A STUDY OF THE BIOMECHANICAL PERFORMANCE OF KNEE-ANKLE-FOOT ORTHOSES IN NORMAL AMBULATORY ACTIVITIES.

by

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Thesis submitted for the Degree of
Doctor of Philosophy in Bioengineering

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January 1985

ACKNOWLEDGEMENTS.

I wish to express my gratitute and appreciation to Professor J.P. Paul for the constructive criticism, the general guidance, and the provision of facilities to undertake this research. Thanks must also be given to Mr. S.E. Solomonidis for his advice and supervision for the past three years.

I am also deeply indebted to Mr. W.D. Spence for his invaluable discussion on lower limb orthotics, and for the fabrication of the test orthoses.

I would also like to extend my appreciation to Mr. D. Robb for machining the transducers, Mr. A.J. Tullis for helping in the strain gauging, and Miss C. Szary for assisting the patient tests.

I am also extremely thankful to my parents for their financial support and encouragements, and to my wife, Ann, for her humour and understanding.

Thanks is also due to David and Morag Black for the use of their photocopier.

Last but not least, I would like to acknowledge the financial assistance given by the Issac Sclar Scholarship, and the ORS Award from the Committee of Vice-Chancellors and Principals of the Universities of the United Kingdom.

ABSTRACT.

Knee-ankle-foot orthoses (KAFO) are prescribed for the management of lower limb instabilies and to reduce loading of anatomical structures. The design of these orthoses has been mainly empirical due to a lack of quantitative orthotic loading data. This project aims to study the loads transmitted by KAFOs when being used in a number of ambulatory activities, with a view to establish realistic design criteria for future KAFO development.

The work undertaken and presented in this thesis includes the design and construction of multi-channel transducers for the monitoring of load actions at the orthotic uprights, and a set of simpler shear force transducers to measure the load actions at strap attachment points. A methodology for the measurement of orthotic loads in a number of ambulatory activities to simulate daily living situations was also devised.

The multi-channel transducers have a total length of 96 mm, and are capable of measuring 3 forces and 3 moments in orthogonal directions. They are able to withstand at least an axial load of 800N, and moments of 25 Nm and 35 Nm in the mediolateral and anterioposterior directions respectively (minimum safety factor of 2.0 based on yield stress).

In the ambulatory tests, orthotic loadings over 6 gait cycles were recorded while the patient performed the activities of walking in a straight line, walking round a double-bend, walking round a U-shaped path, stepping up and down a low platform, and ascending and descending stairs and slopes. In addition to the above, tests involving simultaneous recording of ground reaction loads with a force platform, kinematic data with cine cameras, and the orthotic loadings with the transducers were also performed.

Results of the analyses performed on 8 patients and a normal subject are presented. Graphs showing the variation with time of the orthotic loadings over 5 gait cycles are also presented. The critical bending loads were found to be 33 Nm in the anterioposterior direction, and 15 Nm mediolaterally on the KAFO uprights. It was also shown that this level of loadings can lead to fatigue failure.

C O N T E N T S.

INTRODUCTION.

The knee-ankle-foot orthosis (KAFO) which is more commonly known as the long-leg caliper, has played an active role in the management of lower limb instability since firstly reported by Hugh Owen Thomas in the nineteen century. However, the design of the orthoses have been mainly empirical because of a lack of scientific research relating to the load transmission of the orthoses when actually being used by patients. In order to prevent orthotic breakages or total failure, they were naturally overdesigned, resulting in orthoses that are heavier than they needed to be.

A review has shown that although several researchers have attempted to investigate the loading capabilities of the orthoses, they have failed to produce a comprehensive definition of the loading pattern on the orthoses. Anderson (1974,1977) investigated the axial loads as a function of the length of an ischial weight-bearing KAFO in a static situation. The axial load and the 2 shear forces transmitted by a KAFO were measured at a level just above the ankle joint by Kirkpatrick et al (1969), while Lehmann and coworkers (1970c) considered various KAFO designs with a view to determining their effectiveness in relieving load from anatomical structures. A slightly more comprehensive biomechanical analysis was performed by Lippert (1971). However, instead of instrumenting a standard KAFO, he designed a bulky test orthosis to simulate 19 different orthotic configurations for the purpose of measuring the loads transmitted by them. It is felt that this method of measurement creates an artificial situation which may result in unrealistic loading conditions.

A comprehensive KAFO biomechanical analysis system developed in the Bioengineering Unit of the University of Strathclyde was first reported by Trappitt (1979). The system consists of a set of 4 multi-channel load transducers, 24 'strain gauge amplifiers', and a multiplexer. The results obtained from 2 post-poliomyelitis patients were also reported by Trappitt.

The project described in this thesis aims to study the loads transmitted by KAFOs when being used in a number of ambulatory activities, with a view to establish realistic design criteria for lighter and more comfortable orthoses. An assessment of the methodology and equipment employed by Trappitt (1979) revealed that the transducers

(i.e. Mark I KAFO Transducers) had been working at their maximum loading capacity. It was therefore necessary to design and construct a new set of transducers. The patient testing procedures were also modified to include six different types of ambulatory activities, and for continuous measurement of orthotic loading over 6 or more gait cycles. The objectives of the project can be summarised as:

1. to design and construct a set of 4 multi-channel load transducers (Mark II KAFO Transducers), which were to be shorter and stronger than the Mark I transducers. These were to eventually replace the Mark I transducers in all future tests;
2. to design and construct a set of transducers for the measurement of shear forces at the strap connection points;
3. to develop a methodology for the measurement of loadings experienced by the KAFOs in a variety of ambulatory activities which simulate situations that a KAFO wearer may encounter in his/her daily activities;
4. to obtain a set of results for the orthotic loadings as measured from a number of patients, which may be used as design criteria for future KAFO development.

The content of this thesis is briefly described below: Chapter 1 contains a short introduction to the cause of lower limb disabilities and their consequences. This discussion is continued in Chapter 2, where the orthotic management of lower limb instabilities is briefly described. It includes discussion on the functions of orthoses, the terminology concerned, and a review of the various types of lower limb orthoses that are available. The principles of orthotic design, the prescription, and the patients' response to the KAFO are the subjects of discussion in Chapter 3. This chapter also contains a review of the orthotic load measurement systems employed by various researchers. Chapter 4 presents the design aspects of the KAFO load transducers, while Chapter 5 describes the methodology and instrumentation involved in this study. A brief discussion of the theoretical background and the analysis is given in Chapter 6. Chapter 7 presents the results obtained from 8 patients and a normal subject. Finally, the thesis ends in Chapter 8 with conclusions and recommendations for future work.

CHAPTER 1.

LOWER LIMB DISABILITIES.

1.1 INTRODUCTION.

The inability to walk with relative ease and to stand with adequate security are the principal handicaps encountered by individuals with lower limb incapacities. Restoration of these functions as completely as possible is the main aim of lower limb rehabilitation.

Lower limb disabilities can be caused by abnormalities or disorders in any of the three sub-divisions of the neuromusculoskeletal system, namely, the neurological system, the muscular system, and the skeletal system. These disabilities or deformities may be due to trauma, congenital, hereditary, or virus attack on the body. Although the aetiological origin of the deficiency may be different, the disability produced could be expected to be similar if the affected site in the neuromusculoskeletal system is the same. For example, the same type of flaccid paralysis is experienced irrespective of whether the lower motor neuron lesion is caused by trauma or poliomyelitis.

1.2 NEUROMUSCULOSKELETAL DISORDERS.

The locomotor or musculoskeletal system depends upon its voluntary muscles (or motors) to provide active coordinated movements. A wide variety of clinical disorders and injuries of the nervous system are manifested by disturbances of both form and function of the musculoskeletal system. The two systems are so closely interrelated that in certain circumstances, it may not be easy to classify a disorder to be under musculoskeletal or neuromuscular. For this reason, the discussion that follows will generalise these disorders to be of the neuromusculoskeletal nature. The types of disorders can be broadly classified under the follow headings:

1. common disorders of musculoskeletal system;
2. congenital abnormality;
3. inflammatory disorders;
4. degenerative disorders;
5. motor neuron disorders;
6. cerebovascular accidents; and
7. muscular disorders.

1.2.1 Common Disorders of Musculoskeletal System.

These are disorders of the musculoskeletal system due to lax ligaments, habitual postures, and aging. In children, the disorders may be considered in two main groups based on their underlying causes. Firstly, hypermobility of joints (joint laxity) caused by lax ligaments is relatively common in infancy, but as the child grows older, the condition improves with the ligaments becoming less lax. Flexible flat foot and knock knees (genu valgum) are common manifestation of hypermobile joints. The second type of disorders are speculated to be due to bending and torsional deformities of the growing bones caused by certain sleeping and sitting postures during childhood.

Bow leggedness (genu varum) is a common abnormality in children. Night splints are prescribed for the more serious cases while no treatment is usually given for mild disorders. In adults, these disorders are due to the normal ageing process in the musculoskeletal system which results in gradual decrease of muscle strength and joint motion.

1.2.2 Congenital Abnormalities.

Congenital abnormalities can be caused by a variety of factors including genetic defects and environmental influences, or a combination of the two. The abnormality is 'localised' when the effects are confined to either the foot, long bones, joints or the spine, and 'generalised' when it involves several parts of the musculoskeletal system. Localised abnormalities are manifested if musculoskeletal growth and development is arrested during foetal life or if the normal bony closure of the posterior part of the spinal canal is arrested as in spinal bifida. Talipes equinovarus (club foot) is a common abnormality of the foot, and can be treated with club foot orthoses, or with surgical boots. Congenital joint abnormality occurs when a joint is unstable or dislocated.

Spinal bifida is the commonest congenital abnormality of the spine, and it includes varying degrees of imcomplete bony closure of one or more vertibral arches, frequently in the lumbosacral region. The vertebral abnormality may be uncomplicated by neural defects as in Spina Bifida Occulta where the defect is an incidental radiological finding, or may be associated with meningocele, menigomyelocele, or

myelocele, which is the most serious defect with spinal cord and nerve roots lie completely exposed. Neurological deficit associated with the deformity is significant, ranging from mild muscle imbalance and sensory loss in the lower limb to flaccid paralysis with nerve root involvement, or spastic paralysis when spinal cord is involved. KAFO and HKAFO (see Section 2.5) are used in the orthopaedic treatment of flaccid paralysis in spinal bifida patients.

1.2.3 Inflammatory Disorders.

Inflammatory disorders of the bones and joints, as the name implies, includes a whole range of inflammatory diseases affecting the bones and joints of the body. Among these, rheumatoid arthritis stands out as the prime cause of inflammatory pain and disability. The disorder may involve almost any organ or tissue in the body, however, the main site of pathological process is the synovial membrane of the joints. The affected joints present themselves clinically as painful swelling, warm, and tender when touched. Movements are painful and restricted, with evidence of wasting in the associated muscles. Although no joint is immune, those frequently affected are the small joints of the fingers and the wrist joint.

1.2.4 Degenerative Disorders.

Degenerative disorders of joints and related structures are diseases associated with normal aging, occurring as degenerative arthrosis in synovial joints (osteo-arthritis). The changes tend to be irreversible and progressive because of the limited ability of articular cartilage to regenerate. A local deterioration of articular cartilage can lead to initiation of the disorder, which eventually destroys the cartilage, resulting in hypertrophy and remodelling of the subchondral bone, and a fibrotic and thickened joint capsule.

The hip is a common site for osteo-arthritic changes. Old trauma, infection, abnormalities in the shape of femoral head following Perthes disease, or a minor degree of joint incongruity may possibly be the underlying cause of the disorder. The osteoarthritic patient is usually middle-aged or elderly, and suffering from increasing pain and stiffness. Treatments may involve physiotherapy, therapeutic

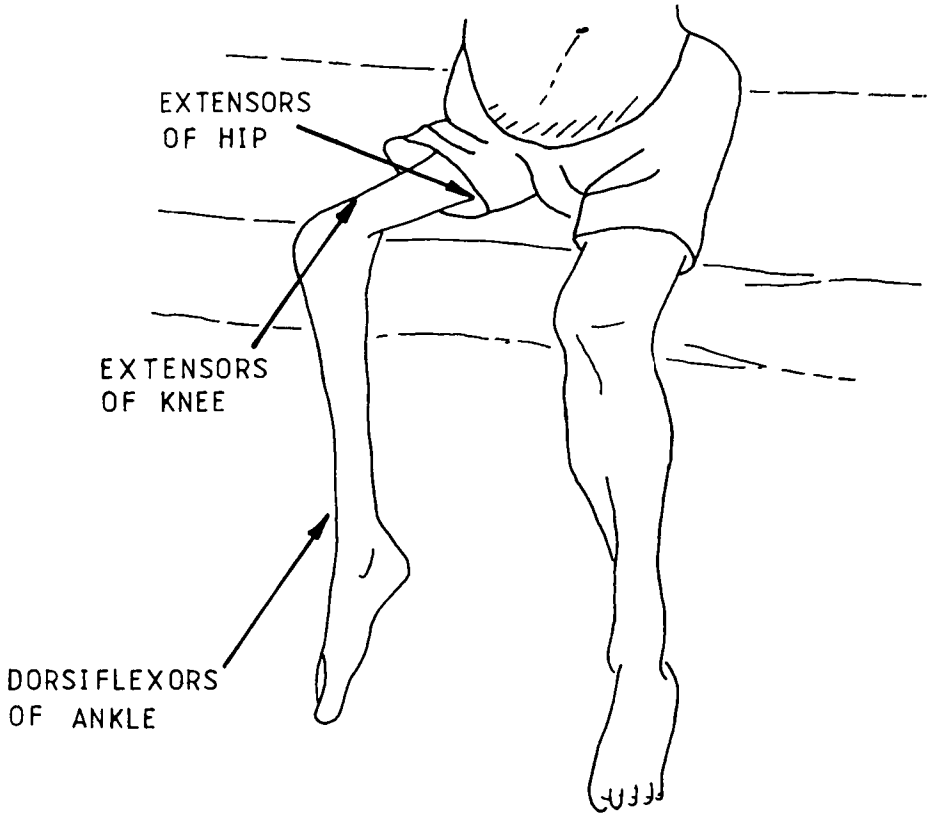
drugs, surgical repairs, and the use of walking aids. KAFO may be necessary for patients with knee instability.

1.2.5 Motor Neuron Disorders.

The initiation and control of voluntary muscle activity are governed by several distinct but inter-related systems of neurons within the central nervous system, namely, the upper motor neuron and cerebellum. Contraction of a muscle is however, stimulated by impulses carried in the peripheral nervous system, i.e. the lower motor neurons. Therefore, pathological processes involving either the upper or lower motor neuron may result in inability of muscular systems to perform their normal function. A lesion in the upper motor neuron will cause varying degrees of continued contraction of the muscle (spasticity) with exaggerated reflexes and increased muscle tone. Damaged neurons in the pyramidal system produce weakness in the pattern of voluntary movements, while extra-pyramidal lesions cause uncontrolled, purposeless movements of an involuntary nature that are aggravated by the patient's emotional tension and attempts at voluntary control.

Flaccid paralysis is the characteristic symptom of lower motor neuron lesion. There is a complete loss of contraction, either voluntary or reflex, in some or all of the fibres of the flaccid muscle, and the muscles remained in a relexed state at all times. Atrophy or wasting of the affected muscles is clearly seen.

Cerebral palsy, as the name implies, refers to a group of motor disorders caused by damage to the motor areas of the brain during foetal life, birth or infancy. It is non-progressive, becomes clinically apparent in early childhood, and persists throughout the patient's life. The three main types of cerebral palsy are: spastic, athetoid and ataxic. Spastic type of the disease is caused by a pyramidal lesion in the cerebral cortex and accounts for 65% of all cerebral palsy incidents. Its characteristic feature is the loss of fine, coordinated muscle action, and an increased muscle tone. When voluntary action is attempted, many muscles contract at the same time, making movements restricted and laborious. A lesion in the extrapyramidal system produces the athetoid type of cerebral palsy (20% of all cases) which is characterised by the involuntary, uncontrollable movements in muscle groups of the face and extremities. Maintenance of posture is severely



(Redrawn from HUCKSTEP, 1973)

FIG 1·1	MUSCLES OF THE LOWER LIMB COMMONLY AFFECTED IN POLIOMYELITIS.

restricted. Finally, the ataxic type, caused by cerebellar and brain stem lesions manifests itself with disturbed coordination of muscle groups and a relative lack of equilibrium or balance. Lower limb orthoses may be used in the treatment to prevent deformity and to enable the child to stand and walk with crutches.

Perhaps poliomyelitis can be considered as the classic example of lower motor neuron lesion, caused by viral infection occurring commonly in childhood. Although it rarely occurs in the Western society as a result of successful immunisation, it is still frequent in developing countries of the tropics and subtropics, especially in the poorer and politically more unstable regions. The poliomyelitis virus invades the body through the oropharyngeal route, multiplies in the gastrointestinal tract lymph nodes before spreading haematogenously to the central nervous system. Anterior-horn ganglion cells of spinal cord are acutely attacked, with those in the lumbar and cervical enlargements being the most vulnerable. The degree of flaccid paralysis depends on the level of spinal cord involved, however, it tends to affect the extensors more than the flexors (Fig 1.1), and the lower extremities twice as frequently as the upper limbs. Quadriceps, glutei, tibialis anterior, medial hamstrings, and hip flexors are the lumbar-innervated muscles most commonly affected, and they eventually show atrophy, fatty infiltration and replacement by connective tissue. As a result of muscle imbalance, contractures, mild valgus deformity at the knee and equinus deformity are common secondary developments of the disorder.

Treatment of patients with residual flaccid paralysis revolves around prevention and correction of deformity. Supporting orthoses, such as KAFOs and AFOs are needed to provide stability for standing and walking. Static joint instability can be controlled easily and indefinitely by orthoses, but active joint instability quickly produces a fixed deformity, making orthotic control difficult and may require surgical intervention.

Another disorder of the central nervous system is multiple sclerosis which causes progressive destruction of the myelin sheaths of the neurons. The sheaths deteriorate to scleroses, which are hardened scars or plaques, in multiple regions, thus interferring with the transmission of impulses from one neuron to another. It may be caused by a virus or other genetic defects, and usually the first symptoms occur

between the ages of 20 and 40. The early symptoms include a lack of coordination in the upper limb, partially paralysed lower limb muscles, double vision, and urinary tract infections. The progressive loss of function is interspersed with remission periods during which the undamaged neurons regain their ability to transmit impulses. As the disease progresses, most voluntary motor control is eventually lost and the patient becomes bedridden. Like other demyelinating diseases, multiple sclerosis is incurable. Electrical stimulation of the spinal cord can improve function in some patients. Lower limb orthoses may be used to support the patient in standing or walking in the early stages of the disease.

1.2.6 Cerebrovascular Accidents.

Cerebrovascular accidents, also known as stroke or cerebral apoplexy, includes all vascular disorders of the brain and is the commonest of all neurological disorders. The most catastrophic complication is sudden and irreversible ischaemia of the brain. Intracerebral haemorrhage, thrombosis, embolism and anterosclerosis of the cerebral arteries can all cause this particularly serious disorder. Its residual effects depends both on the site and extent of the area of cerebral ischaemia. At the onset, the paralysis is flaccid but within a few weeks it becomes spastic as evidenced by hypertonicity, increased deep tendon reflexes and clonus. Treatment includes therapies, functional electro-stimulation, surgical treatments, and use of light braces such as AFOs. Heavier orthoses like KAFOs are only reserved for the very serious cases, and most likely only as a standing aid.

1.2.7 Muscular Disorders.

Included in this category are the progressive muscular dystrophy, with Duchenne type as the most severe and also the most common of all the dystrophies. The Duchenne type of muscular dystrophy has a worldwide distribution and occurs only in males. The disease is present at birth and motor deficiencies generally become evident between 18 and 36 months of age. The child is late in achieving independent ambulation and walks on his toes. He stumbles and falls easily, and is unable to hop, jump, or run normally. Initially, the patient suffers

from selective, symmetrical weakness of the pelvic-girdle muscles, with the hip extensors being affected first. The weakness is progressive, and the proximal muscles are more severely affected than the distal. There is no specific cure for the disease, treatments are geared towards making the remaining years of the patient and his parents more bearable. Active exercise helps to prevent the otherwise inevitable disuse atrophy of muscles not affected, and also to minimise the patient's physical disability as well as to improve his morale. Light KAFO should be provided to assist ambulation, which on average may enable him to keep walking for an additional year or two, before being confined to wheelchair.

1.3 PATHOLOGICAL GAIT.

In normal walking activity, a person uses up only a fraction of the lower-limb potential in strength, joint mobility and coordination. When paralysis or tissue damage restricts the person's physical ability, he spontaneously draws on this reserve so that he can continue to walk as normally as possible. However, if the loss exceeds his ability to adapt, or the effort becomes too strenous, a visible gait abnormality will be evidenced. Walking becomes more precarious as the disparity between loss and substitution capability increases. This may be corrected with an orthosis or other therapeutic procedures, but its effectiveness will depend on the nature and the extent of the patient's pathological conditions.

The cause of pathological gait can be due to structural insufficiency, joint and soft tissue pathology, or neuromuscular disorders. Since structural insufficiency and joint and soft tissue pathology are very much interrelated, making a clearcut differentiation between the two is difficult. Therefore, one may assume that structural insufficiency is mainly concerned with skeletal malformation, such as limb length inequality which is caused by either overgrowth or shortening. For limb length difference of less than 35 mm ($1\frac{1}{2}$ inches), it can be compensated by dropping the pelvis on the affected side, and with exaggerated hip, knee and ankle flexions on the opposite side during swing. The patient will have to walk on tip-toe if the discrepancy is more than 35 mm.

Joint and soft tissue pathological abnormalities include

contractures, limitation of joint range (ankylosis), joint deformities, and joint instability. Muscle imbalance is usually the cause of contractures, with fixed deformity in the direction of the more powerful muscle. If a knee flexion contracture is present, the patient will normally compensate it with a short leg limp. However, for an extension contracture, circumduction, hip hiking, or tietoeing on the contralateral side becomes necessary during swing. Another common contracture is the equinus deformity, which produces a steppage or drop foot gait, i.e. excessive flexion of hip and knee to reduce length of the leg segment.

Ankylosis or restricted joint motion can be caused by articular pathology resulting in stiffening of the joint, or it may be due to prolonged immobilization of the joint. The patient usually possesses a pathological gait similar to those due to contractures.

Ligamentous damage due to trauma, ligamentous laxity and congenital dislocation can all be the underlying factor for joint instability. The pathological gait shows excessive range of motion, abnormal movement of the limb segment, inability to support body weight, and the limb may even buckle suddenly.

Some joint and soft tissue disorders such as joint injury, fracture, tumour, muscle inflammation, rupture tendon, and ligamentous tear or strain can produce a painful limp in the gait pattern. It is characterised by avoidance of weight bearing on the affected side, lengthening of stance duration on the contralateral side, and an attempt to unload the joint as much as possible.

The functional significance of neuron lesion depends on the number of motor units lost and identity of the muscles involved, since functional demands on muscles during walking differ considerably. Hip abductors and ankle plantar flexors must possess good muscle power (MRC Scale 4) for a patient to walk without a limp, whereas a fair hip extensor (Scale 3) and a poor to zero quadriceps may be sufficient (Perry, 1975).

Patients with pure neuromuscular losses have a tremendous ability to rechannel unaffected musculature to perform with maximum advantage. They are able to respond promptly and automatically with the proper adaptation to even minor postural changes. Timing of their muscle actions are altered to absorb considerable loss without limping.

Spontaneous awareness of the alignment needed to maintain stance

stability or to clear the floor during swing leads to characteristic limps in patients with motor losses. During walking, the patient's gait is a mixture of the following gait deviations:

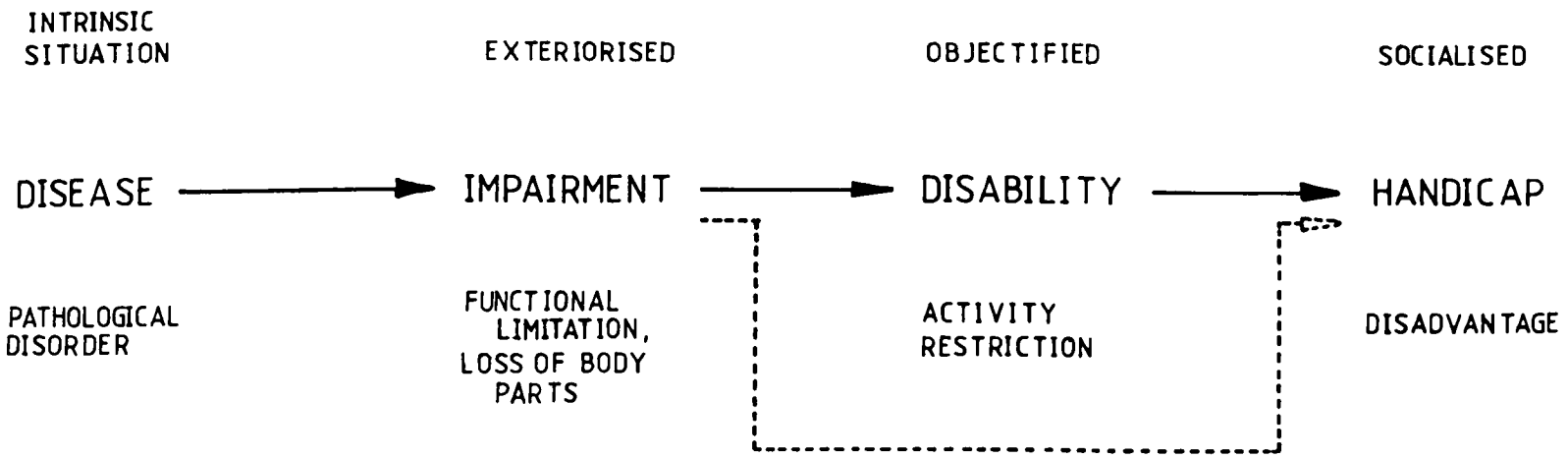
Steppage gait - Excessive hip and knee flexion to lift a drop foot when the ankle dorsiflexor muscles are paralysed. A moderate muscle weakness causes the foot to come down to foot-flat too fast, producing an audible slapping of the foot against the floor.

Circumduction gait - Pelvic hiking with a circumferential or forward flip advances the limb when the hip and knee flexors are inadequate. A patient with severe hip flexor paralysis swings the limb forward using musculature of the opposite hip, throwing the trunk backward. Trunk rotation is transmitted to the swing leg, accelerating the leg forward.

Genu recurvatum - This gait pattern is caused by either a fixed plantar flexion contracture or paralysis of soleus and quadriceps. The patient stabilies his limb with active hip extensor action using the residual gluteus maximus, friction between foot and floor, and tension on a mildly tight iliotibial band. Rapid trunk motion can substitute for inactive gluteal muscles functionally. The knee is forcibly thrown into hyperextension at or preceding heel-strike, with a smooth lurch of the trunk forward immediately after the foot touches the ground.

Gluteus maximus limp - The trunk and pelvis are suddenly thrown backward into hyperextension to lock the hip after heel strike because of the poor hip extensors. The patient walks with excessive hip rotation, and elevates the hip on affected side due to tightly extended knee at mid-stance.

Gluteus medius limp - A patient with moderately weakened gluteus medius has a greater than normal pelvis dropping on the unaffected swing leg. For very weak gluteus medius, the patient shifts his trunk to balance his weight over the stance leg, producing a very prominent lateral deviation or bending of the trunk.



(adapted from WHO, 1980)

FIG 1·2 THE CONSEQUENCES OF DISEASE. (A Conceptual Model)

1.4 DISABILITY AND REHABILITATION.

The World Health Organisation (WHO,1980) defined the following terms relevant to the probable consequence underlying illness related phenomena (Fig 1.2):

Impairment - is any loss or abnormality of phychological, physiological, or anatomical structure or function;

Disability - is any restriction or lack (resulting from an impairment) of ability to perform an activity in the manner or within the range considered normal for a human being;

Handicap - is a disadvantage for a given individual, resulting from an impairment or a disability, that limits or prevents the fulfillment of a role that is normal (depending on age, sex, and social and cultural factors) for that individual.

From the above definitions, it can be seen that impairment is concerned with abnormalities of body structure and appearance, and with organ or system function, resulting from any etiological factor. In principle, it represents disturbance at the organ level, and includes both disorders, loss of a part of the body, and functional limitation. Whereas impairment is concerned with individual functions of the parts of the body, disability is concerned with compound or integrated activities expected of the person or of the body as a whole. Thus it represents disturbances at the level of the person as reflected by the consequences of impairment in terms of functional performance and activity by the individual. Handicap represents the social expression of disease consequences, and is characterised by a discordance between the individual's performance or status and the expectations of the particular group of which he is a member.

The prime concern of locomotor disability is within three areas: namely, ambulation, confinement, and other activities. Ambulation covers walking, transversing uneven terrain, climbing stairs or other obstacles, and running. Transfer to and from a bed or a chair, and getting on or off a vehicle are classified under confinement. Disability prevention can be implemented with a three-tier strategy, aiming at preventing the occurance of disability, and at reducing those that are already present. At the Primary Level, an attempt is made to prevent the initial occurrance of a disease or impairment. Secondary Prevention involves the early detection and treatment of disease or

impairment with a view to return to normal health. The continuing treatment of the disease or disability that is already present to avoid needless progression or complications, and the rehabilitation of the individual in order to minimise any subsequent handicap constitute the Tertiary Prevention.

The term 'rehabilitation' has a Latin origin, meaning restoration. It is a therapeutic programme designed to minimise the consequences of a permanent or protracted disability. This can be achieved by the combined and coordinated use of medical, social, educational, and vocational measures for training or retraining the individual to the highest possible level of functional ability. In sharp constrast with standard curative medicine where the attention is on reversing primary disease processes, rehabilitation concentrates on restoring function, preventing contractures, developing muscle strength, stimulating latent control, training to use residual neuromuscular function effectively, providing assistive devices, and resettlement in the community. Therefore, the ultimate goal may be stated to be 'attainment of the maximum physical and phychological adjustment of each disabled person within the limits of his impairment, and to enable him to live as useful and satisfying a life as is humanly possible' (Rehab Int, 1981).

