration of the gait cycle. Such a phenomenon was observed by Conrad et al.

(1983) who found that cats accommodated to ramp climbing by decreasing their walking speed.

CADENCE

In this study the number of steps taken per minute ranged from a low of 90.2 during Ramp Climbing to a high of 116.0 during Stair Descent. These values lie within the range of 61 to 168 steps per minute reported by Drillis (1958), Grieve and Gear (1966), Murray et al. (1969), Waters et al. (1973), Thurston and Harris (1983), Winter (1983) and Kirtley et al. (1985). It was found that increased cadence was associated with a decrease in the duration of the gait cycle; this is consistent with the findings of Murray (1967), Murray et al. (1969), Winter (1983) and Kirtley et al. (1985).

The cadence during Ramp Climbing was significantly lower than that during the other four types of locomotion. This is consistent with the results of Bobbert (1960) who observed that cadence decreased during climbing when the slope of a treadmill exceeded 8°. This might be explained by the relative unfamiliarity of ramp walking which would lead to a protective locomotor strategy. As with duration of gait cycle, another consideration is the need in ramp walking for the foot to accommodate to a sloped surface; this could result in prolonged bipedal contact and therefore decreased cadence (Grillner, 1981; Conrad et al., 1983).

Cadence was significantly increased during the two descending types of locomotion compared to the other three types of locomotion. This can be explained by the effect of gravity which tends to accelerate the body as the subject walks down a set of stairs or a ramp. As walking speed increases, cadence increases (Murray, 1967; Murray et al., 1969; Winter, 1983; Kirtley et al., 1985). Although not statistically significant, the cadence during Stair Descent was greater than that during Ramp Descent, possibly because descending stairs was generally a more familiar activity than descending a ramp. The subjects might have responded to the increased locomotor demands of the relatively unfamiliar ramp walking by decreasing speed and therefore cadence. This would result in prolonged bipedal contact in order to maintain the centre of gravity between the moving points of support and thus enhace stability (Grillner, 1981; Conrad et al., 1983).

TEMPORAL EVENTS

During Level Walking it was observed that, on average, Right Stance constituted 63±3% of the Right Gait cycle. This finding is congruent with the results of Murray et al. (1964, 1966, 1969), Dubo et al. (1976), Cappozzo (1983), Thurston and Harris (1983), and Kirtley et al. (1985) who found that Stance constituted approximately 60% and Swing approximately 40% of the gait cycle. It was found that the mean percentage of the first Double Support phase was 13$\pm$3% and of the second Double Support phase 12$\pm$3%, comparable to the 10% value for the Double Support phases given by Murray et al. (1964, 1966), Thurston and Harris (1983), and Kirtley et al. (1985). During Level Walking the events of Left End of Weight Bearing and Left Initial Contact occurred at 13$\pm$3% and 50$\pm$2%, respectively of the Right Gait Cycle, compared to the figures of approximately 10% for Left End of Weight Bearing and 50% for Left Initial Contact given by Thurston and Harris (1983). Therefore, in the present study the temporal events of the Right Gait Cycle during Level Walking were consistent with the findings of other investigators.

During Stair Climbing it was observed that, on average, Right Stance constituted 63$\pm$4% and the first and second Double Support phases constituted 14$\pm$2% and 12$\pm$3% respectively, of the Right Gait Cycle. These values differ from the data of Joseph and Watson (1967) who found Stance to be approximately 71% and the two Double Support phases to be 29% of the Right Gait Cycle. Unfortunately, no measures of variation were reported in their study, therefore it is impossible to determine if the results of the present investigation fall within the range of values found by Joseph and Watson (1967).

During Stair Descent it was observed that, on average, Right Stance constituted 61$\pm$5% and the first and second Double Support phases constituted 10$\pm$3% and 14$\pm$9%, respectively of the Right Gait Cycle. The value for Stance is comparable to the 63% found by Joseph and Watson (1967), however the values for the Double Support phases are different from the value of 20% cited by the same investigators. Again, a meaningful comparison is difficult because of the lack of reported measures of variation.

For each type of locomotion in the present study, variation in temporal events between subjects was small, as evidenced by the low standard deviations. When the individual temporal events were compared between different types of locomotion, differences, although statistically significant in Left End of Weight Bearing and Left Initial Contact, were also small. The basic similarity of the temporal events of the gait cycle, regardless of type of locomotion, supports the proposal of Grillner (1981) and of Conrad et al. (1983) that the basic motor act of walking is a rhythmic, stereotyped pattern that can be modified, corrected and adapted in response to changes in the environment and in the volition of the subject.

BODY SEGMENT DISPLACEMENT

LEVEL WALKING

In the present study the mean excursion of the trunk segment was $3 \pm 1°$ which is within the range of values reported by Cappozzo et al. (1978) and Thorstensson et al. (1982, 1984). Although, on average, the trunk segment was held in slight anterior inclination throughout the gait cycle, the position in which this body segment was held varied between subjects and this variability was also observed by Thurston and Harris (1983).

Maximum anterior inclination of the trunk occurred at the end of Right and Left Swing phases as the body fell towards the next point of ground contact. This is consistent with the observations of Carlson and Thorstensson (1981) who found that the point of maximum anterior trunk inclination occurred just after Initial Contact at the end of the Double Support phase, of Thurston and Harris (1983) who found that maximum anterior inclination occurred during Double Support, and of Thorstensson et al. (1984) who found that maximum anterior inclination occurred at Initial Contact.

In this study, the trunk segment did not move into posterior inclination, but attained the neutral (0°) position just before Mid Stance as the trunk moved over the supporting lower limb. This is consistent with the findings of Thurston and Harris (1983) who observed that peak posterior inclination of the trunk occurred at 40% and 80% of the gait cycle at Right and Left Mid Stance, respectively. Thorstensson et al. (1984) reported that peak posterior trunk inclination occurred during Swing which occurs simultaneously with contralateral Stance.

The mean excursion of the pelvic segment was $11 \pm 4°$ which is greater than the value of 6° reported by Lamoreux (1971) and 3° reported by Murray et al. (1964, 1966). Although, on average, the pelvic segment was held in anterior inclination throughout the gait cycle, the position in which this body segment was held varied between subjects and this variablity was also observed by Thurston and Harris (1983).

Maximum anterior inclination of the pelvic segment occurred during the second Double Support phase at the end of Right Stance when the right lower limb was extended behind the body. Murray et al. (1964, 1966) reported that maximum anterior pelvic inclination occurred just before Initial Contact as the trunk inclined forward toward the next area of ground contact. It should be noted that Murray et al. (1964, 1966) measured pelvic inclination from the horizontal plane and careful inspection of the interrupted light photograph in the 1964 paper (p.339) shows that, relative to the vertical, maximum anterior inclination of the pelvic segment occurred at the end of Stance phase. Thurston and Harris (1983) also measured pelvic inclination relative to the horizontal and presented plots of pelvic inclination curves that showed maximum anterior inclination occurring at 40% and 86% of the gait cycle, just after Mid Stance and at Mid Swing with the greatest anterior excursion occurring during Mid Swing. They proposed that pelvic motion was related to the limb in Swing phase in that the inertia of this limb exerted an influence on pelvic movement.

In this study the pelvic segment did not move into posterior inclination; it reached minimum anterior inclination during the first Double Support phase when the right lower limb was positioned in front of the body. Murray et al (1964, 1966) reported that maximum posterior pelvic inclination occurred early in Mid Stance as the trunk moved over the supporting limb. Again, it should be noted that these investigators measured pelvic inclination relative to the horizontal plane and careful inspection of the interrupted light photograph in the 1964 paper (p.339)shows that, relative to the vertical, maximum posterior inclination of the pelvic segment occurred at approximately Initial Contact. Thurston and Harris (1983), who also measured pelvic inclination relative to the horizontal, presented plots of pelvic inclination curves that showed maximum posterior inclination at 16% and 62% of the gait cycle, at the beginning of Stance and Swing phases, with the greatest posterior excursion occurring during Stance.

The apparent discrepancy with respect to pelvic motion between the findings of the present study and those of Murray et al. (1964, 1966) and of Thurston and Harris (1983) could be attributed to the difference in planes of reference and the absence of a precise description of the angle that was measured.

When hip angle is considered, mean maximum hip flexion in this study was $25 \pm 2°$ which is less than the values of 30° and 37° reported by Murray et al. (1964) and Johnson and Smidt (1969), respectively. Mean maximum hip extension was $15 \pm 2°$ which is comparable to the values of 10° and 15° reported by Murray et al. (1964) and Johnson and Smidt (1969), respectively.

In the present study it was found that maximum hip flexion occurred at 85% to 90% of the gait cycle, just before the end of Right Swing. Murray et al. (1964) reported that maximum hip flexion occurred at Initial Contact, but Inman et al. (1981) stated that maximum hip flexion occurred at 85% of the gait cycle.

It was found that hip extension began at 95% of the gait cycle, just before Initial Contact, continued into the next gait cycle, and reached a maximum at 55% of the gait cycle, at the end of Right Stance. Murray et al.(1964) observed that hip extension began as the body started to move over the supporting limb and ended at approximately 50% of the gait cycle, coincident with contralateral Initial Contact. Johnson and Smidt (1969) also found that hip extension began before Initial Contact and gradually increased until the end of Stance phase; hip flexion began immediately before End of Weight Bearing and continued until just before Initial Contact. Nilsson et al. (1985) also observed that hip flexion from the position of maximum extension began during the second Double Support phase.

STAIR CLIMBING

Few reports were found of body segment displacement during stair climbing against which to compare the data from this study. Townsend, Lainhart, et al. (1978) stated that foot clearance during stair climbing was achieved by hip and knee flexion and this is supported by the data from the present study which show that greater hip flexion occurred during Stair Climbing (45°) than during Level Walking (25°) in which the foot does not have to clear the "obstacle"of a step. The pattern of hip angle motion seen in Fig. 11 was also observed by Andriacchi et al. (1980). They reported that maximum hip flexion during Stance was 34° and during Swing was 41°. When the variation in their data is taken into account, these values are comparable to those of 42° for Stance and 45° for Swing found in this study.

STAIR DESCENT

Only one study was found that reported on body segment displacement during stair descent. Andriacchi et al. (1980) observed that the hip was in flexion at Initial Contact, maximum hip flexion occurred at End of Weight Bearing, and the hip moved from flexion into extension during Swing. This was the pattern of motion observed in the present study. Andriacchi et al. (1980) reported that maximum hip flexion during Stance was 13° and during Swing was 23°; these values are comparable to those of 13° for Stance and 35° for Swing found in this study.

RAMP CLIMBING

The study of Gray and Basmajian (1968) was the only one found that reported on body segment displacement during ramp climbing. These investigators proposed that the slope of a ramp acts like an obstacle to the foot and that increased hip or knee flexion is used to ensure foot clearance. This proposal is supported by the data from this study which showed that greater hip flexion occurred during Ramp Climbing (45°) than during Level Walking (25°) in which the foot did not have to clear an "obstacle".

RAMP DESCENT

The study of Tokuhiro et al. (1985) was the only one found that reported on body segment displacement during ramp descent. They observed that there was less hip flexion required in ramp descent than in ramp climbing; this is supported by the finding in this study that maximum hip flexion during Ramp Descent was 22° compared to 45° during Ramp Climbing.

COMPARISON OF BODY SEGMENT DISPLACEMENTS

When the curves of the trunk segment, pelvic segment and hip angle are considered together, it can be seen that in all types of locomotion, the excursion of the trunk segment is smaller compared to those of the pelvic segment and hip. This difference was also observed by Waters et al. (1973), Cappozzo (1981), and Thorstensson et al. (1984). During locomotion, the head must be held in a relatively stable position in order to maintain visual contact with the environment and to provide the optimum conditions for the vestibular apparatus (Thorstensson et al., 1984). It is proposed that the pelvic and trunk segments act sequentially as dampers of the high amplitude oscillations of the lower limbs in order to maintain stable head position during locomotion. If this is the case, it would be expected that the muscles that effect and also restrain sagittal plane motion of the trunk and pelvic segments would be active throughout the gait cycle in order to control the excursion of these segments; this was what was found.

Anterior inclination of both trunk and pelvic segments was greatest during the two climbing types of locomotion, possibly because the anterior displacement of the centre of gravity of these body segments helps to propel the body up the stairs or ramp.

Overall, posterior trunk inclination was seen more frequently in Ramp Descent than in Stair Descent. During Stair Descent the feet are placed on the level surface of the stair treads and the body must make relatively minor accommodations for the effect of gravity. However, during Ramp Descent the feet are placed on a downward sloping surface and equilibrium can only be maintained by a posterior shift of the centre of gravity of the upper body. In addition, posterior trunk inclination with its accompanying posterior shift in centre of gravity may also be used as a strategy to control the rate of fall of the body during Ramp Descent. Vision may also account for the difference in trunk segment inclination between Stair Descent and Ramp Descent. Vision may be more important to foot placement during Stair Descent beacuse of the relatively short length of the stair tread compared to the unbroken sloping surface of a ramp.

Therefore the trunk tends to be held close to neutral (0°) or in anterior inclination during Stair Descent in order for the subject to see where he must place his feet.

The total excursion of the pelvic segment was greatest during Level Walking and least during Stair Descent. The amount of pelvic excursion appeared to be independent of the range of motion that occurred at the hip. The greater amount of pelvic segment excursion during Level Walking may be due to the fact that this was the most familiar of the five types of locomotion. It should be noted that the standard deviations of trunk and pelvic segment displacements and of hip angle were smallest during Level Walking, possibly indicating that this is a more familiar type of locomotion than the other four. It can be argued that during level walking the body segments are relaxed and move freely; during climbing and descending types of locomotion the effect of gravity is increased and an adjustment in locomotor strategy is required in order for the body to move simultaneously in both horizontal and vertical planes and to adapt to different equilibrium requirements. The overall effect is decreased stability during stair and ramp walking and therefore it might be expected that the upper body segments would be held more rigidly during these types of locomotion, as was seen in this study.

The relationship between trunk and pelvic segment inclination was not consistent across subjects although, on average, the trunk segment inclination curves were smaller versions of the pelvic curves. This finding differed from the data of Thurston and Harris (1983) who found that, on average, the trunk and pelvis were 90° out of phase with each other. However, they too reported considerable variation from subject to subject. In addition, they studied the T_{12} to L_4 spinal segment whereas in this investigation, the C_7 to L_5 segment was studied. As noted previously, Thurston and Harris (1983) measured pelvic motion relative to the horizontal and the angle was not defined precisely. Therefore, their data are not strictly comparable to those of the present study. However, given the results of the two studies, it must be agreed with Thurston and Harris (1983) that the link between movements of the pelvis and the lumbar spine still remains unclear. In order to compare meaningfully the results between different studies, there must be agreement among investigators on the methods to be used to measure spinal and pelvic motion. Currently no convention for measuring these body segments has been universally accepted (Winter, 1986).

The patterns of hip excursion in Level Walking, Stair Climbing and Ramp Climbing were similar. Maximum hip flexion occurred at the end of right Swing when the foot had reached its maximum anterior position, but before it contacted the ground. Maximum hip extension occurred at the end of right Stance when the supporting limb was positioned behind the upper body.

The pattern of hip excursion was altered during the two descending types of locomotion. During Ramp Descent maximum hip flexion occurred at the end of right Swing, however the hip did not move into extension, but rather reached a position of minimum flexion at the end of right Stance. The hip may have been maintained in flexion throughout Ramp Descent because hip extension would result in anterior displacement of the centre of gravity of the upper body, likely causing a loss of balance. During Stair Descent maximum hip flexion occurred during Initial Swing as the foot was lifted to clear the stair tread before it could be displaced anteriorly and lowered to the tread next below. Overall, the hip did not move into extension during Stair Descent, but rather minimum hip flexion occurred during Mid Stance as the lower limb supported the upper body while the contralateral limb was lowered to the tread next below.

Comparison of total hip excursion between the five types of locomotion showed that the greatest amount of excursion occurred during the two climbing types and the least amount during the two descending types of locomotion. During both Stair Climbing and Ramp Climbing the foot encounters obstacles - the next highest step in the case of Stair Climbing and the slope of the surface in Ramp Climbing - and foot clearance can be achieved by increasing hip and/or knee flexion. In this study it was found that the amount of hip flexion was significantly greater during Ramp Climbing and Stair Climbing than in the other three types of locomotion. However, the amount of hip extension was significantly greater during Ramp Climbing than during Stair Climbing. This may be due to the fact that during the former type of locomotion the size of step taken can be varied at will, whereas during Stair Climbing the size of step taken is determined by the height of the stair riser. In addition, during Ramp Climbing larger steps may have to be taken in order to maintain the body's centre of gravity between the moving points of support (Grillner, 1981) and this might require greater hip extension. Hip flexion during Ramp Descent was less than in the other types of locomotion, possibly because smaller steps were taken in order to maintain equilibrium. Step and Stride length were not measured in this investigation and should be studied in order to determine if, in fact, the shortest steps are taken during Ramp Descent.

TOTAL MYOELECTRIC ACTIVITY

In the present study it was found that the total amount of electrical activity produced by the erector spinae muscles during Level Walking was 13±7% of the MVC. This value is comparable to the findings of Guth et al. (1979) who reported activity of 6% and 5% in the lumbar and thoracic portions respectively, of the erector spinae muscle. They also found that the level of activity was greater on the right side than on the left; in this study the level of activity in right and left erector spinae muscles during Level Walking was not significantly different.

As noted previously, the trunk and pelvic segments were held in anterior inclination throughout the entire gait cycle during both Stair Climbing and Ramp Climbing. In addition, the point of maximum anterior inclination of these body segments also occurred during the two climbing types of locomotion. Therefore, the centre of gravity of the upper body was displaced furthest anteriorly and the forward bending moment would be greatest during the two climbing types of locomotion than in the other three types. It would be expected that more motor units in the the erector spinae muscles would be recruited during the climbing locomotion types in order to provide a force to balance the forward bending moment and also to control the forward excursion of the trunk. This was found to be the case as the erector spinae muscles showed significantly greater myoelectric activity during Stair Climbing and Ramp Climbing than during any other type of locomotion. The data therefore support the findings of Floyd and Silver (1951) that increasing angles of forward flexion led to incresed levels of myoelectric activity, and also those of Schultz, Andersson, Ortengren, Bjork and Nordin (1982) who found that erector spinae activity closely reflected the magnitude of the net flexor moment.

There was no significant difference between the amount of myoelectric activity in the erector spinae muscles seen during Stair Climbing compared to Ramp Climbing, and it can be concluded that these muscles contract with approximately the same level of intensity regardless of whether the climbing activity involves the use of stairs or a ramp. Therefore, with respect to the erector spinae muscles, no advantage or disadvantage is conferred by using one type of climbing apparatus over another.

For the majority of subjects the rectus abdominis muscles were active throughout the gait cycle of all types of locomotion, however the mean amount of total electrical activity generated was small. This observation differs from the findings of Sheffield (1962) who reported electrical silence in rectus abdominis during level walking. However, it is possible that the instruments he used to record the EMG signal from the muscles were not sufficiently sensitive to detect the low levels of activity in rectus abdominis.

In the present study, the greatest amount of myoelectric activity in rectus abdominis was seen during Ramp Descent; this is consistent with the trunk and pelvic segment inclination data which showed that only during Ramp Descent was there posterior inclination of these segments. The data are in accord with the observations of Floyd and Silver (1950) who found that rectus abdominis was active during trunk extension (posterior inclination) from the upright position, and also support the conclusion of Flint and Gudgell (1965) that rectus abdominis is active to position the trunk and balance the extensor moment produced by trunk extension (posterior inclination).

PHASIC ACTIVITY OF MUSCLES

During Level Walking consistent biphasic activity was found in the erector spinae muscles of all subjects, and in four subjects a third peak of activity was seen in the right erector spinae. Waters and Morris (1972) also found biphasic activity with peaks of back muscle activity occurring consistently at contralateral Initial Contact, but less frequently at ipsilateral Initial Contact. Triphasic activity in paraspinal muscles was observed by Letts et al. (1978).

In this study peaks of myoelectric activity in the two erector spinae muscles occurred symmetrically during the first and second Double Support phases of the gait cycle during Level Walking. These observations are consistent with the findings of Battye and Joseph (1966), Letts et al. (1978), and Carlson and Thorstensson (1981); they differ from the findings of Thorstensson et al. (1982) who found that the peak of activity of the ipsilateral muscle preceded the peak of the contralateral muscle by 5% to 10% of the gait cycle. The data from the present study support the biomechanical model of Cappozzo (1983) which predicted that the trunk extensor muscles would exert peak force just before or just after Initial Contact.