Rehabilitation works most effectively as a comprehensive and multidisciplinary programme. There should be an appropriate blend of medicine, nursing, bioengineering, physiotheraphy, occupational therapy, orthotics and prosthetics, medical social work, clinical psychology, and vocational counselling. Members of the team should work together in harmony at all times, and have a mutual respect of each profession's contribution to the total patient care. An effective exchange of information among the members is an important asset in addition to each member's competence, responsibility, and authority.

Bioengineering is a relative new speciality that has developed specifically to create engineers having a physiological or clinical background to supplement their technical training. It may be defined as the application of engineering techniques to the problems of biology and medicine. One of the main branches of bioengineering is Rehabilitation Engineering, which concerns itself with improving the quality of life of the physically disabled. The work of rehabilitation

engineering includes the designing and fabrication of equipment for clinical or research uses.

The problem of disabilities, especially those caused by poliomyelitis, facing the poorer developing countries of the world are discussed in Appendix 1. Examples of the use of low cost technology in rehabilitation of the physically disabled are also described in the appendix.

CHAPTER 2.

ORTHOTIC MANAGEMENT OF LOWER LIMB INSTABILITY.

2.1 INTRODUCTION.

The terms 'orthotics' and 'orthoses' are of recent introduction to the field of disability and rehabilitation. Originated from Greek, 'ortho' and 'statikos' mean 'straight' and 'to cause to stand' respectively. Orthotics is the branch of technology that deals with the application of force, through an exoskeletal device, to the human body to restore function to individuals with physical impairment. The mechanical device employed to provide these forces to the body is called an orthosis. They assist a dysfunctioning segment of the body principally by controlling motions about the joints and by regulating axial loads in the long bones.

This chapter aims at introducing the reader to the management of unstable lower limbs with the appropriate orthotic devices. A brief account of the historical background is described, followed by a short explanation of the orthotic terminology and classification system. Different types of lower limb orthoses are also reviewed.

2.2 HISTORICAL BACKGROUND.

The history of orthoses and the art of bracing has been a long and honourable one, with fracture splinting being the most probable origin of the field of orthotics. Bones of prehistoric men provide silent testimony of musculoskeletal disorders and injuries, and since then, man has sought ways to alleviate the crippling conditions of his fellow human beings.

As early as 9000BC in the Palaeolithic Age, man had began to use orthoses for weak limbs and broken bones. Skeleton of early Stone Age men show evidence of fracture that healed in good alignment, suggesting that some type of fracture orthosis had been used. Seven thousand years later, the Egyptians developed the concept of the crutch, as demonstrated in some stone carvings. By the 5th century BC, Greek replaced Egypt as the centre of Western Civilization, and was blessed with many 'scientists' and philosophers. Hippocrates, who was one of the greatest Greek physicians, advanced the frontier of orthotics through his many practical ideas concerning the art of bracing. He stressed that splint must fully enclose the hip and knee to achieve complete immobilization of the femur, and further specified that pressure points of the splints should not be over any bony protuberances (Bunch & Keagy, 1976).

The orthoses of the first few centuries AD were probably heavy,

clumsy, and of marginal efficacy. Significant advances were made by the medical school at Bologna in the twelfth century, where the then existing models of the orthoses were standardised, simplified, and lightened. Braces of wood and metal were widely used at that time. Four centuries later, a surgeon by the name of Ambroise Paré (1510-1590 AD) invented special shoe modification for clubfeet. In the same period, Glesson (1597-1677 AD), a Cambridge professor used braces to straighten bowleg deformities. During the mid-1700s, Nicholas Andry devoted much of his work on the correction and prevention of deformities in children. He was the Professor of Medicine in Paris. By the 19th century, the art of bracing had become considerably sophisticated. The most outstanding contribution was by Hugh Owen Thomas who designed the famous fracture splint and other braces for controlling joints of the lower limb.

The care of patients with musculoskeletal disorders has undergone several phases of improvement in the present century. Firstly, there was the 'strap and buckle' phase where various types of orthoses and other mechanical devices were extensively used. This predominantly mechanical era was later replaced by a period of excessive surgical operations, however, not all of which were based on sound surgical or biomechanical principles. In recent years, rehabilitation philosophy has taken on a more logical approach, and with a better understanding of the problems involved. Critical evaluation of various forms of treatment and other experimental researches are constantly carried out in a number of clinics and research centres throughout the world. From these studies, it is hoped that a deeper understanding of both the behaviour of orthotic devices in use by patients, and the physiology and pathology of the musculoskeletal system may lead to a more effective treatment for patients with disabilities.

The development of orthotic devices has long been overshadowed by and trailing behind the more active prosthetic research and development. Great leaps forward in the prosthetic field can be attributed to the Second World War. During the course of the war, a large number of young servicemen lost their limbs, and they consequently demanded to be fitted with effective artificial limbs which would enable them to live as normal a life as possible.This provided an impetus to new developments in the design and construction of modern prostheses. A number of these clever developments had since been 'borrowed' and

incorporated into the design of orthotic devices. The quadrilateral top for weight-bearing KAFOs is one such example.

Orthotic developments in recent years have been advancing with considerable speed, one of the reasons being the acceptance of the engineer as an important member of the rehabilitation team. A direct benefit of this is the application of new materials and modern engineering and production techniques to the design of orthoses. Contributions by engineers are particularly useful when special equipment or devices are required to perform certain functions for the disabled. The progress in orthotics can therefore be summarized as principally owing to:

1. translation of prosthetic principles to orthotics;
2. greater understanding of applied biomechanics; and
3. the advent of new materials and fabrication techniques.

2.3 TERMINOLOGY AND CLASSIFICATION.

Traditionally, orthoses are identified by their proper names, or eponyms derived from the place of origin, the inventor or promotor, or sometimes a name unrelated to either. Many of these eponyms do not give the site of application nor indication of the function provided. Orthoses with very similar functional usage and construction can have several entirely different eponyms in different part of the world. Furthermore, modification from a device would again produce another name but with no real changes.

There have been various attempts to standardize the terminology and to improve communication among the specialists of different fields working in the rehabilitation services. In the early 1970s, representatives from surgeons, orthotists and researchers in the U.S.A. formed a working committee for standardizing the vocabulary involved. An important outcome from the committee was a nomenclature of orthoses that is gaining more and more acceptance throughout the world (McCollough et al,1970; McCollough,1974,1975,1976; Harris,1973).

The basic concept of the nomenclature is to name the orthoses for the segments of the body it emcompasses. In the extremities, (extremities being the preferred name for limbs), the orthoses would be described by the combination of joints it covers. Thus an orthosis embracing the hip,knee, ankle and foot is a 'hip-knee-ankle-foot orthosis', abbreviated to HKAFO. A 'long leg brace' will become a

knee-ankle-foot orthosis or KAFO, and the 'short leg brace' becomes an ankle-foot orthosis or AFO. Orthoses that emcompass only a single joint would also be named accordingly. The common lower extremity orthoses are listed below:

foot orthosis = FO
knee orthosis = KO
ankle-foot orthosis = AFO
knee-ankle-foot orthosis = KAFO
hip-knee-ankle-foor orthosis = HKAFO

To further define an orthosis, the actions the orthosis has on those joints that it emcompasses should also be specified. For this, the qualifiers may be:

1. Flexion/Extension;
2. Abduction/Adduction;
3. Rotation; and
4. Axial Loading.

The second basic concept of the nomenclature is to name the orthosis in terms of its control of anatomical joint motions. There are five types of basic controls used in the system to indicate the effect of an orthosis on joint function, with an additional four supplementary variations on the basic controls. The basic controls are as follows:

FREE - Free motion permitted in any given direction;

ASSIST - Application of an external force to increase range, velocity, or force of a desired motion. The mechanism of this force may be a spring, a motor, or even alignment to make use of gravitational force;

RESIST - Application of an external force to decrease the force, range, or velocity of a desired motion;

STOP - Orthosis to prevent an undesired motion in one direction. Neutral position is assumed if no direction is indicated;

HOLD - Elimination of all motion in one place.

The four supplementary controls are:

Variable Stop - This is usually used with stop to request an orthosis with potentially different end point of joint control;

Joint Locks - An optional device that may be used intermittently;

Range of Motion - Denotes a desired limitation of range, with the

final allowable position in a given direction expressed in degrees;

Percentage Axial Load - Indicates the desired functional weight relief.

In summary, an orthosis is named by the joints or bady segments it emcompasses, and is supplemented by adding the types of control, compensation, or correction to be applied to the body segments involved. For example, a long leg brace may be described as:

KAFO with free flexion, stop extension, and drop locks at the knee, and free dorsiflexion, 90° plantar flexion stop at the ankle.

2.4 FUNCTIONS OF ORTHOSES.

Orthotic treatments are aimed at helping the patient to become physically and socially self-sufficient. Since an orthosis is a mechanical device that applies forces to the limbs or other body segments, its functions can be classified and analysed in mechanical terms. However, for clinical purposes, it will be more logical to describe in terms of the primary goals for which the orthotic treatment was prescribed. To permit safe and effective ambulation, an orthosis is required to be able to perform one or more of the following primary functions, namely, control of motion, support of anatomical structures, compensation for weaknesses, and correction of deformities. In the majority of the cases, these functional classifications are overlapping, with the different functions supplementing each other rather than being mutually exclusive. Many orthoses will therefore serve more than one purpose, and a particular design may be used to achieve different therapeutic goals on different patients.

The most commonly known reason for prescribing orthoses is probably the necessity to control motion of a limb segment with respect to another. A wide spectrum of diseases such as nerve deficits, muscle weakness, loss of ligaments, and other abnormalities of the neuromusculoskeletal system may produce functional deficits which require the service of a motion-control orthosis. In situations where total voluntary control of joint motion is impossible, an immobilizing orthosis is normally used to prevent uncontrolled motion aggreviated by postural forces. Immobilization is also frequently required in the management of delayed union of certain fractures. For other cases where only the

control of excessive movements such as genu recurvatum is needed, the orthosis limits the joint motions to the normal range without unduly restricting other functional movements. Situations may also arise when full motion is undesirable and need to be limited even though the patient may have good joint control and stability. Pain is the most frequent indication for reducing joint excursion. Another benefit of abnormal joint motion control is that the development of deformities or contractures can be prevented.

The other major function of lower limb orthoses is for support of anatomical structures from weight-bearing or other mechanical strains. They are used in the presence of flaccid paralysis, painful joints and structural inadequancies as a result of diseases, traumatic incidents or congenital abnormalities. To a certain extent, supportive orthoses are similar to their immobilization counterparts in that they stabilize weakened, paralysed or unstabled limbs by preventing unwanted joint motions. They also offer protection to and maintain alignment for a diseased painful joint by relieving the weight-bearing pressures at the articulating surfaces.

Functional orthoses are prescribed to compensate for a weakened or absent muscle power. They are characterised by a motor element designed to compensate for the loss of a group of muscle by introducing active function to the limb segment as a substitutive force. The most frequently used motor element is a coil or leaf spring which can be dynamically loaded by the body weight and muscle actions, and subsequently discharges its stored energy during the unloading phase. Their primary use is in the treatment of flaccid paralysis, such as drop-foot AFOs, which permit relatively linear response of joint motion to applied forces. With the failing prime mover compensated, the improved safety level may enable the patient to use his available resources to a better effort elsewhere.

Corrective orthoses are in many respects, surgical tools. Gradual correction of abnormal structural alignment in growing long bones is possible over a period of time by means of night splints specially designed to transmit corrective forces to the affected limb segments. Most corrective orthoses can only be used when the patient is not walking, and are more likely to be effective if applied during infancy. Conditions where the use of corrective orthoses may be required are: congenital club foot, congenital metatarsus varus,

developmental flat foot, congenital dislocation of the hip, and tibial torsion.

The following section attempts to list some of the uses and functions of orthoses as prescribed in a number of pathological conditions.

Arthritis:

1. stabilizes joints to control pain;
2. realignment of the joints to improve weight-bearing and prevent deformities;
3. relieves pressure on the joints.

Cerebral Palsy:

1. controls involuntary motion to prevent poor walking habits in children;
2. control of foot positioning and correction of deformity;
3. control of limb while muscles regain their anti-gravity strength and voluntary control in post-surgical period.

Hemiplegia:

1. to improve functional ambulation;
2. improves proprioception and as a mechanical support.

Hemophilia:

1. immobilize the joint to protect against further haemorrhage;
2. to relieve pain by restricting joint motion.

Muscular Dystrophy:

1. help to keep the child active by prolong the ability to walk and in postponing complete dependence on wheelchair;
2. an orthosis with elastic straps and spring-loaded joints may encourage motions when there is only some muscular function remaining;
3. used as standing pivot transfer after functional standing and walking has become impossible (Hunt, 1981).

Poliomyelitis:

1. stabilizes joints for weight-bearing;
2. realignment of joints to prevent progressive deformity.

Spina Bifida:

1. supports and stabilizes the lower limbs to facilitate standing and walking.

Spinal Cord Injury:

1. stabilizes joints to aid in transfer;

2. for exercise and functional ambulation.

2.5 REVIEWS OF LOWER LIMB ORTHOSES.

Lower limb orthoses are conventionally made from pre-fabricated metal components. The design features of many of these orthoses have remained more or less stagnant for a good number of years. Improvements to the traditional designs were slow and few. However, in recent years there have been a steady increase in the amount of research undertaken in an attempt to produce more efficient and cosmetically more acceptable orthoses. These improvements were partly brought about by the application of engineering skills to orthotic assessment and design, and also partly by the widespread availability of plastic materials suitable for use in these orthoses.

Lower extremity orthoses fabricated from metal components are currently still being used by a large number of patients. The conventional double upright KAFO, which is more commonly known as the long leg caliper continues to hold an important role in enabling patients with weak lower extremities to stand or ambulate functionally. Aluminium alloys are normally used in children's orthoses or in situations where the load is relatively small. When either durability or heavy duty usage is the primary concern, steel components should be used.

Plastic orthoses are much more cosmetic in appearance, have lower mass, offer greater flexibility in design, and are generally better received by the patients. Fabrication and fitting of these orthoses are based on plaster cast impressions of the limb, thus allowing closer fit and more precise control of pressure distribution. Although some plastic orthoses are made from thermosetting plastic laminate, which consists of fabric impregnated with polyester resin, the majority of them are formed thermoplastic materials, especially polypropylene and various derivatives of polyethylene.

There are also a number of designs which employ a combination of both metal and plastic materials. Advantages from both types of materials and designs were skilfully merged in the hybrid orthotic design, where the versatility and cosmetically appealing plastic material is combined with the more functional metallic joints.

Before proceeding to review briefly the various types and versions of lower extremity orthoses, the roles played by the shoes should

be mentioned. Shoes serve a variety of functional and cosmetic purposes, one of which is to minimise the pressure on sensitive or deformed structures of the foot, hence promoting function and comfort. When they are considered as part of the complete lower extremity orthotic system, it serves as a foundation for the application of external forces. Therefore unless they are correctly fitted and appropriately modified, the orthosis will not provide the desired pattern of weight-bearing and gait.

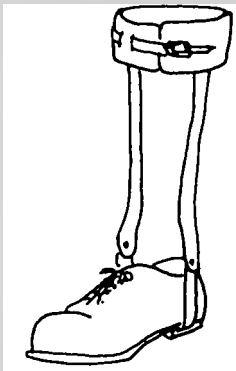
In the discussions and reviews that follow, eponymal names are used extensively to identify the many different versions of a particular class of orthoses.

2.5.1 Ankle-Foot Orthoses.

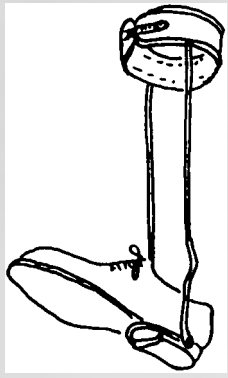
The primary function of an ankle-foot orthosis is to control the alignment and motions of the joints of the foot and ankle, such as to provide dorsiflexion assistance during swing, or to stop or limit plantar flexion. In addition, it can also provide an adequate medio-lateral stabilization of the subtalar joint and foot, either independently or in conjunction with the dorsiflexion-plantar flexion control.

A quick search of the literature will reveal the existance of a wide range of ankle-foot orthotic designs developed over the years by various workers and researchers. They are normally based on either metal components, plastic materials, or a combination of both. Typically, an AFO extends downward from the calf area and terminate underneath the foot, or at the heel of the shoe. Most AFOs, other than those of the plastic design, include two uprights whose proximal ends are connected to a leather-covered or plastic calf band, and the distal ends to the ankle joint mechanism, which in turn anchors itself to the shoe or foot attachments (Fig 2.2a). However, a single bar design may be sufficient for relatively mild cases, with the single upright located either medially, laterally, or posteriorly with respect to the ankle joint.

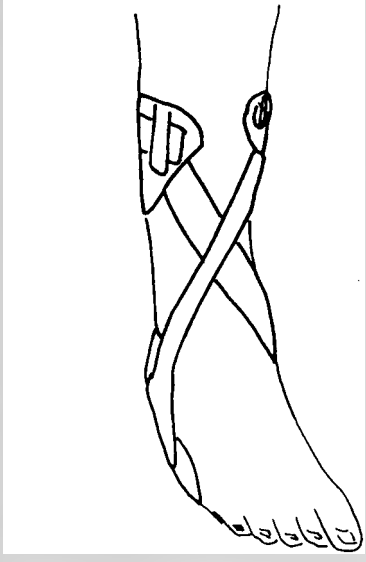
The shoe and foot attachment for the orthoses can be of a stirrup design, a socket and spur construction, or the more sophisticated shoe insert (Fig 2.1). A permanent attachment between uprights and shoe is provided by the stirrup design, and permits fairly accurate correspondance of the mechanical ankle joints with the anatomical joint axis.

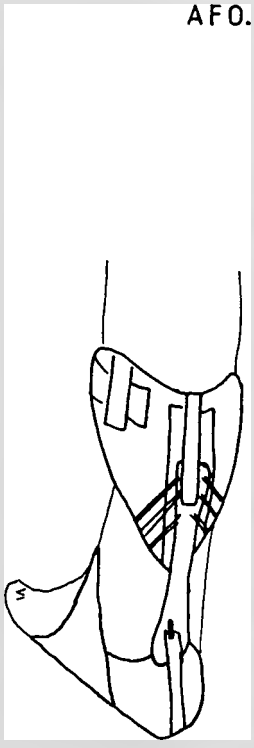


(a) DOUBLE UPRIGHT AFO.



(b) WIRE SPRING AFO.





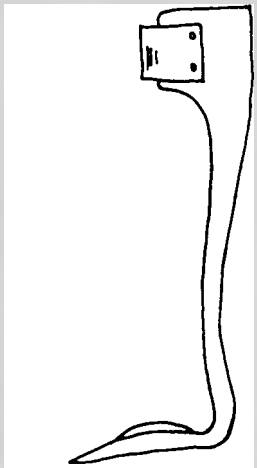
(c) MILNER'S AFO.

FIG 2·2 TYPES OF AFO.

On the other hand, the socket and spur allows easy shoe interchangeability and is economical to build. However, the pivot of the mechanical joint is considerably distal to the anatomical ankle joint, at the level of the shoe heel. The best of the three is the shoe insert (Inman, 1969) which provides both mechanical and anatomical joint congruity as well as shoe interchangeability. It is basically a stirrup incorporated into a shoe insert shaped to the contour of the patient's foot, which in turn fits into the shoe. Maximum control and support of the foot can also be achieved through the closeness of fit.

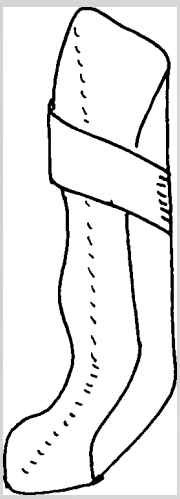
If a mechanical ankle joint is fitted, it is usually of a single axis design which prevents any medio-lateral motion, but allows either free motion, limited motion, or stops the movement completely in the sagittal plane. Plantar and dorsiflexion assistance can also be provided by incorporating a spring mechanism into the ankle joint, where recoiling of the spring aids dorsiflexion in swing. A unique design where ankle motions are assisted without the use of a mechanical joint is the wire spring AFO (Fig 2.2b). The orthosis employs bilateral spring wire uprights and springs at sole level as sources of power to assist motion. Hill & Stube (1969, 1970b) replaced the spring wires with epoxy-fibreglass rods in their attempt to improve the design. These spring wires do not restrict movement to motions around a single axis as most mechanical joints do, but conforms more closely to the movements of the much more complex anatomical ankle and subtalar joints. Attempts to provide two separate sets of motions at the subtalar as well as at the ankle joint had been reported by Inman (1969). He constructed the UC-BL dual-axis ankle control mechanism which out performs the conventional single-axis joints functionally, but cosmetically is less acceptable to the patients. Milner (1982, 1983) took a slightly different approach and dispensed with the analogue of ankle joint all together. He devised a mechanism to transmit the correcting moments between the orthosis and the leg and foot without the normal guidance system (Fig 2.2c). Most of the components were made of glass reinforced plastic and carbon fibres.

With the more widespread availability of plastic materials in recent years, orthoses fabricated from plastics are becoming more and more common. Thermoforming plastics such as polypropylene are generally preferred to polyester and epoxy resin based thermosetting plastics which have a shorter fatigue life. These thermoformed AFOs are vacuum

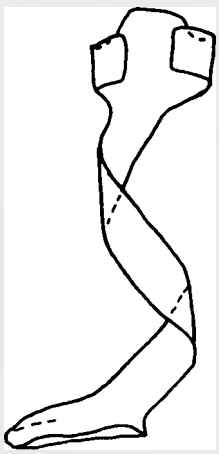


(a) THERMOFORMED POSTERIOR LEAF SPRING AFO.

(b) TIRR AFO.



(c) SOLID ANKLE AFO.



(d) SPIRAL AFO.

FIG 2·3	PLASTIC AFOs.

moulded from a single piece of thermoplastic sheet over a modified plaster cast of the affected limb segments. The advantages of plastics over metal include easy workability, low cost, light weight and a high resistance to material fatigue. The extent to which motions are controlled depends primarily on the relative rigidity of the plastics, which in turn is affected by its chemical composition, thickness, and the geometrical shape.

The most common plastic AFO is probably the thermoplastic type posterior leaf spring AFO (Fig 2.3a). It is characterised by a relatively narrow calf shell between the foot part and calf strap, the dimension of which dictates the permitted range of ankle movement. Although the orthosis is in total contact with the posterior aspect of the leg and the plantar surface of the foot, there is, however, a slight relative movement between the AFO and the enclosed limb at the calf during plantar flexion. This is due to the fact that the instantanous centre of rotation of the orthosis in dynamic activities does not closely follow the anatomical pattern.

Different versions of the polypropylene AFO have been reported by a number of workers, examples of which are the Solid Ankle and the TIRR AFOs. The Solid Ankle AFO (Fig 2.3c) holds the ankle-foot complex in a predetermined position by preventing all plantar and dorsiflexion motions, and at the same time resisting any varus or valgus deviation of the ankle and foot. It functions basically as an immobilization orthosis. The TIRR (Texas Institute for Rehabilitation and Research) polypropylene AFO (Fig 2.3b) is a posterior leaf AFO with incorporated corrugations to provide additional strength and stability in the transition at the foot part or arch support of the malleoli area (Engen, 1972).

Perhaps the most novel designs of all the plastic AFOs are the spiral and hemispiral ankle-foot orthoses developed at the Institute of Rehabilitation (IRM) of New York University Medical Centre (Lehneis, 1969, 1972, 1973, 1980). The spiral AFO consists of a spiral that originates at the medial side of the foot plate, passing round the leg posteriorly, continuing across the anterior aspect and terminating with a calf band at the level of the medial tibial condyle (Fig 2.3d). No straps are required to hold the orthosis in position, and it is also without any metallic joint. The orthosis, as claimed by Lehneis, is able to provided controlled motions in all planes, i.e. it adapts to

transverse rotation as well as to motions in the frontal and sagittal planes. On weight-bearing, the spiral unwinds to permit plantar flexion and it consequently dorsiflexes the foot as it rewinds with the removal of body weight. Winding and rewinding of the spiral also allows the leg to rotate with respect to the foot in the transverse plane.

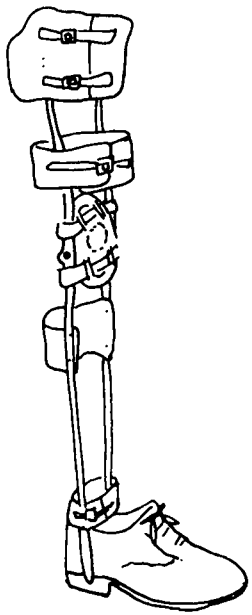
Hemispiral AFOs are similar to the spiral orthoses, except that thay only make a half turn round the leg. The hemispiral originates from lateral part of the foot plate instead of the medial, and passes round the leg in a direction opposite from that of the full spiral AFO. Therefore it provides a greater stiffness with increased resistance against the equinus and varus tendencies.

The Veteran's Administration Prosthetics Centre (VAPC) shoe clasp AFO consists of an epoxy-fibreglass posterior bar with a stainless steel clasp that clipped onto any stiff countered shoe (CPRD, 1972; Rubin & Dixon, 1973). It is designed to correct the dropfoot condition and at the same time allows interchangeability of shoes.

AFOs can also be modified to be weight-bearing for the relieve of chronically painful tarsus and ankle. Kay (1971) used a PTB (Patellar Tendon Bearing) socket with double-upright AFO, while Sweezey (1975) utilised a laced moulded leather calf encasement. Kay's design was later improved by Demopoulos and Eschen (1977) with an all plastic PTB orthosis. Other types of AFO designs include the dynamic club foot splint (Lyquist, 1970), knee-stabilising AFO (Rubin & Danisi, 1975), ankle immobilisation AFO (Kramer & Arnold, 1978), Jointed Plastic AFO (Bensman & Lossing, 1979), and the various types of fracture orthoses. The Jointed Plastic AFO combined the attributes of metal and plastics, using the preferred characteristics of both in a single orthosis. It is an orthosis fabricated primarily of plastic, but uses metal to provide the needed joint motion, hence retaining the light and cosmetic plastic together with the flexibility of a metal joint.

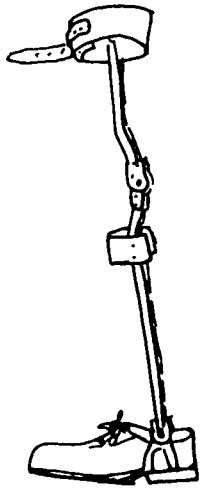
2.5.2 Knee-Ankle-Foot Orthoses.

Orthotic management of knee instability can be provided by a knee-ankle-foot orthosis. The indications for KAFO include weight-relief, mediolateral or anterioposterior stability, and motion control. Typically, the conventional KAFO consists of two metal upright bars



(a) CONVENTIONAL KAFO.

(b) THE ORIGINAL THOMAS SPLINT.



(c) SCOTT-CRAIG KAFO.

FIG 2·4	TYPES OF KAFO.

extending from the foot to the thigh, and may have either fixed or movable knee joints (Fig 2.4a). Distally, the components are the same as those of AFOs, while proximally, two thigh bands hold the upright ends together. A leather apron straps round the knee which retains the lower limb in its desired position. Weight is normally borne by the patient's own skeleton, except for the weight-bearing type.

Single-axis knee joints are commonly prescribed although the anatomical knee has a changing axis of rotation, i.e. it is a combination of a hinge and a sliding joint. Inevitably, some relative movement of the orthosis with respect to the limb will occur during flexion and extension of the knee. This can be minimised by placing the mechanical joint at 1.25cm above the anatomical joint surfaces, and a little posterior to its centre. Free motion, adjustable range of motion, locked, or offset joints are available for the single-axis mechanical knee.

To follow the natural motion of the knee more closely, a polycentric joint which consists of a set of two meshing gears may be used. However, only those patients with significant knee motion during walking can benefit by the joint. Another type of polycentric knee, called the Genucentric Joint, has been reported by Foster and Milani (1979). It consists of a disc pivoted at two points, sandwiched between the proximal and distal sections. The construction permits the instantaneous centre of the genucentric joint to move through a variety of paths, thus allowing the joint to follow the path of an anatomical knee while providing the necessary support.

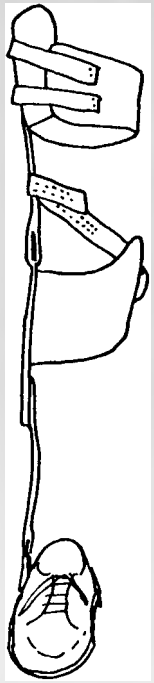
Two other types of knee control systems should also be noted. The first is the hydropneumatic knee-ankle control system devised by Lehneis (1969,1972). It was designed to offer stability of the knee against buckling, to provide controlled fluid resistance to plantar flexion, and to provide swing phase control. Few patients were fitted with the control system, mainly because of problems with weight and cosmesis of the device, and suitability of patients. The other system, an electrically controlled experimental knee joint, was reported by Lehmann and Stonebridge (1978). A pressure sensor in the shoe activates a solenoid which electrically locks and unlocks the joint. The device intends to improve gait pattern and to reduce the energy required for ambulation. However, it was found that energy savings were significant only at high walking speed (73 metres/min) which were

unattainable by patients needing KAFOs.

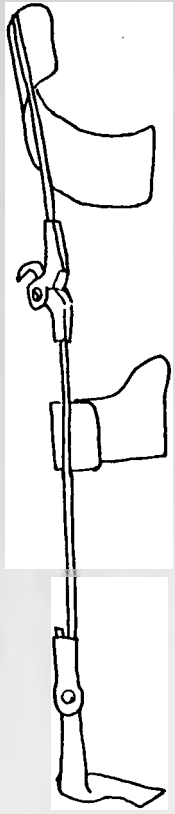
The upper end of a KAFO may be fitted with a ring, cuff, bucket, or quadrilateral socket. The ischial or Thomas ring top consists of a metal ring padded with felt and covered with leather. It is ineffective as a weight bearing device because of the small contact area between the ischium and the ring. The ischial tuberosity may slip into the ring, causing discomfort and possible pressure sores. For non-weight-bearing KAFOs, a padded cuff or band top may be used. Basically, it consists of a broad padded posterior metal thigh band with an adjustable anterior soft leather band. An improvement over the basic ring and cuff tops is the leather bucket top, made by moulding leather over a plaster cast of the thigh. It fits accurately around the upper third of the thigh, and has a posterior curved lip on which the ischial tuberosity rests. The ultimate design of the weight bearing area is the plastic quadrilateral top, an idea taken from the quadrilateral socket of above-knee prosthetic design (Radcliffe, 1955).

The first practical KAFO can be attributed to the Thomas splint designed by Hugh Owen Thomas in the nineteen century (Fig 2.4b), since then, changes to the shape and configuration of the orthosis have been slow and conservative. Design modifications were mainly empirical rather than based on established values of kinetic and kinematic data. One of the reasons being that there have been very little research done on the measurements of these values when actually being used by a large number of patients.

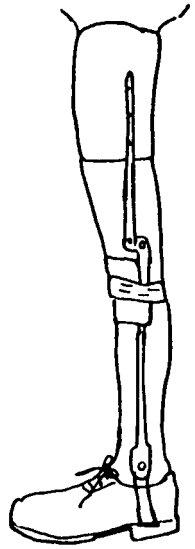
The Scott-Craig KAFO was designed to help spinal cord injury patients to walk more efficiently (Scott BA, 1971, Lehmann et al, 1976). By employing the minimum number of bands and straps, and an alignment of approximately 10° dorsiflexion, the orthosis is able to fulfill two desirable factors at the same time, i.e. to provide both an effective orthotic stabilization system and an ease of application (i.e. donning and doffing). The new design uses locked offset knee joints and conventional double uprights attached to a reinforced T-shaped foot plate, with the enclosed limb held in place by an anterior tibial cuff and a leather thigh band, but without a knee strap (Fig 2.4c). Contradictory to claims based on biomechanical analysis that the orthosis is more energy efficient than conventional KAFO, recent reports had suggested that there is no significant energy savings while standing or walking (Merkel et al, 1981), and similarly while



(d) NITSCHKE SINGLE-BAR KAFO.



(e) VAPC MODULAR SINGLE-BAR KAFO.



(f) UCLA FUNCTIONAL KAFO.

FIG 2·5 TYPES OF KAFO.

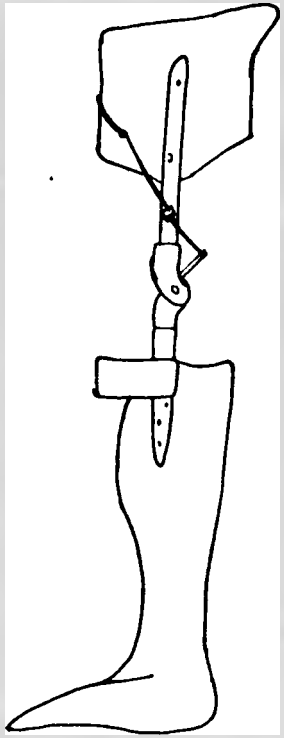
negotiating architectural barriers (Merritt et al, 1981). The only exception is when the orthosis is being used in a swing-through gait.

Nitschke (1971,1976) further reduced the hardware to a single lateral upright, a half stirrup, a modified PTB-shaped pre-tibial cuff, and a thigh band. The orthosis, called Nitschke Single-Bar KAFO (Fig 2.5a), is designed to fit loosely when sitting, snugly while standing, and to maintain proper alignment of the lower limb during walking. Control of valgus or varus may be accomplished by having the pre-tibial cuff shaped with high medial or lateral portion as required. However, due to its design features, heavy-duty use and weight-bearing application should be avoided.

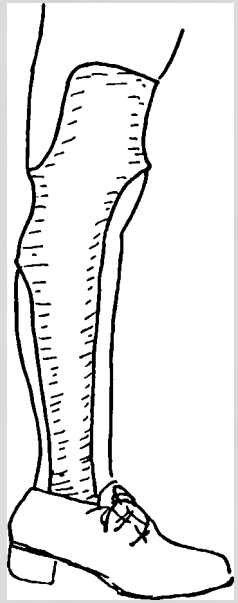
The Veterans Administration Prosthetic Centre (VAPC) had also developed a single lateral bar KAFO (VAPC, 1968; CPRD, 1971). The VAPC Modular Single-Bar KAFO consists of a lateral bar with two metal cuffs: the calf band which extends upwards medially to the tibial condyles, and a thigh cuff that spirals down medially to encompass two-thirds of the thigh (Fig 2.5b). A certain degree of flexibility in choice of components for the KAFO system is permitted in the modular design, for example, different sizes of uprights can be used according to the strength required, with an option for incorporating other facilities such as additional straps.

If patients with unstable lower limbs were allowed to walk with the normal flexed knee gait, a substantial reduction in total energy required for ambulation can be achieved. The orthosis will have to provide stability at the stance phase while allowing knee flexion and maintaining proper alignment of the limb during swing. Both the Rancho Functional KAFO (CPRD, 1971) and the UCLA Functional KAFO (Strohm et al, 1963; Scott CM et al,1971) were designed to perform this task. They have metal uprights with plastic thigh shell, thigh cuff and a shoe insert or a metal stirrup attached to the shoe. The knee joints are posteriorly offset for alignment stability during stance (Fig 2.5c). However, active hip extension power is required to operate these orthoses successfully.

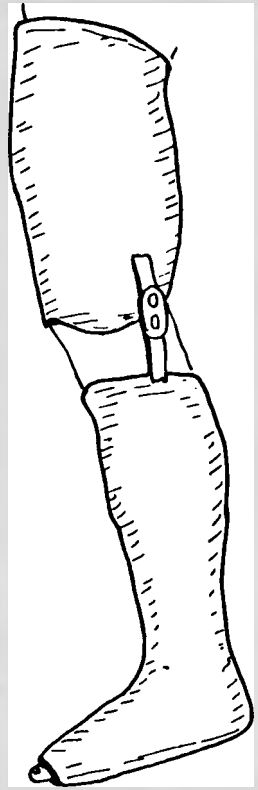
A dual purpose KAFO, which consists of an above-knee extension that fits into each side of a double upright AFO was reoprted by Pathy and Plamer (1977). It is intended for hemiplegic patients whose knee extensors do not improve enough to allow early ambulation. With the orthosis, the patient is able to combine intensive quadriceps exercise



(g) STANMORE COSMETIC KAFO.



(h) SUPRACONDYLAR KAFO.



(I) FRACTURE BRACING.

FIG 2·6	TYPES OF KAFO.

with reeducation in walking. Once the muscles are strong enough, the above-knee section is removed, and the patient immediately has an AFO without delay.

The traditional metal KAFOs have always been seen as cumbersome, heavy, rigid, uncomfortable, and cosmetically unacceptable. Numerous research centres have attempted to improve this situation. The Salford Cosmetic Caliper (Henshaw,1970) is a weight-relieving KAFO, having close fitting metal uprights with an ankle hinge that fits inside the patient's shoe. Patients who have a severe wasting of the calf can have improved cosmesis by the fitting of a suitably shaped calf cover. Tuck (1974) developed an orthosis with polyethylene ("Ortholen") thigh cuff and ankle-foot piece. Aluminium uprights were riveted onto the proximal portion of the ankle-foot piece, which is in fact a plastic AFO. The calf portion was then built up with cellular "Plastazote" to make a cosmetic substitute for the wasted calf, shaped to correspond the contralateral good calf (Fig 2.6a). All metallic parts were given a thin matt coating of plastic material to improve the cosmesis. Tuck claimed that the Stanmore Cosmesis KAFO was very acceptable and successful with the patients.

An all plastic orthosis designed at the IRM, called the Supracondylar KAFO (Fig 2.6b) provides medio-lateral knee stability, prevents genu recurvatum and stabilizes the foot and ankle while allowing knee flexion for walking (Lehneis,1969,1972; Lehneis et al,1973). It is a single component design, with no mechanical knee joints, and immobilizes the ankle in a slight equinus position. The flesh-coloured nylon and fibreglass reinforced polyester laminate provides good cosmesis when standing, however, the proximal supracondylar section protrudes above the knee when the wearer is seated. This ungainly sight may be avoided by introducing conventional metal knee joints into the orthosis, thus alleviating the problem of protrusion by allowing the orthosis to flex at the knee.

In recent years, cast-bracing of femoral shaft fractures has become a more widely accepted method of treatment (Wardlaw,1977; Wardlaw et al,1979,1981). The cast-brace is composed of three basic components: a thigh portion with a quadrilateral shaped top, the knee hinges, and the leg portion (Fig 2.6c). The orthosis holds the two fragments together and controls rotation of the fragments with relation to one another. Thus, optimal function of the limb is achieved without

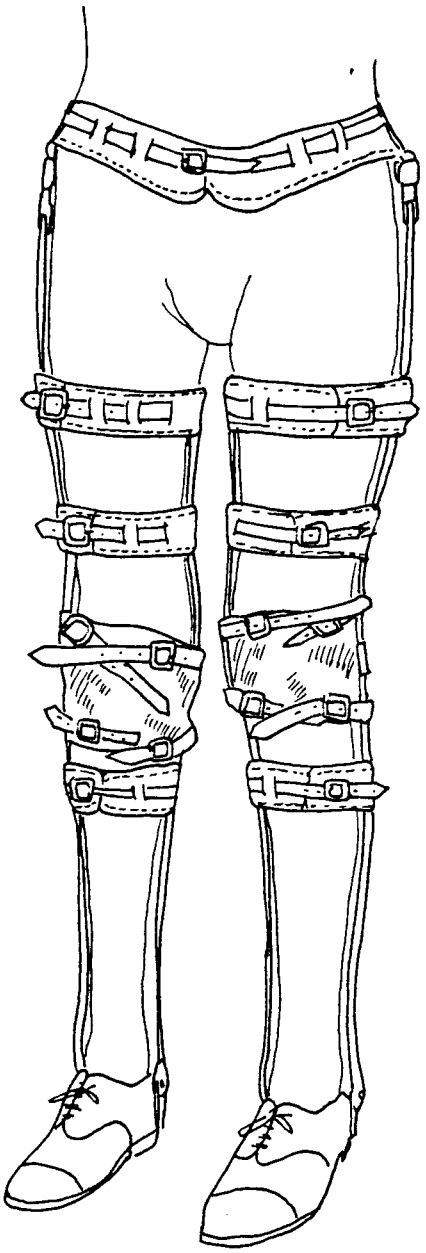


FIG 2·7 BILATERAL HKAFO

sacrifising orthotic control of alignment in the longitudinal directions, and also in the anterioposterior and mediolateral planes.

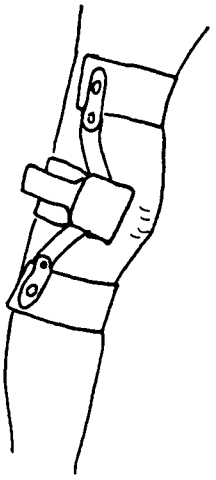
Another group of KAFOs that should be mentioned are those for reducing weight bearing stresses across an affected hip joint, such as in Legg-Perthes-Calves Disease, with the joint stabilized at a desired degree of abduction and internal rotation. The orthosis can either be non-ambulatory or be used for walking. Some of the examples of this type of KAFOs are the Snyder sling (Snyder,1949), Toronto Orthosis (Bobechko et al,1968), Birmingham Splint (Harrison & Turner,1974), Newington Ambulation-Abduction Brace (Curtis et al,1974), and the Pogo Stick Brace by Glimcher and coworkers (1970).

2.5.3 Hip-Knee-Ankle-Foot Orthoses.

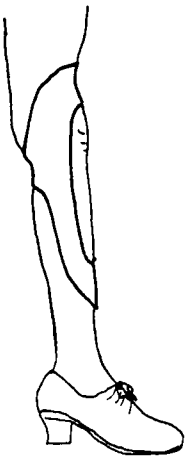
Orthoses for stabilization of the joints of the hip, knee, ankle and foot are commonly used in pediatric orthopaedics, but only rarely in rehabilitation. They are employed mainly for either directional control of joints or weight relief, with the latter being the more common indication. A portion of the force across the hip joint can be relieved with an ischial bearing thigh piece connected to the pelvic band. Orthoses used for this purpose should be as light as possible to avoid increasing the hip forces during swing phase.

Selective directional control of various anatomical joints with a HKAFO can be accomplished by virtue of the design, materials and types of joints used in fabricating the orthosis. In the conventional double-bar, plevic-band long-leg brace (Fig 2.7), selective hip control (medio-lateral and/or anterioposterior) is provided, with an option for stabilization of the knee and ankle as required. An added maximum mediolateral hip and rotation control can also be achieved with the solid metal or plastic pelvic band.

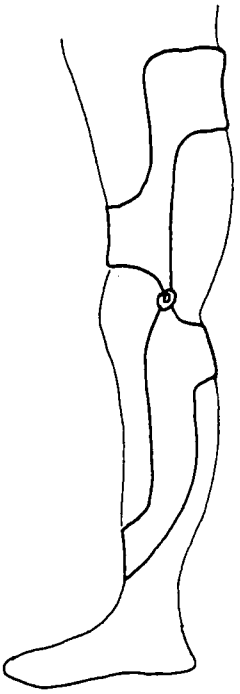
The twister orthosis, fabricated from metal-spring cable, is an attempt to provide twisting moment around long axis of the limb. Although it is effective in assisting external-internal rotation of hip, knee, ankle, or foot, it may cause rotation laxity of knee joint and produce permanent external tibial torsional deformity. Other versions of HKAFO includes a single lateral bar design (Brown et al, 1983), and an appliance to allow patients with failed total hip



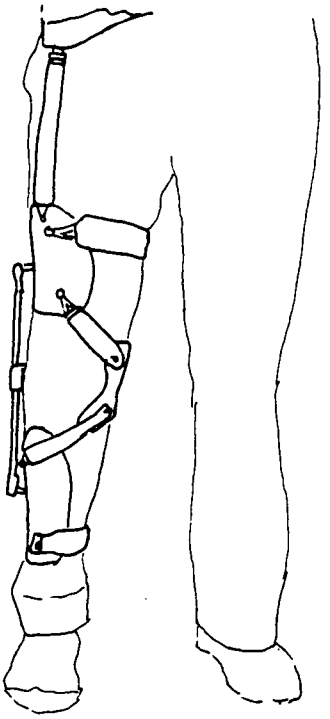
(a) SWEDISH KNEE CAGE.



(b) SUPRACONDYLAR KO.



(c) HINGED SUPRACONDYLAR KO.



(d) CARS - UBC KO.

FIG 2·8 TYPES OF KO.

prothesis to ambulate functionally, i.e without other walking aids (Jacobson & Mason, 1977).

HKAFOs have also been used as aids for standing, or for swing-through or pivoting gait. The orthotic device stabilises the hip, knee and ankle in a neutral position to allow the patient to stand on his two feet, or to ambulate in a swivel fashion. Examples of these types of KAFO are the standing orthosis, parapodium (Motloch,1971), and swivel walker (Motloch & Elliot,1966; Rose & Henshaw,1972,1973; Henshaw,1977; and Stallard, Rose & Farmer,1978).

Swing-through gait in patients with unstable hips tends to require high energy input. To get round this problem, an orthotic method of providing reciprocal walking for high thoracic levels of paralysis has been reported (Rose,1979; Major et al,1981; Butler et al, 1984). The device, called hip guidance orthosis, provides stability and control of the hip joints for reciprocal walking when used in conjunction with crutches.

2.5.4 Other Orthotic Devices.

Orthoses that extend proximally from the foot, and emcompasses more than two joints, have been discussed previously. However, there are also orthoses that span over a single joint or two adjacent joints, such as the foot orthoses, knee orthoses, and hip orthoses. In the following section, knee orthoses and a few other lower limb orthotic aids will be briefly discussed.

Knee orthoses (KO) are intended for patients with good foot and ankle joints, but require support or control of the knee. Typically, a KO consists of thigh and calf bands attached to lateral bars, with either simple or polycentric mechanical joints to allow flexion and extension of the knee. It is usually prescribed for mediolateral knee instability, but may also be used for flexion-extension control.

A device for genu recurvatum control is the Swedish Knee Cage (Lehneis,1968), which consists of a plastic coated aluminium frame with a metal semi-circular posterior band and two anterior webbing straps (Fig 2.8a). Although it has no mechanical joint, the orthosis permits free knee flexion and is cosmetically acceptable when the patient is seated. There have been reports, however, of problems in attachment of the orthosis to the knee. Ambulatory motions of the

knee tend to produce lateral and rotational displacements of the orthosis from its predetermined position. A similar light weight and non-articular knee orthosis for genu recurvatum patients had also been reported by Lehneis (Lehneis,1972; Lehneis et al,1973). The laminated plastic Supracondylar KO (Fig 2.8b) can also be used for mediolateral instability control, however, the greatest drawback of the orthosis is the cosmesis problem, as faced by the Supracondylar KAFO described in Section 2.5.2 . Cassvan et al (1977) overcomed the cosmesis problem by producing a Hinged Supracondylar KO to control both flexion and unwanted recurvatum of the knee (Fig 2.8c).

Other plastic designs include the Knee Cylinder (Pritham & Stills, 1979) for holding the knee firmly in maximum extension, and a knee orthosis using the genucentric knee joint (Foster & Milani,1979). Dixon and Palumbo (1975) reported a hinged all plastic brace that uses a latex suprapatellar strap for suspension. The Dynamic Patellar Brace (Palumbo,1981) also uses elastic straps and sleeves for rotational and anterolateral control. For patients with internally deranged knees, the External-Cruciate-Ligament Orthosis may be used (Martin, 1975).

The CARS-UBC Orthosis (Fig 2.8d) jointly developed by the Canadian Arthritis and Rhematism Society, and the University of British Columbia provides maximum support to the knee when it is extended (Cousins & Foort,1975; Reed,1979). Complete freedom for knee flexion is provided by a telescoping rod between the thigh and shank cuffs. The main drawback is its interference with clothing and this causes cosmetic problems. The TVS (Telescopic Valgus/Varus Support) KO is an attempt to improve the design (Dewar, Chodera & Ackerley,1981a).

There is a need to have paraplegic persons upright for at least a few hours a day, because a person confined to wheelchair or bed may be subjected to a variety of secondary afflictions. Paraplegics with high level of lesion can be mobilised with the use of pneumatic orthoses or pressure suits (Lehmann et al,1977; Strachan & Cook,1983). Such an orthosis consists of a light weight nylon garment with integral pneumatic tubes which, when inflated, form an almost rigid exoskeleton. Assessment studies have shown that the orthosis can be very useful for early mobilization of high level lesion paraplegics. Rabischong et al (1979) equipped a pneumatic orthosis with four electric motors to convert it into a walking machine, called the AMOLL (Active Modular

Orthosis for Lower Limb). An orthosis with a slightly different design, called the Soft-suit Orthosis was reported by Popovic et al (1981). It is a lower limb assisting device made from a jump type suit with pockets for integration of mechanical elements into the suit. The mechanical linkage system allows the orthosis wearer to self propel himself up and down from a sitting position.

CHAPTER 3.

KNEE-ANKLE-FOOT ORTHOSES.

3.1 INTRODUCTION.

The knee-ankle-foot orthosis (KAFO) is indicated for the orthotic management of knee instability, and can be either weight-relieving or knee stabilizing, or a combination of both. Although there have been a large variety of KAFOs reported in the literature, their principles of design are comparatively similar. Only the principal concepts of design which are common to most types of KAFO will be described in the following sections.

A literature review of the orthotic load measurement systems will also be presented, which include the KAFO loadings in both static and dynamic situations, KAFO component testing, and femoral fracture KAFO analysis. For completeness of the review and also for a comparison of the results obtained between the AFO and KAFO loadings, the investigations conducted on AFO loadings will also be discussed.

3.2 PRINCIPLES OF DESIGN.

The design of an orthosis is principally based on the laws of mechanics, which govern the function of the device in controlling body motions. Ideally, an orthosis should have a high degree of safety while at the same time provide effortless ambulation. However, strength and weight are two inter-related, but contradictory parameters where a compromise will be required. A high strength to weight ratio should be achieved in the design.

Duration of use is another parameter to be considered. In the treatment of a transient disorder, a temporary orthosis with simple design is probably all that is required. In situations where an orthosis will be used functionally for a long period of time, it should be designed for strength and durability to reduce the risk of a mechanical breakdown. Details of the design considerations will be discussed in the following sections.

3.2.1 Anatomical Considerations.

An orthosis functions by applying forces through the soft tissues to the musculoskeletal system. To achieve an effective and comfortable control, sites of force application should be carefully selected. Pressures of over 25 mmHg (3.3×10^{-3} MPa) applied continuously over the soft tissues can seriously restrict circulation and cause necrosis.

If applied intermittently, most soft tissues can tolerate higher pressures. However, it was reported by Alcock and Redhead (1970) that a cyclic pressure applied over a long period of time may eventually produce damange, even though it might not produce prohibitive discomfort or pain. Therefore, the controlling forces should be spread over an area as large as possible.

Experienced gained from prosthetic research can also be used in the orthotic design. A large percentage of the body weight can be carried through the ischial tuberosity (Radcliffe,1955), and also through the proximal flares of the tibial condyles and the patellar tendon (Radcliffe & Foort,1961). A small proportion of the vertical force can also carried by the use of a close-fitting thigh corset.

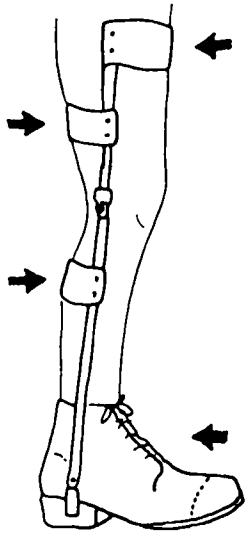
Pressure sensitive areas such as the malleoli and the head and neck of the fibula should be free from any pressure at all times. Pressure should also be avoided over any area of poor quality skin, which may have resulted from burns, injuries, or closure of a spina bifida lesion. If pressure is inevitable, the orthosis must be carefully fitted and adequately padded with no stitching or rivets in contact with the skin.

3.2.2 Kinetic Considerations.

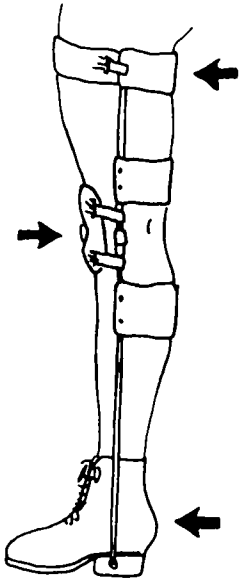
This is perhaps the most important aspect of the orthotic design. Since an orthosis can be considered as a controlling, supporting, correcting, or a compensating force system, the loading pattern of the orthosis will therefore determine the effectiveness of the design.

The design of the orthosis must reflect the need to withstand forces involved both in the static and dynamic situations. Stabilizing forces to maintain equilibrium in standing can be provided with relative ease. In dynamic ambulatory activities, the forces experienced by the orthosis are much more severe than the static case. High loadings are also expected when there is a change of state, such as changing direction during walking. It was reported that swing-through gaits produced even greater forces than those with reciprocal ambulation (Rose, Butler & Stallard,1982).

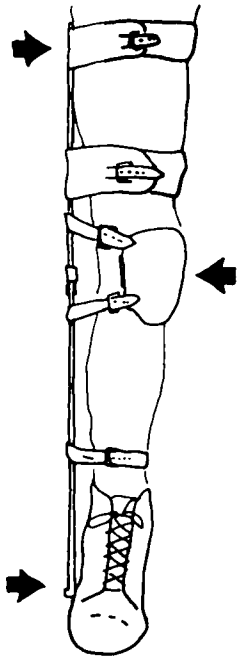
From the engineering point of view, there are essentially 2 types of systems of forces acting on a KAFO: the weight-bearing force system, and the knee-stabilizing force system. The magnitude of the



CONTROL OF
GENU RECURVATUM.



CONTROL OF
KNEE FLEXION.



CONTROL OF
GENU VARUM.

(ARROWS SHOW FORCES ON LIMB)

FIG 3·1

Examples of Knee Stabilizing Forces.

axial force experienced by a weight-bearing KAFO is dependent on the construction and fitting of the orthosis, and may be as high as 1.2 times body weight during normal level walking (Anderson & Henshaw, 1977). In this loading condition, the uprights are acting as struts, therefore, the free length of the uprights should be as short as possible to prevent buckling. Another factor that should be taken into account is the twisting actions frequently associated with weight-bearing KAFOs, which is generated at the ischial-bearing top and transmitted to the uprights as twisting moments.

The 'three-point fixation' force system is the basic requirement of a knee-stabilizing KAFO, which immobilizes the knee and thus allows the body weight to be supported through the skeleton. Since a single 3-point system will control motion only in a particular plane, most orthoses will need more than a 3-point system to safely secure the weakened limb within the orthosis. Additional straps and bands are normally added to the orthosis for strengthening purposes, which effectively transform the 3-point systems into a statically indeterminate structure.

A moment is produced at the joint centre by the 3-point loading to control the angular displacement of the knee. Since moment of a force at a point is the product of the force vector and the perpendicular distance between the line of action of the force and that point, the requisite moment is dependent on the magnitude of the force and the moment arm. For this reason, it is desirable to use the maximum available length of moment arm to reduce the forces on the orthosis. Furthermore, the middle force should be located directly over the affected joint because if the middle force is moved away from the joint toward one end, the forces needed to provide the necessary corrective moment will be increased in the 3-point system. Thus the conflicting interest of mechanical efficiency against cosmesis and functional factors should be carefully compromised. Examples of the use of 3-point system in the control of genu recurvatum, knee flexion, and genu varum/valgum are illustrated in Fig 3.1 .

The forces at the knee strap are dependent on the angle of flexion of the knee, and increase significantly with the degree of knee flexion contracture. Bunch and Keagy (1976) suggested that the maximum knee flexion that can be supported by an orthosis is 25°.

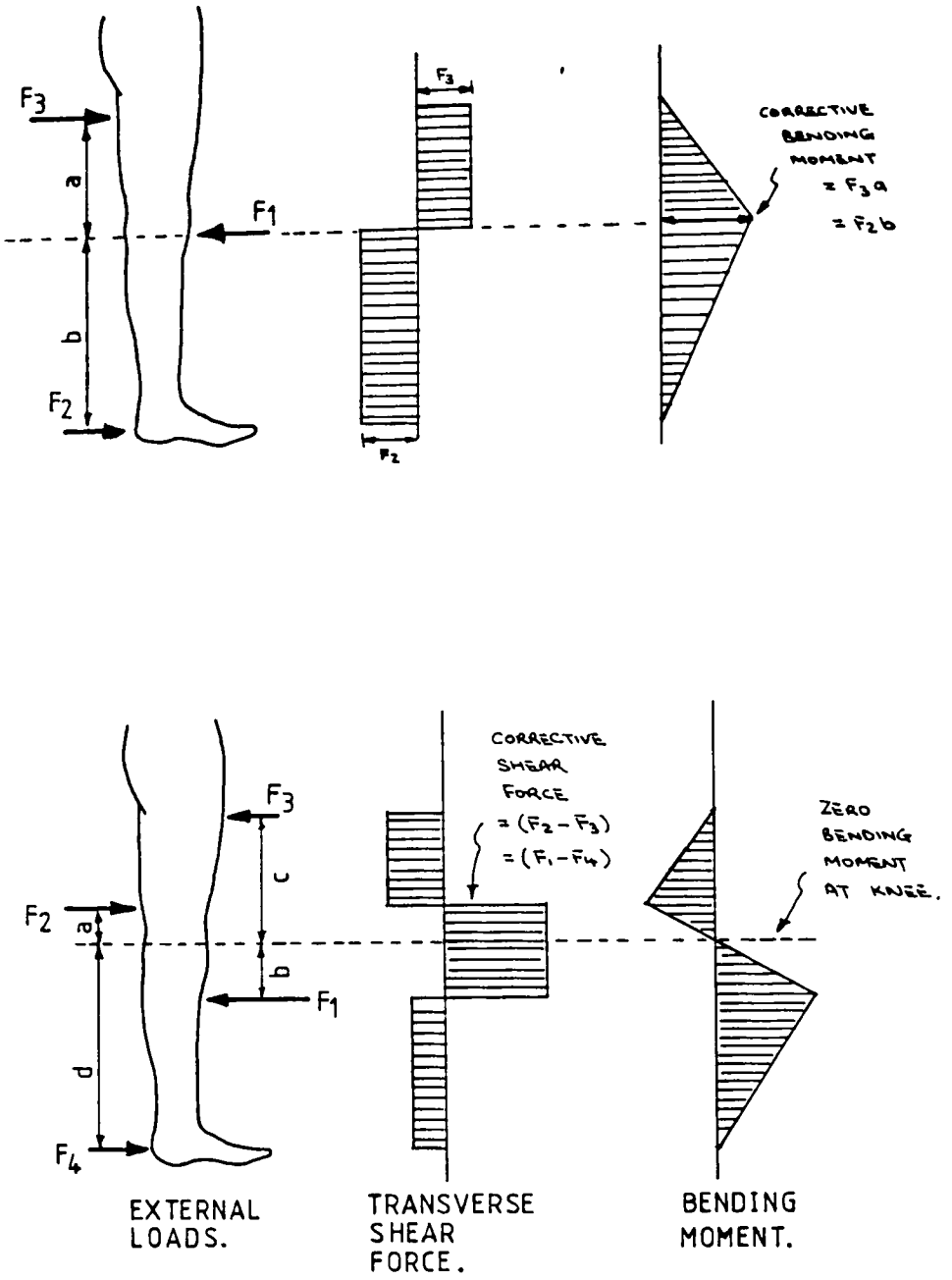


FIG 3·2	Effects of the Placement of Corrective Forces on the Knee.