During Level Walking the first peak of myoelectric activity in the erector spinae muscles occurred at 6% of the Right Gait Cycle, before the point of minimum anterior trunk inclination (10% - 25% of RGC) when the trunk segment was moving in a posterior direction. This confirms the results of Thorstensson et al. (1982) who found that both periods of electrical activity in paraspinal muscles occurred during or at the end of an angular displacement directed backwards in the sagittal plane. They also found that both periods of paraspinal muscle activity coincided with the activation of hip extensor muscles. While hip extensor muscles were not studied in this investigation, it can be assumed that they would be active at the point at which the hip began to move from flexion into extension and at the point of maximum hip extension. In Level Walking these points occurred during the first and second Double Support phases which coincided with the first and second peaks of activity in both right and left erector spinae muscles, supporting the findings of Thorstensson et al. (1982).

When phasic myoelectric activity in rectus abdominis muscle during level walking is considered, Waters and Morris (1972) found that activity occurred simultaneously with that of back muscles. In this study, the first and second peaks of activity of both rectus abdominis and erector spinae muscles occurred during the first and second Double Support phases and, when present, the third peak of activity in both muscles occurred during right Swing.

During Stair Climbing it was found that there was consistent biphasic activity of the erector spinae muscles, with a third peak of activity seen in right and left erector spinae in 13 and three subjects, respectively. The first two peaks of myoelectric activity occurred during the first and second Double Support phases of the gait cycle, a finding that was consistent with the observations of Joseph and Watson (1967). Maximum peak myoelectric activity occurred just after contralateral Initial Contact, also found by Joseph and Watson (1967). In addition, these investigators stated that both erector spinae muscles contracted simultaneously in early Stance when the trunk was inclined forward and the body was displaced vertically; this was supported by the findings of the present study in which it was observed that peak erector spinae activity occurred during the two Double Support phases during which time the trunk was in maximum anterior inclination. The data from this study support the proposal of Joseph and Watson (1967) that bilateral erector spinae contraction controls the amount of forward inclination of the trunk.

During Stair Descent there was consistent biphasic activity of erector spinae with a third peak of activity seen in the right and left muscles in eight and three subjects, respectively. The first and second peaks of myoelectric activity occurred at the end of the first and during the second Double Support phases of the gait cycle at the points of minimum anterior inclination of the trunk segment. This is consistent with the findings of Joseph and Watson (1967) and supports their proposal that erector spinae contracts bilaterally to prevent trunk flexion (anterior inclination).

During ramp climbing Waters and Morris (1970) observed that the myoelectric activity of erector spinae and rectus abdominis was similar to that seen in level walking. Examination of the data from the present study shows that the differences in the points of peak myoelectric activity of the erector spinae muscles between Ramp Climbing and Level Walking were small with only Peak 1 activity in the left erector spinae occurring significantly earlier during Level Walking compared to Ramp Climbing. In addition, the difference between the points of peak myoelectric activity of the rectus abdominis muscles between Ramp Climbing and Level Walking were also small, with only Peak 3 activity in the left rectus abdominis occurring significantly earlier in Level Walking compared to Ramp Climbing.

POSSIBLE ROLE OF THE TRUNK MUSCLES DURING LOCOMOTION

Letts et al. (1978) proposed that reciprocal activity of the paraspinal muscles during level walking may play a major role in stabilization of the spine. Carlson and Thorstensson (1981) and Thorstensson et al. (1982) postulated that erector spinae activity prevents trunk flexion, lateral flexion and rotation. They also proposed that erector spinae controls the stiffness of the trunk which restricts movement, especially in the sagittal plane, and helps to maintain upright posture and equilibrium. Gracovetsky (1985) suggested that ipsilateral contraction of erector spinae forces the vertebral column to flex to the ipsilateral side, engaging the ipsilateral vertebral facet joints. This results in a flexion force because of coupling (Lumsden & Morris, 1968; White & Panjabi, 1978) and ultimately gives rise to an axial torque which initiates counter-rotation of pelvis and shoulder girdle.

In the present study it was found that maximum peak myoelectric activity in erector spinae did not occur bilaterally at the same point in the gait cycle, but instead occurred consistently in the ipsilateral muscle during the second Double Support phase and in the contralateral muscle during the first Double Support phase.

When motion in the sagittal plane is considered, the body is in a controlled forward fall during walking. This fall is interrupted by ground contact of the advancing limb. Momentum and inertia would tend to keep the trunk segment moving forward and downward (anterior inclination) and this motion must be controlled in order to avoid loss of balance. The forward motion is restricted by bilateral contraction of erector spinae. However, unopposed bilateral contraction of erector spinae would cause the trunk segment to move backward and upward, therefore control of trunk motion in the sagittal plane requires co-ordinated interplay between trunk extensors and flexors. Consequently, bilateral contraction of both erector spinae and rectus abdominis is required at the two Initial Contact events and this is seen in the data from the present study. However, the asymmetry of the maximum peak myoelectric activity in erector spinae might be due to motion of body segments in the coronal and transverse planes as well as in the sagittal plane.

In the coronal plane, according to Inman et al. (1981), the pelvis drops to the ipsilateral side just before Right End of Weight Bearing (approximately 55% of the RGC), and returns to level during right Swing (approximately 80% of the RGC). This dropping of the pelvis is controlled by eccentric contraction of the contralateral hip abductors, but eccentric contraction of the ipsilateral erector spinae may also occur to allow a controlled descent of the pelvis. This would require a burst of ipsilateral erector spinae activity at approximately 55% of the gait cycle, possibly accounting for the maximum peak activity during the second Double Support phase seen in the results of all five types of locomotion in this study. As the pelvis drops to the ipsilateral side, the trunk segment must be kept upright and this would require concentric contraction of the contralateral erector spinae at approximately 55% of the gait cycle. To return the pelvis to the level position during ipsilateral swing, concentric contraction of the contralateral hip abductors occurs. It may also be that there is a contralateral isometric contraction of erector spinae to stabilize the trunk segment, allowing the ipsilateral erector spinae to contract concentrically to assist in raising the pelvis. This could account for bilateral erector spinae peaks at approximately

80% of the gait cycle. As the pelvis returns to level, the trunk must remain upright and this would require either a contralateral eccentric contraction of erector spinae or an ipsilateral concentric contraction of erector spinae, or both.

At Initial Contact the body must be balanced on the ipsilateral hip joint in preparation for contralateral Swing. At this point a torque exists which tends to pull the pelvis down to the side of the Swing limb. While it is known that the ipsilateral hip abductors play a major role in preventing excessive pelvic tilt, the erector spinae muscle may also be involved and this could account for the peak of erector spinae activity seen bilaterally at Initial Contact. The ipsilateral erector spinae could contract to stabilize the column of the lower limb and trunk, while the contralateral erector spinae could work with reverse origin and insertion to prevent pelvic drop.

In the transverse plane, according to Inman et al. (1981), the pelvis at Initial Contact is maximally rotated to the contralateral side while the shoulder girdle and upper body are maximally rotated to the ipsilateral side. This ipsilateral rotation of the upper body could be the result of concentric ipsilateral contraction of erector spinae. However, maximum peak activity in erector spinae occurred on the contralateral side immediately after Initial Contact possibly because the contralateral muscle must contract to "derotate" pelvis and shoulder girdle from their extreme positions.

The small amount of myoelectric activity seen in rectus abdominis in this study suggests that these muscles have a synergistic as opposed to a prime mover function during locomotion. Using the classification of muscle function outlined by Gowitzke and Milner (1980), it is proposed that the rectus abdominis muscles act as stabilizing, neutralizing and conjoint synergists during locomotion.

The rectus abdominis muscles may work as stabilizing synergists to steady the mobile vertebral column and allow erector spinae and possibly the oblique abdominal muscles to effect pelvic and shoulder girdle counter-rotation, thus accounting for the peaks of activity seen during both first and second Double Support phases when pelvis and shoulder girdle are at their maximum excursions in the transverse plane. Rectus abdominis may work as a neutralizing synergist to prevent unwanted motion. Because erector spinae is an extensor, side flexor and rotator of the trunk, neutralizing synergy will be required to maintain the trunk in the upright position and this could account for the observation that peaks of erector spinae and rectus abdominis activity tended to occur close to the same points in the gait cycle. The ipsilateral rectus abdominis may also work as a conjoint synergist with the ipsilateral erector spinae to translate the upper body in the coronal plane over the weight-bearing limb at Mid Stance, seen during left Stance. Rectus abdominis may also work conjointly with erector spinae, like guy wires on a tent, to maintain the trunk at the correct inclination during locomotion, possibly accounting for the fact that peaks of erector spinae and rectus abdominis activity tended to occur close to the same points in the gait cycle.

APPLICATION

This study was carried out to measure and compare duration, cadence, temporal events, body segment displacement in the sagittal plane and myoelectric activity during a Right Gait Cycle in five different types of locomotion. From these data it was hoped to be able to clarify the role of erector spinae and rectus abdominis during the types of locomotion commonly required in the workplace. It was found that the two climbing types of locomotion were similar in duration of cycle, cadence, body segment displacements, amount of myoelectric activity, and phasic activity of trunk muscles. Likewise, the two descending types of locomotion were similar except for the pattern of hip motion. The characteristics of Level Walking lay on a continuum between the climbing types of locomotion on the one hand, and the descending types on the other.

Because a linear relationship has been found between myoelectric activity in paraspinal muscles and intradiscal pressure (Ortengren et al., 1981), thus giving an indirect measurement of back load, it can be concluded from this study that Stair Climbing and Ramp Climbing cause greater loading of the intervertebral disc a than do Level Walking, Stair Descent and Ramp Descent. Therefore, jobs that require large amounts of stair or ramp climbing may be considered to place higher demands on back muscles than would jobs that require only walking on a level surface. Therefore in job analysis, it is recommended that the percentage of time spent in specific types of locomotion be measured in order to determine an index of back muscle load for a particular job. Materials handling jobs involve carrying objects of various weights and dimensions from one location to another. It is recommended that the type of locomotion involved be taken into consideration as well as the characteristics of the load. A light object carried up a ramp may give rise to the same load on the intervertebral disc as a heavier object carried across a level surface. Further research must be conducted to study the kinematics and muscle activity in the various types of locomotion while carrying objects of different weights and dimensions.

The normal data derived from this study can be used to compare data from investigations of body segment motion and trunk muscle activity in back injured workers during different types of locomotion. Relative comparisons of total myoelectric activity can be made by comparing the difference in the EMG signal of an individual muscle between different types of locomotion. It would be expected that differences in trunk and pelvic segment excursion, total myoelectric activity and phasic muscle activity might be found between normals and back injured subjects. These differences might have implications for rehabilitation of the injured worker with respect to exercise and job conditioning programs. The locomotor demands of the injured worker's job should be built into the rehabilitation program. If, for example, the worker's job has a high frequency of stair walking, the goals of the rehabilitation program should include increasing the strength of the back muscles, increasing the mobility of the trunk, pelvis and hip, and inclusion of stair walking in work simulation and work conditioning.

Knowledge of body segment excursion and of trunk muscle activity during different types of locomotion can also be used to modify the workplace in order to enable an injured worker to return to his/her job as soon as possible. For example, from the results of this study it can be recommended that stair and ramp climbing be kept to a minimum for workers who have sustained injury to back muscles. However, pain in low-back injury may be the result of ligament or joint capsule irritation and, in this case, extremes of trunk and pelvic motion may have to be avoided. If anterior inclination of the trunk and/or pelvis must be restricted, prolonged or repeated climbing of stairs or ramps may have to be decreased or eliminated from a job. If the posterior inclination of trunk and/or pelvis must be restricted, stair descent in the workplace would be preferable to ramp descent. If the total amount of trunk excursion must be restricted, stair and ramp walking may have to be decreased or eliminated from a job. If the total amount of pelvic excursion must be restricted, a decrease in the amount of level walking would be required and, if climbing activities are necessary components of a job, stair climbing and descending would be preferable to ramp climbing and descending.

Results from this study might also be used in prevention of low-back injury. The load on the back is increased during stair and ramp climbing and this must be taken into consideration when establishing recommended limits for weight and dimensions of objects to be carried in a particular job. There is no apparent difference between stairs and ramps with respect to demands upon the trunk musculature, however stability of footing, equilibrium of the total body, ability of the worker to see the placement of his/her feet, and location of the centre of gravity of the body over the points of support must all be taken into consideration when making recommendations for modifications of a specific job to prevent back injury.

The role of the trunk muscles in locomotion is complex. Simultaneous three-dimensional kinematic, kinetic and EMG studies need to be carried out to clarify body segment motion and trunk muscle activity in normals. In addition, studies of persons with low-back injury should be conducted to determine if differences in locomotor function exist as a result of back injury. Studies of the effects of changes in ramp inclination and stair dimensions on body segment displacement and trunk muscle activity also need to be conducted.

To date, the lower limb has been the focus of the majority of gait studies, but the upper body would appear to play a significant role in human locomotion. Given the prevalence and economic cost of back injury, clarification of the role of trunk muscles in normal functional motion, including locomotion, is essential. Knowledge of the normal should result in adaptation of the environment in order to assist disabled individuals to return to work and also to prevent injuries that might have long-term social and economic impact.

CHAPTER VI

SUMMARY

The incidence of back injury is a significant problem in heavy industry, particularly in jobs that require frequent bending, twisting, lifting and carrying. Walking on the level as well as climbing and descending stairs and ramps are components of many jobs, however little is known about the load on the back produced by these types of locomotion. Therefore, the general purposes of this study were to measure and compare displacement of the upper body segments in the sagittal plane, and the amount and pattern of myoelectric activity of erector spinae and rectus abdominis during level walking, stair climbing and descent, and ramp climbing and descent.

Electromyography was used to analyze the activity of erector spinae and rectus abdominis in 18 normal male subjects. Foot switches were used to identify the temporal events in the gait cycle, and simultaneous high speed cinefilm was used to record the displacement of trunk and pelvic segments and hip angle in the sagittal plane. Each subject was assigned a random sequence of locomotion types and each type was performed three times.

EMG data were processed to give a linear envelope configuration. The processed EMG signal, footswitch signals and a film synchronization signal were A/D converted and stored on magnetic disc. Using custom-designed software, temporal events were defined and the EMG signals were then related to these events. Using a photo-optical analyzer and digitizing tablet, body segment co-ordinates were extracted from the cinefilm and stored on a magnetic disc. The raw co-ordinate data were subsequently filtered and converted to give trunk and pelvic inclination values relative to the vertical and hip angle data.

All temporal and displacement data were related to the right gait cycle, and the average of the total electrical activity of each muscle over a right gait cycle was calculated and expressed as a percentage of a maximum voluntary contraction of that muscle. For each parameter comparisons were made between locomotion types using a repeated measures analysis of variance.

Slight differences in timing of the principal phases of the gait cycle were observed between each type of locomotion. Significant differences in duration of cycle, cadence, body segment displacement and excursion, and the amount and pattern of myoelectric activity were evident, with climbing and descending at opposite ends of a continuum and level walking in the middle.

Duration of cycle was longer in the climbing than in the descending locomotion types. Cadence was higher in Stair and Ramp descent, lower in Ramp Climbing. Anterior inclination of trunk and pelvic segments was significantly greater in Stair and Ramp climbing than in descending. Maximum hip flexion occurred in the climbing locomotion types while maximum hip extension occurred in Level Walking and Ramp Climbing; on average the hip did not move into extension during Stair and Ramp Descent. Total excursion of the trunk segment was small, but was largest during Ramp Climbing. Total excursion of the pelvic segment was greater than that of the trunk segment and maximum excursion occurred during Level Walking. Maximum excursion of the hip occurred during Ramp Climbing. The myoelectric activity of erector spinae was significantly higher during the climbing locomotion types and lowest during the descending types. The myoelectric activity of rectus abdominis was low in all types of locomotion; only in Ramp Descent was it higher than in the other types. Triphasic myoelectric activity was observed in both erector spinae and rectus abdominis. On average, the maximum peak of erector spinae activity occurred during the second Double Support phase of the gait cycle at the time of ipsilateral End of Weight Bearing and contralateral Initial Contact. Besides motion in the sagittal plane, which was investigated in this research, this pattern of erector spinae activity may be related to control of trunk motion in the coronal and transverse planes. The timing of the peak activity of rectus abdominis was closely related to that of erector spinae. The maximum peak activity of rectus abdominis was less consistent than that of erector spinae, but this muscle appears to have multiple synergistic functions which may contribute to trunk control, co-ordination and equilibrium.

Because myoelectric activity has been shown to be an indirect measure of back load, it can be concluded that Stair and Ramp Climbing give rise to greater loads on the spinal mechanism than do Stair and Ramp Descent or Level Walking. Therefore, jobs that involve a high frequency of climbing activities may place higher demands on the back than would jobs that involve only walking on a level surface. It is recommended that the locomotor demands of a particular job be analyzed in order to assist back injured workers to return to full or modified employment as soon as possible after injury. Knowledge of the physical demands of the locomotor components of a job can also be used to prevent back injuries in the workplace.

BIBLIOGRAPHY

ADAMS, M.A., HUTTON, W.C. STOTT, J.R.R. (1980). The resistance to flexion of the lumbar intervertebral joint. Spine 5: 245-253.

ADAMS, M.A. and HUTTON, W.C. (1985). The effect of posture on the lumbar spine. J. Bone Joint Surg. 67B: 625-629.

AMERICAN ACADEMY OF ORTHOPAEDIC SURGEONS (1965). Joint Motion: Method of Measuring and Recording. Chicago: American Academy of Orthopaedic Surgeons.

ANDERSON, J.A.D. and SWEETMAN, B.J. (1975). A combined flexi-rule/hydrogoniometer for measurement of lumbar spine and its sagittal movement. Rheumatol. Rehabil. 14: 173-179.

ANDERSSON, G.B.J. (1981). Epidemiologic aspects of low-back pain in industry. Spine 6: 53-60.

ANDERSSON, G.B.J. (1983). The biomechanics of the posterior elements of the lumbar spine: introductory comments. Spine 8: 326.

ANDERSSON, G.B.J. and ORTENGREN, R. (1974). Myoelectric back muscle activity during sitting. Scand. J. Rehabil. Med., Suppl. 3: 73-90.

ANDERSSON, G.B.J. and ORTENGREN, R. (1984). Assessment of back load in assembly-line work using EMG. Ergonomics 27: 1157-1168.

ANDERSSON, G.B.J., ORTENGREN, R., HERBERTS, P. (1977). Quantitative electromyographic studies of back muscle activity related to posture and loading. Orthop. Clin. North Am. 8: 85-96.

ANDERSSON, G.B.J., ORTENGREN, R., NACHEMSON, A. (1976). Quantitative studies of back loads in lifting. Spine 1: 178-185.

ANDRIACCHI, T.P., ANDERSSON, G.B.J., FERMIER, R.W., STERN, D., GALANTE, J.O. (1980). A study of lower-limb mechanics during stair-climbing. J. Bone Joint Surg. 62A: 749-757.

ANDRIACCHI, T.P., HAMPTON, S.J., SCHULTZ, A.B., GALANTE, J.O. (1979). Three-dimensional co-ordinate data processing in human motion analysis. J. Biomech. Eng. 101: 279-283.

ASMUSSEN, E. and KLAUSEN, K. (1962). Form and function of the erect human spine. Clin. Orthop. 25: 55-63.

BARTELINK, D.L. (1957). The role of abdominal pressure in relieving the pressure on the lumbar intervertebral discs. J. Bone Joint Surg. 39B: 718-725.

BASMAJIAN, J.V. (1973). Electrodes and electrode connectors. In: New Developments in Electromyography and Clinical Neurophysiology, Vol. 1, J.E. Desmedt (ed.). Basel: Karger.

BASMAJIAN, J.V. (1978). Muscles Alive (4th ed.). Baltimore: Williams & Wilkins.

BASMAJIAN, J.V. (ed.) (1980). Grant's Method of Anatomy (10th ed.). Baltimore: Williams & Wilkins.

BASMAJIAN, J.V., CLIFFORD, H.C., MCLEOD, W.D., NUNNALY, A.N. (1975). Signal processing. In: Computers in Electromyography. London: Butterworths.

BASMAJIAN, J.V. and DE LUCA, C.J. (1985). Muscles Alive (5th ed.). Baltimore: Williams & Wilkins.

BATTYE, C.K., JOSEPH, J. (1966). An investigation by telemetering of the activity of some muscles in walking. Med. Biol. Eng. 4: 125-134.

BEARN, J.G. (1961). The significance of the activity of the abdominal muscles in weight lifting. Acta Anat. 45: 83-89.

BEEVOR, C.E. (1903). Croonian lectures on muscular movements and their representation in the central nervous system. Lancet 1: 1715-1724.

BEJJANI, F.J., GROSS, C.M., PUGH, J.W. (1984). Model for static lifting: relationship of loads on the spine and the knee. J. Biomech. 17: 281-286.

BENDIX, T., KROHN, L., JESSEN, F., AARAS, A. (1985). Trunk posture and trapezius muscle load while working in standing, supported-standing and sitting positions. Spine 10: 433-439.