Fig 3.2 shows the effects of the placement of corerctive forces on the knee. With the basic 3-force system, a considerable amount of bending moemnt can be generated at the knee, while by adding an extra strap, the moments at the knee may be reduced to zero to protect the weakened knee joint, ligaments, or for any other application.

3.2.3 Alignment.

Since none of the joints of the lower limb is a simple unaxial joint, the alignment of the mechanical joints with respect to the anatomical joints will also determine the performance of the orthosis. Ideally, the mechanical joint centre of rotation should correspond precisely with the anatomical joint axis throughout the motion range. However, simple hinge joints are normally used in orthotic designs, and thus precise alignment is impossible.

The concepts of orthotic fit and alignment are very interrelated. Fit is concerned with the relationship between the orthosis, anatomical landmarks and body contours, while alignment covers the angular relationship of the orthotic components to each other and to a reference line relating the orthosis to the body as a whole. Thus, a change in either fit or alignment will inevitably affect the other.

Anatagonistic forces which cause undesirable linear and rotational displacements of the orthosis with respect to the enclosed limb may be generated from a malaligned situation. This not only causes discomfort, but also produces tissue damage due to the shearing effect. Incorrect alignment is due to either non-parallel or linearly displaced axes systems, where the former can induce a twisting of the body segment in the orthosis, while the latter case causes sliding of orthotic component over the body segment.

Although knee alignment is critical in functional KAFOs, it is not so important in KAFOs with locked knees, since the main function of the knee joints will be to facilitate sitting. However, most KAFOs allow a certain degree of ankle motion in the sagittal plane, and this should be carefully aligned. The common practice in the U.K. is to place the ankle joint at the heel of the shoe, using the spur and socket attachment. This arrangement produces a situation where the axes may be aligned parallel but displaced from one another. Severe limitation of ankle motion can be caused by this out-of-alignment axes

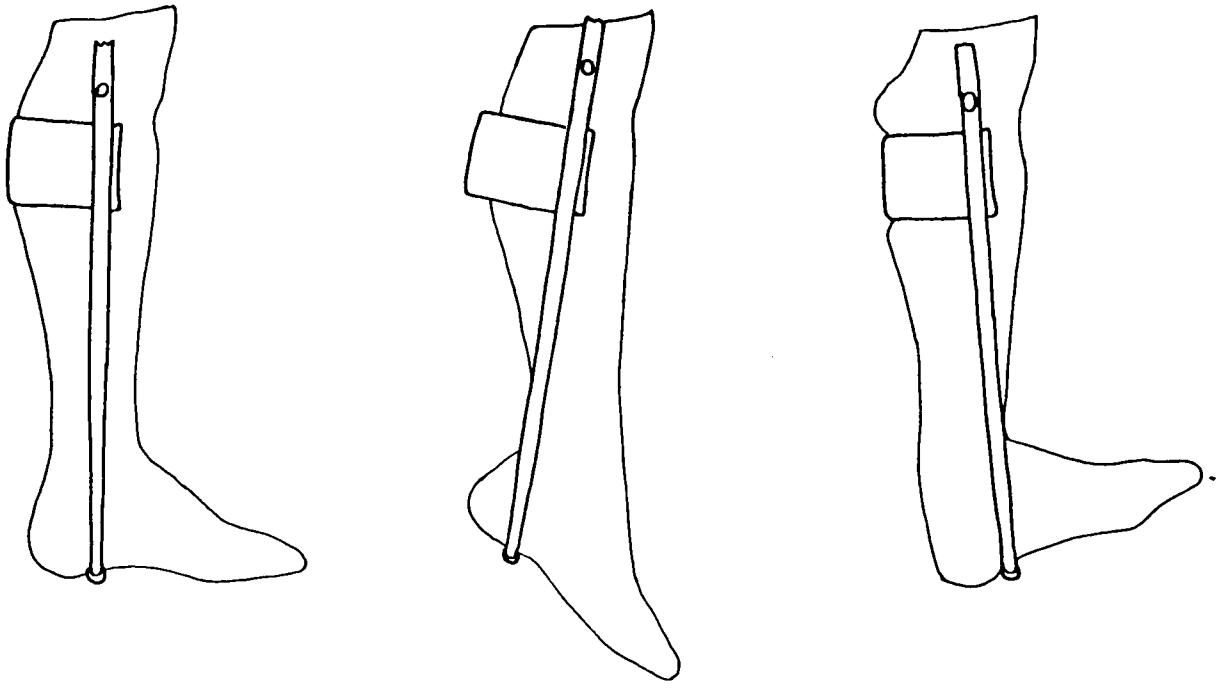


FIG 3·3	Malalignment Caused by the Spur and Socket System.

system, and may result in undesirable stresses between the orthosis and the limb (Fig 3.3). Moreover, the uprights need to be spread laterally at each change of shoe, which can cause considerable distortion of the fit and alignment.

Another important alignment parameter is the amount of tibial torsion and degree of toe-out. It affects both the gait pattern and comfort in a seated position.

3.2.4 Material and Strength Requirements.

The important factors in orthotic material selection are: strength, weight, stiffness, fatigue strength, resistance to wear and corrosion, ease of fabrication, and cost. Among these, strength and weight should be carefully considered.

In general, higher energy expenditure is needed for patients with heavy orthoses, and thus the lighter the KAFO, the more acceptable it will be to the patient. However, the size and weight of an orthosis required by a physically handicapped person usually increases with the degree of disability. This means that the most severely handicapped, especially the smallest and lightest persons, will be further handicapped by the additional weight of their appliances. Nelham (1978) estimated that the HKAFOs used for spina bifida patients amounted to 7 - 15% of the child's body weight, and around 6% for that of an adult.

Inadequate design and poor material specification can often lead to failure by a ductile, brittle, or fatigue mechanism. Ductile failure is caused by single-cycle overload generally arising from an emergency or misuse, and suggests an under-designed orthosis. It is identified by structural distortion at the region of fracture, where there is a reduction in the cross-sectional area. Brittle failure is a sudden failure under a single abruptly applied stress field, and is initiated at a discontinuity such as a sharp edge, an imperfection in the material, or a hole in the structural member which produces significant local stress concentration. The metal may actually yield in the immediate vicinity of the discontinuity, but the overall failure resembles a brittle failure.

Probably the most frequent mode of failure is that due to seemingly moderate levels of repeated or fluctuating stresses. This fatigue failure also originates at a discontinuity or stress

Table 3.1 Material Specification for KAFO Components.

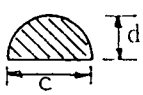
KAFO Components	Material Specification. *	Dimensions, Remarks.
Side-members	070M55 (En 9) 635A14 (En351)	Side-member & lower adjustable pieces.**
	070M55 (En9) 070M20 (En3B)	Upper adjustable pieces.**
	HE15TB (HE15WP) (2014A)	Rectangular section 20 x 6 (mm) for adult 12 x 6 (mm) for children.
	410S21 (En56A)	(Kellie,1982) Modular side bars.
Ring-top	070M20 (En3B)	Steel rod, Large: 8 mm ϕ Medium: 6.4 mm ϕ Small: 4.8 mm ϕ
Calf & thigh bands	HS50 [High carbon HS60 steel] 070M55	25 mm wide (adult) 20 mm wide (children) min thickness 1.2 mm.
	HS15TB HC15TB	width as above, 2.03 mm thickness.
Knee & ankle joints.	070M55 (En9) 635A15 (En351) 220M07 (En1A)	For joint members & locking rings.
	HE15TB ANC1B (Stainless steel).	(Kellie,1982)
Spurs & sockets	220M07 (En1A) 070M55 (En9)	Spurs, large: 7.9 mm medium: 6.3 mm small: 4.7 mm
	070M55 (En9)	Socket.

* All material specifications are according to BS 2574,BS 1932 & BS 3563, unless otherwise stated.

** Dimensions of D-sections are:

Adult: 

a = 12.7 mm
b = 6.35 mm

Children: 

c = 11.1 mm
d = 5.6 mm

concentration point, and may be responsible for 90% of all orthotic breakages (Murphy & Burstein,1975). These failures may be avoided by using sound engineering design, and by having the working stress below the endurance limit for the material, which is recommended to be a third of the ultimate tensile stress of mild steel. For high tensile steel, it should be less than half the ultimate stress. Aluminium alloys, however, do not possess such endurance limit and have a finite life in service.

In attempts to standardise the KAFO components and to reduce the incidences of failure, the British Standard Institution has published specifications and guidelines on the types of material and dimensions of components to be used for orthotic construction (BS 2574, BS 2931, BS 2932 & BS 3563). A list of the KAFO components with their appropriate material specifications are given in Table 3.1 , while Table 3.2 provides a brief description of the material properties as obtained from BS 970, BS 1474 and also from Smithells and Braudes (1976).

In recent years, the role of sheet thermoplastics materials has become more and more prominent in orthotics. Their advantages include high strength-weight ratio, easy to work with, good fatigue characteristics, inexpensive, and hygenically and cosmetically more acceptable to the patient. High density polypropylene has performed satisfactory where large loads are involved, and can withstand several million cycles of flexures before showing any sign of failure (Yates,1974). Although high density polyethylene is not very suitable as a load carrying structure in lower limb orthotics, it is often superior for use in cuffs and other auxiliary components.

The ability of thermoplastic orthoses to support large loads depends on 3 major variables, namely, the properties of the polymer, the geometrical shape, and the cross-sectional area. The geometrical shape is mainly determined by the anatomical configuration of the enclosed limb segment, while the cross-sectional area is dependent on the thickness of the material used and the placement of the trimlines. Greater strength can also be achieved by reinforcing with strips of polypropylene at the weaker points before the orthosis is being vacuum formed (Shower & David,1983).

An improvement in the strength-to-weight ratio can be achieved by the use of Carbon Fibre Reinforced Plastics (CFRP). In a series of experimental work, struts that are compatible in strength and stiffness

Table 3.2 Properties of Materials used for KAFO side-members.

Material Specification.	070M55	635A14	070M20	410S21	HE15TB
Other Designations.	En9, 55 Carbon Steel	En351, ¾% Nickel Chromium Steel	En3B, 0.2 Carbon Steel	En56A, Martensitic 13% Chromium Steel	2014A, HE15WP, Duralumin
Tensile Strength (MPa)	700	700	540 - 700	540 - 700	410
Yield Strength (MPa)	355	355	355	370	-
0.2% Proof Strength (MPa)	-	-	-	-	255
Elongation on $5.65\sqrt{So}$ (%)	12	12	20	20	10
Hardness (HB)	202-255	201-255	152-207	152-207	115
Fatigue Strength (smooth specimen, room temperatures)	±293 MPa @ 10×10^6 cycles	-	±193 MPa @ 10×10^6 cycles	±340 MPa @ 10×10^6 cycles	170 MPa @ 50×10^6 cycles
Heat treatment	Hot forged at 1150-850°C Normalised at 810-840°C Air quenched	Hot forged at 1150-900°C Hardened at 850°C Oil quenched Tempered at 580-620°C	-	(Condition P) Hardened at 950-1020°C Tempered at 650-750°C	Heat treatmed at 450-540°C Solution quenched Naturally aged in room temperature.

to steel uprights, but only 25% of the weight were obtained by combining CFRP with I-shaped aluminium extrusion beams (Nelham,1978,1981; Ring & Florence,1979). However, CFRP is relatively costly to produce, and may fail by delamination under high bending or torsional stresses. Johnson (1978b) improved the brittle nature of CFRP with a carbon/glass-fibre hybrid laminate, and claimed that it functioned well in the Stanmore KAFOs.

3.2.5 Control Mechanisms.

Undesirable ankle motions can be controlled by the use of mechanical ankle joints with stops. For the spur and socket design, an ankle strap is needed to retain the spur piece in the heel socket during ambulation. However, the strap can also seriously interfere with the free motion of the foot.

A leather T-strap provides ankle stability control, and substitutes for paralysed or weakened invertor or evertor muscles. The vertical limb of the 'T' is attached to the shoe, while the long tongues at the upper end of the strap encircles both the ankle and the side bars, thus providing a counter-acting force. At the knee, a somewhat similar principle may be used for genu valgum or varum, where a leather medial or lateral strap buckles around one upright and pulls the knee towards that upright with a medially or laterally directed correction force.

Knee flexion control can be provided by a variety of devices. A common British design is the knee cap strap (knee apron) which applies forces directly onto the patella. Other knee retaining straps include the patellar tendon and the supra-patellar straps. A calf band is needed in conjunction with all the above cases. For effective knee stabilization, it is also necessary to keep the knee locked in extension, however, the patient's energy expenditure can increase drastically during ambulation. Circumduction of the swing leg and vaulting on the stance leg are the normal compensatory actions for the locked knee.

3.2.6 Cosmesis.

Although frequently neglected by the rehabilitation team, the appearance of an orthosis is normally a very important consideration for an adult patient. The orthosis supplied should be as inconspicuous as possible. Within the constraint of resources and pathological and anatomical considerations, plastic moulded tops and ankle-foot pieces should be used to improve the appearance. The uprights and their mechanical joints should closely follow the contour of the enclosed limb. The external shoe attachment points can be replaced by a cosmetically more acceptable insert. Finally, metals should be coated with a thin layer of matt plastics to match the patient's skin colour.

3.3 PRESCRIPTION AND SUPPLY.

The fundamental aim of orthotic prescription and supply is the provision of orthosis capable of generating rational and predicted effects directly related to the medical needs, and with a minimum of physical and cosmetic side effects. In addition, they should also be reliable, safe, and requiring the minimum of maintenance. To prescribe an effective orthosis, the physician concerned should have a working knowledge of the capabilities and limitations of the orthotic appliance. His responsibility would be to analyse the patient's physical and functional deficits, and to specify the controls required to overcome or minimise these disabilities. On the other hand, the orthotist measures the patients and records these values so that unambiguous information can be transferred to the orthotic technicians responsible for the manufacture of the appropriate device. Therefore he should be proficient in physiology and medicine to understand the clinical reasons for the prescription.

3.3.1 Orthotic Prescription.

Fundamental to a sound prescription is an accurate and complete analysis of the patient's physiological condition and functional impairments. It involves assessments and gait observation to establish the nature and degree of the functional loss, which provides a basis for matching the patient's requirements with an appropriate orthosis.

Therefore, the approach is founded not on the disease, but primarily on the biomechanical deficits present. However, there is a limit on the ability of orthoses to improve pathological gait, thus calling for a compromise between the expectation and capability of an orthosis.

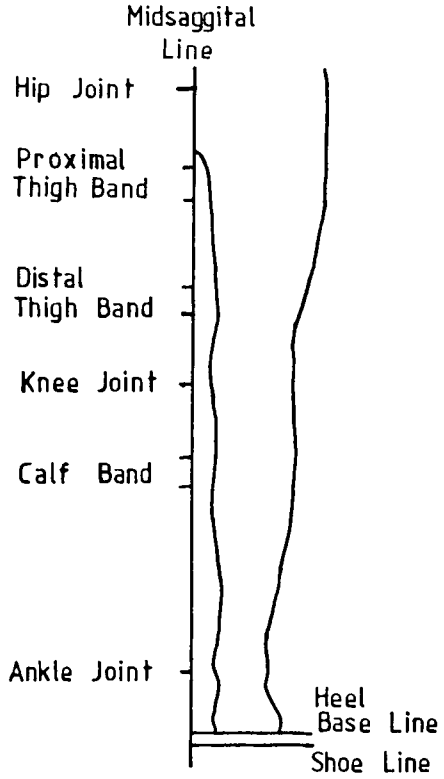
Some of the orthotic prescription factors are summarised below:

1. patient's history, including home environment and general health conditions;
2. physical status of the patient;
3. physical demands of the patient's vocational and recreational pursuits;
4. the gait characteristics;
5. the specific functional losses to be substituted;
6. the functional deficits that may interfere with orthotic application, such as impaired or absent of selective motor control, dysequilibrium, and body-image deficit; and
7. psychological aspects.

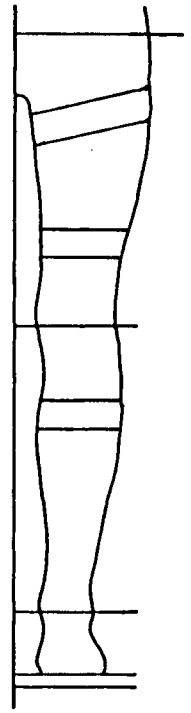
In attempts to organise a systematic approach to orthotic prescription, several research centres had developed their own prescription forms or charts to match the biomechanical benefits of orthosis (CPRD, 1972; McCollough,1975). Among the charts available, the AAOS Lower Limb Technical Analysis Form has been widely accepted (see Appendix 2). On this form, the patient's sensory, motor and skeletal deficits are detailed graphically for easy and quick interpretation. With proper use, the chart can quickly identify the biomechanical requirements and the controls desired, thus enabling an orthosis to be prescribed on the basis of the patient's functional disability.

Training on the maintenance of orthoses, ambulatory activity exercises, and gait training are often needed to overcome problems in the control of the orthosis, and to acquire the maximum benefit from its use. The patient should also be educated in the functions of the orthosis prescribed to him.

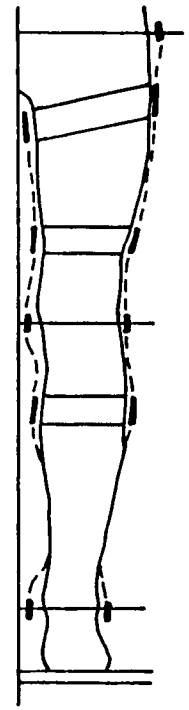
As the patient receives his new orthotic device, a systematic examination of the patient with the orthosis as a biomechanical entity should be carried out. It involves checking for fit, alignment, and general performance of the orthosis. Thereafter, periodic follow-up examinations should be carried out to ensure that maximum functional efficiency of the device is maintained.



LOCATION OF JOINTS AND BANDS.



HORIZONTAL LAYOUT.



VERTICAL LAYOUT.

(Adapted from N.Y.U.,1981)

FIG 3·4 The Layout of a KAFO.

3.3.2 Fabrication of Orthosis.

The fabrication of a KAFO principally consists of measurement and casting of the limb, and layout and construction of the orthosis. At the measurement stage, important landmarks on the limb which can serve as reference points for bands and joints of the orthosis are loacted. These landmarks locations, together with the outlines of the limb are transferred onto a tracing paper, which will be used as the basis for laying-out of the orthotic components.

In some situations, plaster-of-paris bandage casts may be required to define accurately the shapes of the quadrilateral brim and the ankle-foot component. The correction desired is moulded onto the cast by shaping and holding the limb segment in the required position while the bandage sets. Positive moulds are subsequently made from these cast impressions, and the moulds modified, if necessary, to enhance the correction desired. Thermoplastic quadrilateral brim or ankle-foot piece can then be vacuum formed from the positive casts.

The various components are cut and shaped to match the tracing obtained previously from the patient (Fig 3.4). In addition to conforming to the contour of the limb, the metal uprights are also required to accommodate the tibial torsion. With the mechanical knee joints mounted on a jig aligned closely with that of the tracing, the complete KAFO is then fabricated.

3.3.3 Patients' Response.

For an orthosis to be accepted by the patient, it should be prescribed on both the factors of biomechanical functions and its adaptability into daily living. Comfort, appearance, simplicity, efficiency, reliability, and ease of application and removal are all integral factors influencing a patient's acceptance of an orthosis.

Kaplan et al (1966) conducted a follow-up study of 179 patients with spinal cord dysfunction after completion of their inpatient rehabilitation stay. They reported that a very low percentage (21%) of the patients were consistently using the KAFOs supplied to them. It was also found that patients with poliomyelitis disability tend to accept the orthosis more readily than those with traumatic or other disability. The reason for rejection of the orthoses include discomfort, difficulty in the use of the orthosis, and inability to perform

daily living activities with the orthosis.

An investigation of the patients' experience with their National Health Service (NHS) supplied KAFOs was carried out by Jay and Dunne (1976). They sampled 39 patients from a number of London hospitals, and reported that 30 patients (i.e. 77%) wore them regularly. The main causes of complaint were discomfort, poor reliability, long delivery time, and cosmetic unacceptability. Most patients found their KAFOs uncomfortable on first wearing, and 40% of them had tried to alter their own orthoses. A quarter of them still found the orthosis too heavy, even after its initial 'break-in' period. An exceptionally poor record of reliability was also reported, with 70% requiring major repairs, and more than 50% having performed minor repairs themselves. One particular patient had his orthosis broken 3 times in $2\frac{1}{2}$ years. The delivery time ranging from a month or so to more than 9 months was also considered to be unacceptable. Another common complaint was the 'old fashioned and ugly' appearance of the orthoses. It is therefore evident that cosmesis is far more important to patient than had been assumed.

The findings of Jay and Dunne were confirmed by the Royal Association for Disability and Rehabilitation (RADAR) with their report on the availability and supply of orthotic appliances in the NHS (RADAR, 1983). The general view portrayed was one of most unsatisfactory orthotic provision, with very high rejection rate, and frequent do-it-yourself type of alteration by the patients. RADAR also reported that the appearance of the orthoses is the major source of dissatisfaction.

Therefore, the orthoses supplied to the patients should not only be functional and reliable, but also of a cosmetic quality that will improve the appearance, rather than drawing attention to their disabilities.

3.4 REVIEWS OF ORTHOTIC LOAD MEASUREMENT SYSTEMS.

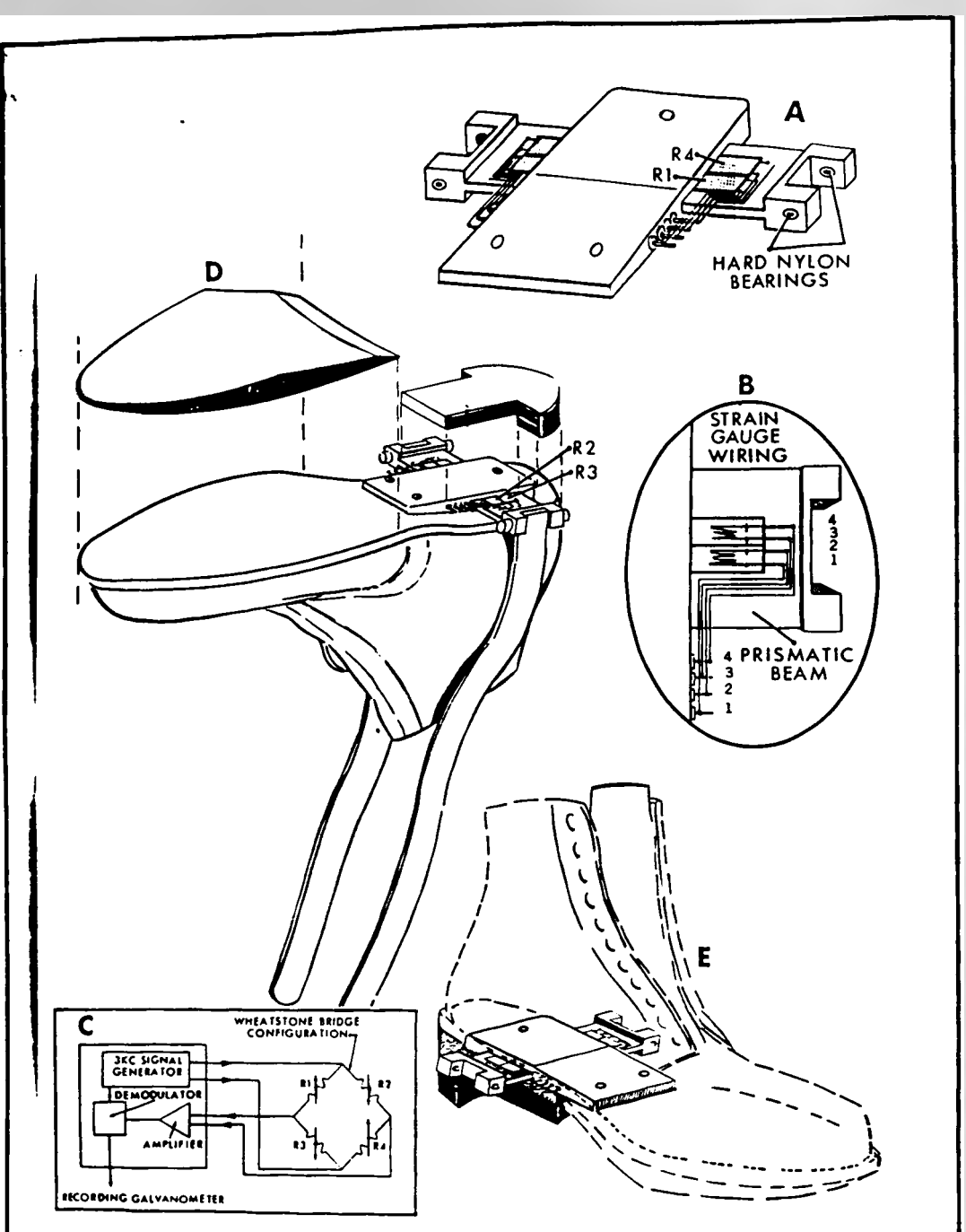
The evaluation of an orthotic system is a two-fold process, comprising clinical as well as biomechanical assessments. The clinical evaluation is a relatively straight forward assessment to determine whether the device is acceptable to the patient, and may also include simple qualitative gait analysis, with and without the orthosis.

However, the biomechanical evaluation is principally a quantitative analysis of the mechanical characteristics and biomechanical function of the device. The mechanical characteristics can be derived from loading tests performed on the orthosis, while for biomechanical assessment, measurements are taken with the orthosis actually being used by the patient.

A search of the literature revealed a considerable lack of information on comprehensive KAFO load analyses. The types of investigation carried out by various researchers over the years may be conveniently divided into the following aspects, namely, femoral fracture KAFO analysis, KAFO component testing, KAFO loading in static situations, and the performance of KAFO while being used by patients. For completeness, AFO load measurement studies are also reviewed.

3.4.1 AFO Loading Investigations.

The researchers at Hadassah University Hospital in Jerusalem, Isreal, had performed a series of investigations on the performance of several types of AFO. Miniature strain gauges were attached to the duralumin stirrup beneath the ankle joints of conventional double upright AFOs, and tests were carried out on 3 normal healthy subjects (Magora et al,1968). They reported that the highest loads were found in the outer uprights, although the actual load distribution between the two uprights varies according to the gait pattern. A total orthotic axial load in excess of the body weight of the subject was also recorded when the subject walked with forced inversion in his foot. Robin and coworkers (1968) from the same centre, also investigated seven types of AFO, with gauges located on the uprights at two different levels, and the results were obtained from a single normal subject. The highest stresses were recorded in posterior spring AFOs, where they reached a value of 300MPa. This suggested that even if the orthosis were to be made from aircraft high quality aluminium alloy, it would still have an estimated life of less than 1,000,000 cycles, i.e. a life-span of approximately a year for an average person. Peak stresses of 110 MPa were obtained in unilateral upright AFOs. For the bilateral upright AFOs, a much smaller stress level was recorded, with the lateral upright consistently demonstrating higher stresses than its medial counterpart.



Schematic representation of instrumented weight-bearing braces. *A*, Cantilever beam with strain gauges attached; *B*, close-up of the strain gauges on the beam; *C*, strain-gauge circuitry; *D*, *E*, installation of instruments in orthopaedic shoe.

(from Staros and Peizer, 1964)

FIG 3·5	An Instrumented Stirrup For Axial Load Measurements.

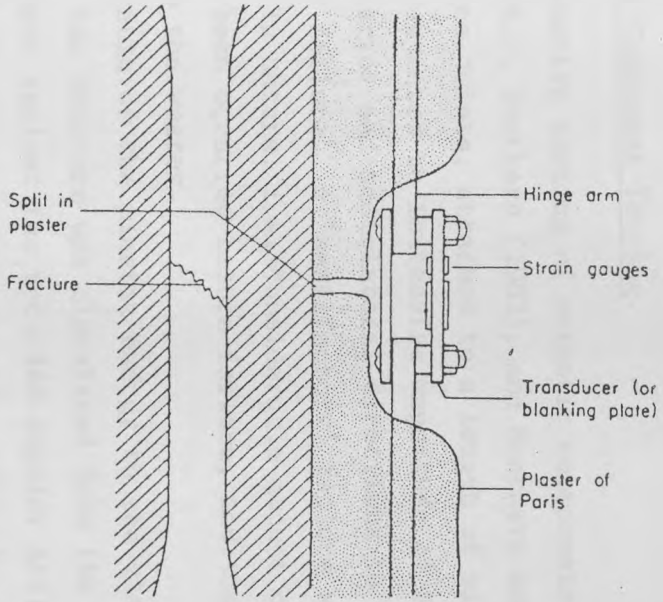
Robin and Magora (1969) continued the investigation using a similar set-up, but on 4 post-poliomyelitis patients with unilateral drop-foot. For double upright AFOs, the anterioposterior forces were found to be similar to the normals, while the mediolaterally directed forces were about twice as high as those of a non-paralysed limb. The same 4 patients were subsequently tested with polyester AFOs, where the gauges were fitted to the metal reinforcing plates below the stainless steel ankle joints (Robin et al,1971). The relationship between muscle actions and mechanical stresses in AFOs were also investigated by Simkim and coworkers (1973).

To measure the unloading of the leg in VAPC weight-bearing AFOs, Staros and Peizer (1964) instrumented the orthosis stirrup so that the axial load carried by both the uprights could be recorded (Fig 3.5). When used in conjunction with a force platform and cyclography, the weight-bearing of the limb can be computed from the ground reaction force and the load carried by the brace. The force actions between the calf band of an AFO and its enclosed limb had also been measured with calf band transducers (Lee & Johnson,1974). When tested with 10 normal subjects, these forces were found to be very small for orthoses with either free ankle motion or dorsiflexion assist mechanisms. However, for an orthosis with 90° plantar-flexion stop, the forces at the calf band reached 90N at heel strike, and 55N at push-off.

Lehmann et al (1980) studied the AFO functions in patients with flaccid paralysis of muscles which were innervated by the peroneal and tibial nerves. Eight normal adults, wearing double-upright AFOs were used as subjects, with nerve blocks performed aseptically by a physician. The highest loadings in the orthoses were noted in subjects with both peroneal and tibial blocks, where the dorsiflexion moment and plantar-flexion moment reached 28 Nm and 20 Nm respectively.

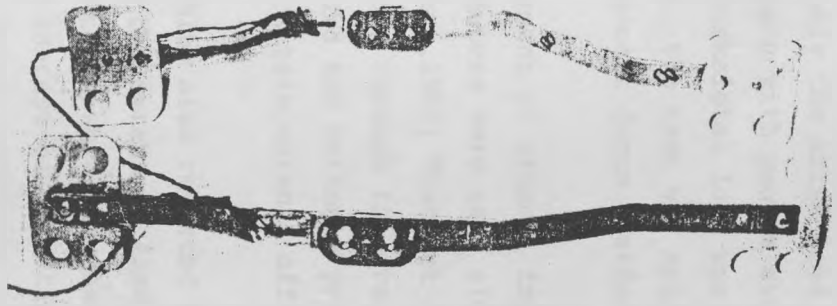
3.4.2 Femoral Fracture KAFO Load Measurements.

In recent years, as a result of the collaboration between the University of Aberdeen and the Aberdeen Royal Infirmary, a number of publications on the biomechanical studies of femoral fracture KAFOs have been presented. They investigated both the static weight-bearing of a 'cast-brace' and the dynamic fracture loadings during gait. For static load measurement, detachable strain-gauged transducers were



(a) Mounting of transducer in hinge arm and division of cast at fracture level

(from Wardlaw et al , 1979)



(b) *A pair of knee hinges with strain gauges attached.*

(from Hardy , 1981)

FIG 3·6	Measurements of Axial Loads in Femoral Fracture KAFO.

mounted at the level of fracture, and also at the upper arms of the knee hinges (Wardlaw et al,1979,1981). The single component transducers, which measure only the axial load, each consists of a flat rectangular duralumin plate with gauges attached on both sides (Fig 3.6a). To ensure that all exoskeletal load was transmitted to the transducers at the fracture level, the cast was split circumferentially at that level. A single-component force platform recorded the total load on the limb.

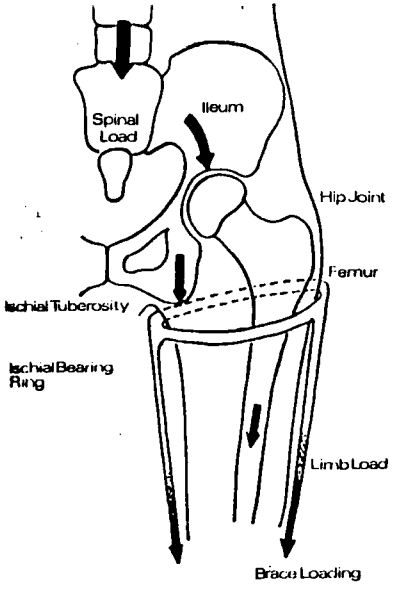
For the measurement of dynamic fracture loading during gait, multicomponent transducers were used simultaneously with TV and force platform systems (Pratt,1981; Pratt et al,1982). The results obtained indicated that the load through the fracture progressively increased as union proceeded, and the orthosis provided a mechanism whereby the patient could, to a certain extent, off-load the weight at fracture site.

Dewar et al(1981b) also reported on an instrumented femur fracture KAFO to measure the extent of load-bearing provided by the orthosis, where four load cells were attached to the above-knee section of the cast. A similar concept of investigation as the Aberdeen researchers was adopted by Hardy (1981) who strain gauged the aluminium knee hinges (Fig 3.6b). Ground reaction was measured by standing scales.

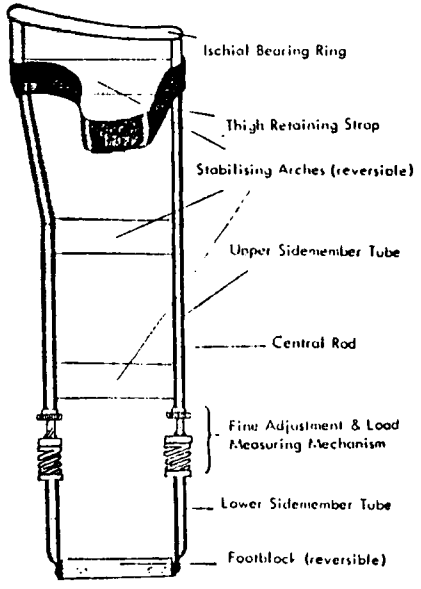
3.4.3 KAFO Component Testing.

Destructive testing of orthotic components has been reported by Johnson (1982), Scothern (1982), and Scothern and Johnson (1984). Orthotic knee joints, attached to a length of side member on either end, were subjected to increasing magnitude of bending moment in the anterioposterior and mediolateral directions to identify their strength and mode of failure. A 'satisfactory' failure was taken as one that is ductile in nature, i.e. considerable amount of plastic deformations would have been occurred before failure, and thus less likely to cause injuries to the wearer.

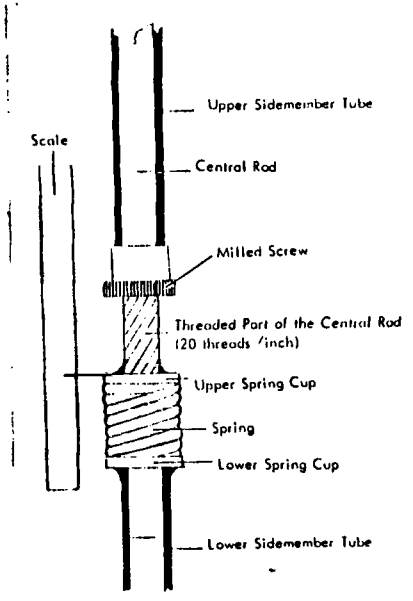
The ratio of the plastic energy to the elastic energy (PE ratio) leading to the fracture was claculated from the graph of the applied bending moment against the recorded angular deflection. This ratio was taken as the basis for quantification of failures. The higher the value of this ratio, the more ductile will be the failure. From



The ischial loading of an orthosis.



The apparatus.



Fine adjustment and load measuring mechanism.

(from Anderson , 1977)

FIG 3·7	KAFO Axial Load Measurement.

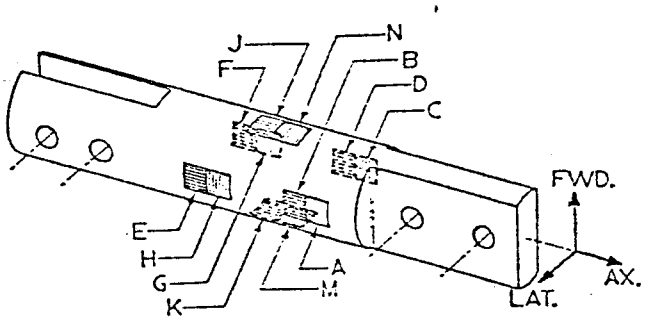
examination of a group of results and their modes of failure, the investigators concluded that a PE ratio of less than 5 represents a totally unaccepted failure while a ratio of greater than 10 will be acceptable. They also reported that if the first component to fail were to be a casting, the failure would not be a satisfactory one. The majority of satisfactory failures were associated with components machined from wrought materials.

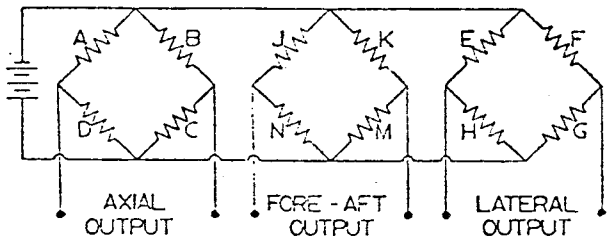
3.4.4 KAFO Loadings in Static Situations.

The axial load of a KAFO in the static phase was measured by Anderson (1974,1977) in an attempt to relate the length of the orthosis with its weight-bearing capability. The experimental orthosis was based on a standard adult "Thomas' Splint", with adjustable length and thigh circumference to suit every patient. A set of calibrated springs act as the load measuring devices, and were incorporated into each side member (Fig 3.7). Ground reaction forces were measured with calibrated bathroom scales. Anderson reported that loading was not directly proportional to the length of the orthosis, and that it unloaded the leg by approximately 50% when the orthosis had the same length as the limb (i.e. ischial tuberosity to floor). He also reported that the loads on the lateral side-member of the KAFO were smaller than on the medial upright. However, Robin and coworkers (1969) obtained a results showing the lateral uprights were more highly stressed than the medial uprigths of AFOs (see Section 3.4.1).

3.4.5 Dynamic Loadings of KAFO.

Comprehensive measurements of KAFO loadings while actually being used by patients have been undertaken in the U.S.A. by the researchers at the University of Washington in Seattle. Kirkpatrick et al (1969) investigated the distribution of loads between an ischial load-bearing KAFO and the patient's lower limb during normal walking cycle. Forces in the orthosis were measured at a level just above the ankle joint, where a short section of each sidemember had been replaced by a transducer that was capable of measuring forces in 3 orthogonal directions (Fig 3.8). Force platform and cinephotography were used simultaneously with the instrumented KAFO. The results obtained from a single patient





Location and wiring of strain gages on leg-brace transducers

(from Kirkpatrick et al, 1969)

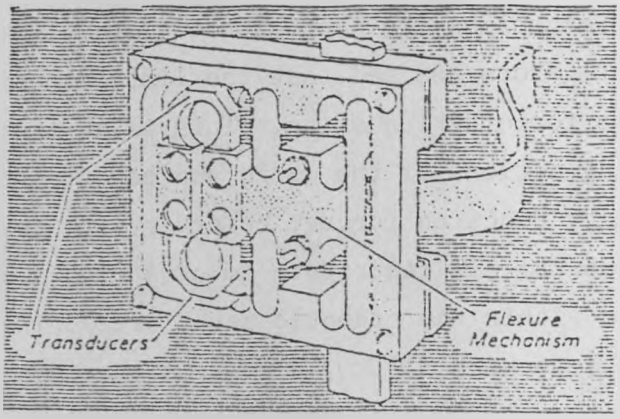
FIG 3·8	Transducer for KAFO Load Measurement.

showed that the axial load borne by the orthosis did not exceed 30% of the body weight. Distribution of loadings between the two sidemembers was not reported. Instead of using the orthosis in its unloaded state (i.e. before the patient put on the orthosis, and with the orthosis carrying its own weight) as the 'zero force' datum line for the transducers, the investigators assumed that the limb-orthosis complex was in a state of zero force during swing. This is an incorrect assumption since the strapping actions of the cuffs and bands, and the gravitational and deformity reaction forces, together with any residual muscle actions in the limb can all contribute to producing a considerable force distribution within the orthotic structure during swing.

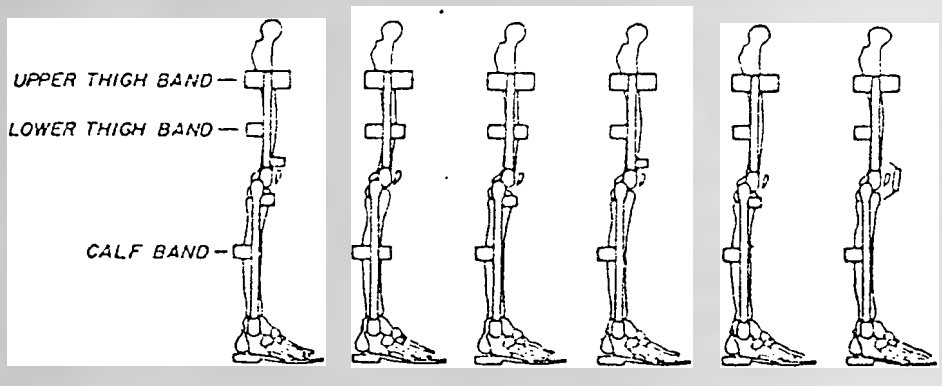
Lehmann and coworkers(1970c), also from the same centre, continued the investigation of ischial weight-bearing KAFO by considering the various designs and their effectiveness in relieving axial loading of the skeletal system. Using a similar set-up as that of Kirkpatrick, they reported that the axial loads experienced by the orthosis were at the maximum at foot-flat, and decreased toward toe-off. Their findings can be summarised below, with the orthosis designs listed in increasing effectiveness, and the amount of weight-bearing obtained expressed as percentage of body weight against each design description:

Loosely-fitted Thomas ring, fixed knee, free ankle 18%
Close-fitting Thomas ring, fixed knee, free ankle 42%
Quadrilateral shell, fixed knee, free ankle 61%
Quadrilateral shell, fixed knee, fixed ankle 76%
Quadrilateral shell, fixed knee, fixed ankle, rocker bottom..95%
Quadrilateral shell, fixed knee, fixed ankle, patten bottom.100%.

Hydraulic pressure transducers were also installed on the posterior rim of the quadrilateral shell to determine the amount of force transmission at the ischial-orthosis interface. It was found that a large proportion of the loadings were transmitted through the soft tissues, with less than half of the force taken by the posterior rim of the shell. Another important observation was that voluntary muscle action can vary the amount of weight-bearing by the orthosis, and with proper training, the patient can significantly increase the effectiveness of the device. Lehmann and Warren (1973) also discussed and compared the weight-bearing functions of ischial KAFOs and patellar-tendon bearing AFOs.



(a) Mechanism used to limit transmission of force to one plane. Transducers to measure tension and compression are incorporated within this flexure mechanism.



(b) Schematic representation of the six orthotic configurations studied.

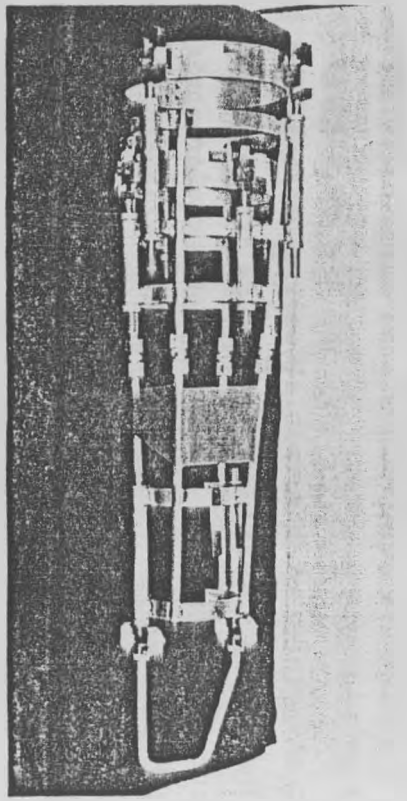
(from Lehmann and Warren , 1976)

FIG 3·9	(a) Strap Transducer Assembly. (b) Six KAFO Designs.

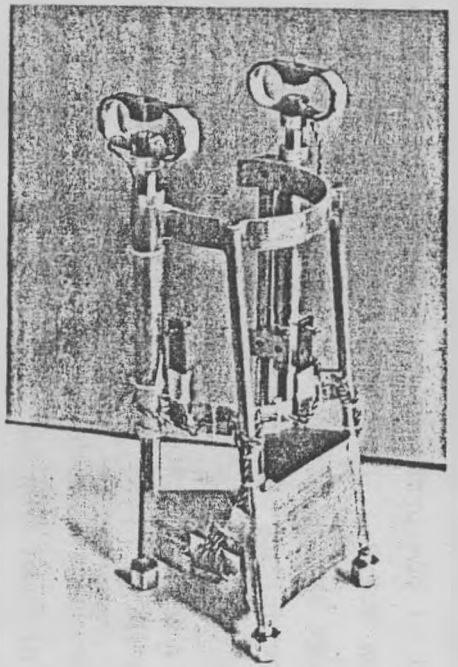
A biomechanical evaluation of 6 common designs of double upright KAFO (Fig 3.9b) worn by spinal cord injuried persons ambulating with a swing-through gait was carried out by Lehmann and Warren (1976). Transducers were placed just above the ankle joint and just beneath the thigh band. In addition, a transducer assembly (Fig 3.9a) designed to measure the anterioposteriorly directed forces was mounted at the strap attachment point on the upright. It was reported that the lowest total force required to stabilize the knee was produced when the restraining forces provided by the orthosis were applied closed to the knee axis, as in the knee-cap strap. However, the knee-cap strap was also found to produce the highest amplitude of shear. The suprapatellar strap was considered to be the most favourable when both shear measurement and stabilizing forces were taken into the account. It was also reported that the forces measured were substantially greater when the knee was flexed in the orthosis, a finding which confirmed the discussion in section 3.2.2 . Lehmann and Warren also suggested that the pressure produced by the straps and bands were likely to exceed capillary pressure during normal ambulation.

Another investigation based on the similar instrumentation as described above, but on the evaluation of Scott-Craig KAFO had also been reported by Lehmann et al (1976). From the above review, it was noted that the investigors at the Washington University had concentrated primarily on the axial load carried by the orthosis, instead of the more critical bending loadings experienced by the uprights.

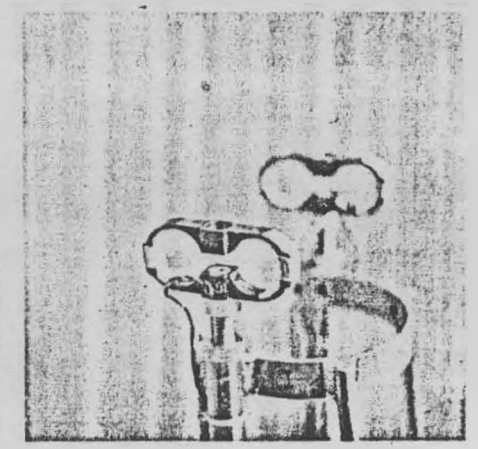
A comprehensive force measurement of a KAFO had also been carried out by Lippert (1971) at Stockholm, Sweden. Instead of instrumenting standard KAFOs, he designed a test orthosis to simulate 19 different prescriptions and at the same time measure the loading experienced by specific orthotic structures of the ischial weight-bearing KAFO combination chosen. The orthosis was constructed from duralumin circular ring sections securely attached to 4 longitudinal tubular uprights (Fig 3.10), and its mass was 2 - 3 kg heavier than standard designs. The forces generated at the limb-orthosis interfaces during ambulation were measured by 3 sets of transducers (Fig 3.10 and 3.11). Results from a single normal subject indicated that the orthosis was carrying approximately 50% of the body weight, with the average anteriopesterior (AP) and mediolateral (ML) forces acting on the orthosis at 50N and



Complete brace without wiring. Couplings between the the upper and lower segments are seen above the stiffner plates.



Lower segment upside down showing the ankle transducers, calf band beam transducers and couplings for attachment to the upper segment.



Ankle transducer without foot assembly.

(from Lippert, 1971)

FIG 3·10	Lippert's Experimental Orthosis for KAFO Load Measurements.

35N respectively. However, the magnitude of bending moment experienced by the orthosis was calculated only at two locations, namely, the ischial seat platform and the thigh band corset, with the loading at the ischial location the higher of the two. Averages of the maximum AP and ML moments recorded at the ischial seat were 10.3 Nm and 3.7 Nm respectively. When used in patten bottom combination, the AP moment at the ischial seat increased to 22 Nm. The normal valgus thrust of the knee together with the abducted gait associated with KAFO wearers were reported to be the main contributors of ML bending moments.

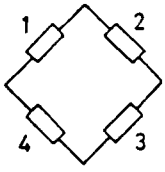
Lippert also suggested that the greatest bending moments were recorded when the axial loads were at the highest. He further commented that lighter KAFO designs using a single medial upright or lighter lateral upright might be possible because of the much lower lateral upright loadings. The axial forces on the medial upright in an abducted gait was 3.7 times that of an non-abducted gait. Contradictory to Lehmann's findings discussed earlier, Lippert reported that orthosis with fixed ankle assemblies were less weight-bearing than the free-ankle types.

Investigators at the Bioengineering Unit of the University of Strathclyde had also been carrying out detailed measurement and analysis of KAFO loadings in adult post-poliomyelitis patients (Trappitt, 1979; Berme & Trappitt,1981; Trappitt & Berme,1981). Tests were also being done on children who need KAFOs for ambulations (Szary,1985). Trappitt employed 4 six-channel load transducers, one above and one below the knee joint of each side-member of the conventional double-upright KAFOs (see Section 4.4.1 for further detailed design of the transducers). Szary used a similar experimental set-up as Trappitt, but with another set of shorter transducers to ensure that they can be adequately accommodated in the much shorter uprights of children's KAFOs (see Section 4.4.4).

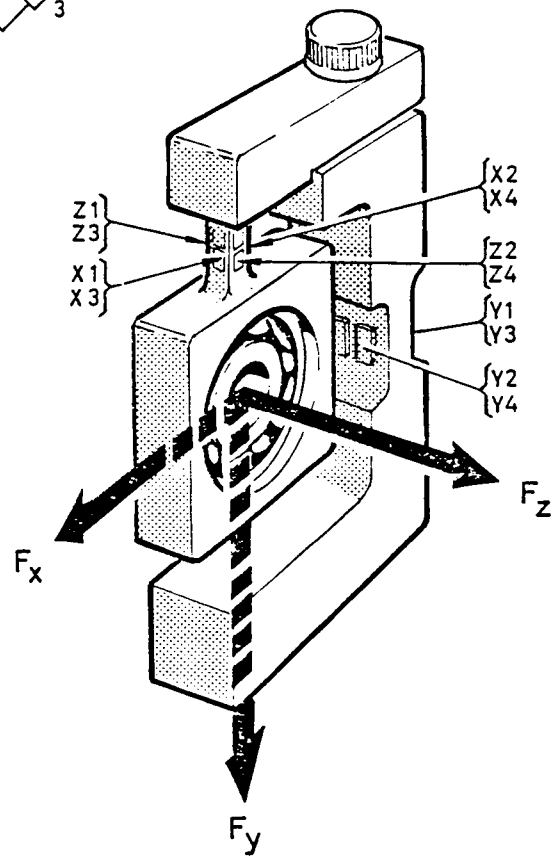
Trappitt reported that the medial proximal section of the orthoses experienced the highest loads, and that the medial upright carried a higher proportion of the axial load, which agrees with Anderson (1974,1977) and Lippert (1971). The maximum AP and ML forces recorded were 100N and 50N respectively. He also suggested that the more important type of loading exerted on the orthoses were those due to bending moments, which might be as high as 23 Nm in the AP direction and 13 Nm mediolaterally. He further indicated that the generally

STRAIN GAGES POSITION IN THE WHEATSTONE BRIDGE

X=Y=Z



STRAIN GAGES:
X = Z, DIRECTION
MICRO-MEASUREMENT
EA-06-062-AQ 350 (= 350 Ω)
Y, DIRECTION
MICRO-MEASUREMENT
EA-06-063-AQ 350



3 AXIS FORCE TRANSDUCER

(from Lippert, 1971)

FIG 3·11	Force Transducer for KAFO Load Measurements.

held opinion that conventional KAFOs were over-designed could be disputed.

Other centres who had attempted orthotic load measurements include the University of California at Los Angleles (Scott et al,1971), and the Chailey Heritage Hospital in Sussex (Reeve,1978). Scott and coworkers' measurement of orthotic loading was carried out as part of the biomechanical analysis for the design of the functional KAFO. Strain gauges were attached onto the orthosis to measure the axial load and the AP bending moments. Reeve (1978) reported on the work at Chailey Heritage where they instrumented a HKAFO for the investigation of orthotic loadings with crutch-assisted locomotion. However, details of their measurement transducers and the experimental methods were not available, nor their results reported.

Lew and coworkers (1983) quantitatively evaluated the relative efficiency of 4 orthotic knee joint designs by comparing their tendency to cause migration or pistoning action during ambulation. The longitudinal pistoning restraint forces in each side-member were measured by transducers. They reported that there was a wide variation of results, with no one orthotic joint showing superiority over the others.

CHAPTER 4.

DESIGN OF ORTHOTIC LOAD MEASUREMENT TRANSDUCERS.

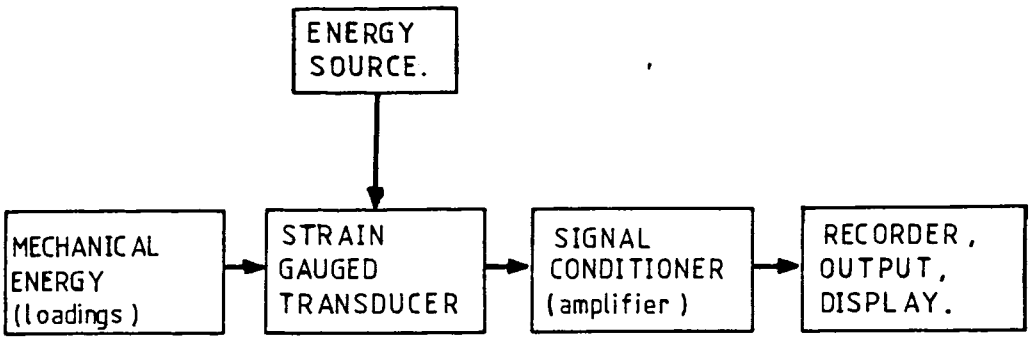


FIG 4·1 General Features of a Transducer Measurement System.

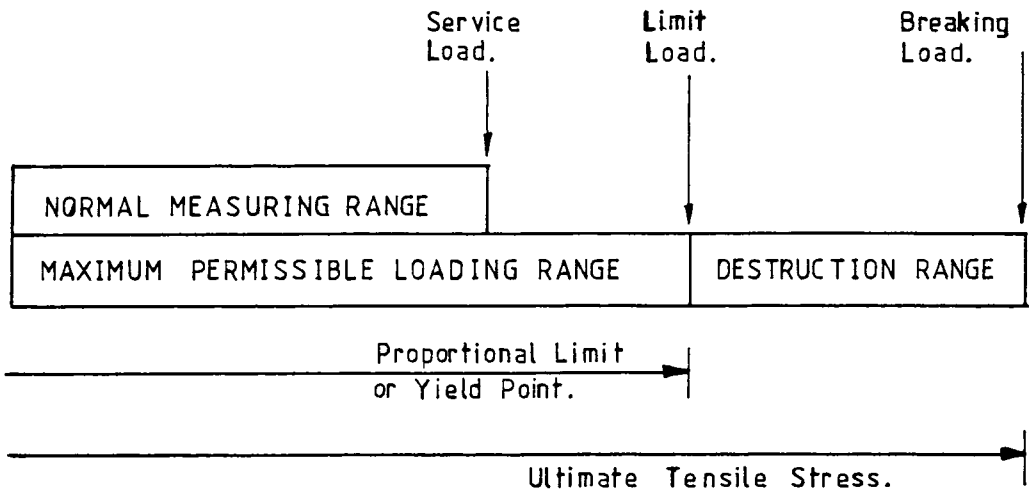


FIG 4·2 The Mechanical Strength of a Transducer.

4.1 INTRODUCTION.

A large proportion of all the orthotic designs to date has been empirically derived. To establish realistic design criteria, comprehensive biomechanical data of the orthotic structures under normal ambulatory conditions are needed. This requires the measurement of the loads actually being experienced by the structural members of the orthosis.

The transducers or sensing devices used in the measurement of orthotic loadings are normally of the electro-mechanical type, which translates a mechanical energy input in the form of forces and moments, to equivalent electrical signals (Fig 4.1). These devices can be located either at the points of application of the external loads, or at members of the structure which transmit the external loads to other parts of the system. In this respect, the transducers can be divided into 3 general classes, viz.:

1. Sensing elements permanently built into the structure, as used by the researchers at Haddassah University Hospital in Jerusalem (see Section 3.4.3);
2. transducers built into removable components of the orthosis, as adopted by Hardy (1981);
3. separate multi-channel force transducers that attach to and replace sections of the orthotic structure, as used by Kirkpatrick et al (1969), Lippert (1971) and Trappitt (1979).

The first method involves mounting strain gauges onto the structural members themselves. It is costly and time-consuming, especially in multi-channel measurements involving more than one patient. Since the orthoses have to be custom built for each individual patient, it will not be realistic to employ the second method of measurement, i.e. instrumenting orthotic components. For the third measurement to be successful, the load transducers should be small, light, and at the same time will not significantly alter the original loading patterns of the orthosis.

This chapter attempts to outline the design considerations and the calibration of multi-channel load transducers, and to describe the transducers designed and constructed for the monitoring of orthotic loadings.

4.2 DESIGN CONSIDERATIONS.

The design of orthotic load transducers is principally based on strain gauge measurement of load-induced surface strains. A good transducer should therefore have repeatable response, good linearity, and small 'cross-talk' or interference between the measuring bridge circuits. The device must also possess sufficient mechanical stiffness so that it will not interfere with the normal performance of the orthosis. In addition, a reasonably large load sensitivity of measuring channels will minimise the noise-to-signal ratio, thus improving the accuracy. However, complexity of the transducer increases rapidly with the number of load components required to measure accurately under the limitation of weight, size, and other geometrical constraints.

4.2.1 Mechanical Requirements.

The mechanical stability of a load transducer is closely related to the material, design, and manufacture of the transducer elastic element. High strength, low hysteresis, linear response, low creep, and predictable low thermal effects are some of the qualities required of a transducer material. The transducer designed should be free of mechanical defects such as backlash, friction and clamping effects, and should be sufficiently rugged and shock-resistant to withstand the experimental environment.