BENDIX, T., SORENSON, S.S., KLAUSEN, K. (1984). Lumbar curve, trunk muscles and line of gravity with different heel heights. Spine 9: 223-227.

BENN, R.T. and WOOD, P.H.N. (1975). Pain in the back. Rheumatol. Rehabil. 14: 121-128.

BERKSON, M., SCHULTZ, A., NACHEMSON, A., ANDERSSON, G. (1977). Voluntary strengths of male adults with acute low back syndromes. Clin. Orthop. 129: 84-95.

BIGLAND-RITCHIE, B., JOHANSSON, R., LIPPOLD, O.C.J., WOODS, J.J. (1983) Contractile speed and EMG changes during fatigue of sustained maximal voluntary contractions. J. Neurophysiol. 50: 313-324.

BIGOS, S.J., SPENGLER, D.M., MARTIN, N.A., ZEH, J., FISHER, L., NACHEMSON, A., WANG, M.H. (1986). Back injuries in industry: a retrospective study II. Injury factors. Spine 11: 246-251.

BLACKBURN, S.E. and PORTNEY, L.G. (1981). Electromyographic activity of back musculature during Williams' flexion exercise. Phys. Ther. 61: 878-885.

BOBBERT, A.C. (1960). Energy expenditure in level and grade walking. J. Applied Physiol. 15: 1015-1021.

BOGARDH, E. and RICHARDS, C.L. (1981). Gait analysis and relearning of gait control in hemiplegic patients. Physiotherapy Canada 33: 223-230.

BOGDUK, N. (1983). The innervation of the lumbar spine. Spine 8: 286-293.

BOGDUK, N. and MACINTOSH, J.E. (1984). The applied anatomy of the thoracolumbar fascia. Spine 9: 164-170.

BOOTH, D., CHENNELLS, M., JONES, D., PRICE, A. (1980). Assessment of abdominal muscle exercises in non-pregnant, pregnant and postpartum subjects using electromyography. Aust. J. Physiother. 26: 177-197.

BRANDELL, B.R. (1977). Functional roles of the calf and vastus muscles in locomotion. Am. J. Phys. Med. 56: 59-74.

BROWN, D.M., DE BACHER, G.A., BASMAJIAN, J.V. (1979). Feedback goniometers for hand rehabilitation. Am. J. Occup. Ther. 33: 458-463.

BRUCE, F.M., FLOYD, W.F., WARD, J.S. (1967). Oxygen consumption and heart rate during stair climbing. J. Physiol. 191: 90-92.

CAILLIET, R. (1981). <u>Low Back Pain Syndrome</u> (3rd ed.). Philadelphia: F.A. Davis.

CAMPBELL, J.M. (1952). An electromyographic study of the role of the abdominal muscles in breathing. J. Physiol. 117: 222-233.

CAMPBELL, E. and GREEN, J. (1953). The respiratory function of the abdominal muscles in man. An EMG study. J. Physiol. 120: 409-418.

CAPPOZZO, A. (1981). Analysis of the linear displacement of the head and trunk during walking at different speeds. J. Biomech. 14: 411-425.

CAPPOZZO, A. (1983). The forces and couples in the human trunk during level walking. J. Biomech. 16: 265-277.

CAPPOZZO, A., FIGURA, F., LEO, T., MARCHETTI, M. (1978). Movements and mechanical energy changes of the upper part of the human body during walking. In: Biomechanics VI A, E. Asmussen and K. Jorgensen (eds.). Baltimore: University Park Press.

CAPPOZZO, A., FIGURA, F., MARCHETTI, M. (1976). The interplay of muscular and external forces in human ambulation. J. Biomech. 9: 35-43.

CARLSON, H. and THORSTENSSON, A. (1981). Control of the human trunk during locomotion. Acta Physiol. Scand. 114: 14A.

CHAPMAN, M.W. and KUROKAWA, K.M. (1969). Some observations on the transverse rotations of the human trunk during locomotion. Bull. Prosth. Res. Spring: 38-59.

CHAMPAN, A.E. and TROUP, J.D.G. (1969). The effect of increased maximal strength on the integrated electrical activity of lumbar erectores spinae. Electromyography 9: 263-280.

CHAZAL, J., TANGUY, A., BOURGES, M., GAUREL, G., ESCANDE, G., GUILLET, M., VANNEUVILLE, G. (1985). Biochemical properties of spinal ligaments and a histological study of the supraspinal ligament in traction. J. Biomech. 18: 167-176.

CLEMENTE, C. (ed.) (1985). Gray's Anatomy. Philadelphia: Lea & Febiger.

CLOSE, J.R., NICKLE, E.D., TODD, F.N. (1960). Motor-unit action potential counts: Their significance in isometric and isotonic contractions. J. Bone Joint Surg. 42A: 1207-1222.

COATES, J.E. and MEADE, F. (1961). The energy expenditure and mechanical energy demand in walking. Ergonomics 4: 97-117.

CONRAD, B., BENECKE, R., CARNEHL, J., HOHNE, J., MEINCK, H.M. (1983). Pathophysiological aspects of human locomotion. In: Advances in Neurology (Vol.39) Motor Control Mechanisms in Health and Disease. New York: Raven Press.

COOPER, J.E. (1982). An electromyographic and electrogoniometric analysis of the role of the lateral abdominal musculature and the erector spinae muscle during thoracolumbar rotation. M.Sc. Thesis, University of Manitoba, Winipeg, Canada.

CORLETT, E.N., HUTCHESON, C., DELUGAN, M.A., ROGOZENSKI, J. (1972). Ramps and stairs: the choice using physiological and biomechanical criteria. Applied Ergonomics 3: 195-201.

CRELIN, E.S. (1981). Development of the musculoskeletal system. Clinical Symposia 33: 1-36.

CURRIER, D.P. (1984). <u>Elements of Research in Physical Therapy</u> (2nd ed.). Baltimore: Williams & Wilkins.

DANIELS, L. and WORTHINGHAM, C. (1980). <u>Muscle Testing: Techniques of Manual Examination</u>. Toronto: W.B. Saunders.

DAVIS, P.R. and TROUP, J.D.G. (1964). Pressures in the trunk cavities when pulling, pushing and lifting. Ergonomics 7: 465-474.

DAVIS, P.R., TROUP, J.D.G., BURNARD, J.H. (1965). Movements of the thoracic and lumbar spine when lifting: a chronocyclophotographic study. J. Anat. 99: 13-26.

DEAN, G.A. ((1965). An analysis of the energy expenditure in level and grade walking. Ergonomics 8: 31-47.

DE SOUSA, O.M. and FURLANI, J. (1974). Electromyographic study of the m. rectus abdominis. Acta Anat. 88: 281-298.

DIETZ, V., SCHMIDT BLEICHER, D., NOTH, J. (1979). Neuronal mechanism of human locomotion. J. Neurophysiol. 42: 1212-1222.

DRILLIS, R.J. (1958). Objective recording and biomechanics of pathological gait. Ann. N. Y. Acad. Sci. 74: 86-109.

DUBO, H.I.C., PEAT, M., WINTER, D.A., QUANBURY, A.O., HOBSON, D.A., STEINKE, T., REIMER, G. (1976). Electromyographic temporal analysis of gait: normal human locomotion. Arch. Phys. Med. Rehabil. 57: 415-420.

EBERHART, H. and INMAN, V.T. (1951). An evaluation of experimental procedures used in a fundamental study of human locomotion. Ann. N.Y. Acad. Sci. 51: 1213-1228.

EKHOLM, J., ARBORELIUS, U.P., NEMETH, G. (1982). The load on the lumbo-sacral joint and trunk muscle activity during lifting. Ergonomics 25: 145-161.

ESCH, D. and LEPLEY, M. (1971). Evaluation of Joint Motion: Methods of Measurement and Recording. Minneapolis: University of Minnesota Press.

FARFAN, H.F. (1973). Movements of the lumbar spine. In: Mechanical Disorders of the Low Back. Philadelphia: Lea & Febiger.

FARFAN, H.F. (1975). Muscular mechanism of the lumbar spine and the position of power and efficiency. Orthop. Clin. North Am. 6: 135-144.

FARFAN, H.F. (1978). The biomechanical advantage of lordosis and hip extension for upright activity. Man as compared with other anthropoids. Spine 3: 336-342.

FITCH, J.M., TEMPLER, J., CORCORAN, P. (1974). The dimensions of stairs. Scientific American 231: 82-90.

FLECK, H. (1962). Action potentials from single motor units in human muscle. Arch. Phys. Med. Rehabil. 43: 99-107.

FLINT, M.M. and GUDGELL, J. (1965). Electromyographic study of abdominal muscular activity during exercise. Res. Quart. 36: 29-37.

FLOYD, W.F. and SILVER, P.H.S. (1950). Electromyographic study of patterns of activity of the anterior abdominal wall muscles in man. J. Anat. 84: 132-145.

FLOYD, W.F. and SILVER, P.H.S. (1951). Function of erectores spinae in flexion of the trunk. Lancet, Jan.20: 133-134.

FLOYD, W.F. and SILVER, P.H.S. (1955). The function of the erectores spinae muscles in certain movements and postures in man. J. Physiol. 129: 184-203.

FRANKEL, V.H. and NORDIN, M. (1980). Basic Biomechanics of the Skeletal System. Philadelphia: Lea & Febiger.

FRYMOYER, J.W., POPE, M.H., COSTANZA, M.C., ROSEN, J.C., GOGGIN, J.E., WILDER, D.G. (1980). Epidemiologic studies of low-back pain. Spine 5: 419-423.

GALANTE, J.O. (1967). Tensile properties of human lumbar annulus fibrosus. Acta Orthop. Scand. (Suppl.) 100: 1-91.

GANS, C and GORNIAK, G.C. (1980). Electromyograms are repeatable: Precautions and limitations. Science 210: 795-797.

GHADIALLY, F.N. (1978). Fine Structure of the joints. In: The Joints and Synovial Fluid, Vol.I. L. Sokoloff, Ed. New York: Academic Press.

GOLDING, J.S.R. (1952). Electromyography of the erector spinae in low back pain. Postgrad. Med. J. 28: 401.

GORDON, A.M. (1982). Muscle. In: Physiology and Biophysics IV: Excitable Tissues and Reflex Control of Muscle. T.Ruch & H.D. Patton (eds.). Philadelphia: W.B. Saunders.

GOWITZKE, B.A. and MILNER, M. (1980). Understanding the Scientific Bases of Human Movement (2nd ed.). Baltimore: Williams & Wilkins.

GRACOVETSKY, S. (1985). An hypothesis for the role of the spine in human locomotion: a challenge to current thinking. J. Biomech. 7: 205-216.

GRACOVETSKY, S., FARFAN, H.F., LAMY, C. (1977). A mathematical model of the lumbar spine using an optimized system to control muscles and ligaments. Orthop. Clin. North Am. 8: 135-153.

GRACOVETSKY, S., FARFAN, H.F., LAMY, C. (1981). The mechanism of the lumbar spine. Spine 6: 249-262.

GRAY, E.G., BASMAJIAN, J.V. (1968). Electromyography and cinematography of leg and foot ('normal' and flat) during walking. Anat. Rec. 161: 1-16.

GREGERSEN, G.G. and LUCAS, D.B. (1967). An in vivo study of axial rotation of the human thoraco-lumbar spine. J. Bone Joint Surg. 49A: 247-262.

GREW, N.D. (1980). Intra-abdominal pressure response to load applied to the torso in normal subjects. Spine 5: 149-154.

GRIEVE, D.W. (1976). Electromyography. In: Techniques for the Analysis of Human Movement, D.W. Grieve, D.I. Miller, D. Mitchelson, J.P. Paul, A.J. Smith (eds.). Princeton, NJ: Princeton Book Co.

GRIEVE, D.W. and GEAR, R.J. (1966). The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. Ergonomics 5: 379-399.

GRILLNER, S. (1981). Control of locomotion in bipeds, tetrapods and fish. In: Handbook of Physiology, Section 1. The Nervous System, Vol. II Motor Control, Part 2. Bethesda: American Pysiological Society.

GROSSMAN, W.I. and WEINER, H. (1966). Some factors affecting the reliability of surface electromyography. Psychosom. Med. 28: 78-83.

GUTH, V., ABBINK, F., THEYSOHN, H. (1979). Electromyographic investigations on gait. Electromyogr. Clin. Neurophysiol. 19: 305-323.

HALBERTSMA, J.M. and DE BOER, R.R. (1981). On the processing of electromyograms for computer analysis. J. Biomech. 14: 431-435.

HAMILTON, W.J. and MOSSMAN, H.W. (1972). <u>Hamilton, Boyd and Mossman's Human Embryology</u> (4th ed.). Baltimore: Williams & Wilkins.

HATAMI, T. (1961a). Electromyographic studies of influence of pregnancy on activity of the abdominal wall muscles: I. Tohoku J. Exp. Med. 75: 71-80.

HATAMI, T. (1961b). Electromyographic studies of influence of pregnancy on activity of the abdominal wall muscles: II. Tohoku J. Exp. Med. 75: 81-88.

HATZE, H. (1974). The meaning of the term "biomechanics". J. Biomech. 7: 189-190.

HAYES, K., NORMAN, R., WINTER, D. (undated). Kinesiological electromyography. Dept. of Kinesiology, University of Waterloo.

HEMBORG, B., MORITZ, U., LOWING, H. (1985). Intra-abdominal pressure and trunk muscle activity during lifting, IV. The causal factors of the intra-abdominal pressure rise. Scand. J. Rehab. Med. 17: 25-38.

HIRSCH, C. (1955). The reaction of intervertebral discs to compression forces. J. Bone Joint Surg. 37A: 1188-1191.

HIROSE, K., UONO. M., SOBUE, I. (1974). Quantitative EMG comparison between manual values and computer ones on normal subjects. Electromyogr. Clin. Neurophysiol. 14: 315-320.

HOLLINSHEAD, W.H. (1976). Functional Anatomy of the Limbs and Back (4th ed.). Philadelphia: W.B. Saunders.

INMAN, V.T., RALSTON, H.J., TODD, F. (1981). Human Walking. Baltimore: Williams & Wilkins.

INMAN, V.T., SUNDERS, J.B.C.M., ABBOTT, L.C. (1944). Observations on the function of the shoulder joint. J. Bone Joint Surg. 26A: 1-30.

JAMES, W.V. and ORR, J. (1982). A clinical electrogoniometry system. Eng. Med. 11: 123-124.

JOHNSTON, R.C., SMIDT, G.L. (1969). Measurement of hip-joint motion during walking. J. Bone Joint Surg. 51A: 1083-1094.

JONSSON, B. (1968). Electrode problems in electromyographic kinesiology. Electromyography (Supp. 1): 27-29.

JONSSON, B. (1970). The functions of individual muscles in the lumbar part of the spinae muscle. Electromyography 10: 5-21.

JONSSON, B. and KOMI, P.V. (1973). Reproducibility problems when using wire electrodes in electromyographic kinesiology. In: New Developments in Electromyography and Clinical Neurophysiology, Vol. 1. J.E. Desmedt (ed.). Basel: Karger.

JOSEPH, J. and MCCOLL, I. (1961). Electromyography of muscles of posture: posterior vertebral muscles in man. J. Physiol. 157: 33-37.

JOSEPH, J. and WATSON, R. (1967). Telemetering electromyography of muscles used in walking up and down stairs. J. Bone Joint Surg. 49B: 774-780.

KAPANDJI, I.A. (1974). The Physiology of the Joints, Vol.3 London: Churchill Livingstone.

KAZARIAN, L.E. (1975). Creep characteristics of the human spinal column. Orthop. Clin. N. Am. 6: 3-18.

KELLEY, D.L. (1971). Kinesiology: Fundamentals of Motion Description. Englewood Cliffs, NJ: Prentice-Hall.

KIPPERS, V. and PARKER, A.W. (1984). Posture related to myoelectric silence of erectores spinae during trunk flexion. Spine 9: 740-745.

KIRTLEY, C., WHITTLE, M.W., JEFFERSON, R.J. (1985). Influence of walking speed on gait parameters. J. Biomed. Eng. 7: 282-288.

KLAUSEN, K. (1965). The form and function of the loaded human spine. Acta Physiol. Scand. 65: 176-190.

KOMI, P.V. and BUSKIRK, E.R. (1970). Reproducibility of electromyographic measurements with inserted wire electrodes and surface electrodes. Electromyography 10: 357-368.

KORKALA, O., GRONBLAD, M., LIESI, P., KARAHARJU, E. (1985). Immunohistochemical demonstration of nociceptors in the ligamentous structures of the lumbar spine. Spine 10: 156-157.

KRYWULAK, W. (1983). Personal Communication.

KUMAR, S. and DAVIS, P.R. (1983). Spinal loading in static and dynamic postures: EMG and intra-abdominal pressure study. Ergonomics 26: 913-922.

LAMOREUX, L.W. (1971). Kinematic measurements in the study of human walking. Bull. Prosthet. Res. BPR: 10-15.

LANGEBARTEL, D.A. (1977). The Anatomical Primer. Baltimore: University Park Press.

LANGRANA, N.A., LEE, C.K., ALEXANDER, H., MAYOTT, C.W. (1984). Quantitative assessment of back strength using isokinetic testing. Spine 9: 287-293.

LANSING, R.W. and MYERINK, L. (1981). Load compensating responses of human abdominal muscles. J. Physiol. (Lond.) 320: 253-268.

LAST, R.J. (1978). Anatomy: Regional and Applied (6th ed.). London: Churchill Livingstone.

LETTS, R.M., QUANBURY, A.O., VOJNIC, C.D., MONSON, R. (1978). Paraspinal muscle activity during sports. Sports Medicine 6: 80-90.

LETTS, R.M., WINTER, D.A., QUANBURY, A.O. (1975). Locomotion studies as an aid in clinical assessment of childhood gait. CMA Journal 112: 1091-1094.

LEVEAU, B. (1977). Williams and Lissner: Biomechanics of Human Motion. Toronto: W.B. Saunders.

LIBERSON, W.T. (1965). Biomechanics of gait: A method of study. Arch. Phys. Med. 46: 37-48.

LIN, H.S., LIU, Y.K., ADAMS, K.H. (1978). Mechanical response of the lumbar intervertebral joint under physiologic (complex) loading. J. Bone Joint Surg. 60A: 41-55.

LINDH, M. (1980). Biomechanics of the lumbar spine. In: Basic Biomechanics of the Skeletal System. V.H. Frankel and M. Nordin (eds.). Philadelphia: Lea & Febiger.

LINTON, M. and GALLO, P.S. (1975). The Practical Statistician: Simplified Handbook of Statistics. Monterey, CA: Brooks/Cole.

LIPETZ, S. and GUTIN, B. (1970). Electromyographic study of four abdominal exercises. Med. Sci. Sports 2: 35-38.

LOCKHART, R.D., HAMILTON, G.F., FYFE, F.W. (1965). Anatomy of the Human Body (2nd ed.). London: Faber & Faber.

LOEBL, W.Y. (1967). Measurement of spinal posture and range of spinal movement. Ann. Phys. Med. 9: 103-110.

LORENZ, M., PATWARDHAN, A., VANDERBY, R. (1983). Load-bearing characteristics of lumbar facets in normal and surgically altered spinal segments. Spine 8: 122-130.

LYONS, K., PERRY, J., GRONLEY, J.K., BARNES, L., ANTONELLI, D. (1983). Timing and relative intensity of hip extensor and abductor muscle action during level and stair ambulation. Phys. Ther. 63: 1597-1605.

LUMSDEN, R.M. and MORRIS, J.M. (1968). An <u>in vivo</u> study of axial rotation and immobilization at the lumbosacral joint. J. Bone Joint Surg. 50A: 1591-1602.

MACCONAILL, M.A. and BASMAJIAN, J.V. (1977). <u>Muscles and Movements</u>. Huntington, NY: R.E. Kreiger.

MAGORA, A. and GONEN, B. (1978). Computer editing of electromyographic recordings. Electromyogr. Clin. Neurophysiol. 18: 35-43.

MAIRIAUX, P.H., DAVIS, P.R., STUBBS, D.A., BATY, D. (1984). Relation between intra-abdominal pressure and lumbar moments when lifting weights in the erect posture. Ergonomics 27: 883-894.

MARKOLF, K.L. (1972). Deformation of the thoracolumbar intervertebral joints in response to external loads: biomechanical study using autopsy material. J. Bone Joint Surg. 54A: 511-533.

MARRAS, W.S., KING, A.I., JOYNT, R.L. (1984). Measurements of loads on the lumbar spine under isometric and isokinetic conditions. Spine 9: 176-187.

MAYER, T.G., SMITH, S., KONDRASKE, G., GATCHEL, R.J., CARMICHAEL, T.W., MOONEY, V. (1985). Quantification of lumbar function: Part 3: Preliminary data on isokinetic torso rotation testing with myoelectric spectral analysis in normal and low back pain subjects. Spine 10: 912-920.

MAYER, T.G., TENCER, A.F., KRISTOFERSON, S., MOONEY, V. (1984). The use of noninvasive techniques for quantification of spinal range-of-motion in normal subjects and chronic low-back dysfunction patients. Spine 9: 588-595.

MAYHEW, T.P., NORTON, B.J., SAHRMANN, S.A. (1983). Electromyographic study of the relationship between hamstring and abdominal muscles during a unilateral straight leg raise. Phys. Ther. 63: 1769-1773.

MCLEOD, W.D. (1973). EMG instrumentation in biomechanical studies: Amplifiers, recorders and integrators. In: New Developments in Electromyography and Clinical Neurophysiology, Vol. 1. J.E. Desmedt (ed.). Basel: Karger.

MILLER, D.J. (1985). Comparison of electromyographic activity in the lumbar paraspinal muscles of subjects with and without chronic low back pain. Phys. Ther. 65: 1347-1354.

MILNER, M., BASMAJIAN, J.V., QUANBURY, A.O. (1971). Multifactorial analysis of walking by electromyography and computer. Am. J. Phys. Med. 50: 235-258.

MIZRAHI, J., SUSAK, Z., HELLER, L., NAJENSON, T. (1982). Variation of time-distance parameters of the stride as related to clinical gait improvement in hemiplegics. Scand. J. Rehabil. Med. 14: 133-140.

MOORE, K.L. (1982). <u>The Developing Human</u> (3rd ed.). Toronto: W.B. Saunders.

MORRIS, J.M., BENNER, G., LUCAS, D.B. (1962). An electromyographic study of the intrinsic muscles of the back in man. J. Anat. (London) 96: 509-520.

MORRIS, J.M., LUCAS, D.B., BRESLER, B. (1961). Role of the trunk in stability of the spine. J. Bone Joint Surg. 43A: 327-351.

MURRAY, M.P. (1967). Gait as a total pattern of movement. Am. J. Phys. Med. 46: 290-333.

MURRAY, M.P., DROUGHT, A.B., KORY, R.C. (1964). Walking patterns of normal men. J. Bone Joint Surg. 46A: 335-360.

MURRAY, M.P., KORY, R.C., CLARKSON, B.H. (1969). Walking patterns in healthy old men. J. Gerontol. 24: 169-178.

MURRAY, M.P., KORY, R.C., CLARKSON, B.H., SEPIC, S.B. (1966). Comparison of free and fast speed walking patterns of normal men. Am.J. Phys. Med. 45: 8-24.

NACHEMSON, A. (1978). Quantitative studies on lumbar spine loads: implications for the scientist and the clinician. In: <u>Biomechanics VI B</u>, E. Asmussen and K. Jorgensen (eds.). Baltimore: University Perk Press.

NACHEMSON, A. (1981). Disc pressure measurements. Spine 6: 93-97.

NACHEMSON, A. and ANDERSSON, G.B.J. (1982). Classification of low-back pain. Scand. J. Work Environ. Health 8: 134-136.

NACHEMSON, A. and MORRIS, J.M. (1964). <u>In vivo</u> measurements of intradiscal pressure. Discometry, a method for the determination of pressure in the lower lumbar discs. J. Bone Joint Surg. 46A: 1077-1092.

NILSSON, J., THORSTENSSON, A., HALBERTSMA, J. (1985). Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans. Acta Physiol Scand. 123: 457-475.

NORDIN, M., ORTENGREN, R., ANDERSSON, G.B.J. (1984). Measurements of trunk movements during work. Spine 9: 465-469.

O'CONNELL, A.L. and GARDNER, E. (1963). The use of electromyography in kinesiological research. Res. Quart. 34: 166-184.

ONO, K. (1958). Electromyographic studies of the abdominal wall muscles in visceroptosis. I. Analysis of patterns of activity of the abdominal wall muscles in normal adults. Tohoku J. Exper. Med. 68: 347-354.

ORTENGREN, R. and ANDERSSON, G.B.J. (1977). Electromyographic studies of trunk muscles, with special reference to the functional anatomy of the lumbar spine. Spine 2: 44-52.

ORTENGREN, R., ANDERSSON, G.B.J., NACHEMSON, A.L. (1981). Studies of relationship between lumbar disc pressure, myoelectric back muscle activity, and intra-abdominal (intragastric) pressure. Spine 6: 98-103.

PANJABI, M.M. (1977). Experimental determination of spinal motion segment behavior. Orthop. Clin. N. Am. 8: 169-180.

PANJABI, M.M., GOEL, V.K., TAKATA, K. (1982). Physiologic strains in the lumbar spinal ligaments. Spine 7: 192-203.

PARE, E.B., STERN, J.T., SCHWARTZ, J.M. (1981). Functional differentiation within the tensor fasciae latae. A telemetered electromyographic analysis of its locomotor roles. J. Bone Joint Surg. 63A: 1457-1471.

PARTRIDGE, M.J. and WALTERS, C.E. (1959). Participation of the abdominal muscles in various movements of the trunk on man. Phys. Ther. Rev. 39: 791-800.

PATWARDHAN, A., VANDERBY, R., KNIGHT, G.W., GOGAN, W.J., LEVINE, P.D. (1985). Biomechanics of the spine. In: Atlas of Orthotics (2nd ed.), W.H. Bunch, R. Keagy, A.E. Kritter, L.M. Kruger, M. Letts, J.E. Lonstein, E.B. Marsolais, J.C. Matthews, L.R. Pedegana (eds.). St. Louis: Mosby.

PAULY, J.E. (1966). An electromyographic analysis of certain movements and exercises I. Some deep muscles of the back. Anat. Rec. 155: 223-234.

PAULY, J.E. and STEELE, R.W. (1966). Electromyographic analysis of back exercises for paraplegic patients. Arch. Phys. Med. 47: 730-736.

PEARCY, M., PORTEK, I., SHEPHERD, J. (1984). Three-dimensional X-ray analysis of normal movement in the lumbar spine. Spine 9: 294-297.

PEARSON, K. (1976). The control of walking. Sci. Am. 235: 72-86.

PEAT, M., GRAHAME, R., FULFORD, R., QUANBURY, A.O. (1976). An electrogoniometer for the measurement of single plane movements. J. Biomech. 9: 423-424.

PERRY, J. (1985). Normal and pathological gait. In: Atlas of Orthotics (2nd ed.), W.H. Bunch, R. Keagy, A.E. Kritter, L.M. Kruger, M. Letts, J.E. Lonstein, E.B. Marsolais, J.C. Matthews, L.R. Pedegana (eds.). St Louis: Mosby.

PERRY, J., EASTERDAY, C.S., ANTONELLI, D.J. (1981). Surface versus intramuscular electrodes for electromyography of superficial and deep muscles. Phys. Ther. 61: 7-15.

PIMENTAL, N.A., SHAPIRO, Y., PANDOLF, K.B. (1982). Comparison of uphill and downhill walking and concentric and eccentric cycling. Ergonomics 25: 373-380.

PORTNOY, H. and MORIN, F. (1956). Electromyographic study of postural muscles in various positions and movements. Amer. J. Physiol. 186: 122-126.

QUANBURY, A.O. (1981). Personal communication.

QUANBURY, A.O. (1985). Personal communication.

RANCHO LOS AMIGOS HOSPITAL (1978). Normal and Pathological Gait Syllabus. Downey, CA: Pathokinesiology Service and Physical Therapy Dept., Rancho los Amigos Hospital.

REID, J.G. (1984). Physical properties of the human trunk as determined by computed tomography. Arch. Phys. Med. Rehabil. 65: 246-250.

RODGERS, M.M. and CAVANAGH, P.R. (1984). Glossary of biomechanical terms, concepts and units. Phys.Ther. 64: 82-98.

ROMANES, G.J. (ed.) (1976). Cunningham's Manual of Anatomy (14th ed.). London: Oxford University Press.

ROSENFALCK, A. (1960). Evaluation of the electromyogram by mean voltage recording. In: Medical Electronics, C.N. Smyth (ed.). London: Iliffe & Sons.

SADLER, T.W. (1985) Langman's Medical Embryology (5th ed.). Baltimore: Williams & Wilkins.

SAUNDERS, J.B. de C.M., INMAN, V.T., EBERHART, M.D. (1953). The major determinants in normal and pathological gait. J. Bone Joint Surg. 35A: 543-558.

SCHULTZ, A.B., ANDERSSON, G.B.J., ORTENGREN, R., BJORK, R., NORDIN, M. (1982). Analysis and quantitative myoelectric measurements of loads on the lumbar spine when holding weights in standing postures. Spine 7: 390-397.

SCHULTZ, A.B., ANDERSSON, G.B.J., ORTENGREN,R., HADERSPECK, K., NACHEMSON, A. (1982). Loads on the lumbar spine: validation of a biomechanical analysis by measurements of intradiscal pressures and myoelectric signals. J. Bone Joint Surg. 64A: 713-720.

SCOTT, A.D. and TROMBLY, C.A. (1983) Evaluation. In: Occupational Therapy for Physical Dysfunction (2nd ed.). Baltimore: Williams & Wilkins.

SENSENIG, E.C. (1949). The early development of the human vertebral column. Contrib. Embryol. 33: 21-41.

SHEFFIELD, F.J. (1962). Electromyographic study of the abdominal muscles in walking and other movements. Am. J. Phys. Med. 41: 142-147.

SIEGLER, S., HILLSTOM, H.J., FREEDMAN, W., MOSKOWITZ, G. (1985). Effect of myoelectric processing on the relationship between muscle force and processed EMG. Am. J. Phys. Med. 64: 130-149.

SMIDT, G.L. (1971). Hip motion and related factors in walking. Phys. Ther. 51: 9-21.

SMITH, A.J. (1975). Photographic analysis of movement. In: Techniques for Analysis of Human Movement. D.W.Greive, D. Mitchelson, J.P. Paul, A.J. Smith (eds.). London: Lepus Books.

SNOOK, S.H. (1983). Back and other musculoskeletal disorders. In: Occupational Health: Recognizing and Preventing Work-Related Disease. B.L. Levy and D.H. Wegman (eds.). Toronto: Little Brown.

SODERBERG, G.L. (1986). Kinesiology: Application to Pathological Motion. Baltimore: Williams and Wilkins.

SODERBERG, G.L. and BARR, J.O. (1983). Muscular function in chronic low-back dysfunction. Spine 8: 79-85.

SODERBERG, G.L. and COOK, T.M. (1984). Electromyography in biomechanics. Phys. Ther. 64: 1813-1820.

SOKAL, R.R. and ROHLF, F.J. (1981). Biometry (2nd ed.). San Francisco: W.H. Freeman.

SPENGLER, D.M., BIGOS, S.J., MARTIN, N.A., ZEH, J., FISHER, L., NACHEMSON, A. (1986). Back injuries in industry: a retrospective study. I. Overview and cost analysis. Spine 11: 241-245.

STOCKWELL, R.A. (1979). Biology of Cartilage Cells. London: Cambridge University Press.

STROHL, K.P., MEAD, J., BANZETT, R.B., LORING, S.H., KOSCH, P.C. (1981). Regional differences in abdominal muscle activity during various maneuvers in humans. J. App. Physiol. 51: 1471-1476.

STUBBS, D.A. (1981). Trunk stresses in construction and other industrial workers. Spine 6: 83-89.

SUTHERLAND, D.H. and HAGY, J.L. (1972). Measurement of gait movements from motion picture film. J. Bone Joint Surg. 54A: 787.

TATA, J.A. (1980). An electrogoniometric analysis of knee joint movement and electromyographic study of the peak activity of the thigh muscles during the stair cycle in normals and Syme amputees. Ph.D. Thesis, University of Manitoba, Winnipeg, Canada.

THORSTENSSON, A., CARLSON, H., ZOMLEFER, M.R., NILSSON, J. (1982). Lumbar back muscle activity in relation to trunk movements during locomotion in man. Acta Physiol. Scand. 116: 13-20.

THORSTENSSON, A., NILSSON, J., CARSON, H., ZOMLEFER, R. (1984). Trunk movements in human locomotion. Acta Physiol. Scand. 121: 9-22.

THORSTENSSON, A., ODDSON, L., CARLSON., H. (1985). Motor control of voluntary trunk movements in standing. Acta Physiol. Scand. 125: 309-321.

THURSTON, A.J. (1985). Spinal and pelvic kinematics in osteoarthritis of the hip joint. Spine 10: 467-471.

THURSTON, A.J. and HARRIS, J.D. (1983). Normal kinematics of the lumbar spine and pelvis. Spine 8: 199-205.

THURSTON, A.J., WHITTLE, M.W., STOKES, I.A.F. (1981). Spinal and pelvic movement during walking - a new method of study. Eng. Med. 10: 219-222.

TICHAUER, E.R. (1971). A pilot study of the biomechanics of lifting in simulated industrial work situations. J. Safety Res. 3: 98-114.

TKACZUK, H. (1968). Tensile properties of human lumbar longitudinal ligaments. Acta Orthop. Scand. (Suppl.) 115.

TOKUHIRO, A., NAGASHIMA, H., TAKECHI, H. (1985). Electromyographic kinesiology of lower extremity muscle during slope walking. Arch. Phys. Med. Rehabil. 66: 610-613.

TOWNSEND, M.A. and TSAI, T.C. (1976). Biomechanics and modeling of bipedal climbing and descending. J. Biomech. 9: 227-239.

TOWNSEND, M.A., LAINHART, S.P., SHIAVI, R., CAYLOR, J. (1978). Variability and biomechanics of synergy patterns of some lower-limb muscles during ascending and descending stairs and level walking. Med. Biol. Eng. Comput. 16: 681-688.

TOWNSEND, M.A., SHIAVI, R., LAINHART, S.P., CAYLOR, J. (1978). Variability in synergy patterns of leg muscles during climbing, descending and level walking of highly-trained athletes and normal males. Electromyogr. Clin. Neurophysiol. 18: 69-80.

TWOMEY, L.T. and TAYLOR, J.R. (1983). Sagittal movements of the human lumbar vertebral column: a quantitative study of the role of the posterior vetebral elements. Arch. Phys. Med. Rehabil. 64: 322-325.

Van Nostrand's Scientific Encyclopedia (4th ed.). (1968). Toronto: Van Nostrand.

WATERLAND, J.C. and SHAMBES, G.M. (1969). Electromyography: One link in the experimental chain of kinesiological research. Phys. Ther. 49: 1351-1356.

WATERS, R.L. and MORRIS, J.M. (1970). Effect of spinal supports on the electrical activity of muscles of the trunk. J. Bone Joint Surg. 52A: 51-60.

WATERS, R.L. and MORRIS, J.M. (1972). Electrical activity of muscles of the trunk during walking. J. Anat. 111: 191-199.

WATERS, R.L., MORRIS, J., PERRY, J. (1973). Translational motion of the head and trunk during normal walking. J. Biomech. 6: 167-172.

WEIS, E.B. (1975). Stresses at the lumbosacral junction. Orthop. Clin. N. Am. 6: 83-91.

WHITE, A.A. and HIRSCH, C. (1971). The significance of the vertebral posterior elements in the mechanics of the thoracic spine. Clin. Orthop. 81: 2-14.

WHITE, A.A. and PANJABI, M.M. (1978). The basic kinematics of the human spine: a review of past and current knowledge. Spine 3: 12-20.

WILLIAMS, P.L. and WARWICK, R. (eds.) (1980). <u>Gray's Anatomy</u> (36th ed.). London: Churchill Livingstone.

WILLISON, R.G. (1963). A method of measuring motor unit activity in human muscle. J. Physiol. (London) 168: 35P-36P.

WINTER, D.A. (1979). <u>Biomechanics of Human Movement</u>. New York: John Wiley & Sons.

WINTER, D.A. (1982). Camera speeds for normal and pathological gait analyses. Med. Biol. Eng. & Comput. 20: 408-412.

WINTER, D.A. (1983). Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. Clin. Orthop. Rel. Res. 175: 147-154.

WINTER, D.A. (1984). Pathologic gait diagnosis with computer-averaged electromyographic profiles. Arch. Phys. Med. Rehabil. 65: 393-398.

WINTER, D.A. (1986). Definitions, terms and conventions related to human gait (Draft #4). University of Waterloo: Dept. of Kinesiology.

WINTER, D.A., GREENLAW, R.K., HOBSON, D.A. (1972a). Television computer analysis of kinematics of human gait. Comput. Biomed. Res. 5: 498-504.

WINTER, D.A., GREENLAW, R.K., HOBSON, D.A. (1972b). A microswitch shoe for use in locomotion studies. J. Biomech. 5: 553-554.

WINTER, D.A., QUANBURY, A.O., REIMER, G.D. (1976). Analysis of instantaneous energy of normal gait. J. Biomech. 9: 253-257.

WOLF, S.L. and BASMAJIAN, J.V. (1978). Assessment of paraspinal electromyographic activity in normal subjects and in chronic back pain patients using a muscle biofeedback device. In: <u>Biomechanics VI B</u>, E. Asmussen and K. Jorgensen (eds.). Baltimore: University Park Press.

WORKERS COMPENSATION BOARD OF MANITOBA (1982). <u>Annual Report</u>. Winnipeg, Manitoba, Canada.

WORKERS COMPENSATION BOARD OF MANITOBA (1985). <u>Annual Report</u>. Winnipeg, Manitoba, Canada.

YACK, H.J. (1984). Techniques for clinical assessment of human movement. Phys. Ther. 64: 1821-1830.

YANG, J.F. and WINTER, D.A. (1983). Electromyography reliability in maximal and submaximal isometric contractions. Arch. Phys. Med. Rehabil. 64: 417-420.

YETTRAM, A.L. and JACKMAN, M.J. (1980). Equilibrium analysis for the forces in the human spianl column and its musculature. Spine 5: 402-411.

ZUNIGA, E.N., TRUONG, X.T., SIMONS, D.G. (1970). Effects of skin electrode position on averaged electromyographic potentials. Arch. Phys. Med. Rehabil. 51: 264-272.

APPENDIX I

PRERUN CHECK LIST

1. Turn on force plate and multiplexer.

2. Set up Stairs, Ramp and camera according to the apparatus sequence designated for the subject. Make sure Ramp is adjusted to second level from the top.

3. Check paper and channels on chart recorder.
   - set sensitivity of Channels 1 and 6 to 10 mV/div
   - set gain on Channels 3 and 4 to 2,500.

4. Select leads and attach them to junction box.

   Leads 1 and 6 = Footswitches

   Leads 2 to 5 = EMG

5. Make sure correct belt is attached to junction box.

6. Check output from footswitches.

7. Prepare electrodes - 4 pairs bipolar, 1 reference.

8. Cut tape for electrodes.

9. Put alcohol swabs, tape measure and skin marker with electrodes.

10. Check lens on cine camera.

11. Load film in camera.

12. Attach cine camera to cart or tripod (depending on apparatus to be used first).
    - if camera is on tripod, level it and check tripod height.
    - position synchronizing light

13. Check trunk targets - C_7 - tape.

- L_5 - tape.
- Greater trochanters (2) - tape.
- Knee joints (2) - tape.

14. Adjust run number holder and place with first apparatus.

15. Attach sequence holder to ramp and stairs, make sure sequence cards are in correct order.

16. Load computer program - enter subject number and date.

EXPERIMENT ROUTINE

1. Meet subject, explain procedure, show equipment.

2. Ask subject to sign consent form.

3. Ask subject to change into shorts/bathing suit and running shoes.

4. Enter subject height into computer.

5. Have subject stand on force plate to enter weight into computer.

6. Fill in data sheet.

7. Place subject in supine position on mat, prepare skin over mid abdomen and lower chest by rubbing vigorously with alcohol swab.

8. Mark bony landmarks (symphysis pubis, xiphoid process), lateral border of rectus abdominis muscle and draw guide lines.

9. Apply electrodes over right and left rectus abdominis, apply reference electrode immediately below xiphoid process.

10. Photograph subject.

11. Attach leads - 3 = right rectus abdominis, 4 = left rectus abdominis.

12. Check to ensure that gain on channels 3 and 4 is set at 2,500.

13. Have subject perform maximum voluntary isometric contraction: supine, knees bent, feet flat on mat, apply

maximum resistance immediately below clavicles. Subject flexes upper trunk 15 cm off mat and told "Hold, don't let me push you down". Subject holds position for 5 seconds. Three trials with a 30 second rest between trials. Record all three trials on chart recorder, record third trial on computer (5 sec sample).