Within the transducer's service range of loadings (Fig 4.2), it is essential to have a linear stress-strain relationship in both tension and compression. An identical stress-strain path should also be obtained in both loading and unloading cycles, i.e. of minimum hysteresis. If the transducer were to be used in static or quasi-static measurements, then it should also possess good creep characteristics, i.e. the strain remains essentially constant with time under the same load.

The more conventionally cited structural properties such as proportional limits and yield strength do not directly affect the performance of a transducer. However, they are very important with respect to its overload capacity. The fatigue strength should also be taken into account when designing a transducer for dynamic load measurements.

The thermal or heat transfer characteristics of the transducer will also play an important role in the accuracy and repeatability of the measurement. Although at room temperatures, it may not be a great cause of concern, nevertheless, the thermal conductivity and expansion coefficient of the transducer material can seriously affect the stability of the output signal.

Machinability and availability of the material are the main manufacturing considerations of the design. The material chosen should be readily available commercially in a usable form and shape, and be able to accept intricate machining without any additional heat treatment.

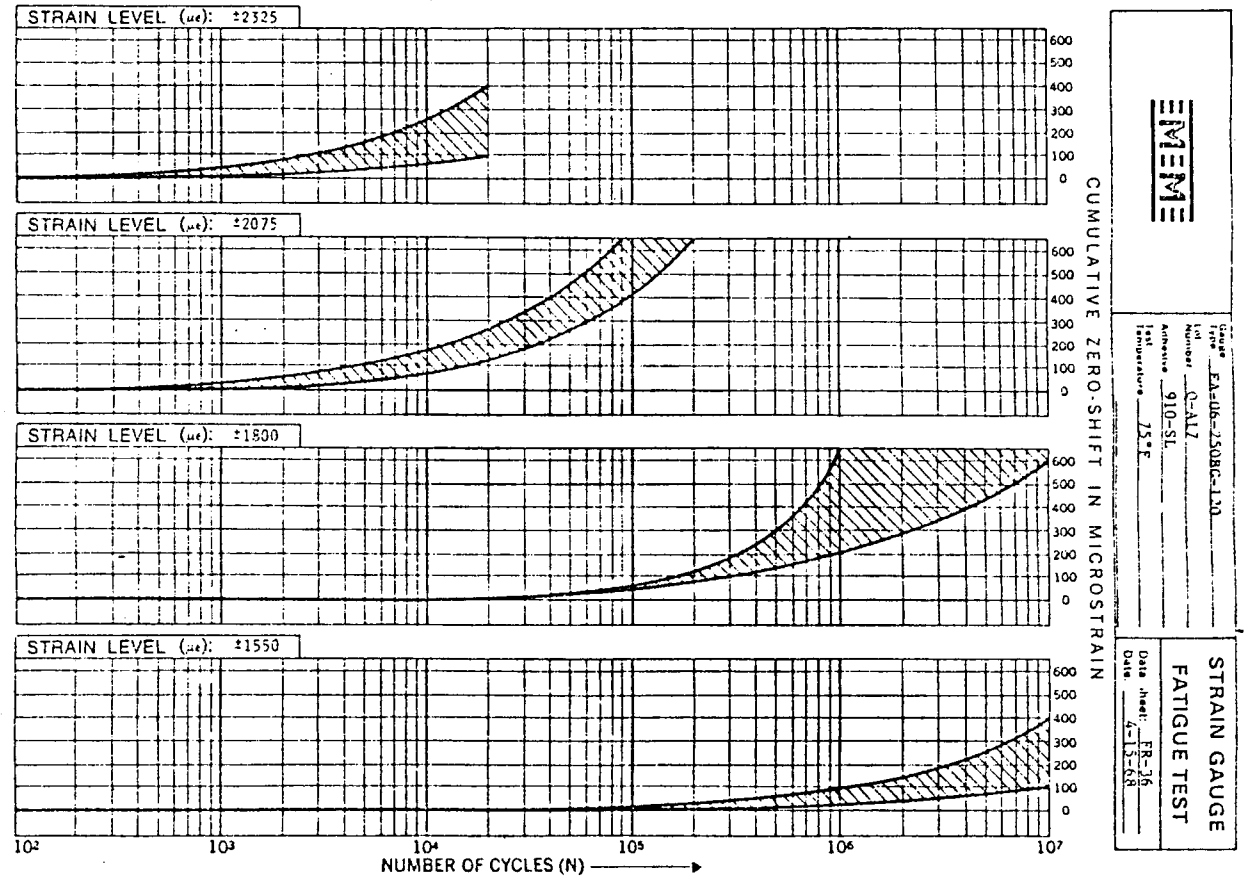
The widely used materials include heat-treatable low alloy steels with relatively high carbon content of up to 0.5% (Reeve,1978; Elliot, 1980), AISI 4140 tool steel (Measurement Group,1983), stainless steel (Measurement Group,1983), aluminium alloys (Cunningham & Brown,1952; Kirkpatrick et al,1969; Berme et al,1976), and the rather costly beryllium copper alloy.

4.2.2 Transducer Elastic Elements.

The elastic element (or spring element) is the most critical component of a strain gauged transducer. It serves to focus the effects of the applied load into an area where the strains may be measured. Ideally, the element should have a simple and symmetrical design with a highly efficient energy transfer characteristic. It should also possess a proper geometrical configuration to facilitate the prevention of cross-talk and to attain a high degree of mechanical and thermal stability. As far as possible, the element should be of an integral design with the rest of the transducer structure, i.e. machined from a solid piece of material to eliminate mechanical couplings that cause nonlinearity and hysteresis.

Sensitivity, strength, and dynamic response are the three parameters that demand careful consideration in the design of the spring element. However, it is complicated by the fact that a compromise between the factors is needed since not all of them are complementary to each other. An example of this is the contradictory requirements of sensitivity and strength.

Sensitivity is directly related to the strain level at the gauge



(from Micro-Measurements Tech. Bulletin TN-130-2)

FIG 4·3

Fatigue Characteristics of a Strain Gauge.

locations, and it should be as high as possible within the limits imposed by the transducer material and the strain gauge stability. This can be partially achieved by reducing the cross-sectional area in selected portions of the element. If the sensor is of the multi-channel integral construction type, then the same cross-section must also withstand other components of the applied load. Generally, a working range of 0-1500 micro-strain is used in the design (Window, 1970; Measurement Group,1982). A nominal output signal of 3 millivolts per bridge excitation voltage can be obtained from the 1500 $\mu\varepsilon$ strain level when 4 strain gauges of gauge-factor 2.0 are connected in a full Wheatstone bridge circuit (see Appendix 3).

Load sensitivity also has to be balanced with the fatigue characteristics of the strain gauges used. Fatigue damage in the form of "zero-shift", i.e. a permanent change in the unstrained resistance of the gauge may occur with cyclic loadings at sufficiently high magnitude. Micro-Measurement technical literature (TN-130) reported a nominal fatigue life (100 $\mu\varepsilon$ zero shift) for their constantan-polyimide gauges (EA series) at 10^8 cycles when subjected to $\pm$1200 $\mu\varepsilon$ loading. However, a shorter life of 10^6 cycles was indicated under a higher strain level of $\pm$1500 $\mu\varepsilon$. The useful life reduces drastically with increasing strain level, and at $\pm$2300 $\mu\varepsilon$, a mere 3000 cycles will push the zero-shift beyond the 100 $\mu\varepsilon$ mark (Fig 4.3).

Although sensitivity is an important design criterion, it is equally essential to ensure that the stresses remain well below the elastic limit of the material. Normally, the gauged area is designed to be the most highly strained region of the transducer. In the event of an accidental overload, the device should be able to accommodate at least twice the rated load without damage, and a 400-500% overload before destruction. A benefit of limiting the working stress to 30% of the yield stress will be the achievement of minimal hysteresis (Pople, 1980). For transducers made from wrought steel, it should additionally be ensured that the endurance stress level will not be exceeded (Spotts,1978). The effect of residual stresses due to manufacturing or other processes can be significant, but are usually neglected. However, in some cases, while load stresses are well below the elastic limit, the total stress may reach the yield point when the residual stresses are included.

The response time, or the time required to reach the equilibrium of a transducer subjected to dynamic loading is a function of damping and natural frequency. The natural frequency of the transducer should be as high as possible, and at least more than 4 times the maximum exciting frequency of the applied load (Loewen et al,1951). For normal human movement, the rate of change of forces is approximately 20 cycles per second, thus a natural frequency in the region of 200 Hz for the transducer is required. If the measurand approaches the natural frequency of the transducer, the resulting resonance will make interpretation of the data difficult, if not impossible. Structurally, a high natural frequency requires a rigid, light, and low compliant elastic element.

The size and configuration of the elastic element is also very much dependent on the magnitude and direction of the applied load, and also on the geometrical constraints. The most popular, and probably the most successful multi-channel elastic element has been the cylindrical columns (or pylons) as used by Cunningham and Brown (1952), Paul (1967), Harper et al (1967), Kirkpatrick et al (1969), Berme et al (1976), Boenick et al (1979), and Pratt (1981), to name but a few. Octagonal rings had also been employed by Schultz and Galante (1969), and Kirkpatrick and coworkers (1969). Trappitt (1979) used a combination of extended octagonal ring and cylindrical columns.

4.2.3 Strain Gauging.

The strain gauge is probably the most universal sensing device for electrical measurement of mechanical quantities, and it can be grouped into the wire, foil, or semi-conductor type according to its construction and operating principle. In recent years, wire gauges have been mostly replaced by the more advanced foil gauges. Typically, a foil gauge consists of a grid measuring element bonded to a polymer backing which is in turn cemented to the test specimen. It operates on the principle that deformations of the specimen are transmitted to the measuring grid element which changes its electrical resistance proportionally with strain. This relationship between the specific change in resistance and the induced strain of the measuring element is called the Gauge Factor, K, and can be represented mathematically as:

$$K = [\Delta R/R] \; / \; [\Delta L/L]$$

where R and L are the initial values of resistance and length of the gauge respectively, and the symbol Δ signifies an increment in R or L.

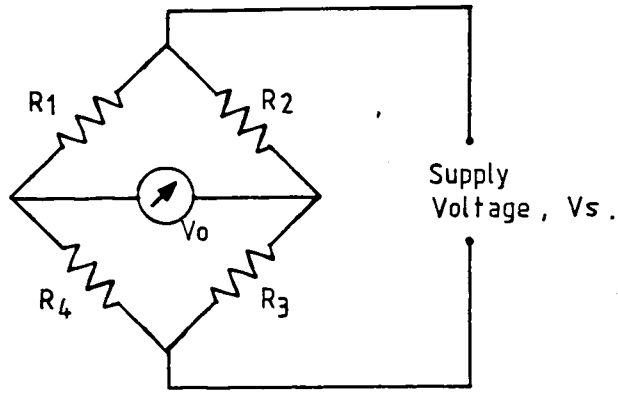
Semi-conductor gauges offer a much higher gauge factor of up to 180, as compared with 2.0 for most foil gauges, and therefore have an advantage of greater signal output per unit deformation. However, its temperature depedent K factor, shorter fatigue life, lower breaking strain, and inferior linearity all contributed to its lower popularity in general measurement usage.

The choice of gauge and adhesive (or cement) determines the quality of the results, and should be considered with the aim to obtaining accurate and reliable measurement, optimizing the gauge performance, and ease of installation. In general, gauges are selected according to their strain-sensing alloy, the backing materials, the grid dimensions and configurations, the bonding material, and the gauge protection required.

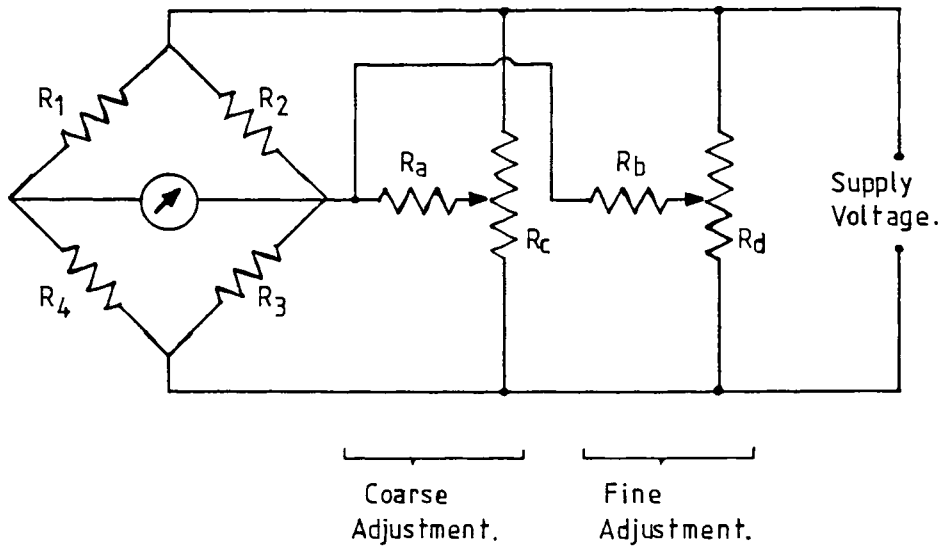
Special alloys are used in the strain-sensing grid for modern gauges. Constantan (45-55 nickel-copper) is most suitable for transducer application over moderate temperature ranges because of its low apparent strain due to temperature effects, good linearity, high ductility, and long fatigue life. It is available in self-temperature-compensated (STC) forms to match a range of expansion coefficients. The backing material is usually polyimide, a tough and flexible polymer. Selection of gauge length and grid pattern should be considered on the factors of strain gradient present, accuracy required, heat dissipation, space for installation, and the shape and orientation of grid needed for the measurement.

Ideally, the strain gauge should form an integral part of the transducer elastic element, and this can be achieved through careful gauge installation. A solvent thinned epoxy, or an epoxy-phenolic adhesive may be used in the gauge mounting. The procedure for gauge installation may be summarized as followed:

1. degreasing and removal of soluble contamination;
2. surface abrading to remove oxides and other adherents, and to develop a suitable surface finish;
3. application of gauge location layout lines;
4. surface conditioning to remove any contaminants;



(a) The Bridge Circuit.



(b) Circuit for Initial Balance of the Bridge.

FIG 4·4	The Wheatstone Bridge.

5. neutralizing the surface condition;
6. application of strain gauges; and
7. application of waterproof protective coatings.

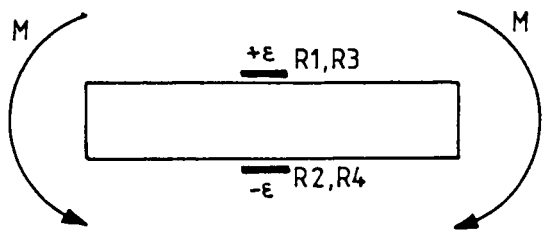
4.2.4 The Measuring Circuits.

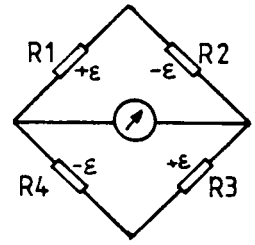
Since the output of a strain gauge is typically less than 1.0 mV per bridge voltage, a simple but accurate way of measuring these small signals is needed. This can be accomplished by the use of the Wheatstone bridge, with strain gauges connected across one or more arms of the circuit (Fig 4.4a). Two types of bridge systems are generally available, namely, the null-balance system and the deflection system. Although the null-balance method gives no direct reading and requires frequent rebalancing, nevertheless, it is widely used for static strain measurement because of its simple and reliable design. For dynamic measurement, the deflection system where the output is directly proportional to the strain level is normally used. However, it is less precise than the null-balance method, and needs a stabilised power supply to reduce the output fluctuations.

In transducer applications, bridges are normally made up of gauges from the same batch to minimise the errors of batch variations. Ideally, there would be a zero voltage at the output terminals when no load is being applied. But in most cases, a residual output exists at the unloaded state, which can be 'zero-adjusted' using the circuit as shown in Fig 4.4b.

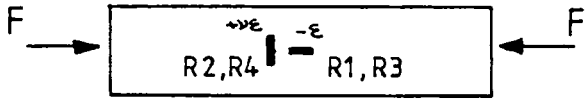
Another important factor of the measuring circuit is the bridge supply voltage, which should be as high as possible to obtain a maximum output per micro-strain, and to achieve a reasonable signal-to-noise ratio. The level to which the supply voltage can be raised depends on the heat dissipation properties of the gauge and the mounting surface. If the temperature rise is excessive, hysteresis, creep, thermocouple effects, and instability may occur.

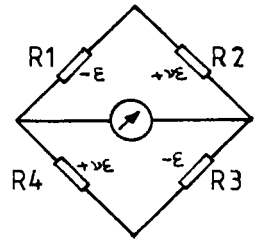
The mounting positions of the strain gauges on the transducer and their relative locations in the Wheatstone bridge circuit are the main deciding factors of load sensitivity and cross-effects between measuring channels of the device. For a beam subjected to bending moments, the gauges mounted as shown in Fig 4.5a, and connected into a full bridge will give an output 4 times as large as the corresponding output



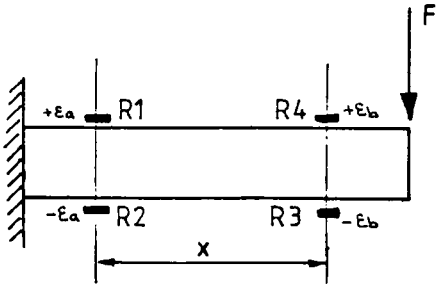


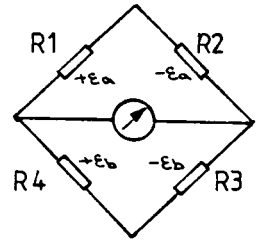
(a) Bending load.





(b) Axial load.





(c) Transverse shear.

$$F = \frac{Ma - Mb}{x}$$

FIG 4·5	The Measuring Circuits.

of a single gauge (Appendix 3). It will, at the same time be independent of all other loadings on the transducer.

Forces acting along the long axis of the elastic element can be measured by utilizing 2 longitudinally and centrally placed gauges to sense the direct strain of the forces, and 2 transverse gauges for the Poisson strain (Fig 4.5b). The output voltage will be $2(1+\nu)$ times that of a single longitudinal gauge, where ν is the value of the Poisson's ratio. To further reduce the errors caused by any eccentrical loadings, a set of four 90° rosettes equi-spaced around the periphery may be used, as by Berme and coworkers (1976). The Wheatstone bridge circuit will also have to be expanded to contain a series of two gauges in each arm.

The measurement of transverse shear forces can be realised by using Wheatstone bridge to obtain the difference of the bending strains from 4 strain gauges placed at a distance 'x' apart (Fig 4.5c). The output is independent of the line of application of the transverse force, but is directly proportional to the distance x between the 2 gauge levels. Hence the distance x should be as large as permissible by the size of the transducer.

Transverse shear forces can also be accurately estimated through the measurement of the principal strains produced by the shearing actions. These principal axes are at ±45° to the longitudinal axis of the transducer elastic element. Fig 4.6 shows the position of the gauges and the bridge circuit used. The gauges are aligned at 45° to the longitudinal transducer axis, with the grid line directions on the back face of the spring element perpendicular to those on the front face to cancel the effects due to torque and bending.

Torque applied about the long axis of the transducer can also be measured with 90° rosettes aligned at 45° to the long axis (Fig 4.7). However, in constrast to the shear gauges, torque gauges on both the front and back surfaces of the spring element are placed along the same 45° helix with each other. The torque bridge thus sums the torque strain, and substract those due to shear.

4.2.5 Sources of Error.

The measurement errors associated with the characteristics of a multi-channel load transducer may be caused by one or more of the

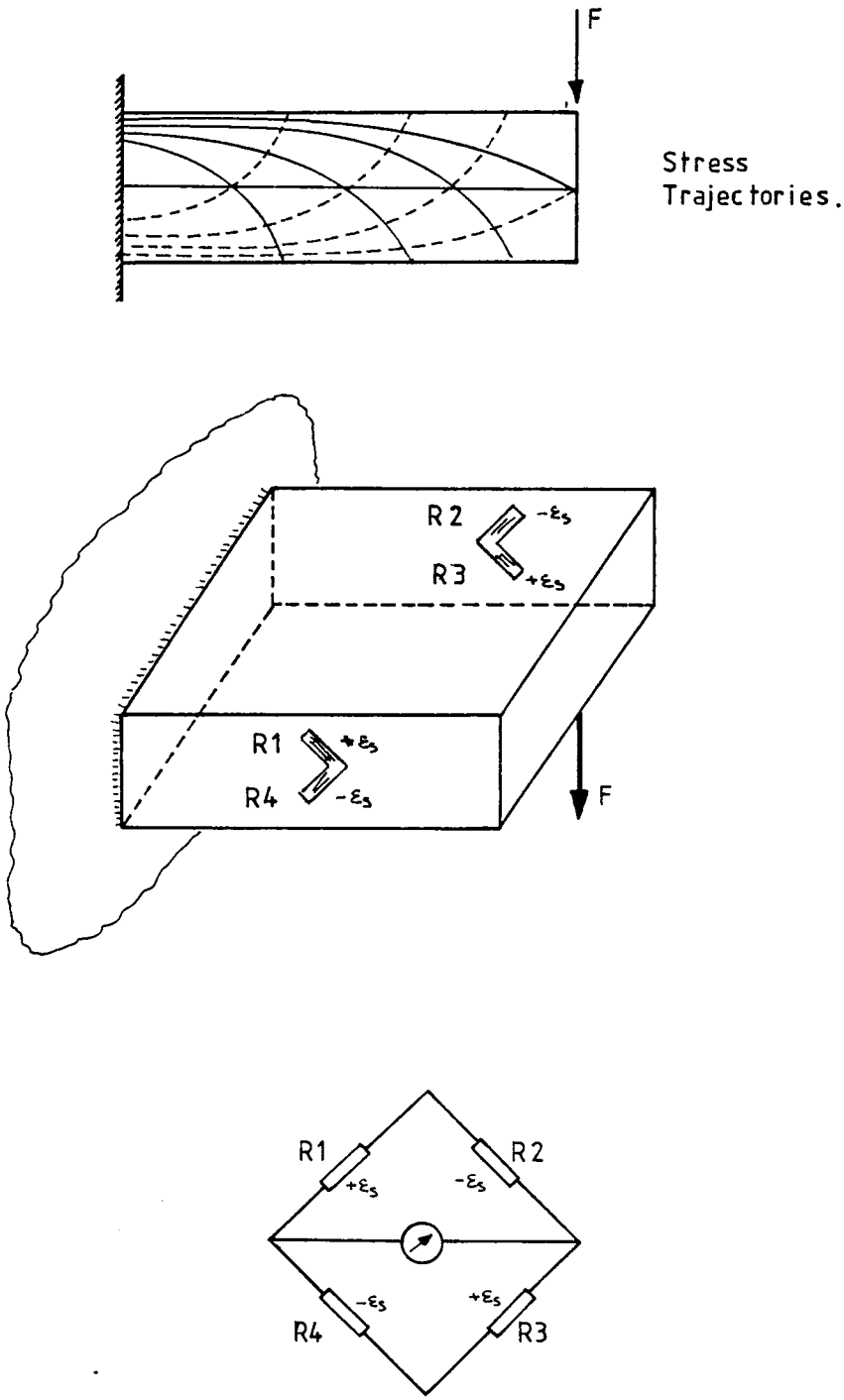


FIG 4·6	Measurement of Transverse Shear by Principal Strains.

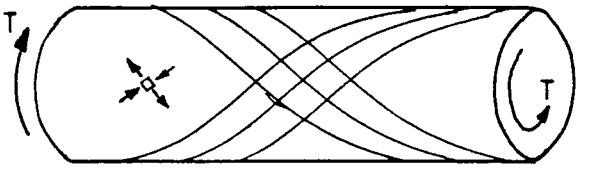
following factors, namely, electrical errors, temperature induced errors, improper transducer geometry, and faulty gauge mounting or selection. Sources of error concerned with other components of the KAFO biomechanical analysis will, however, be dealt with in the appropriate sections later in the thesis.

Electrically induced errors are caused by Wheatstone bridge nonlinearity and the presence of electrical or magnetic fields around the strain measurement system. When a deflection type of Wheatstone bridge is used, the output is nonlinear , but the nonlinearity is small for small [$\Delta R/R$]. However, it is usually ignored in general transducer applications. The electrical noise generated by the main power lines, transformers, and other heavy electrical equipments can be a serious source of errors when long supply and signal leads are used. To minimise these errors, leads should be of the same length and type, and with a properly grounded conductive shield over it. Twisting of the signal leads together can also reduce the magnetic noise.

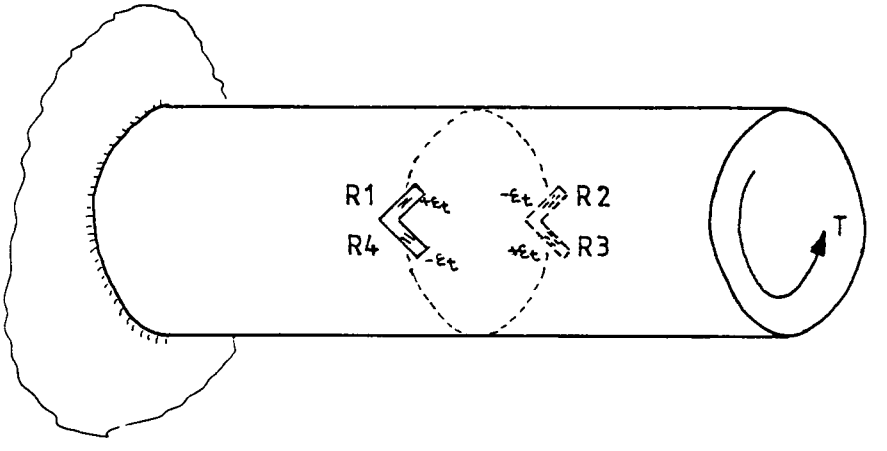
Although temperature-induced apparent strain is the most serious source of error at elevated temperature, it is generally small and can be compensated successfully at room temperatures. This temperature sensitivity is mainly caused by the differential thermal expansion between the strain gauge and the transducer. It can be compensated by the use of a full Wheatstone bridge with STC gauges.

In the design of a multi-channel transducer, errors due to the 'cross-talk' effect are inevitable and care should be taken to keep it to the minimum. This is due to the inter-sensitivity of the measuring channels, and is manifested by an unloaded channel showing a reading when another channel is being loaded. Libertiny (1975) listed the major causes of cross-talk as being due to improper transducer geometry, improperly located gauges, improper calibration, and boundary conditions that changes with the applied loads. The combined problem of complexity in transducer geometry design and difficulties in precise gauge alignment made it almost impossible to eliminate the error completely. Anderson (1975) suggested a way to eliminate the cross-talk by reducing the undesired load sensitivies with resistors parallelly connected to the gauges. However, it also desensitizes all the other channels.

Other possible sources of error are creep, hysteresis, non-repeatability, non-linearity, zero shift, and poor resolution. Creep



Stress
Trajectories.



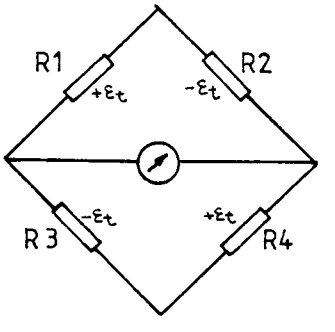


FIG 4·7	Measurement of Torque.

is the reduction of strain in the gauge as compared with that on the substrate, which is due to yielding of the cement and polymer backing under continuous load. The energy absorbed by the transducer during loading may not be fully recovered when unloaded, resulting in hysteresis and zero-shift (Fig 4.8a).

A factor that is closely related to creep and hysteresis is repeatability, which is the ability of the device to reproduce its reading accurately under a repeated load. Linearity of transducer response ensures that the sensitivity with respect to an applied load is independent of the input amplitude of the measurand (Fig 4.8b). Another often overlooked source of error is the resolution of the output, which is the smallest increment of the measurand that can be detected with certainly by the transducer.

4.3 TRANSDUCER CALIBRATION.

Calibration is the process of comparing an unknown with a standard, and determining the value of the unknown from the accepted value of the standard. In transducer development, calibration essentially consists of applying known loads along a specific axis of the transducer and recording the output signal thus produced. It is normally carried out on a 'calibration bench' in a similar environment to that in which the transducer will be used. Calibration loads may be produced either by known masses under gravity (dead-weight calibration), or by a force-generating machine of close accuracy, such as an Instron Mechanical Testing Machine (comparison calibration).

For many transducer applications, calibration is the only method to predict its operating characteristics because of the complexity in the design. Each transducer is calibrated individually, and the sensitivity coefficient of each channel, magnitude of the cross-talk, and linearity and repeatability of the response are noted for each device. During calibration, some error sources 'built-into' the transducer with its construction are also being eliminated since the effects will already have reflected in the output obtained.

The accuracy of a transducer is closely associated with that of the calibration device, where the inconsistency is a cumulative function of the errors in the calibration fixtures, measuring tools, instruments, and calibration techniques used. Generally, the transducer

should be calibrated against standards of 4 to 10 times the accuracy expected from the transducer (Levi,1969; Evans,1979). A recalibration will be required if an overload situation has occurred during its normal usage. However, a transducer which performs satisfactory during calibration may produce erroneous results on an installation where the measurand varies widely and rapidly, or where the signal transmission distance is much longer than in calibration.