14. Check tracing of third trial on chart recorder. If it is acceptable, STORE the trial.

15. Repeat procedure in #13 and #14 for left rectus abdominis.

16. Change gain on Channels 3 and 4 to 4,000.

17. Detach leads.

18. Place subject in standing, prepare the skin over the lower back by rubbing vigorously with an alcohol swab.

19. Mark bony landmark (L_3 spinous process) and lateral border of erector spinae muscle, draw guide lines.

20. Apply electrodes over right and left erector spinae.

21. Photograph subject.

22. Attach leads - 2 = right erector spinae, 5 = left erector spinae.

23. Have subject perform maximum voluntary isometric contraction: prone lying, stabilize the lower limbs, apply maximum resistance in middle of back immediately inferior to the scapulae. Subject extends upper trunk 15 cm from the mat

and told " Hold, don't let me push you down". Subject holds position for 5 seconds. Three trials with a 30 second rest between trials. Record all three trials on the chart recorder, record third trial on the computer (5 sec sample).

24. Check tracing of third trial on chart recorder. If it is acceptable, STORE the trial.

25. Repeat procedure in #23 and #24 for left erector spinae.

26. Detach leads.

27. Tape footswitches to subject's shoes, check footswitch output on chart recorder, adjust if necessary.

28. Ask subject to put on shoes.

29. Attach body markers - C_7, L_5, right greater trochanter, left greater trochanter, right knee joint line, left knee joint line.

30. Apply and adjust belt with junction box. Make sure they do not interfere with the L_5 marker.

31. Attach all leads - 1 = Right Foot Switch, 2 = Right Erector Spinae, 3 = Right Rectus Abdominis, 4, Left Rectus Abdominis, 5 = Left Erector Spinae, 6 = Left Foot Switch.

32. Check footswitch output on chart recorder, adjust if necessary.

33. Secure all leads with tape.

34. Turn on TV tracking camera and monitor on mobile cart.

35. Position subject in front of first apparatus, check body markers, adjust if necessary.

36. Turn on flood lights.

37. Give subject commands for first apparatus:

LEVEL - "When I say 'Go', walk to the end of the runway and stop. Turn, and when I say 'Go', walk back. Walk at your usual speed." Repeat this twice to give a total of three trials. Change trial sequence cards after each trial. STORE DATA AFTER <u>EACH</u> TRIAL.

OR STAIRS - "When I say 'Go', walk up the stairs and stop. Turn around at the top and, when I say 'Go', walk down the stairs and stop. Walk at your usual speed." Repeat this twice to give a total of three UP trials and three DOWN trials. Change trial sequence cards after each trial. STORE DATA AFTER <u>EACH</u> TRIAL.

OR RAMP - "When I say 'Go', walk up the ramp and stop. Turn around at the top and, when I say 'Go', walk down the ramp and stop. Walk at your usual speed." Repeat this twice to give a total of three UP trials and three DOWN trials. Change trial sequence cards after each trial. STORE DATA AFTER <u>EACH</u> TRIAL.

38. Ensure that the following sequence is used - camera on, computer starts to sample, synch light on, subject begins walk.

39. Turn off flood lights.

40. After trials on each apparatus are complete, ask subject to rest while the next apparatus is being prepared. Offer subject a backless stool to sit on.

41. Change apparatus.

a) Walkway to Stairs/Ramp - Unplug and move camera to tripod.

- Move transformer and plug camera in.
- Check camera alignment with spirit level.
- Unplug, move and align flood lights.
- Remove synch light from walkway backdrop and attach to Stairs/Ramp.

b) Stairs/Ramp to Walkway - Unplug camera and remove from tripod.

- Move camera to mobile cart and attach to camera mount.
- Move transformer and plug camera in.
- Unplug, move and align flood lights.
- Remove synch light from Stairs/Ramp and attach to walkway backdrop.

c) Stairs to Ramp/Ramp to Stairs - Remove synch light.

- Move used apparatus out of the way.
- Move new apparatus into position. Check alignment with floor markings.
- Attach synch light to new apparatus.

42. When all three pieces of apparatus have been used, detach leads from electrodes and footswitches, remove belt and junction box, remove body markers, remove electrodes, remove footswitches. Thank subject.

APPENDIX II

Table 1. Mean Percentages of Stance for Right and Left Gait Cycles During Level Walking

SUBJECT	RIGHT STANCE	LEFT STANCE
01	65	64
02	65	62
03	62	65
04	61	60
05	61	59
06	66	65
07	58	61
08	58	59
09	60	60
10	63	63
11	63	59
12	64	64
13	68	69
14	64	65
15	62	65
16	65	62
17	60	62
18	66	62
$\overline{X}$	63	63
SD	3	3

Table 2. Mean Percentages of Stance for Right and Left Gait Cycles during Stair Climbing

SUBJECT	RIGHT STANCE	LEFT STANCE
01	62	66
02	61	68
03	60	63
04	63	65
05	68	60
06	69	58
07	60	65
08	60	62
09	66	64
10	56	59
11	64	66
12	64	65
13	67	65
14	61	65
15	68	63
16	66	61
17	63	63
18	65	60
$\overline{X}$	64	63
SD	3	3

Table 3. Mean Percentages of Stance for Right and Left Gait Cycle during Stair Descent

SUBJECT	RIGHT STANCE	LEFT STANCE
01	60	65
02	67	68
03	61	62
04	61	57
05	77	74
06	58	63
07	61	66
08	59	65
09	57	59
10	55	60
11	61	65
12	61	64
13	60	64
14	62	60
15	60	65
16	65	60
17	62	60
18	59	63
$\bar{X}$	61	63
SD	5	4

Table 4. Mean Percentages of Stance for Right and Left Gait Cycles during Ramp Climbing

SUBJECT	RIGHT STANCE	LEFT STANCE
01	66	67
02	67	64
03	62	66
04	61	66
05	66	64
06	59	64
06	65	60
08	55	68
09	62	65
10	65	65
11	66	66
12	65	65
13	68	68
14	65	58
15	67	66
16	65	63
17	61	63
18	64	67
$\overline{X}$	64	63
SD	3	3

Table 5. Mean Percentages of Stance for Right and Left Gait Cycles during Ramp Descent

SUBJECT	RIGHT STANCE	LEFT STANCE
01	66	67
02	63	64
03	63	63
04	58	58
05	58	62
06	58	58
07	57	66
08	62	58
09	63	59
10	61	63
11	61	61
12	65	64
13	63	70
14	56	67
15	62	60
16	63	58
17	57	51
18	61	65
$\bar{X}$	61	62
SD	3	5

Table 6. Mean Duration of Right Gait Cycle (Seconds)

SUBJECT	TYPE OF LOCOMOTION L	SC	SD	RC	RD
01	1.3	1.6	1.3	1.7	1.2
02	1.0	1.3	1.0	1.3	1.1
03	1.4	1.9	1.7	1.9	1.3
04	1.2	1.4	1.2	1.5	1.0
05	1.2	1.2	1.0	1.2	1.0
06	1.2	1.2	1.1	1.3	1.1
07	1.1	1.2	1.2	1.3	1.1
08	1.3	1.2	0.8	1.2	0.9
09	1.2	1.2	1.0	1.5	1.1
10	1.4	1.2	0.9	1.3	1.0
11	1.2	1.2	1.0	1.2	1.0
12	1.3	1.5	1.3	1.5	1.2
13	1.3	1.3	1.2	1.5	1.3
14	1.2	1.2	1.1	1.3	1.0
15	1.2	1.1	1.0	1.4	1.0
16	1.5	1.1	1.0	1.2	1.2
17	1.1	1.0	0.8	1.0	1.0
18	1.3	1.1	0.9	1.4	1.1
$\overline{X}$	1.2	1.3	1.1	1.4	1.1
SD	0.1	0.2	0.2	0.2	0.1

KEY: L = Level Walking
SC = Stair Climbing   RC = Ramp Climbing
SD = Stair Descent   RD = Ramp Descent

Table 7. Mean Cadence (Number of Steps per Minute)

SUBJECT	L	SC	SD	RC	RD
	TYPE OF LOCOMOTION				
01	95.6	74.1	94.0	73.3	98.5
02	112.7	95.3	120.1	91.1	113.0
03	84.2	68.5	78.2	70.0	94.5
04	96.6	88.3	101.7	83.8	108.1
05	104.6	94.9	116.9	95.9	117.0
06	100.9	105.6	115.8	92.4	116.6
07	105.0	100.0	103.8	92.8	112.6
08	99.8	101.5	153.4	99.3	140.4
09	100.3	96.3	119.3	84.1	108.8
10	85.3	108.8	134.5	95.5	119.8
11	100.7	99.7	120.0	97.3	117.7
12	89.6	83.0	90.9	78.9	99.0
13	93.0	91.4	102.6	80.4	93.0
14	97.5	97.9	115.4	96.1	120.0
15	102.6	110.3	125.1	87.8	117.7
16	84.9	109.6	119.2	99.4	104.6
17	113.2	126.9	146.4	118.6	129.2
18	96.1	108.5	130.8	87.8	108.4
$\overline{X}$	97.9	97.8	116.0	90.2	112.6
SD	8.5	13.8	18.9	11.2	12.0

KEY: L = Level Walking
SC = Stair Climbing   RC = Ramp Climbing
SD = Stair Descent   RD = Ramp Descent

Table 8. Mean Point of Temporal Events during Level Walking (% of RGC)

	TEMPORAL EVENTS		
SUBJECT	LEWB	LIC	REWB
01	14	51	65
02	12	50	65
03	11	46	62
04	11	51	61
05	11	53	61
06	14	50	66
07	9	49	58
08	8	50	58
09	10	50	60
10	13	50	63
11	13	54	63
12	14	50	64
13	18	50	68
14	15	50	64
15	15	50	62
16	14	52	65
17	11	49	60
18	16	54	66
$\bar{X}$	13	51	63
SD	3	2	3

KEY: LEWB = Left End of Weight Bearing
LIC = Left Initial Contact
REWB = Right End of Weight Bearing

Table 9. Mean Point of Temporal Events during Stair Climbing (% of RGC)

SUBJECT	TEMPORAL EVENTS LEWB	LIC	REWB
01	15	49	62
02	16	48	61
03	15	51	60
04	15	50	63
05	13	53	68
06	11	54	69
07	13	48	60
08	14	51	60
09	15	51	66
10	8	49	56
11	16	50	64
12	16	51	64
13	15	50	67
14	14	50	61
15	18	56	68
16	17	56	66
17	13	50	63
18	14	54	65
$\bar{X}$	14	51	64
SD	2	2	3

KEY: LEWB = Left End of Weight Bearing
LIC = Left Initial Contact
REWB = Right End of Weight Bearing

Table 10. Mean Point of Temporal Events during Stair Descent (% of RGC)

SUBJECT	TEMPORAL EVENTS LEWB	LIC	REWB
01	10	45	60
02	11	44	67
03	10	47	61
04	10	53	61
05	5	31	77
06	11	48	58
07	14	48	61
08	9	43	59
09	7	48	57
10	7	47	55
11	14	49	61
12	13	49	61
13	10	46	60
14	13	53	62
15	11	46	60
16	12	52	65
17	9	49	62
18	12	49	59
$\bar{X}$	10	47	61
SD	2	5	5

KEY: LEWB = Left End of Weight Bearing
LIC = Left Initial Contact
REWB = Right End of Weight Bearing

Table 11. Mean Point of Temporal Events during Ramp Climbing (% of RGC)

	TEMPORAL EVENTS		
SUBJECT	LEWB	LIC	REWB
01	18	51	66
02	19	55	67
03	17	51	62
04	16	50	61
05	15	51	66
06	13	49	59
07	13	53	65
08	14	47	55
09	15	50	62
10	16	51	65
11	17	51	66
12	16	51	65
13	15	47	68
14	10	52	65
15	19	53	67
16	15	52	65
17	9	47	61
18	14	48	64
$\overline{X}$	15	51	64
SD	3	2	3

KEY: LEWB = Left End of Weight Bearing
LIC = Left Initial Contact
REWB = Right End of Weight Bearing

Table 12. Mean Point of Temporal Events during Ramp Descent (% of RGC)

SUBJECT	TEMPORAL EVENTS LEWB	LIC	REWB
01	16	49	66
02	12	48	63
03	11	47	63
04	7	49	58
05	13	52	58
06	10	52	58
07	12	45	57
08	8	50	62
09	10	51	63
10	10	47	61
11	9	48	61
12	13	49	65
13	16	46	63
14	13	47	56
15	10	50	62
16	16	57	63
17	2	52	57
18	12	46	61
$\overline{X}$	11	49	61
SD	3	3	3

KEY: LEWB = Left End of Weight Bearing
LIC = Left Initial Contact
REWB = Right End of Weight Bearing

Table 13. Mean Degrees of Trunk Inclination from the Vertical during Level Walking (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	3	4	3	2	2	2	2	2	1	2	2
02	-1	0	2	2	1	0	0	-1	-2	-2	-2
03	-2	-2	-2	-1	-1	-2	-2	-2	-3	-3	-2
04	1	1	1	0	0	0	0	-1	-1	-1	-1
05	1	1	2	3	3	3	2	1	1	0	1
06	-3	-3	-4	-4	-4	-3	-3	-3	-3	-3	-3
07	-4	-3	-3	-2	-1	-2	-2	-2	-2	-3	-3
08	3	3	3	4	4	5	3	3	3	2	2
09	1	2	3	4	4	3	3	1	-1	-1	-1
10	1	2	2	2	2	2	2	2	1	2	1
11	-8	-7	-7	-8	-8	-8	-8	-8	-8	-8	-8
12	-5	-5	-4	-4	-4	-4	-5	-5	-6	-6	-6
13	-1	0	1	2	3	2	2	1	1	0	0
14	0	1	2	2	2	2	2	1	1	1	1
15	-3	-3	-3	-3	-3	-4	-3	-4	-4	-4	-4
16	-3	-2	-2	-3	-3	-3	-3	-3	-3	-5	-5
17	-2	-2	-1	-1	-1	-2	-2	-1	-1	-1	-1
18	1	1	2	2	1	1	1	1	1	1	0
$\bar{X}$	-1	-1	0	0	0	0	-1	-1	-1	-2	-2
SD	3	3	3	3	3	3	3	3	3	3	3

KEY: - = Anterior Inclination + = Posterior Inclination

Table 13 (cont). Mean Degrees of Trunk Inclination from the Vertical during Level Walking (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	2	2	3	2	3	3	4	3	3	3
02	-1	0	0	0	0	0	-1	-1	-1	-1
03	-2	-1	-1	-1	-1	-2	-2	-3	-3	-3
04	-1	-1	-1	-1	-2	-2	-2	-2	-2	-2
05	1	2	3	3	2	2	2	1	1	1
06	-3	-3	-3	-3	-3	-3	-3	-3	-2	-2
07	-2	-2	-2	-2	-1	-2	-2	-3	-3	-3
08	2	3	3	4	3	3	2	1	1	1
09	-1	0	1	1	1	0	0	0	0	0
10	2	2	2	2	2	1	2	2	1	2
11	-8	-8	-8	-8	-8	-8	-8	-8	-8	-8
12	-6	-6	-6	-6	-5	-5	-6	-6	-7	-7
13	1	2	2	2	1	1	0	1	1	1
14	1	2	2	2	2	1	1	0	0	1
15	-4	-3	-2	-2	-2	-2	-2	-2	-2	-2
16	-4	-3	-4	-4	-4	-3	-4	-4	-5	-5
17	-1	-1	-5	-4	-4	-3	-2	-2	-2	-2
18	0	1	1	1	1	1	1	0	0	0
$\overline{X}$	-1	-1	-1	-1	-1	-1	-1	-2	-2	-2
SD	3	3	3	3	3	3	3	3	3	3

KEY: - = Anterior Inclination + = Posterior Inclination

Table 14. Mean Trunk Inclination from the Vertical during Stair Climbing (at 5% Intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	-5	-5	-5	-5	-5	-5	-5	-6	-7	-8	-9
02	-11	-11	-11	-11	-10	-10	-10	-11	-12	-12	-12
03	-8	-8	-8	-8	-7	-6	-6	-7	-8	-9	-9
04	-13	-13	-13	-12	-11	-10	-10	-10	-11	-11	-12
05	-7	-7	-6	-6	-6	-6	-6	-6	-6	-6	-7
06	-17	-18	-18	-18	-18	-17	-16	-16	-17	-18	-19
07	-19	-19	-20	-20	-20	-20	-20	-20	-19	-19	-19
08	-11	-11	-10	-9	-8	-7	-8	-9	-10	-10	-11
09	-7	-6	-6	-6	-5	-5	-5	-5	-6	-7	-6
10	-12	-11	-10	-10	-10	-9	-8	-8	-9	-10	-10
11	-22	-22	-22	-23	-23	-23	-22	-22	-22	-22	-22
12	-13	-13	-13	-14	-13	-13	-14	-14	-14	-14	-14
13	-11	-11	-12	-11	-10	-9	-9	-9	-9	-10	-10
14	-6	-6	-7	-7	-7	-7	-7	-7	-8	-7	-7
15	-16	-16	-16	-16	-17	-18	-18	-18	-18	-18	-17
16	-10	-10	-10	-9	-9	-9	-9	-9	-10	-11	-11
17	-9	-9	-9	-10	-10	-11	-12	-13	-14	-14	-14
18	-12	-11	-11	-11	-10	-10	-10	-10	-10	-11	-11
$\bar{X}$	-12	-12	-12	-11	-11	-11	-11	-11	-12	-12	-12
SD	5	5	5	5	5	5	5	5	5	5	5

KEY: - = Anterior Inclination + = Posterior Inclination

Table 14 (cont.). Mean Degrees of Trunk Inclination from the Vertical during Stair Climbing (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-8	-8	-8	-6	-5	-3	-3	-3	-4	-4
02	-11	-10	-9	-8	-7	-6	-6	-6	-6	-7
03	-10	-10	-11	-10	-10	-9	-9	-10	-10	-10
04	-12	-12	-11	-11	-11	-10	-11	-12	-12	-12
05	-7	-7	-7	-6	-6	-7	-7	-7	-7	-7
06	-20	-20	-20	-20	-19	-19	-19	-19	-19	-20
07	-19	-19	-19	-18	-17	-16	-16	-17	-17	-17
08	-10	-10	-9	-9	-8	-8	-9	-10	-10	-10
09	-6	-5	-4	-2	-1	-1	-1	-2	-3	-4
10	-11	-10	-10	-9	-9	-9	-9	-10	-10	-11
11	-22	-23	-23	-22	-21	-20	-19	-18	-18	-18
12	-13	-13	-12	-11	-10	-10	-10	-10	-11	-12
13	-10	-9	-9	-9	-8	-7	-6	-6	-7	-7
14	-7	-7	-6	-5	-4	-4	-3	-4	-4	-4
15	-17	-17	-17	-17	-17	-16	-16	-15	-15	-15
16	-11	-11	-11	-10	-10	-9	-9	-9	-9	-9
17	-13	-13	-13	-12	-11	-11	-10	-10	-10	-10
18	-11	-11	-10	-9	-8	-7	-7	-7	-8	-9
$\overline{X}$	-12	-12	-12	-11	-10	-10	-9	-10	-10	-10
SD	5	5	5	5	5	5	5	5	5	5

KEY: – = Anterior Inclination + = Posterior Inclination

Table 15. Mean Degrees of Trunk Inclination from the Vertical during Stair Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	2	2	1	1	1	1	2	2	3	4	4
02	-2	-2	-2	-3	-3	-3	-4	-4	-3	-2	-2
03	1	2	2	1	-1	-2	-2	-2	-2	-1	0
04	0	0	-1	-2	-3	-3	-4	-4	-4	-3	-3
05	1	1	1	0	0	-1	-1	-1	-1	0	0
06	-3	-3	-5	-7	-7	-7	-6	-6	-5	-4	-3
07	-4	-4	-5	-6	-6	-7	-7	-7	-6	-5	-5
08	-1	-1	-2	-3	-5	-5	-4	-3	-2	-1	0
09	3	3	2	2	2	2	2	2	2	3	3
10	2	2	2	1	0	-1	-1	-1	0	0	0
11	-7	-7	-7	-9	-10	-11	-12	-12	-12	-11	-9
12	-4	-4	-4	-5	-6	-7	-7	-7	-7	-5	-3
13	0	0	0	0	0	-1	-2	-2	-2	-1	-1
14	0	0	0	-1	-1	-2	-2	-2	-2	-1	0
15	-1	-1	-1	-2	-3	-3	-4	-4	-4	-3	-2
16	-2	-1	-1	-2	-3	-3	-3	-4	-3	-3	-2
17	2	2	2	1	0	-1	-1	-1	-1	0	2
18	1	2	2	1	0	1	1	1	1	2	2
$\bar{X}$	-1	-1	-1	-2	-2	-3	-3	-3	-3	-2	-1
SD	3	3	3	4	3	3	3	3	4	3	3

KEY: - = Anterior Inclination + = Posterior Inclination

Table 15 (cont.). Mean Degrees of Trunk Inclination from the Vertical during Stair Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	4	3	2	1	1	1	1	1	2	2
02	-2	-2	-3	-3	-3	-3	-3	-2	-2	-2
03	0	0	-1	-2	-2	-2	-1	0	0	0
04	-2	-3	-3	-4	-4	-3	-3	-2	-1	-1
05	1	1	1	0	0	0	0	1	2	2
06	-4	-5	-6	-6	-7	-6	-6	-5	-4	-4
07	-5	-5	-5	-6	-6	-6	-7	-7	-6	-6
08	0	0	-1	-1	-1	0	0	0	1	0
09	3	2	1	1	1	1	2	2	3	3
10	0	-1	-1	-1	-1	0	1	3	4	5
11	-8	-9	-10	-11	-12	-12	-11	-10	-8	-8
12	-2	-3	-4	-4	-5	-5	-5	-4	-4	-4
13	0	0	-1	-1	0	0	-1	-1	-1	0
14	0	0	-1	-1	-2	-2	-3	-2	-1	-1
15	-2	-2	-3	-3	-4	-4	-3	-3	-2	-2
16	-1	-1	-2	-2	-3	-3	-3	-3	-3	-3
17	2	2	1	1	1	1	2	2	3	3
18	2	1	1	1	2	2	2	2	3	3
$\bar{X}$	-1	-1	-2	-2	-3	-2	-2	-2	-1	-1
SD	3	3	3	3	4	3	4	3	3	3

KEY: – = Anterior Inclination   + = Posterior Inclination

Table 16. Mean Degrees of Trunk Inclination from the Vertical during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	-5	-5	-6	-7	-7	-7	-7	-7	-7	-7	-8
02	-12	-13	-14	-14	-13	-13	-14	-15	-16	-16	-16
03	-9	-9	-9	-8	-7	-6	-6	-7	-7	-8	-8
04	-13	-13	-14	-14	-13	-12	-11	-11	-12	-12	-12
05	-6	-6	-6	-6	-6	-6	-6	-6	-6	-7	-7
06	-19	-20	-22	-24	-24	-23	-22	-21	-20	-19	-20
07	-15	-15	-16	-17	-18	-18	-18	-19	-19	-20	-20
08	-10	-10	-10	-10	-10	-9	-8	-7	-7	-7	-9
09	-6	-5	-3	-3	-4	-3	-3	-3	-3	-4	-5
10	-10	-11	-11	-12	-12	-11	-11	-11	-12	-12	-11
11	-20	-20	-21	-22	-22	-23	-23	-22	-21	-21	-21
12	-13	-12	-12	-12	-12	-13	-13	-14	-14	-15	-14
13	-8	-8	-7	-6	-5	-5	-4	-4	-5	-6	-7
14	-8	-9	-10	-11	-11	-10	-9	-9	-8	-9	-10
15	-12	-12	-13	-14	-14	-15	-14	-14	-14	-13	-13
16	-15	-15	-15	-15	-15	-15	-14	-13	-13	-13	-14
17	-9	-10	-11	-11	-12	-11	-11	-11	-11	-11	-11
18	-12	-12	-13	-13	-14	-14	-14	-13	-13	-13	-13
$\overline{X}$	-11	-11	-12	-12	-12	-12	-12	-12	-12	-12	-12
SD	4	4	5	5	5	6	6	5	5	5	5

KEY: − = Anterior Inclination + = Posterior Inclination

Table 16 (cont.). Mean degrees of Trunk Inclination from the Vertical during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-9	-10	-11	-11	-10	-9	-8	-7	-7	-8
02	-15	-14	-13	-12	-12	-11	-12	-12	-13	-14
03	-9	-10	-11	-12	-11	-11	-10	-10	-9	-9
04	-13	-14	-15	-15	-14	-14	-14	-13	-13	-12
05	-7	-6	-6	-6	-5	-5	-5	-6	-6	-5
06	-22	-23	-22	-21	-20	-17	-15	-13	-12	-12
07	-20	-17	-18	-17	-17	-16	-16	-16	-15	-15
08	-12	-14	-17	-18	-16	-14	-11	-9	-8	-8
09	-5	-5	-5	-4	-3	-3	-2	-2	-2	-3
10	-11	-12	-13	-13	-12	-12	-11	-10	-9	-9
11	-22	-22	-22	-22	-22	-21	-19	-19	-18	-18
12	-14	-14	-14	-15	-15	-15	-15	-15	-15	-15
13	-7	-7	-7	-6	-5	-4	-5	-5	-6	-5
14	-10	-11	-12	-11	-9	-8	-7	-6	-7	-7
15	-13	-13	-13	-16	-17	-16	-16	-15	-14	-14
16	-15	-15	-14	-14	-12	-11	-9	-10	-10	-10
17	-11	-11	-11	-11	-11	-10	-9	-8	-9	-9
18	-13	-13	-14	-14	-14	-12	-11	-10	-10	-10
$\bar{X}$	-13	-13	-13	-13	-13	-12	-11	-10	-10	-10
SD	5	5	5	5	5	5	4	4	4	4

KEY: - = Anterior Inclination + = Posterior Inclination

Table 17. Mean Degrees of Trunk Inclination from the Vertical during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	0	0	0	0	0	1	1	2	2	2	1
02	0	1	1	1	1	1	1	1	1	1	1
03	-1	0	0	-1	-1	-1	-1	-1	-1	0	0
04	-2	-2	-2	-3	-4	-4	-4	-5	-5	-5	-5
05	3	3	4	4	4	4	4	4	4	4	4
06	1	0	-1	-2	-2	-1	-1	-1	0	0	0
07	-4	-4	-4	-5	-5	-5	-6	-6	-5	-5	-4
08	3	3	3	2	2	2	2	1	1	2	2
09	3	4	5	5	5	5	4	4	3	3	4
10	3	3	3	2	2	1	1	1	1	1	1
11	-6	-6	-6	-6	-7	-7	-7	-8	-8	-7	-7
12	-2	-2	-3	-3	-3	-2	-2	-2	-2	-3	-3
13	1	1	2	2	2	2	1	1	1	0	0
14	3	3	3	3	3	3	3	3	4	5	5
15	-3	-3	-3	-3	-3	-3	-3	-3	-3	-2	-2
16	-4	-4	-3	-4	-4	-3	-3	-2	-2	-1	0
17	3	3	3	2	2	2	2	3	3	4	5
18	1	1	1	0	0	0	0	0	1	1	1
$\overline{X}$	0	0	0	0	0	0	0	0	0	0	0
SD	3	3	3	3	3	3	3	3	3	3	3

KEY: – = Anterior Inclination + = Posterior Inclination

Table 17 (cont.). Mean Degrees of Trunk Inclination from the Vertical during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	1	0	1	1	2	3	3	3	3	3
02	1	1	2	2	2	3	3	3	3	3
03	0	0	-1	-1	-2	-2	-3	-2	-2	-2
04	-5	-6	-6	-6	-5	-4	-4	-3	-3	-2
05	4	4	4	4	4	4	5	5	5	6
06	0	-1	-1	-1	-1	-1	0	1	2	2
07	-5	-5	-5	-5	-5	-6	-6	-7	-7	-7
08	2	3	2	2	3	3	3	3	3	3
09	6	6	6	6	5	5	4	4	4	4
10	2	2	2	2	2	3	3	3	4	3
11	-6	-6	-7	-7	-7	-8	-8	-7	-7	-6
12	-3	-3	-4	-4	-4	-4	-4	-4	-4	-3
13	0	0	0	0	0	0	0	0	0	0
14	5	4	3	3	2	2	2	2	3	3
15	-2	-2	-2	-2	-1	-1	0	0	1	1
16	1	1	2	1	1	1	1	1	2	2
17	4	2	-1	0	0	1	1	2	3	3
18	1	1	0	0	0	0	0	1	1	0
$\overline{X}$	0	0	0	0	0	0	0	0	1	1
SD	3	3	4	3	3	4	4	4	4	3

KEY: - = Anterior Inclination + = Posterior Inclination

Table 18. Mean Degrees of Pelvic Inclination from the Vertical during Level Walking (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	1	1	1	0	0	−1	−2	−3	−5	−9	−11
02	3	4	4	3	3	1	0	−1	−3	−5	−7
03	−1	−1	0	0	−1	−2	−4	−5	−5	−7	−8
04	−3	−2	−2	−1	−1	−1	−1	−2	−2	−3	−4
05	−1	0	0	1	1	0	−3	−6	−9	−11	−12
06	−7	−7	−7	−8	−8	−8	−8	−9	−9	−10	−11
07	−3	−1	0	0	0	−2	−4	−7	−9	−10	−11
08	−3	−3	−4	−5	−8	−5	−4	−4	−5	−6	−7
09	−2	−2	−3	−3	−3	−3	−5	−6	−7	−9	−10
10	−4	−3	−3	−3	−3	−5	−6	−8	−10	−13	−15
11	−1	0	−2	−4	−4	−5	−6	−7	−8	−9	−10
12	1	1	−1	−3	−3	−4	−5	−7	−8	−9	−9
13	0	1	1	1	1	0	−2	−3	−6	−9	−12
14	8	8	9	9	9	7	4	0	−2	−4	−6
15	−1	0	−1	−2	−3	−5	−6	−8	−10	−13	−16
16	−8	−7	−7	−6	−6	−6	−7	−8	−9	−11	−13
17	−3	−2	−2	−2	−3	−4	−6	−8	−10	−13	−16
18	−2	−2	−2	−2	−1	−1	−3	−4	−5	−8	−9
$\overline{X}$	−1	−1	−1	−1	−2	−2	−4	−5	−7	−9	−10
SD	4	4	4	4	4	3	3	3	3	3	3

KEY: − = Anterior Inclination + = Posterior Inclination

Table 18 (cont.) Mean Degrees of Pelvic Inclination from the Vertical during Level Walking (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-13	-12	-9	-7	-3	-1	-1	0	2	4
02	-8	-7	-5	-4	-3	-1	-1	0	0	2
03	-7	-8	-8	-6	-6	-5	-6	-6	-6	-6
04	-5	-4	-4	-4	-4	-2	-2	-3	-3	-3
05	-12	-12	-11	-9	-8	-6	-4	-3	-2	-1
06	-12	-12	-11	-10	-8	-6	-5	-4	-5	-6
07	-10	-10	-7	-5	-4	-3	-3	-4	-4	-3
08	-6	-5	-5	-5	-5	-5	-4	-4	-5	-5
09	-10	-9	-8	-8	-7	-7	-7	-6	-6	-6
10	-17	-18	-17	-15	-13	-12	-11	-11	-10	-10
11	-10	-8	-7	-6	-5	-3	-3	-3	-3	-2
12	-9	-9	-10	-10	-10	-10	-10	-10	-9	-8
13	-12	-12	-10	-9	-8	-7	-6	-5	-5	-5
14	-5	-4	-3	-2	1	4	6	6	6	6
15	-17	-14	-10	-6	-5	-3	-2	-2	-3	-2
16	-15	-15	-13	-9	-8	-8	-7	-7	-6	-4
17	-16	-14	-13	-10	-8	-6	-5	-4	-4	-4
18	-9	-9	-8	-7	-7	-5	-4	-4	-5	-5
$\overline{X}$	-11	-10	-9	-7	-6	-5	-4	-4	-4	-3
SD	4	4	4	3	3	4	4	4	4	4

KEY: - = Anterior Inclination + = Posterior Inclination

Table 19. Mean Degrees of Pelvic Inclination from the Vertical during Stair Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	-4	-4	-6	-6	-7	-6	-6	-7	-8	-8	-9
02	-6	-6	-5	-5	-4	-4	-5	-5	-5	-5	-4
03	-17	-16	-14	-11	-12	-10	-8	-8	-10	-10	-9
04	-9	-10	-10	-9	-9	-8	-8	-8	-8	-9	-9
05	-14	-12	-12	-11	-11	-12	-12	-13	-13	-14	-14
06	-19	-19	-20	-20	-20	-19	-18	-18	-19	-20	-19
07	-12	-11	-9	-7	-7	-8	-9	-10	-10	-10	-10
08	-23	-23	-22	-20	-19	-18	-17	-15	-14	-14	-15
09	-21	-21	-19	-16	-15	-12	-11	-11	-11	-11	-11
10	-14	-13	-12	-11	-10	-10	-10	-10	-11	-12	-12
11	-9	-8	-8	-8	-8	-9	-9	-9	-10	-10	-10
12	-20	-20	-19	-18	-17	-16	-14	-13	-13	-12	-13
13	-3	-4	-4	-4	-4	-4	-4	-4	-6	-7	-7
14	-12	-10	-9	-7	-6	-5	-5	-4	-5	-6	-6
15	-4	-5	-7	-8	-8	-7	-7	-8	-9	-10	-11
16	-15	-14	-14	-14	-13	-12	-12	-13	-12	-12	-13
17	-19	-19	-17	-13	-12	-11	-10	-10	-10	-10	-10
18	-9	-10	-10	-9	-7	-7	-7	-8	-8	-8	-8
$\bar{X}$	-13	-13	-12	-11	-11	-10	-10	-10	-10	-10	-11
SD	6	6	5	5	5	4	4	4	3	3	4

KEY: − = Anterior Inclination + = Posterior Inclination

Table 19 (Cont.). Mean Degrees of Pelvic Inclination from Vertical during Stair Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-8	-9	-8	-7	-3	-3	-4	-2	-3	-1
02	-3	-2	0	1	2	3	2	1	-1	-1
03	-10	-11	-11	-11	-13	-16	-18	-18	-20	-17
04	-10	-10	-9	-8	-8	-9	-10	-10	-8	-9
05	-13	-13	-12	-12	-11	-11	-13	-14	-15	-15
06	-18	-19	-19	-19	-19	-19	-19	-19	-19	-20
07	-10	-11	-10	-8	-8	-8	-9	-9	-9	-8
08	-14	-14	-13	-14	-15	-16	-17	-17	-17	-19
09	-11	-11	-11	-11	-11	-12	-14	-16	-16	-16
10	-12	-10	-9	-8	-8	-9	-10	-10	-12	-14
11	-10	-10	-10	-10	-9	-8	-8	-9	-9	-9
12	-13	-14	-15	-15	-16	-17	-19	-21	-21	-20
13	-7	-7	-6	-5	-4	-4	-4	-4	-3	-2
14	-5	-4	-3	-4	-6	-8	-8	-8	-8	-7
15	-11	-10	-9	-8	-7	-6	-7	-9	-10	-10
16	-13	-14	-14	-14	-14	-14	-15	-18	-18	-19
17	-10	-9	-9	-10	-10	-12	-14	-16	-17	-16
18	-8	-8	-8	-8	-8	-8	-7	-8	-8	-8
$\overline{X}$	-10	-10	-10	-10	-9	-10	-11	-12	-12	-12
SD	3	4	4	5	5	6	6	6	6	6

KEY: − = Anterior Inclination + = Posterior Inclination

Table 20. Mean Degrees of Pelvic Inclination from Vertical during Stair Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	−4	−4	−4	−4	−4	−5	−5	−5	−4	−3	−2
02	−4	−5	−5	−4	−4	−4	−5	−5	−5	−4	−4
03	−3	−4	−4	−4	−5	−5	−4	−3	−3	−4	−4
04	−6	−3	−4	−4	−6	−6	−7	−7	−8	−8	−6
05	0	0	0	0	1	1	1	0	0	1	1
06	−1	−1	−1	0	−2	−3	−3	−2	−2	−3	−3
07	2	3	4	6	6	5	3	−3	3	4	5
08	8	8	8	7	7	8	8	7	6	6	6
09	−8	−7	−6	−3	−1	−2	−3	−4	−4	−4	−4
10	−4	−4	−4	−5	−6	−6	−6	−7	−7	−8	−7
11	−4	−4	−4	−3	−3	−3	−4	−4	−5	−4	−4
12	3	3	2	2	2	2	2	3	5	5	5
13	−7	−6	−6	−5	−5	−6	−6	−8	−8	−8	−9
14	4	5	6	7	7	6	4	4	3	2	2
15	−3	−3	−4	−4	−3	−3	−3	−3	−4	−4	−5
16	−5	−5	−5	−4	−5	−6	−8	−8	−8	−8	−8
17	−10	−10	−9	−8	−8	−8	−8	−9	−9	−9	−10
18	−8	−7	−7	−6	−6	−6	−7	−8	−9	−8	−8
$\bar{X}$	−3	−2	−2	−2	−2	−2	−2	−3	−3	−3	−3
SD	5	5	5	5	5	5	5	5	5	5	5

KEY: − = Anterior Inclination + = Posterior Inclination

Table 20 (cont). Mean Degrees of Pelvic Inclination from Vertical during Stair Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	0	2	3	3	2	0	-2	-2	-5	-6
02	-4	-4	-3	-4	-4	-5	-6	-6	-6	-6
03	-4	-5	-2	-2	-2	-2	-4	-4	-5	-6
04	-5	-3	-1	0	-1	-3	-5	-6	-6	-7
05	2	2	2	2	1	1	0	-1	-1	-1
06	3	3	3	2	1	0	-1	-2	-3	-4
07	6	10	10	9	8	6	4	2	1	1
08	6	6	6	6	5	5	5	7	8	8
09	-2	-1	0	0	-2	-4	-5	-6	-8	-8
10	-6	-6	-7	-8	-9	-10	-10	-11	-12	-12
11	-5	-5	-4	-3	-4	-5	-7	-6	-6	-6
12	5	6	6	7	6	5	3	2	2	2
13	-9	-8	-6	-6	-7	-8	-8	-8	-8	-8
14	2	3	4	4	3	2	2	3	3	3
15	-6	-4	-1	1	0	-2	-4	-4	-6	-7
16	-8	-7	-7	-6	-6	-7	-7	-8	-8	-8
17	-9	-8	-8	-8	-10	-11	-13	-13	-14	-14
18	-7	-7	-7	-8	-9	-10	-10	-10	-10	-10
$\overline{X}$	-2	-1	-1	-1	-2	-3	-4	-4	-5	-5
SD	5	6	5	5	5	5	5	5	6	6

KEY: - = Anterior Inclination + = Posterior Inclination

Table 21. Mean Degrees of Pelvic Inclination from Vertical during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	1	1	0	−1	−2	−2	−3	−5	−5	−6	−8
02	0	−1	−4	−5	−5	−4	−4	−5	−5	−5	−6
03	−18	−18	−17	−15	−12	−9	−7	−7	−7	−7	−7
04	−11	−9	−9	−8	−7	−7	−6	−7	−7	−8	−9
05	−10	−8	−7	−8	−9	−10	−11	−12	−12	−13	−13
06	−17	−17	−18	−19	−19	−19	−18	−17	−17	−18	−18
07	−4	−4	−3	−3	−6	−7	−9	−11	−13	−13	−14
08	−20	−20	−20	−21	−20	−18	−16	−14	−13	−13	−13
09	−16	−16	−16	−15	−14	−12	−10	−9	−9	−9	−10
10	−7	−6	−8	−10	−10	−10	−11	−12	−13	−14	−15
11	−6	−5	−5	−6	−7	−7	−6	−6	−7	−7	−8
12	−14	−14	−14	−14	−15	−15	−15	−14	−14	−14	−14
13	−2	0	0	−1	−1	−1	−3	−3	−4	−4	−6
14	−6	−5	−4	−5	−7	−6	−4	−3	−4	−6	−8
15	−6	−6	−7	−7	−9	−11	−12	−11	−11	−11	−12
16	−14	−13	−14	−15	−14	−13	−13	−13	−15	−17	−18
17	−12	−12	−11	−10	−9	−10	−10	−10	−11	−12	−14
18	−10	−9	−8	−9	−9	−10	−10	−10	−10	−10	−11
$\bar{X}$	−10	−9	−9	−10	−10	−10	−9	−9	−10	−10	−11
SD	6	6	6	6	5	5	5	4	4	4	4

KEY: − = Anterior Inclination + = Posterior Inclination

Table 21 (cont.). Mean Degrees of Pelvic Inclination from Vertical during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	−11	−13	−13	−10	−7	−6	−6	−4	−2	−1
02	−7	−6	−5	−5	−5	−6	−6	−5	−4	−3
03	−8	−9	−11	−12	−14	−12	−13	−15	−16	−16
04	−10	−12	−11	−11	−11	−12	−11	−11	−10	−9
05	−14	−15	−14	−12	−11	−10	−9	−9	−8	−7
06	−18	−18	−15	−13	−11	−11	−9	−7	−6	−5
07	−16	−14	−15	−15	−15	−14	−12	−9	−6	−5
08	−19	−18	−18	−15	−13	−13	−14	−15	−16	−17
09	−11	−12	−14	−12	−10	−9	−10	−11	−10	−11
10	−16	−17	−16	−14	−13	−13	−13	−11	−8	−4
11	−9	−9	−8	−7	−6	−5	−5	−5	−4	−3
12	−15	−15	−16	−16	−15	−15	−16	−16	−16	−14
13	−7	−8	−8	−8	−8	−6	−5	−5	−5	−5
14	−8	−8	−8	−8	−8	−7	−7	−6	−5	−4
15	−13	−14	−15	−13	−9	−6	−6	−5	−4	−4
16	−19	−20	−19	−17	−15	−14	−14	−14	−13	−10
17	−15	−17	−15	−13	−13	−13	−15	−15	−14	−11
18	−13	−14	−14	−14	−14	−14	−13	−14	−13	−11
$\overline{X}$	−13	−13	−13	−12	−11	−10	−10	−10	−9	−8
SD	4	4	4	3	3	3	4	4	5	5