In a 6-channel transducer, the output signal S_i (i=1,6) is a function of the six components of the applied load F_j (j=1,6). Using a second order model, the signal of the i-th channel, i.e. S_i, can be represented mathematically as (Levi,1971; Dubois,1981):

$$S_i = \sum_{j=1}^{6} a_{ij} F_j + \sum_{j=1}^{6} \sum_{k=j}^{6} b_{ijk} F_j F_k$$

where the coefficients of sensitivity a_{ij} and b_{ijk} corresponds to the linear and quadratic response of the i-th channel respectively. The main coefficient of the channel is a_{ii} (where i=j), with the a_{ij} terms representing the cross-sensitivities. Since the quadratic interactions b_{ijk} are due to elastic deformations that change the transducer geometry under load (Dubois,1981), any reasonably well designed transducer would have a first-order response, i.e.

$$S_i = \sum_{j=1}^{6} a_{ij} F_j \qquad (i=1,6)$$

This can be represented in a matrix form as:

$$[S] = [C][F]$$

where [S] = a column matrix containing the output signal voltages;

[F] = a column matrix containing the input loads;

[C] = the calibration matrix, which is a square matrix containing the elements c_{ij} obtained from the calibration data.

When using the transducer to measure unknown loads, the output signal voltages S_i are measured and the unknown loads F_j from:

$$[F] = [C]^{-1}[S]$$

where $[C]^{-1}$ is the inverse of the calibration matrix.

In general, the response of a 6-component transducer can be written matrically as:

$$\begin{bmatrix} SF_x \\ SF_y \\ SF_z \\ SM_x \\ SM_y \\ SM_z \end{bmatrix} = \begin{bmatrix} C_{11} & C_{12} & C_{13} & C_{14} & C_{15} & C_{16} \\ C_{21} & C_{22} & C_{23} & C_{24} & C_{25} & C_{26} \\ C_{31} & C_{32} & C_{33} & C_{34} & C_{35} & C_{36} \\ C_{41} & C_{42} & C_{43} & C_{44} & C_{45} & C_{46} \\ C_{51} & C_{52} & C_{53} & C_{54} & C_{55} & C_{56} \\ C_{61} & C_{62} & C_{63} & C_{64} & C_{65} & C_{66} \end{bmatrix} \begin{bmatrix} F_x \\ F_y \\ F_z \\ M_x \\ M_y \\ M_z \end{bmatrix}$$

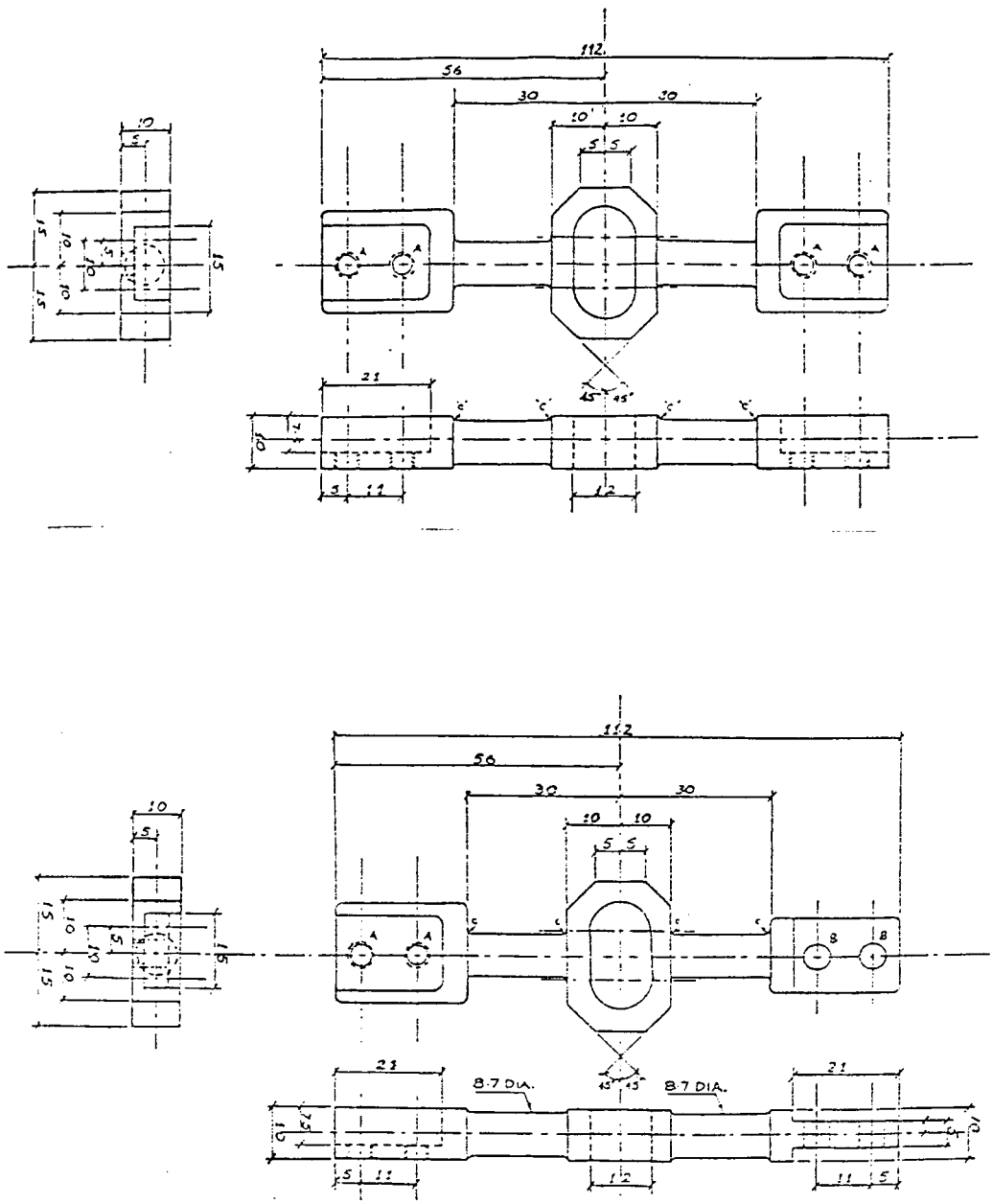
The signal SF_x, for example, can be expressed in the more familiar mathematical equation as:

$$SF_x = C_{11}F_x + C_{12}F_y + C_{13}F_z + C_{14}M_x + C_{15}M_y + C_{16}M_z$$

The horizontal row of the calibration matrix [C] is related to the applied load, while the vertical column of the matrix contains the signals obtained from each applied load component. Thus, if a pure axial load F_y is applied to the transducer, the element C_{22} of the matrix [C] represents the principle sensitivity coefficient of the load, with its magnitude (in units of uV/N/supply voltage) obtained from the gradient of the graph relating the bridge output signal to the applied load. All the other elements in the second column, i.e. C_{12}, C_{32}, C_{42}, C_{52}, and C_{62} are the cross-talk components corresponding to the F_y load.

4.4 STRATHCLYDE KAFO LOAD TRANSDUCERS.

This section describes the application of the design considerations in the construction of specialised load transducers for the measurement of orthotic loadings. The performance of the original set of transducers was assessed and the significance of its findings were discussed. These devices were found to be rather inadequate for proper KAFO tests, and a new set of stronger and shorter transducers were designed. In order to differentiate between the two sets of transducer, the first set was designated as the Mark I transducers while the new ones as the Mark II version. The design and calibration of Mark IIs will be discussed in details. Also included in this section is the design of a set of simpler transducers for the measurement of knee strap forces.



MATERIAL: 10 mm PLATE STEEL: Sy > 400 MPa.

HOLES: 'A' DRILLED AND TAPPED 5mm THREAD.
'B' 5mm CLEARANCE.

RADII: 'C' 1mm
: ALL OTHERS 1·5 mm.

(Adapted from Trappitt, 1979)

FIG 4·9

Mark 1 K.A.F.O. Transducers, with Two Different End Mounting Designs.

4.4.1 Mark I KAFO Transducer.

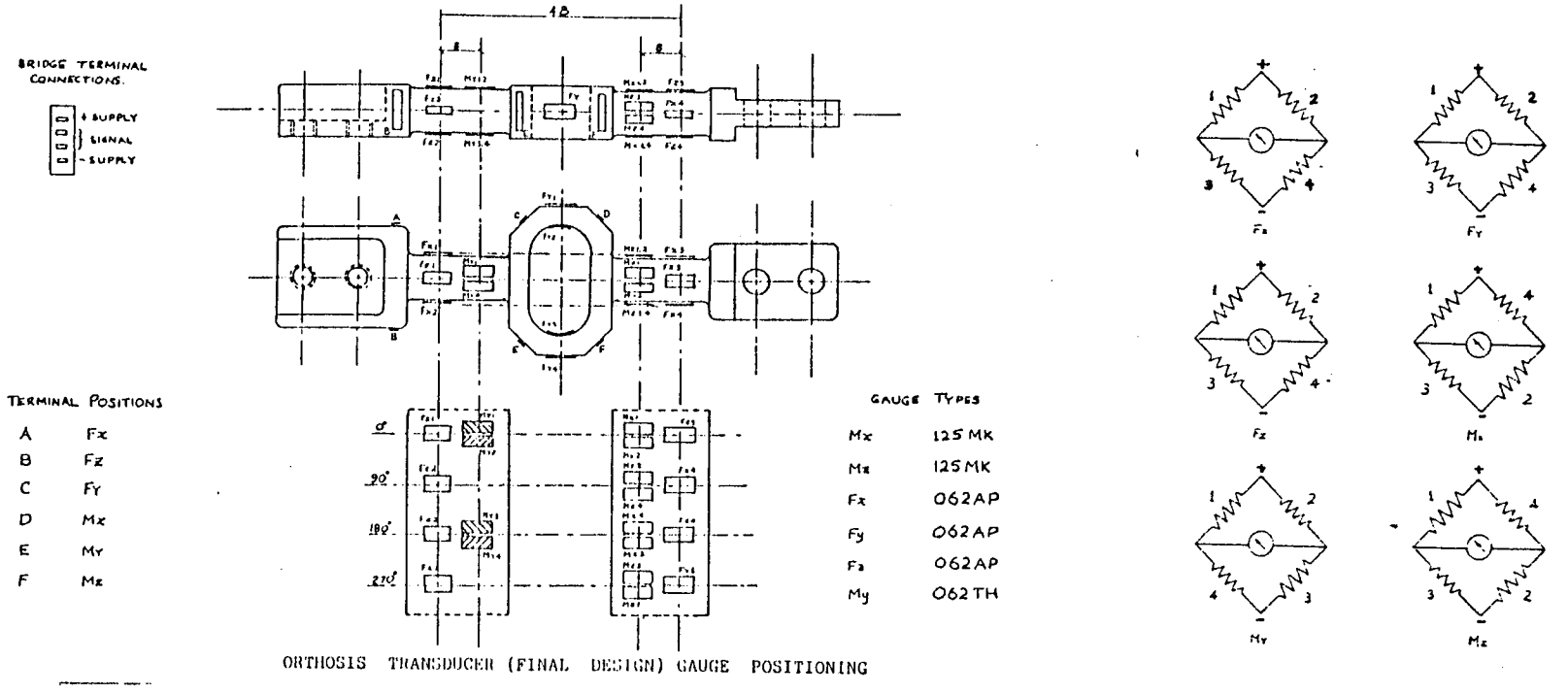
4.4.1.1 The transducer design: The Mark I transducer was designed by Trappitt (1979), and subsequently reported in Berme & Trappitt (1981), and Trappitt & Berme (1981). The transducer replaces a small section of the upright in the orthosis, and by measuring all 3 force and 3 moment components, the loads experienced by the orthosis at that level can be fully known. The transducers were machined from a 10 mm low alloy high carbon 'Stub Steel' plate, and the material has an empirically determined yield strength of "more than 400 MPa" (Trappitt,1979). The reported design loads were 800N for the axial force, and 15Nm for moments.

The axial force sensing element is an extended ring with an octagonal outer surface (Fig 4.9). Shearing forces in the AP and ML directions are obtained from measurement of bending moments at two levels, and electrically substracting the signals of one level from the other through the Wheatstone bridge to yield a signal proportional to the shearing force (see Section 4.2.4). Details of the gauge positions and bridge configurations are shown in Fig 4.10.

The transducer has an overall length of 112 mm, and is 30 mm wide at its broadest point at the ring section. All the 4 transducers produced are identical, except for the end mountings, where 2 of the transducers have one end of the device designed to couple directly onto a knee or ankle joint of the orthosis while the other end attaches to a 16 mm upright (Fig 4.9). The other 2 transducers have orthotic upright coupling devices on both ends.

Calibration matrices of the transducers are given in Table 4.1 . The elements of the matrices are in units of micro-volts per newton per bridge voltage (for forces), and micro-volts per newton-metre per bridge voltage (for moment channels).

4.4.1.2 Assessment of the transducers: The loadings recorded by the transducers in walking tests involving post-poliomyelitis patients was reported by Trappitt (1979) to be much higher than expected, and in some cases, had even surpasses the design load ratings of the device (Table 4.2). The situation was further complicated by a lack of detailed design data of the permissible overload capacity of the transducer. To ensure that the safety of the patient and the accuracy of



(Adapted from Trappitt, 1979)

FIG 4·10

Strain Gauge Positioning and Bridge Configurations of Mark 1 Transducer.

Table 4.1 Calibration Matrices of Mark I KAFO Transducers.
(from Trappitt, 1979)

Transducer 1:

	F_x	F_y	F_z			
F_x	3.726	0	0	4.833	0.246	0.132
F_y	-0.019	-0.809	-0.064	2.720	0.407	0.671
F_z	-0.069	0	3.777	0	-2.370	2.960
M_x	0	0	0	-152.9	0	1.170
M_y	-0.092	0	0.209	-4.770	91.8	1.470
M_z	0	0	0	1.810	-3.350	150.6

Transducer 2:

-3.601	0	0.067	-6.900	2.000	-0.622
-0.055	-0.831	0	0	0	-0.853
0.149	0	-3.666	0	0	154.0
0	0	0	-158.1	0	-3.550
0	0	-0.439	4.610	-91.7	0
0	0	0	4.630	0	-154.9

Transducer 3:

-3.722	0	0.084	3.720	-0.520	1.050
0	0.879	0	1.400	1.070	-0.097
0.137	0	-3.162	-1.387	0	-3.270
0	0	0	-155.4	0	-0.924
0	0	-0.132	3.510	-91.4	0
0	0	0	4.52	1.69	-154.8

Transducer 4:

-3.738	0	-0.020	-3.890	-1.050	0
0	-0.825	0.076	-5.490	-0.800	0
0	0	3.767	0	0	3.870
0	0	0	154.3	0	5.566
0	0	-0.291	4.960	-90.1	0
0	0	0	4.160	0	-153.7

** In units of $\mu V/N/V$ and $\mu V/Nm/V$.

Table 4.2 Design Loads of KAFO Transducers.

Channel	Mark I KAFO Transducers.		Mark II KAFO Transducers.	
	Maximum Load Recorded in Patient Tests	Design Load.	Maximum Service Load Allowed.	Minimum Overload Capacity.
Fx	100N	100N	200N	400N
Fy	245N	800N	800N	1600N
Fz	46N	40N	200N	400N
Mx	13Nm	15Nm	25Nm	50Nm
My	8Nm	5Nm	15Nm	30Nm
Mz	25Nm	15Nm	35Nm	70Nm

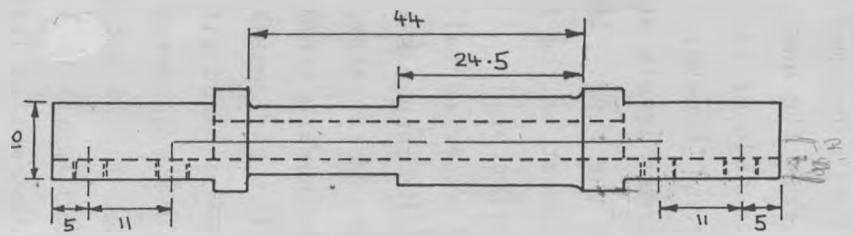
332 31

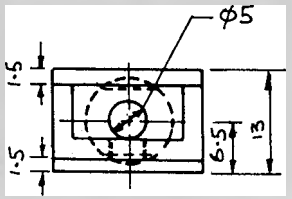
the transducers can be protected at all times, it became clear that an assessment of the transducer would be required to establish the maximum allowable loadings that can be imposed onto the device. This will indicate the types of patient and activity that can be carried out with confidence.

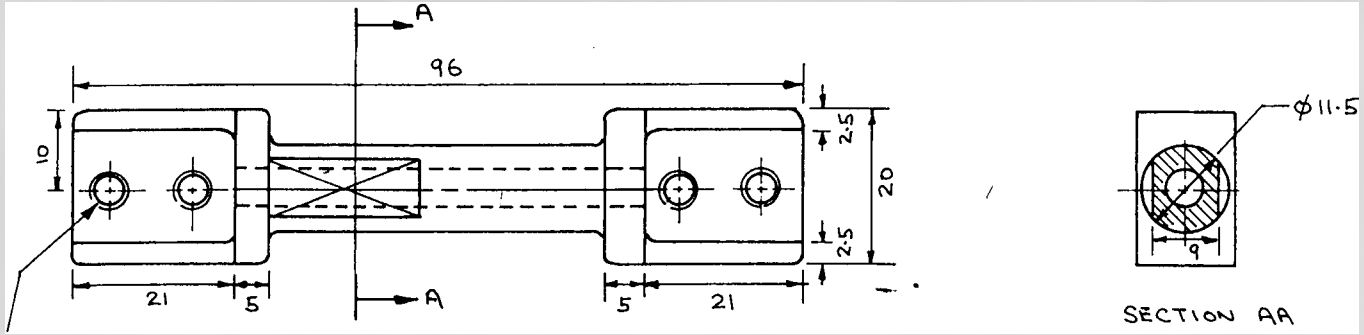
In Trappitt's design analysis, he used a photoelastic model and a circular ring theorem to estimate the strength of the octagonal extended ring. Two inconsistencies were discovered. Firstly, the use of an ordinary circular ring theorem would over-estimate the strength of the extended ring. Secondly, the photoelastic model employed was geometrically dissimilar in shape to the strain-gauged section of the transducer. In the model, the strain-gauged areas were rectangular cross-sectionally while those in the transducer were circular. This would produce an invalid comparison between the model and the prototype since the sectional moduli had not been taken into considerations.

The assessment consisted of a theoretical and an experimental analysis to establish the maximum allowable load. From the analysis shown in Appendix 4, it can be concluded that the transducer is capable of withstanding at least 3000N in the axial direction. However, under a bending condition, it was found that the stress levels at the beam sections are some 3 times higher than those at the ring. It was also found that the critical magnitude of bending moment on the transducer is 29 Nm, boyond which irreversible damage may occur.

Each of the four transducers were also tested experimentally in a loading jig to assess their accuracies. Only the Mz channels were recalibrated, since it was this channel that had been most severely loaded during patient tests. The rest of the channels were considered to be satisfactory. Results of the test revealed that the sensitivities of the Mz channels have changed by some 2.3% as compared with those reported by Trappitt. A hysteresis that may cause an error of 0.3 to 0.5 Nm in the output reading was also recorded in all but one of the transducers. With the exception of Transducer 3, all the devices exhibit good linearity. Nonlinearity was observed in Transducer 3 as loading reached 17 Nm, which may imply that the device had been loaded beyond the limiting load of 29 Nm. Possible causes of the transducer damage include mishandling of the device, excessive mounting stresses induced by misalignment of the orthotic upright and transducer, accidental overloading during patient tests, or a combination of the above factors.







4 HOLES
DRILL AND
TAP FOR M5 X φ·8

FIG 4·11	The Mark II K.A.F.O. Transducer.

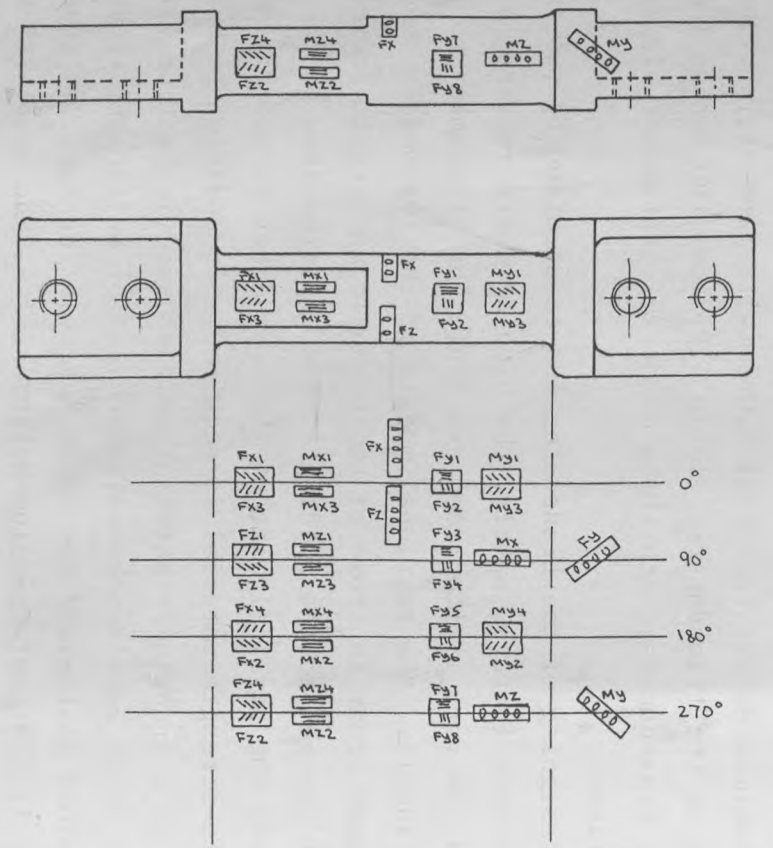
4.4.2 Mark II KAFO Transducers.

As discussed in the previous section, the KAFO transducers designed by Trappitt (1979) have been shown to be working at or very near to its loading capacity. For this reason, the choice of patients and the activities to be undertaken were severely restricted to those that will not exert excessive loading on the device. Furthermore, due to the length of the transducer (112 mm), a few otherwise potentially suitable patients had to be rejected because of the lack of room for mounting the transducers. Therefore, it was decided to design a set of transducers that are not only shorter in length, but will also be able to withstand any normal ambulatory activities undertaken by a KAFO wearer.

4.4.2.1 Transducer design: The design loads of the transducer were drawn up with reference to the data obtained from Mark I KAFO transducers. In normal usage, the loadings on the transducer should always be within the designed maximum load ratings (Fig 4.2). There should also be a minimum safety factor of 2.0 to withstand any accidental overloading of the transducers. Table 4.2 shows the design loads of the Mark II transducers, with those of Mark I transducer also tabulated as a comparison.

The designing of the transducer was constrained by two rather noncomplementary factors, namely, the high strength required of the device, and the very limited room available for attachment to the orthosis. Ideally, the sectional moduli of the transducer should also be in the same order as those of the orthotic upright to prevent any significant changes to the loading stress distributions of the orthosis.

To keep the physical size of the transducer to the minimum, a high tensile-strength material is required. In addition to having the properties of a transducer material as discussed in Section 4.2, it should also be of good shock resistance and of sufficient reserve-strength beyond post-yield for added safety. The material chosen was a low-alloy '1½% Nickel-Chromium-Molybdenum' 817M40 steel (BS 970), heat and tempered to the 'U' condition. It is more commonly known as the 'En24U', and supplied in heat-treated bright 1¼ inch diameter bar. The material has a yield strength of 755 MPa, and an ultimate tensile strength of 930 MPa (minimum). It also possesses good ductility and



GAUGE TYPES.

(MICRO - MEASUREMENT GAUGES).

			MAX
A/P SHEAR	Fx	EA-06-062TH-120	200N
AXIAL	Fy	EA-06-062TT-120	800N
M/L - BENDING	FZ	EA-06-062TH-120	300N
A/P - BENDING	MX	EA-06-125MK-120	25Nm
AXIAL TORSION	My	EA-06-062TH-120	15Nm
Mediolateral M/L Bending	MZ	EA-06-125MK-120	.35Nm

FIG 4·12	Strain Gauge Positioning of Mark II K.A.F.O. Transducer.

toughness (12% elongation, and 35 ft lbf Izod). In its heat treated form, the machinability is approximately 40% of that for mild steel (Woolman & Mottram,1964).

The geometrical configuration chosen for the transducer elastic element was a hollow cylindrical design, with the external circular surface along approximately half the length of the transducer machined off to form a section with opposite sides flattened in a direction perpendicular to the Z-axis (Fig 4.11). This cross-section provides the high strength required for the Mz moments, but at the same time increases the sensitivity of the much lower Mx moments. The remaining cylindrical sections are used for the measurement of axial load, and also for torque about the long axis. A brief theoretical analysis of the design is contained in Appendix 5.

Three types of constantan strain gauge rosettes with polyimide backings (supplied by Welwyn Strain Measurement Ltd, Basingstoke) were used in the construction of the transducers. In the transverse shear force channels(Fx and Fz), and also in the torsional moment channel (My), the principal strains were measured by rosette gauges containing a pair of sensing elements aligned at 45° to the main axis of the gauge. Although this method of measurement for shear force is less sensitive than the monitoring of bending strains method (as discussed in Section 4.2.4 and Appendix 3), it was used because a shorter transducer length can be achieved. Rosettes with sensing elements at right angles to the main gauge axis were connected with 2 rosettes in each arm of a full bridge to monitor the direct compressive and 'Poisson' strains produced by the axial load Fy. Lastly, moments in the mediolateral and anterioposterior directions (Mx and Mz) were obtained from the measurement of surface strains at a plane normal to the direction of the applied moment. Rosettes with a pair of parallel sensing elements were used in these channels. Details of the gauge positioning and the bridge configurations are given in Fig 4.12 & 4.13 .

As a result of the severe limitation of surface area on the transducer, the backings of the gauges were trimmed to a perimeter 1.5 to 2 mm away from the sensing element. The effects of trimming on strain gauges accuracy have been reported by Chalmers (1966), Kerr (1981), and Window (1982). They commented that errors will arise only with the trimming of very small gauges (i.e. 0.032 inch gauge length or less). It therefore can be concluded that this trimming of gauges

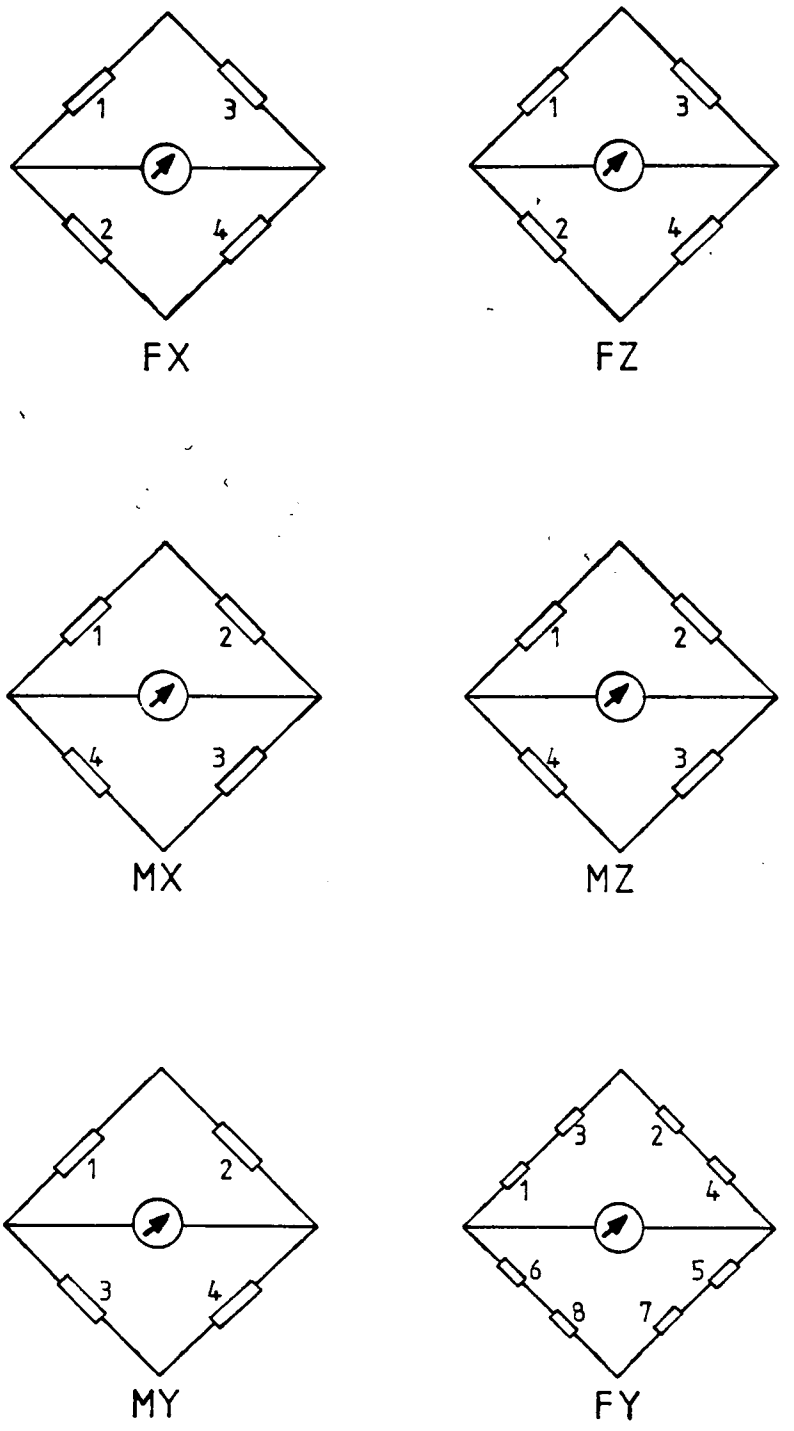


FIG 4·13	Bridge Configurations of Mark II K.A.F.O. Transducer.

will not affect the accuracy and the repeatability of the results.