KEY: − = Anterior Inclination + = Posterior Inclination

Table 22. Mean Degrees of Pelvic Inclination from Vertical during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	−7	−6	−5	−4	−4	−4	−4	−5	−5	−6	−8
02	0	0	1	2	2	1	0	−2	−2	−3	−4
03	−6	−6	−6	−5	−3	−2	−1	−1	0	0	0
04	−8	−8	−8	−6	−6	−6	−6	−8	−9	−10	−10
05	−2	−2	−1	0	1	0	−2	−3	−5	−7	−8
06	3	3	3	5	5	5	5	4	3	1	1
07	10	10	10	11	11	10	9	8	6	5	5
08	−6	−6	−4	−1	0	−2	−4	−5	−5	−6	−6
09	0	0	1	1	2	2	0	−1	−2	−3	−3
10	−7	−6	−7	−6	−5	−3	−3	−4	−6	−8	−8
11	−1	−1	0	2	2	2	3	3	3	1	1
12	−7	−7	−4	−1	−2	−3	−3	−5	−6	−8	−8
13	3	3	5	7	9	8	4	1	−2	−6	−7
14	−3	−2	−2	0	0	−1	−1	−1	−1	−2	−4
15	−6	−6	−6	−4	−4	−5	−7	−10	−13	−15	−18
16	−13	−12	−10	−9	−9	−9	−11	−13	−14	−16	−16
17	−1	0	0	1	1	1	0	−1	−3	−4	−5
18	−2	−1	0	1	2	3	3	3	2	1	0
$\bar{X}$	−3	−3	−2	0	0	0	−1	−2	−3	−5	−5
SD	5	5	5	5	5	5	5	5	5	6	6

KEY: − = Anterior Inclination + = Posterior Inclination

Table 22 (cont.). Mean Degrees of Pelvic Inclination from Vertical during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-8	-9	-8	-7	-6	-6	-6	-7	-7	-7
02	-4	-4	-2	-1	-1	0	0	-1	-1	-1
03	-1	-1	-2	-3	-3	-3	-4	-4	-4	-4
04	-9	-8	-6	-6	-7	-7	-8	-8	-9	-9
05	-8	-8	-9	-8	-7	-5	-5	-5	-5	-4
06	2	4	3	1	1	3	4	4	4	3
07	1	3	4	5	4	2	1	1	0	0
08	6	7	7	8	8	8	8	8	7	7
09	-6	-7	-6	-5	-4	-4	-4	-5	-5	-5
10	-4	-5	-4	-3	-2	-2	-3	-4	-5	-4
11	-8	-8	-6	-4	-4	-5	-5	-5	-5	-5
12	0	0	1	2	2	1	0	-2	-4	-5
13	-8	-8	-7	-5	-5	-5	-5	-5	-7	-8
14	-6	-5	-3	-1	2	4	4	4	3	4
15	-4	-3	-1	1	2	2	2	1	1	1
16	-19	-18	-16	-15	-14	-13	-11	-10	-10	-9
17	-16	-22	-22	-20	-17	-15	-13	-13	-13	-14
18	-5	-4	-2	-2	-2	-3	-4	-5	-4	-4
$\bar{X}$	-5	-5	-4	-4	-3	-3	-3	-3	-4	-4
SD	6	7	7	7	6	6	5	5	5	5

KEY: - = Anterior Inclination + = Posterior Inclination

Table 23. Mean Hip Angle (°) during Level Walking (at 5% Intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	22	22	20	17	11	5	0	-6	-10	-15	-18
02	22	20	19	15	11	5	1	-5	-8	-13	-16
03	23	22	19	15	11	8	4	0	-4	-8	-10
04	22	22	22	20	15	8	2	-4	-10	-13	-16
05	25	25	25	22	16	10	3	-3	-6	-12	-15
06	24	23	22	18	14	9	5	2	-2	-6	-9
07	24	24	22	19	14	10	5	0	-5	-10	-13
08	20	20	17	15	9	4	-1	-5	-7	-11	-14
09	23	21	18	14	10	5	1	-2	-6	-10	-13
10	22	21	18	14	9	4	1	-3	-6	-9	-13
11	25	23	21	18	13	8	4	-2	-6	-11	-14
12	20	19	17	14	10	5	1	-3	-8	-11	-14
13	19	18	16	11	7	3	0	-3	-6	-10	-14
14	22	20	18	14	8	3	-3	-7	-11	-15	-19
15	30	30	29	27	22	16	8	1	-4	-10	-13
16	24	24	24	21	17	11	6	1	-3	-8	-11
17	29	29	29	27	22	15	7	-2	-9	-15	-19
18	25	25	24	21	16	10	4	-3	-8	-12	-15
$\overline{X}$	23	23	21	18	13	8	3	-2	-7	-11	-14
SD	3	3	4	4	4	4	3	3	2	3	3

KEY: + = Flexion − = Extension

Table 23 (cont). Mean Hip Angle (°) during Level Walking (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-17	-13	-3	8	15	19	23	23	22	21
02	-16	-12	-6	5	13	20	23	23	20	18
03	-11	-7	0	9	14	19	23	24	24	23
04	-15	-12	-4	6	17	22	24	23	22	21
05	-15	-10	-3	4	14	21	25	26	26	24
06	-11	-8	-1	8	16	22	26	26	25	23
07	-13	-9	0	9	17	22	26	26	25	23
08	-16	-13	-7	2	10	18	23	24	22	21
09	-14	-10	-2	8	17	22	26	26	24	21
10	-14	-12	-4	8	17	24	27	27	23	21
11	-14	-9	0	10	17	22	24	24	22	22
12	-15	-12	-2	8	16	22	25	24	21	18
13	-16	-15	-9	0	11	19	24	25	23	20
14	-19	-16	-7	2	13	19	24	24	23	20
15	-15	-13	-5	5	17	26	30	30	30	28
16	-13	-11	-4	6	15	22	25	24	24	24
17	-19	-14	-7	2	13	21	26	28	28	27
18	-16	-12	-4	1	10	17	23	25	24	23
$\overline{X}$	-15	-12	-4	6	15	21	25	25	24	22
SD	2	2	3	3	2	2	2	2	2	3

KEY: + = Flexion − = Extension

Table 24. Mean Hip Angle (°) during Stair Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	45	41	36	30	24	16	9	3	-1	-3	-5
02	42	38	33	26	19	11	6	1	-2	-4	-7
03	41	38	33	26	18	11	7	4	2	0	-2
04	45	43	39	34	30	25	19	13	7	3	0
05	43	40	37	33	28	23	18	13	9	6	4
06	44	41	36	31	25	21	15	10	6	3	0
07	44	40	33	28	20	14	8	3	-2	-6	-7
08	38	34	28	22	17	12	7	3	0	-3	-5
09	40	35	29	22	17	12	6	1	-2	-4	-5
10	37	35	30	27	24	20	15	10	5	2	0
11	43	40	36	32	26	19	13	6	2	-1	-3
12	39	37	32	27	22	16	10	4	0	-3	-6
13	33	30	25	18	13	7	2	-2	-5	-6	-8
14	48	43	36	29	23	16	9	3	0	-8	-9
15	52	49	45	40	34	29	22	16	11	7	4
16	40	38	34	29	24	17	12	7	3	0	-2
17	43	40	36	30	25	17	12	7	2	-3	-5
18	36	32	27	22	16	9	2	-3	-7	-9	-10
$\overline{X}$	42	39	34	28	23	16	11	6	2	-2	-4
SD	4	4	5	5	5	6	6	5	5	4	4

KEY: + = Flexion − = Extension

Table 24 (cont.). Mean Hip Angle (°) during Stair Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-7	-6	2	12	24	33	41	44	46	44
02	-6	0	8	18	28	36	44	47	47	43
03	-3	-5	-1	8	18	27	38	44	45	43
04	-2	-3	1	9	19	28	37	42	43	42
05	0	-3	-1	8	21	31	39	43	44	42
06	-2	0	8	17	29	36	44	48	49	47
07	-6	3	14	25	34	41	48	52	53	50
08	-7	-3	4	13	21	28	35	39	40	37
09	-4	2	11	14	28	36	43	46	46	44
10	0	3	9	18	30	38	44	45	41	37
11	-4	-1	6	17	29	37	44	48	49	47
12	-6	-3	5	14	23	33	40	44	45	43
13	-8	-6	1	9	19	27	35	40	41	38
14	-6	2	13	26	36	44	50	52	50	46
15	2	2	6	13	25	34	45	49	50	47
16	-4	-4	0	8	16	25	33	39	41	40
17	-4	1	10	19	30	38	44	47	47	45
18	-10	-7	-2	8	16	26	33	38	41	40
$\overline{X}$	-4	-2	5	14	25	33	41	45	45	43
SD	3	3	5	6	6	6	5	4	4	4

KEY: + = Flexion − = Extension

Table 25. Mean Hip Angle (°) during Stair Descent (at 5% intervals of the RGC)

SUBJECT	0	5	10	15	20	25	30	35	40	45	50
	INTERVALS OF RIGHT GAIT CYCLE										
01	11	11	13	13	13	10	9	9	10	12	17
02	6	6	7	7	7	6	6	6	8	11	17
03	14	11	8	7	6	5	5	7	11	14	15
04	12	10	10	12	13	11	8	8	9	10	12
05	17	16	16	19	20	19	18	18	18	18	19
06	13	13	14	17	17	15	12	11	11	11	14
07	13	13	14	15	15	15	14	14	14	13	15
08	4	4	5	7	8	7	7	6	6	7	10
09	17	16	16	17	16	13	12	10	10	11	13
10	15	14	14	16	18	18	18	17	16	17	21
11	19	18	18	20	20	18	17	17	17	18	19
12	10	8	7	7	7	5	4	4	6	9	12
13	11	10	10	11	10	9	9	10	11	13	15
14	10	9	10	11	10	8	6	5	6	9	12
15	15	13	13	15	17	17	17	15	15	14	16
16	15	14	14	15	15	14	12	11	11	13	15
17	15	13	12	13	15	15	15	14	13	13	15
18	17	16	16	17	17	15	13	12	12	14	17
$\bar{X}$	13	12	12	13	14	12	11	11	11	13	15
SD	4	4	4	4	5	5	5	4	4	3	3

KEY: + = Flexion − = Extension

Table 25 (cont.). Mean Hip Angle (°) during Stair Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	21	28	32	32	29	24	18	14	10	8
02	24	28	31	31	28	23	16	10	7	6
03	17	24	32	37	37	33	26	20	16	11
04	16	23	31	36	35	31	25	19	15	11
05	22	29	36	39	39	37	32	24	19	16
06	21	31	39	41	38	33	27	21	17	15
07	18	26	33	37	36	32	27	20	16	11
08	15	18	23	25	23	21	17	12	8	5
09	18	25	30	32	32	30	26	21	16	13
10	28	34	39	39	36	30	25	18	15	13
11	22	27	36	39	40	38	32	27	21	17
12	14	20	27	32	31	28	23	17	12	8
13	18	24	31	35	35	32	28	22	16	12
14	16	23	30	34	33	30	24	17	12	8
15	19	27	34	38	39	36	29	23	17	14
16	18	23	29	33	24	32	28	23	18	15
17	20	28	34	38	38	35	26	20	15	11
18	23	29	35	35	33	29	24	20	16	14
$\overline{X}$	19	26	32	35	34	31	25	19	15	12
SD	4	4	4	4	5	5	4	4	4	3

KEY: + = Flexion − = Extension

Table 26. Mean Hip Angle (°) during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	43	43	41	36	30	21	12	4	−2	−9	−14
02	41	39	34	25	17	8	1	−3	−7	−10	−13
03	33	34	32	27	20	14	9	6	3	−1	−5
04	47	46	42	35	28	20	10	2	−4	−8	−13
05	44	43	39	34	28	21	13	6	0	−5	−9
06	54	53	47	38	28	19	9	1	−7	−12	−14
07	54	53	49	42	34	26	17	8	0	−7	−11
08	49	47	43	33	25	15	7	−2	−9	−14	−17
09	45	44	39	35	27	20	12	5	0	−4	−8
10	51	48	43	37	30	19	10	2	−4	−9	−13
11	47	45	41	34	27	18	11	4	−2	−7	−9
12	40	37	33	27	21	14	9	2	−2	−6	−9
13	36	32	27	19	11	4	−2	−6	−9	−11	−14
14	46	44	37	30	22	13	5	−4	−10	−14	−18
15	48	49	50	47	43	36	27	17	10	3	−3
16	47	46	43	37	29	21	12	4	−2	−8	−13
17	49	47	43	39	31	21	11	2	−8	−16	−21
18	40	39	34	28	20	10	2	−5	−10	−14	−17
$\bar{X}$	45	44	40	34	26	18	10	3	−4	−8	−12
SD	6	6	6	7	7	7	6	5	5	5	5

KEY: + = Flexion − = Extension

Table 26 (cont.). Mean Hip Angle (°) during Ramp Climbing (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	-18	-20	-13	-1	14	26	37	42	46	47
02	-12	-6	1	10	18	28	35	42	46	49
03	-8	-9	-5	3	14	24	33	38	40	39
04	-15	-14	-5	8	21	33	41	48	51	51
05	-10	-7	0	10	19	28	34	38	39	38
06	-14	-10	1	12	25	37	44	48	48	44
07	-14	-16	-14	-6	7	21	34	45	52	54
08	-18	-19	-13	-5	8	17	29	39	43	44
09	-11	-12	-9	-2	8	19	29	36	42	45
10	-15	-14	-8	3	20	32	43	50	48	44
11	-9	-6	3	13	23	30	36	39	39	36
12	-11	-9	-2	9	20	29	38	43	44	41
13	-16	-15	-10	-3	8	18	26	33	36	34
14	-19	-16	-6	6	17	28	36	42	43	39
15	-9	-12	-11	-1	13	29	42	50	53	54
16	-16	-15	-8	2	13	24	33	38	40	39
17	-23	-19	-7	8	20	31	42	49	51	51
18	-18	-15	-4	5	17	27	36	40	41	38
$\bar{X}$	-14	-13	-6	4	16	27	36	42	45	44
SD	4	4	5	6	6	5	5	5	5	6

KEY: + = Flexion − = Extension

Table 27. Mean Hip Angle (°) during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF THE RIGHT GAIT CYCLE										
SUBJECT	0	5	10	15	20	25	30	35	40	45	50
01	15	15	15	15	14	13	12	11	12	8	7
02	12	13	15	16	14	11	8	6	4	2	2
03	17	17	16	16	15	13	12	11	11	10	9
04	15	16	18	20	20	18	14	11	9	7	5
05	18	19	20	22	21	18	16	13	10	8	7
06	14	15	18	20	20	18	14	10	7	5	1
07	19	20	25	23	23	20	17	14	12	10	8
08	9	10	12	12	11	6	2	-2	-5	-8	-10
09	18	17	17	16	13	10	9	7	6	4	3
10	18	19	20	21	20	18	15	13	11	9	5
11	18	19	22	25	24	21	18	14	12	9	7
12	13	13	15	16	15	12	9	7	6	5	3
13	13	13	14	14	12	9	8	6	6	6	6
14	12	12	14	14	13	9	4	0	-5	-9	-12
15	20	22	23	23	21	16	12	6	2	-3	-8
16	19	19	21	20	17	14	10	6	3	0	-4
17	20	22	26	27	25	20	14	8	1	-3	-5
18	18	19	21	22	21	19	15	11	9	7	5
$\overline{X}$	16	17	18	17	18	15	12	8	6	4	2
SD	3	4	4	4	4	5	4	4	5	6	7

KEY: + = Flexion − = Extension

Table 27 (cont.). Mean Hip Angle (°) during Ramp Descent (at 5% intervals of the RGC)

	INTERVALS OF RIGHT GAIT CYCLE									
SUBJECT	55	60	65	70	75	80	85	90	95	100
01	8	12	17	20	22	22	19	16	15	15
02	4	8	15	20	22	22	20	16	13	13
03	8	9	13	17	20	21	20	18	17	16
04	7	12	19	24	26	25	21	16	14	14
05	9	13	20	26	29	29	27	23	20	18
06	-1	0	5	11	16	20	22	21	20	18
07	9	13	18	23	26	28	27	24	22	21
08	-10	-7	-2	4	9	12	13	13	11	10
09	2	5	10	15	18	20	19	17	15	14
10	3	3	9	14	20	23	23	20	19	19
11	7	10	16	21	23	24	22	19	18	17
12	4	9	15	20	21	20	17	13	11	11
13	6	9	14	17	20	21	20	18	16	13
14	-11	-5	5	11	16	18	17	14	11	11
15	-9	-5	3	12	19	24	25	23	20	19
16	-5	-4	3	11	19	22	22	19	18	19
17	-3	-9	-4	4	14	20	23	22	20	19
18	5	7	13	19	24	27	27	24	21	20
$\overline{X}$	2	5	11	18	20	22	21	19	17	16
SD	7	7	7	5	5	4	4	4	4	3

KEY: + = Flexion − = Extension

Table 28. Mean Amount of Total Muscle Activity during Level Walking (% of MVC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	27	29	2	2
02	22	19	8	5
03	13	9	7	12
04	12	16	2	3
05	15	15	6	8
06	5	5	0	1
07	7	9	3	3
08	6	6	1	1
09	4	8	3	4
10	6	12	5	4
11	12	16	3	3
12	13	12	1	1
13	3	4	2	2
14	8	7	1	1
15	12	10	2	3
16	20	18	0	0
17	19	13	11	4
18	15	27	3	7
$\bar{X}$	12	13	3	4
SD	7	7	3	3

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 29. Mean Amount of Total Muscle Activity during Stair Climbing (% of MVC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	17	24	0	3
02	36	47	1	3
03	18	25	8	2
04	31	25	5	4
05	23	52	3	4
06	15	11	3	1
07	5	13	3	1
08	15	14	2	1
09	5	3	9	8
10	3	20	6	3
11	11	31	4	2
12	33	24	2	5
13	13	9	2	3
14	35	34	1	1
15	23	35	1	2
16	41	74	1	1
17	24	35	2	5
18	40	55	4	3
$\overline{X}$	22	30	3	3
SD	12	18	2	2

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spina LRA = Left Rectus Abdominis

Table 30. Mean Amount of Total Muscle Activity during Stair Descent (% of MVC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	18	8	3	1
02	24	19	2	1
03	9	8	7	22
04	15	8	5	1
05	12	5	5	2
06	4	4	0	2
07	6	12	1	3
08	6	8	2	2
09	7	15	4	8
10	8	8	7	4
11	21	7	1	4
12	14	14	2	1
13	1	3	2	2
14	10	20	1	0
15	11	5	1	1
16	26	9	1	0
17	16	11	6	3
18	13	9	9	4
$\overline{X}$	12	10	3	3
SD	7	5	3	5

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 31. Mean Amount of Total Electrical Activity during Ramp Climbing (% of MVC)

	MUSCLE			
SUBJECT	RES (N=18)	LES (N=18)	RRA (N=18)	LRA (N=18)
01	20	30	0	3
02	35	53	1	3
03	24	27	10	2
04	25	46	4	9
05	23	46	2	4
06	16	13	3	1
07	10	10	0	1
08	13	10	1	1
09	9	3	9	6
10	1	17	7	1
11	10	31	4	2
12	30	20	2	4
13	9	6	1	3
14	32	26	1	1
15	18	28	1	2
16	45	87	2	1
17	28	38	4	7
18	45	44	3	2
$\bar{X}$	22	30	3	3
SD	12	21	3	2

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 32. Mean Amount of Total Muscle Activity during Ramp Descent (% of MVC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	23	12	3	1
02	15	17	7	6
03	15	10	7	21
04	14	14	1	3
05	3	3	7	6
06	3	3	1	7
07	4	3	1	0
08	7	8	2	2
09	5	16	2	8
01	7	16	8	3
11	20	9	1	4
12	14	11	2	1
13	1	1	3	3
14	3	10	1	1
15	13	8	7	6
16	23	13	4	4
17	37	30	10	4
18	17	13	11	3
$\overline{X}$	12	11	4	5
SD	9	7	3	5

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 33. Mean Amount of Total Erector Spinae Muscle Activity in All Types of Locomotion (% of MVC)

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	28	22	13	25	18
02	20	41	21	44	16
03	11	22	9	25	13
04	14	28	12	35	14
05	15	37	8	35	3
06	5	13	4	14	3
07	8	9	9	10	3
08	6	15	7	12	8
09	6	4	11	6	11
10	9	12	8	9	12
11	14	21	14	21	15
12	13	28	14	25	13
13	3	11	2	8	1
14	8	34	15	29	6
15	11	29	8	23	11
16	19	57	18	66	18
17	16	30	13	33	33
18	21	48	6	45	15
$\overline{X}$	13	26	11	26	12
SD	7	14	5	16	8

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 34. Mean Amount of Total Rectus Abdominis Muscle Activity in All Types of Locomotion (% of MVC)

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	2	2	2	2	2
02	6	2	2	2	7
03	9	5	15	6	14
04	3	4	3	6	2
05	7	4	3	3	7
06	0	2	1	2	4
07	3	2	2	0	1
08	1	1	2	1	2
09	4	8	6	8	5
10	5	5	5	4	6
11	3	3	3	3	2
12	1	3	2	3	1
13	2	2	2	2	3
14	1	1	1	1	1
15	3	2	1	2	7
16	0	1	0	1	4
17	8	4	4	6	7
18	5	3	6	2	7
$\overline{X}$	4	3	3	3	5
SD	3	2	3	2	3

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 35. Mean Point of Peak 1 EMG Activity during Level Walking (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=11)	LRA(N=6)
01	6	7	-	-
02	8	5	23	23
03	3	3	-	1
04	5	7	-	-
05	3	6	-	17
06	5	4	12	-
07	5	5	-	-
08	4	6	16	-
09	7	5	8	-
10	5	5	-	-
11	7	6	15	-
12	8	7	19	21
13	5	9	8	-
14	7	6	18	-
15	5	6	15	-
16	6	11	16	16
17	6	4	17	11
18	10	11	-	-
$\overline{X}$	6	6	15	15
SD	2	2	4	8

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 36. Mean Point of Peak 1 EMG Activity during Stair Climbing (% of RGC)

	MUSCLE			
Subject	RES(N=18)	LES(N=18)	RRA(N=16)	LRA(N=15)
01	4	11	7	16
02	2	10	8	19
03	5	9	–	–
04	5	9	15	10
05	2	8	–	–
06	2	8	9	7
07	4	8	9	11
08	4	19	8	19
09	3	20	9	10
10	7	11	9	6
11	16	10	11	17
12	3	11	14	15
13	3	10	12	–
14	5	11	3	9
15	6	11	12	13
16	3	9	21	12
17	4	8	8	12
18	3	10	10	16
$\overline{X}$	5	11	10	13
SD	3	3	4	4

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 37. Mean Point of Peak 1 EMG Activity during Stair Descent (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=12)	LRA(N=15)
01	12	15	-	11
02	11	3	6	3
03	12	3	-	-
04	9	6	8	12
05	5	5	12	12
06	15	5	-	-
07	12	15	-	-
08	19	14	4	3
09	6	25	25	2
10	11	11	-	2
11	18	8	11	16
12	4	4	3	8
13	3	16	8	16
14	10	23	10	6
15	25	6	13	14
16	13	20	9	14
17	15	14	-	15
18	13	7	11	7
$\overline{X}$	12	11	10	9
SD	6	7	6	5

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 38. Mean Point of Peak 1 EMG Activity during Ramp Climbing (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=15)	LRA(N=14)
01	9	13	1	24
02	6	15	6	-
03	10	11	3	14
04	7	9	-	-
05	5	11	-	-
06	3	9	10	9
07	6	10	6	8
08	5	10	8	10
09	6	4	-	10
10	6	12	14	9
11	5	11	27	23
12	3	11	15	13
13	4	9	19	-
14	5	10	2	2
15	7	14	17	18
16	4	8	13	15
17	2	4	8	16
18	4	14	18	7
$\overline{X}$	5	10	11	13
SD	2	3	7	6

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 39. Mean Point of Peak 1 EMG Activity during Ramp Descent (% of RGC)

SUBJECT	RES(N=18)	LES(N=18)	RRA(N=13)	LRA(N=15)
	MUSCLE			
01	5	7	2	2
02	5	5	-	16
03	2	2	-	7
04	9	4	14	10
05	5	4	14	18
06	4	4	-	-
07	9	4	-	-
08	10	9	11	4
09	16	4	3	9
10	5	3	29	20
11	16	3	13	12
12	15	5	22	5
13	7	17	10	11
14	7	5	22	2
15	7	5	21	-
16	19	19	18	18
17	7	3	4	6
18	11	7	-	6
$\overline{X}$	9	6	14	10
SD	5	5	8	6

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 40. Mean Point of Peak 1 EMG Activity of Right Erector Spinae Muscle (% of RGC)

SUBJECT	TYPE OF LOCOMOTION L	SC	SD	RC	RD
01	6	4	12	9	5
02	8	2	11	6	5
03	3	5	12	10	2
04	5	5	9	7	9
05	3	2	5	5	5
06	5	2	15	3	4
07	5	4	12	6	9
08	4	4	19	5	10
09	7	3	6	6	16
10	5	7	11	6	5
11	7	16	18	5	16
12	8	3	4	3	15
13	5	3	3	4	7
14	7	5	10	5	7
15	5	6	25	7	7
16	6	3	13	4	19
17	6	4	15	2	7
18	10	3	13	4	11
$\bar{X}$	6	5	12	5	9
SD	2	3	6	2	5

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 41. Mean Point of Peak 1 EMG Activity of Left Erector Spinae Muscle (% of RGC)

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	7	11	15	13	7
02	5	10	3	15	5
03	3	9	3	11	2
04	7	9	6	9	4
05	6	8	5	11	4
06	4	8	5	9	4
07	5	8	15	10	4
08	6	19	14	10	9
09	5	20	25	4	4
10	5	11	11	12	3
11	6	10	8	11	3
12	7	11	4	11	5
13	9	10	16	9	17
14	6	11	23	10	5
15	6	11	6	14	5
16	11	9	20	8	19
17	4	8	14	4	3
18	11	10	7	14	7
$\overline{X}$	6	11	11	10	6
SD	2	3	7	3	5

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 42. Mean Point of Peak 1 EMG Activity of Right Rectus Abdominis Muscle (% of RGC)

SUBJECT	TYPE OF LOCOMOTION L	SC	SD	RC	RD
01	–	7	–	21	2
02	–	8	6	6	–
03	23	–	–	3	–
04	–	15	8	–	14
05	–	–	12	–	14
06	12	9	–	10	–
07	–	9	–	6	–
08	16	8	4	8	11
09	8	9	25	–	3
10	–	9	–	14	29
11	15	11	11	27	13
12	19	14	3	15	22
13	8	12	8	19	10
14	18	3	10	2	22
15	15	12	13	17	21
16	16	21	9	13	18
17	17	8	–	8	4
18	–	10	11	18	–
$\overline{X}$	15	10	10	11	14
SD	4	4	6	7	8

KEY: L = Level Walking
SC = Stair Climbing   RC = Ramp Climbing
SD = Stair Descent   RD = Ramp Descent

Table 43. Mean Point of Peak 1 EMG Activity of Left Rectus Abdominis Muscle (% of RGC)

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	–	16	11	24	2
02	23	19	3	–	16
03	1	–	–	14	7
04	–	10	12	–	10
05	17	–	12	–	18
06	–	7	–	9	–
07	–	11	–	8	–
08	–	19	3	10	4
09	–	10	2	10	9
10	–	6	2	9	20
11	–	17	16	23	12
12	21	15	8	13	5
13	–	–	16	–	11
14	–	9	6	2	2
15	–	13	14	18	–
16	16	12	14	15	18
17	11	12	15	16	6
18	–	16	7	7	6
$\overline{X}$	15	13	9	13	10
SD	8	4	5	6	6

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 44. Mean Point of Peak 2 EMG Activity during Level Walking (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	56	56	37	43
02	57	57	48	51
03	51	49	45	41
04	56	57	46	44
05	54	52	62	60
06	56	51	58	45
07	54	55	28	23
08	56	56	66	32
09	55	56	49	32
10	56	55	33	36
11	59	61	31	36
12	55	49	68	65
13	59	58	40	31
14	57	56	45	42
15	56	55	45	68
16	58	58	67	60
17	57	57	47	45
18	62	58	63	39
$\overline{X}$	56	55	49	44
SD	2	3	13	12

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 45. Mean Point of Peak 2 EMG Activity during Stair Climbing (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=17)	LRA(N=18)
01	55	51	48	59
02	55	49	51	54
03	56	55	51	30
04	60	53	-	62
05	61	55	63	66
06	58	47	57	57
07	56	47	56	56
08	58	50	65	63
09	60	68	57	52
10	60	44	57	47
11	61	42	63	67
12	59	45	63	61
13	57	46	47	43
14	57	51	57	56
15	62	45	62	62
16	59	51	64	64
17	58	46	57	53
18	59	46	58	63
$\bar{X}$	58	50	57	56
SD	2	6	6	9

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 46. Mean Point of Peak 2 EMG Activity during Stair Descent (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=17)	LRA(N=17)
01	55	46	47	40
02	45	35	40	43
03	53	53	–	–
04	53	39	50	46
05	35	23	50	60
06	56	64	53	48
07	53	48	42	50
08	58	68	47	49
09	51	60	70	44
10	51	54	39	38
11	56	53	55	66
12	57	37	51	53
13	52	58	57	68
14	74	62	66	62
15	53	60	51	57
16	56	56	50	60
17	54	75	49	54
18	64	56	46	43
$\bar{X}$	54	53	51	52
SD	8	13	8	9

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 47. Mean Point of Peak 2 EMG Activity during Ramp Climbing (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=18)	LRA(N=18)
01	59	58	40	56
02	64	61	66	71
03	54	41	58	44
04	57	45	39	55
05	59	52	60	53
06	56	47	50	49
07	60	56	34	41
08	55	54	49	38
09	58	48	31	55
10	59	55	53	52
11	58	47	69	72
12	57	49	63	61
13	62	58	51	35
14	58	54	54	50
15	63	61	56	55
16	58	57	51	64
17	55	52	52	53
18	60	56	60	56
$\bar{X}$	58	53	52	53
SD	3	6	11	10

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 48. Mean Point of Peak 2 EMG Activity during Ramp Descent (% of RGC)

	MUSCLE			
SUBJECT	RES(N=18)	LES(N=18)	RRA(N=17)	LRA(N=18)
01	55	55	41	43
02	53	29	30	46
03	56	77	43	37
04	44	54	51	35
05	55	42	58	58
06	54	55	42	40
07	53	49	43	45
08	56	60	58	57
09	54	52	51	46
10	52	54	58	60
11	58	60	38	55
12	55	41	62	44
13	50	52	52	32
14	54	37	62	50
15	54	61	42	56
16	57	58	–	60
17	53	53	70	53
18	57	57	39	48
$\overline{X}$	54	53	49	49
SD	3	11	11	9

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 49. Mean Point of Peak 2 EMG Activity of Right Erector Spinae (% of RGC) in All Types of Locomotion

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	56	55	55	59	55
02	57	55	45	64	53
03	51	56	53	54	56
04	56	60	53	57	44
05	54	61	35	59	55
06	56	58	56	56	54
07	54	56	53	60	53
08	56	58	58	55	56
09	55	60	51	58	54
10	56	60	51	59	52
11	59	61	56	58	58
12	55	59	57	57	55
13	59	57	52	62	50
14	57	57	74	58	54
15	56	62	53	63	54
16	58	59	56	58	57
17	57	58	54	55	53
18	62	59	64	60	57
$\overline{X}$	56	58	54	58	54
SD	2	2	8	3	3

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 50. Mean Point of Peak 2 EMG Activity of Left Erector Spinae (% of RGC) in All Types of Locomotion

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	56	51	46	58	55
02	57	49	35	61	29
03	49	55	53	41	77
04	57	53	39	45	54
05	52	55	23	52	42
06	51	47	64	47	55
07	55	47	48	56	49
08	56	50	68	54	60
09	56	68	60	48	52
10	55	44	57	55	54
11	61	42	53	47	60
12	49	45	37	49	41
13	58	46	58	58	52
14	56	51	62	54	37
15	55	45	60	61	61
16	58	51	56	57	58
17	57	46	75	52	53
18	58	46	56	56	57
$\overline{X}$	55	50	53	53	53
SD	3	6	13	6	11

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 51. Mean Point of Peak 2 EMG Activity of Right Rectus Abdominis (% of RGC) in All Types of Locomotion

	TYPES OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	37	48	47	40	41
02	48	51	40	66	30
03	45	51	–	58	43
04	46	–	50	39	51
05	62	63	50	60	58
06	58	57	53	50	42
07	28	56	42	34	43
08	66	65	47	49	58
09	49	57	70	31	51
10	33	57	39	53	58
11	31	63	55	69	38
12	68	63	51	63	62
13	46	47	57	51	52
14	45	57	66	54	62
15	45	62	51	56	42
16	67	64	50	51	–
17	47	57	49	52	70
18	63	58	46	60	39
$\bar{X}$	49	57	51	52	49
SD	13	6	8	11	11

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 52. Mean Point of Peak 2 EMG Activity of Left Rectus Abdominis (% of RGC) in All Types of Locomotion

SUBJECT	TYPE OF LOCOMOTION L	SC	SD	RC	RD
01	43	59	40	56	43
02	51	54	43	71	46
03	41	30	–	44	37
04	44	62	46	55	35
05	60	66	60	53	58
06	45	57	48	49	40
07	23	56	50	41	45
08	32	63	49	38	57
09	32	52	44	55	46
10	36	47	38	52	60
11	36	67	66	72	55
12	65	61	53	61	44
13	31	43	68	35	32
14	42	56	62	50	50
15	68	62	57	55	56
16	60	64	60	64	60
17	45	53	54	53	53
18	39	63	43	56	48
$\overline{X}$	44	56	52	53	49
SD	12	9	9	10	9

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 53. Mean Point of Peak 3 EMG Activity during Level Walking (% of RGC)

SUBJECT	RES(N=4)	LES(N=0)	RRA(N=16)	LRA(N=15)
		MUSCLE		
01	–	–	88	82
02	–	–	87	79
03	–	–	91	96
04	–	–	88	86
05	–	–	–	–
06	97	–	93	63
07	–	–	84	83
08	–	–	91	72
09	85	–	86	89
10	–	–	77	79
11	83	–	61	63
12	88	–	91	96
13	–	–	90	88
14	–	–	92	88
15	–	–	93	–
16	–	–	–	–
17	–	–	89	70
18	–	–	85	80
$\overline{X}$	88	–	87	81
SD	6	–	8	10

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA= Left Rectus Abdominis

Table 54. Mean Point of Peak 3 EMG Activity during Stair Climbing (% Of RGC)

SUBJECT	MUSCLE RES(N=13)	LES(N=3)	RRA(N=11)	LRA(N=8)
01	84	–	86	85
02	–	–	93	88
03	73	–	89	–
04	95	95	–	–
05	–	–	–	–
06	93	–	96	–
07	–	–	–	–
08	98	–	–	–
09	89	97	98	–
10	81	–	96	–
11	96	87	87	87
12	88	–	–	83
13	83	–	89	88
14	–	–	–	98
15	94	–	–	–
16	96	–	94	95
17	96	–	97	–
18	–	–	93	91
$\overline{X}$	90	93	93	89
SD	8	5	4	5

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 55. Mean Position of Peak 3 EMG Activity during Stair Descent (% of RGC)

	MUSCLE			
SUBJECT	RES(N=8)	LES(N=3)	RRA(N=16)	LRA(N=10)
01	–	76	88	84
02	–	81	96	96
03	–	99	85	84
04	74	–	95	–
05	74	–	–	88
06	–	–	89	90
07	72	–	95	–
08	–	–	–	–
09	95	–	94	–
10	75	–	91	84
11	90	–	98	94
12	89	–	80	92
13	74	–	87	91
14	–	–	98	–
15	–	–	85	92
16	–	–	67	–
17	–	–	95	–
18	–	–	85	–
$\overline{X}$	80	85	89	90
SD	9	12	8	4

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 56. Mean Point of Peak 3 EMG Activity during Ramp Climbing (% of RGC)

SUBJECT	MUSCLE RES(N=4)	LES(N=1)	RRA(N=10)	LRA(N=9)
01	–	–	–	–
02	–	–	–	–
03	–	–	95	98
04	–	–	–	–
05	–	–	–	–
06	–	–	96	81
07	–	–	77	93
08	–	–	–	–
09	77	96	86	–
10	85	–	97	97
11	–	–	–	–
12	84	–	–	82
13	–	–	90	–
14	–	–	93	91
15	87	–	99	99
16	–	–	84	97
17	–	–	–	–
18	–	–	83	85
$\overline{X}$	83	96	90	91
SD	4	0	7	7

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 57. Mean Point of Peak 3 EMG Activity during Ramp Descent (% of RGC)

	MUSCLE			
SUBJECT	RES(N=8)	LES(N=3)	RRA(N=15)	LRA(N=14)
01	–	–	86	86
02	–	–	76	82
03	–	–	90	92
04	81	–	83	93
05	–	–	–	–
06	–	–	91	91
07	98	–	84	87
08	89	88	96	–
09	94	–	94	97
10	96	–	83	85
11	78	–	77	88
12	86	–	95	68
13	96	81	79	71
14	–	85	–	98
15	–	–	95	94
16	–	–	–	–
17	–	–	98	–
18	–	–	85	81
$\bar{X}$	90	85	87	87
SD	8	4	7	9

KEY: RES = Right Erector Spinae RRA = Right Rectus Abdominis
LES = Left Erector Spinae LRA = Left Rectus Abdominis

Table 58. Mean Point of Peak 3 EMG Activity of Right Erector Spinae (% of RGC) in All Types of Locomotion

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	–	84	–	–	–
02	–	–	–	–	–
03	–	73	–	–	–
04	–	95	74	–	81
05	–	–	74	–	–
06	97	93	–	–	–
07	–	–	72	–	98
08	–	98	–	–	89
09	85	89	95	77	94
10	–	81	75	85	96
11	83	96	90	–	78
12	88	88	89	84	86
13	–	83	74	–	96
14	–	–	–	–	–
15	–	94	–	87	–
16	–	96	–	–	–
17	–	96	–	–	–
18	–	–	–	–	–
$\bar{X}$	88	90	80	83	90
SD	6	8	9	4	8

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 59. Mean Point of Peak 3 EMG Activity of Left Erector Spinae (% of RGC) in All Types of Locomotion

	TYPE OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	–	–	76	–	–
02	–	–	81	–	–
03	–	–	99	–	–
04	–	95	–	–	–
05	–	–	–	–	–
06	–	–	–	–	–
07	–	–	–	–	–
08	–	–	–	–	88
09	–	97	–	96	–
10	–	–	–	–	–
11	–	87	–	–	–
12	–	–	–	–	–
13	–	–	–	–	81
14	–	–	–	–	85
15	–	–	–	–	–
16	–	–	–	–	–
17	–	–	–	–	–
18	–	–	–	–	–
$\overline{X}$	–	93	85	96	85
SD	–	5	12	0	4

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 60. Mean Point of Peak 3 EMG Activity of Right Rectus Abdominis (% of RGC) in All Types of Locomotion

SUBJECT	TYPE OF LOCOMOTION L	SC	SD	RC	RD
01	88	86	88	–	86
02	87	93	96	–	76
03	91	89	85	95	90
04	88	–	95	–	83
05	–	–	–	–	–
06	93	96	89	96	91
07	84	–	95	77	84
08	91	–	–	–	96
09	86	98	94	86	94
10	77	96	91	97	83
11	61	87	98	–	77
12	91	–	80	–	95
13	90	89	87	90	79
14	92	–	98	93	–
15	93	–	85	99	95
16	–	94	67	84	–
17	89	97	95	–	98
18	85	93	85	83	85
$\overline{X}$	87	93	89	90	87
SD	8	4	8	7	7

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent

Table 61. Mean Point of Peak 3 EMG Activity of Left Rectus Abdominis (% of RGC) in All Types of Locomotion

	TYPES OF LOCOMOTION				
SUBJECT	L	SC	SD	RC	RD
01	82	85	84	–	86
02	79	88	96	–	82
03	96	–	84	98	92
04	86	–	–	–	93
05	–	–	88	–	–
06	63	–	90	81	91
07	83	–	–	93	89
08	72	–	–	–	–
09	89	–	–	–	97
10	79	–	84	97	85
11	63	87	94	–	88
12	96	83	92	82	68
13	88	88	91	–	71
14	88	98	–	91	98
15	–	–	92	99	94
16	–	95	–	97	–
17	70	–	–	–	–
18	80	91	–	85	81
$\bar{X}$	81	89	90	91	87
SD	10	5	4	7	9

KEY: L = Level Walking
SC = Stair Climbing RC = Ramp Climbing
SD = Stair Descent RD = Ramp Descent