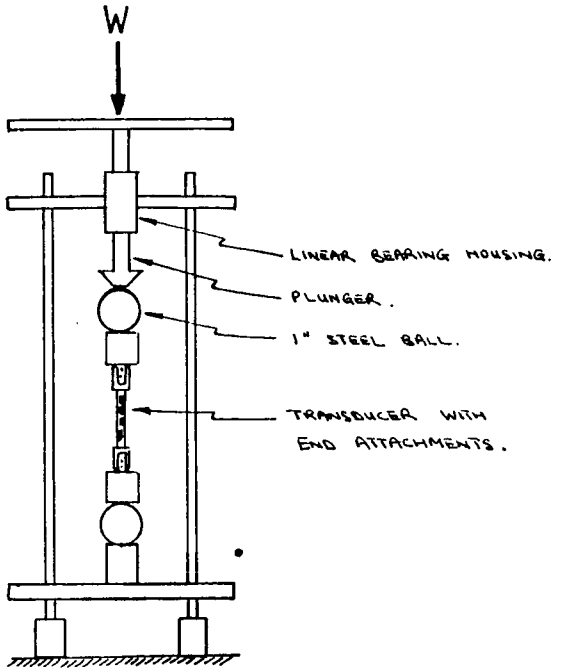
4.4.2.2 Calibration of transducers: Before the calibration of a transducer, the magnitudes of its bridge supply voltages have to be carefully established, since they not only affect the magnitudes of the output signal, but also have a considerable influence on the stability of the device. The optimal bridge supply voltages for the transducers were determined experimentally as 6.0 volts for the transverse force and axial force channels, and 3.0 volts for the moment channels.

Calibration was carried out using dead weights with accuracy better than 0.15%, and the required components of the load were applied through a system of calibration jigs and fixtures. The calibration loads were referred to a point called the 'transducer origin' (Levi, 1969), which is an imaginary point on the transducer where the line of action of the resultant force passes through. For the transducers, the origin is assumed to be situated at the centre of the cylinder, at the level of the Mx and Mz gauges. It is further assumed that the transducer measures the applied loads at its distal end.

The transducer was cycled a few times at a strain level approximately equal to that of the maximum service load prior to calibration. This aims to remove any irregularities in the output signal of a newly built transducer. Pure loads, i.e. loadings containing only the desired component, were applied one at a time, and the output from all the bridge circuits were recorded in order to isolate and quantify the cross-effects between the measuring channels. Loadings were repeated 3 times to obtain an average value for the sensitivity. Repeatability, hysteresis, and other behaviours of the transducer can also be monitored and rectified if necessary.

Axial loads were applied only in the compressive direction, since this is the type that will normally be experienced by the device. The transducers' response under tensile load was experimentally verified to behave with the same sensitivity as in compression. The axial load was applied through a linearly frictionless plunger in a rigid framework (Fig 4.14a). For shear force calibration, the line of action of the force was arranged to be at the level of the moment gauges (Fig 4.14b), in doing so, eliminating the induced bending moment at the level. However, this is not a 'pure shear' situation because a small amount of bending was induced at the shear gauges due to the different

(a) Fy :



(b) Fx , Fz : (rotate 90°

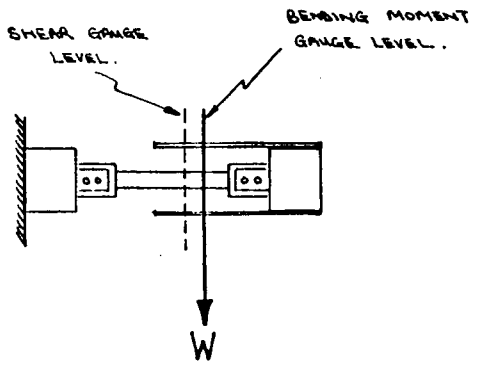
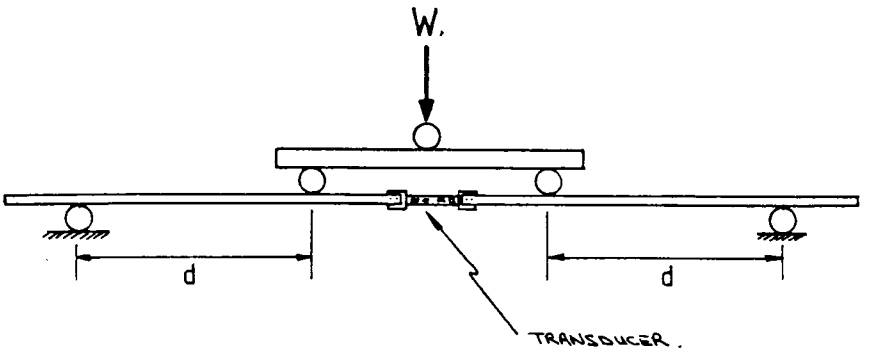


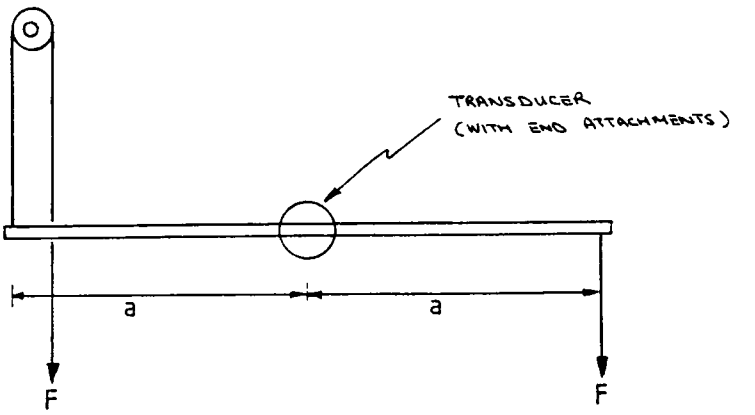
FIG 4·14	Calibration of the Transducer.

(a) Mx, Mz :



Bending Moment = $\frac{W}{2} * d$

(b) My :



Torsional Moment = $2F * a$

FIG 4·15	Calibration of the Transducer.

Table 4.3 Calibration Matrices of Mark II KAFO Transducers.

Transducer 1:

$$\begin{bmatrix} -30.8 & 0 & 0.9 & 80.6 & -441.1 & -350.0 \\ 0 & -6.9 & -1.4 & 79.6 & 14.3 & -24.8 \\ -1.1 & 0 & 20.7 & -249.5 & 129.0 & 179.1 \\ 0 & 0 & 1.9 & 8974.6 & 123.4 & 0 \\ 0 & 0 & 0 & 38.2 & -4250.2 & 46.2 \\ 0 & -2.0 & 0 & 0 & 0 & 7078.9 \end{bmatrix}$$

Transducer 2:

$$\begin{bmatrix} -34.4 & 0 & 0.8 & 180.0 & -308.0 & -328.1 \\ 0 & -7.2 & 0 & 0 & -39.2 & 0 \\ 0 & 0 & -23.6 & 212.8 & 149.5 & 43.1 \\ 0 & 0 & -12.6 & 9143.0 & 111.2 & 0 \\ 0 & 0 & 0 & 0 & -4410.7 & 0 \\ 10.3 & -1.5 & 0 & 83.9 & 185.7 & 7079.0 \end{bmatrix}$$

Transducer 3:

$$\begin{bmatrix} -33.9 & 0 & 0 & -101.4 & 0 & -305.0 \\ 1.9 & -7.0 & -1.3 & 62.5 & 19.7 & 105.0 \\ 1.3 & 0 & -22.4 & 297.2 & 148.4 & 181.3 \\ 0 & 0 & -7.3 & 8716.7 & 16.6 & -23.1 \\ 3.9 & 0 & -3.1 & 8.3 & -4324.6 & 0 \\ 5.2 & -1.0 & 0 & 184.7 & 0 & 6818.8 \end{bmatrix}$$

Transducer 4:

$$\begin{bmatrix} -31.4 & 0 & 1.9 & 107.6 & -880.6 & -548.3 \\ 0 & -7.1 & -1.8 & 117.8 & 142.8 & -22.6 \\ -0.7 & -0.7 & -21.6 & 347.9 & 199.3 & 216.9 \\ -1.0 & -2.3 & -5.7 & 8500.0 & 154.9 & -76.3 \\ -5.4 & 0 & -6.1 & -37.5 & -4335.4 & 32.5 \\ 4.6 & -5.9 & 0 & 45.1 & 41.0 & 6937.5 \end{bmatrix}$$

** In units of 10^{-2} μV/N/V or 10^{-2} μV/Nm/V.

bending and shear gauge levels. The effect of this can be considered to be negligible since the gauges concerned are only 8 mm apart. Bending moment calibration was performed with the use of a 4-point loading system to obtain a constant moment over the whole length of the transducer (Fig 4.15a). Torsional moment was applied with the aid of a system of levers and pulley as shown in Fig 4.15b.

During the calibration process, it is important to ensure that there is no relative motion at the coupling between the transducer and the loading jig. Any relative movement at the mechanical interface may lead to a 'permanent set' situation, which could produce a zero-shift when fully unloaded. This error can be prevented by applying a full loading cycle immediately prior to calibration in the same direction.

The calibration matrices for the four transducers are shown in Table 4.3, where the columns represent the applied loads and the signal output components are in the rows, as explained in Section 4.3 . The response of the transducers, especially the main channels, showed very good linearity throughout the service loading range. However, some nonlinear output were observed in the cross-talk channels, and these were most likely to be caused by the small relative displacement at the calibration jig.

4.4.3 KAFO Strap Transducers.

A set of 4 KAFO knee strap transducers that are sensitive to only the anterioposterior (Fx) and mediolateral (Fz) forces were also designed and constructed (Fig 4.16). They were designed for a service load of 250N in both the Fx and Fz channels, and can withstand an over-loaded condition of up to 200% without seriously affecting their performance. The fracture laoding is rated at a minimum level of 700N.

The transducers were designed for attachment onto the proximal and distal ends of an 'Otto Bock' knee joint assembly for 16 mm uprights. The device was fixed onto the orthotic joint assembly with 2 M5 bolts which at the same time secure the uprights to the joint assembly. Since the strap attachment points on the transducers are directed away from the joint centre, two versions of the transducers were required, one with the strain gauged section on the left and one on the right hand sides of the main transducer body respectively. The

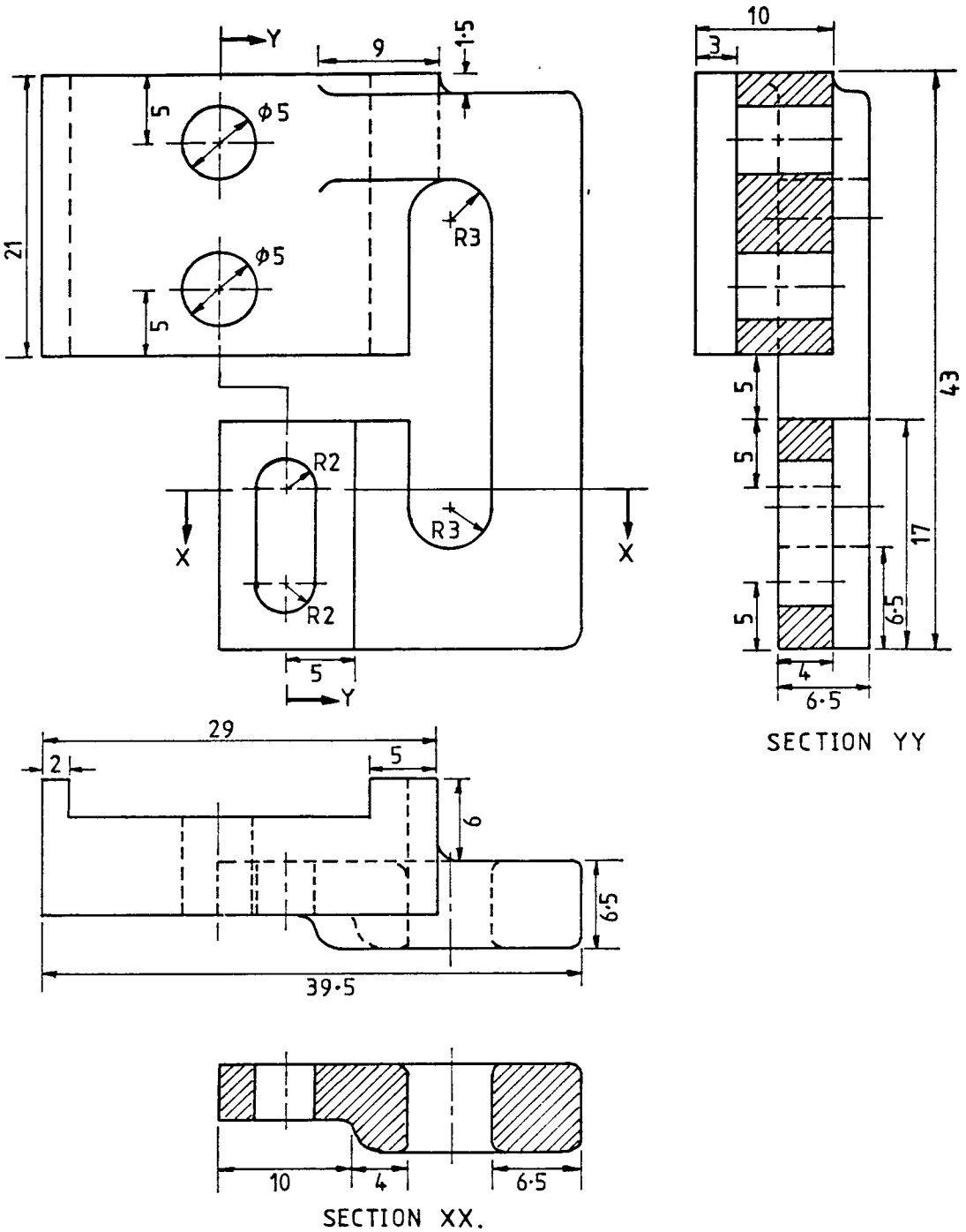


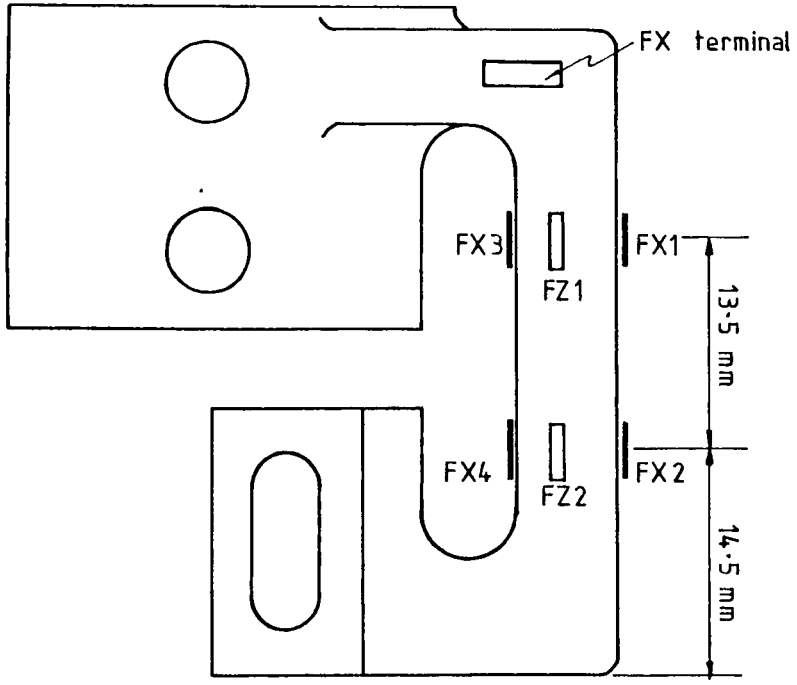
FIG 4·16	The Knee Strap Transducer.

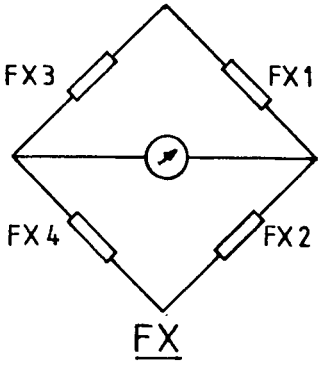
transducers were machined as integral components from a light-weight high-strength aluminium alloy (HE15TF) which possesses an ultimate tensile strength of more than 470 MPa, and an elastic limit of at least 340 MPa.

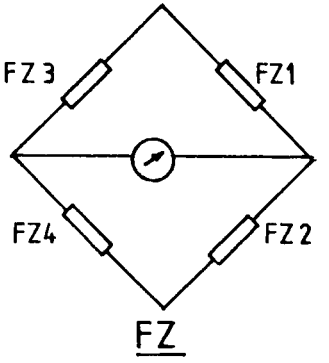
The shear forces are measured through strain gauges wired into a fully active Wheatstone bridge, using the principle of difference of bending moments at two levels to obtain the shearing forces (Fig 4.17) (see Appendix 3). Dead weight calibration method was also used to quantify the performance of the transducers, the results of which show very good linearity (1%) in the response, with negligible hysteresis or cross-effects (Table 4.4). A good sensitivity in the primary channels (averaging to 4.45 μV/N/V) and good repeatability in the out-out signals were also observed.

4.4.4 Shorter Transducers for Children's Orthoses.

The Mark I and Mark II KAFO transducers described previously were designed for adult patients, and were considered to be unsuitable for use in the load measurement of children's light-weight orthoses as they too long. For this reason, a set of shorter and lighter transducers have been designed and constructed by Szary (1985). Although they are not directly concerned with the study reported in this thesis, they will be very briefly described here for completeness in the discussion of the Strathclyde orthotic load transducers. The 6-channel transducers were constructed from high tensile strength aluminium alloy, and are capable of measuring forces and moments about an orthogonal axis system. The active load sensing system consists of a hollow cylindrical elastic element, 10 mm in diameter and 40 mm long. On both ends of the cylindrical section are orthotic upright attachment devices similar in design as those in the Mark I and II transducers. The completed device has a total length of 80 mm, some 32 mm and 16 mm shorter than the Mark I and Mark II transducers respectively.







STRAIN GAUGES USED:

FX, FZ : EA-13-062AA-120.

(MICRO-MEASUREMENT GAUGES).

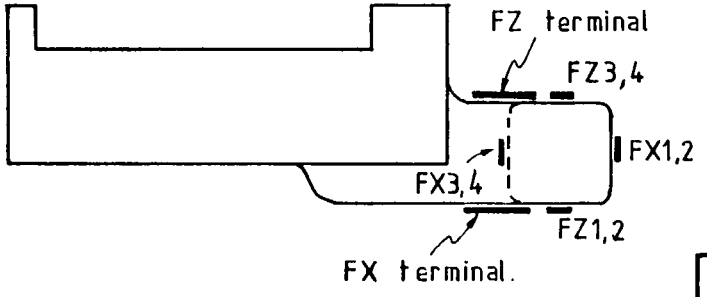


FIG 4·17

Strain Gauge Positioning of Strap Transducer.

Table 4.4: Calibration Matrices of KAFO Strap Transducers.

Transducer S1:

	Fx	Fz
Fx	4.272	0.033
Fz	0.070	4.167

Transducer S2:

	Fx	Fz
Fx	4.209	0.239
Fz	-0.310	4.141

Transducer S3:

	Fx	Fz
Fx	4.473	0.111
Fz	-0.200	4.415

Transducer S4:

	Fx	Fz
Fx	4.856	-0.239
Fz	-0.128	4.533

** In units of μV/N/bridge voltage.

CHAPTER 5.

METHODOLOGY AND INSTRUMENTATION.

Table 5.1 Patient Profile.

Patient Code	Height (cm)	Mass (kg)	Sex (F/M)	Age (years)	Affected Side	Other Walking Aids	Muscle Power (Oxford Scales)												Remarks
							Hip Flexors		Hip Extensors		Knee Flexors		Knee Extensors		Ankle Dorsi-Flexors		Ankle Plantar Flexors		
							L	R	L	R	L	R	L	R	L	R	L	R	
DC	173	67	M	36	Left	-	3	4	3	4	2	5	0	4	0	4	0	4	Teacher Very active
DR	152	50	F	36	Left	-	3	4	4	4	1	5	0	0	0	3	3	5	Factory worker Very active
EM	157	61	F	58	Right	-	5	5	5	5	5	4	5	4	4	0	5	2	Retired Fairly active
FR	166	63	M	65	Left	Double Crutches	1	5	1	5	0	4	0	5	0	0	0	0	Retired Fairly active
MC	169	76	M	37	Right	Stick	4	3	4	3	5	1	5	3	0	3	0	0	Unemployed Low back pain Very active
MN	160	83	M	41	Left	-	3	4	4	4	0	4	0	4	4	5	3	5	Unemployed Fairly active
MR	163	70	M	50	Right	Stick (sometimes)	5	4	5	4	3	0	4	0	3	0	5	0	Administrator Very active
WI	163	70	F	60	Left	-	3	5	1	5	1	5	0	5	0	5	0	5	Retired, Diabetic, Very active
JC	163	55	F	23	Right	-	N O R M A L												Student Very active

5.1 INTRODUCTION.

This chapter describes the methodology and instrumentation employed in the tests. A brief profile of the subjects participating in the tests is discussed, and a description of the various components in the data gathering system will also be provided in the sections that follows.

There are generally two types of tests involving the subjects, namely, the 'normal level walking tests' using the force platform, transducers, and cinephotography; and also the 'limited range free walking tests' where the patients are requested to perform a number of ambulatory activities. In the free walking tests, orthotic loading data are collected over a number of gait cycles, and neither the force platform nor the cinephotographic techniques are used in these tests. The experimental procedures for both types of the tests will be described in detail.

The data analysis and computation of results will also be discussed. A copy of the KAFO test protocol is included in Appendix 6, while Appendix 7 contains samples of the KAFO survey form and the test recording sheets. The flow charts of the data processing system are also given in Appendix 8.

5.2 THE TEST SUBJECTS.

The test subjects participating in the KAFO research were selected from the orthopaedic departments of the collaborating hospitals. The selection criteria are summarised below:

1. the patient should have the ability to walk a distance of at least 5 metres in a straight line without the use of any types of walking aid other than the prescribed devices;
2. the patient should be able to walk for a distance of approximately 80 metres, with a rest of 3 to 5 minutes in between each 10 metre walk without becoming unduly tired;
3. the patient's mass should be less than 80 kg, and the "free length" between any two strap attachment points on the upright of the orthosis, especially in the above knee section, should be at least 150 mm;
4. the lower limb should be free from any major contractures which might affect the patient's ability to stand in a

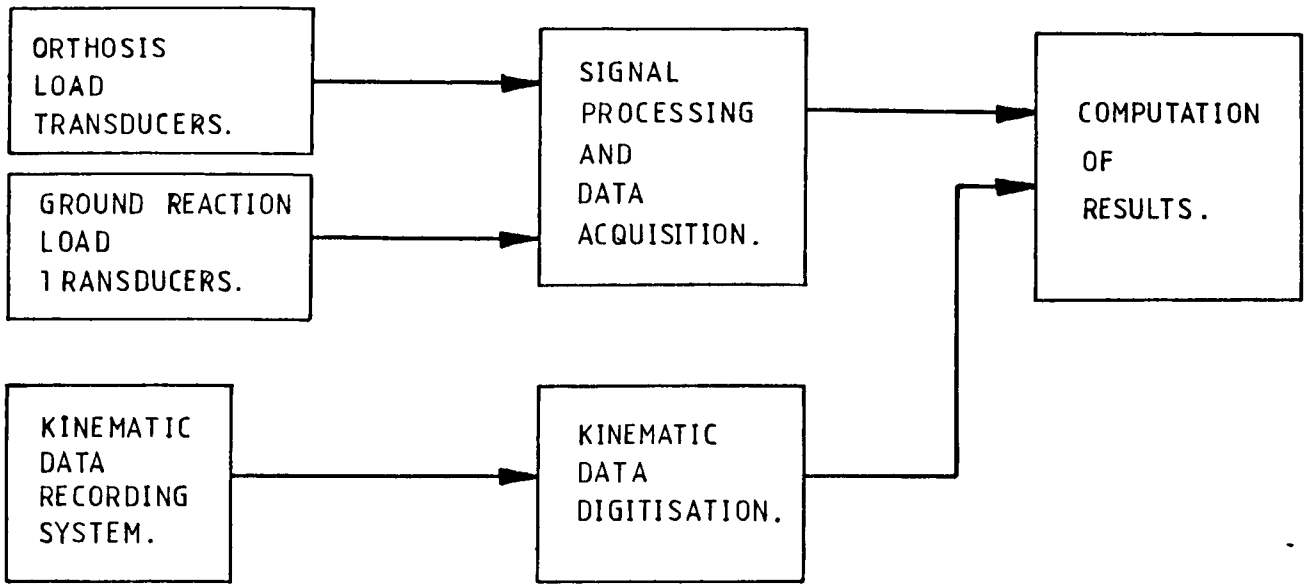


FIG 5·1 The Data Gathering System.

"normal" upright posture.

Eight post-poliomyelitis patients were selected for the walking tests, and of these, 5 were independent of any walking aids other than the prescribed orthoses. For the other 3 subjects, 2 used walking sticks while the other required double crutches for ambulation. A normal subject was also fitted with an orthosis to participate in the investigation.

Each patient was fitted with an instrumented orthosis similar to his/her own KAFO. The muscle power around the major joints of the lower limbs, on both the affected and the contralateral sides, was also assessed and recorded in the Technical Analysis Form (a sample of which is given in Appendix 2). In an attempt to sample the patient's opinions and their experiences with the prescribed KAFOs, they were requested to fill in a KAFO Survey Questionnaire (Appendix 7). Table 5.1 gives a summarised profile of the patients participating in the tests.

5.3 THE DATA GATHERING SYSTEM.

The data gathering system essentially consists of 3 sub-systems, which can be classified as:

1. A load measurement system responsible for monitoring the loadings exerted on the orthotic device and on the ground;
2. A signal processing and data acquisition system to condition, sample, and store the acquired load data; and
3. A kinematic data recording system to sample the 3-dimensional positions of the limb-orthosis complex.

A block diagram of the complete data gathering system is shown in Fig 5.1 .

5.3.1 Load Measurement System.

The load measurement system employed in this study can be subdivided into those transducers responsible for the measurement of loadings on the orthosis, and the device for monitoring the ground reaction loads. Orthotic load measurements were carried out using 3 different sets of custom built transducers, i.e. Mark I and II KAFO transducers, and the strap transducers. The design and construction of these

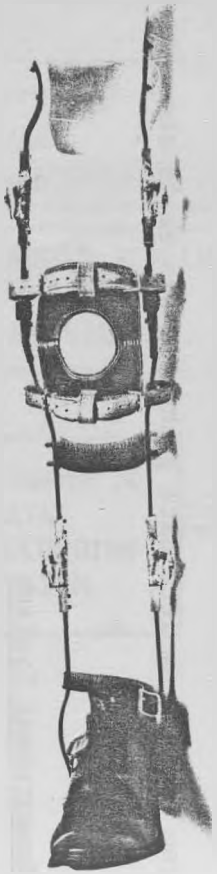


FIG 5.2	An orthosis Instrumented with Mark II KAFO Transducers.

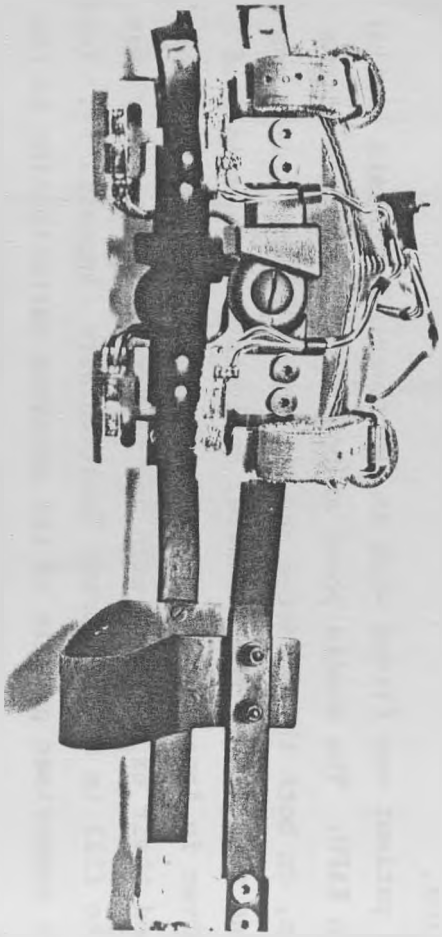


FIG 5.3	Close-up View of a KAFO Instrumented With Knee Strap Transducers.

transducers has already been discussed in Chapter 4.

The Mark I transducers were employed in most of the tests while the Mark II version was still being designed, constructed, and calibrated. These transducers were incorporated into the uprights of the orthosis, with one transducer fitted to each section of the uprights, i.e. proximally at the medial and lateral sides, as well as distally at the medial and lateral sides (Fig 5.2). The strap transducers were mounted onto the proximal and distal attachment points of the upright with the orthotic knee joint assembly (Fig 5.3).

The ground reaction loadings were measured with a "Kistler" multicomponent force platform Type 9261A. It consists of four 3-component quartz transducers prestressed between the top and base plates of surface area of 600 x 400 mm^2. Since the platform is mounted flush with the laboratory floor, and finished with the same type of floor covering material, it presence is not obvious to the participating patients.

5.3.2 Signal Processing and Load Data Acquisition.

The signal processing and load data acquisition system consists of the following components, viz.,

1. 24 bridge units which perform the dual-function of supplying the bridge excitation voltages for the transducers, and amplifying the bridge output voltages of the strain gauge transducers to a level suitable for data acquisition;
2. the charge amplifiers and signal conditioning equipment for the force platform;
3. a multiplexer unit which is capable of converting 30 signal channels from the transducers and the force platform to a single output channel without any loss of information; and
4. a minicomputer for data acquisition and storage purposes.

5.3.2.1 Strain gauge amplifiers: The custom built strain gauge amplifiers for the processing of KAFO load transducer signals were designed to give low drift, low noise, and long term stability measurements. A separate amplifier is connected to each channel of the transducers, and 6 amplifiers are grouped into a bank, sharing a common power supply unit. The amplifier is centred around a CIL Electronics

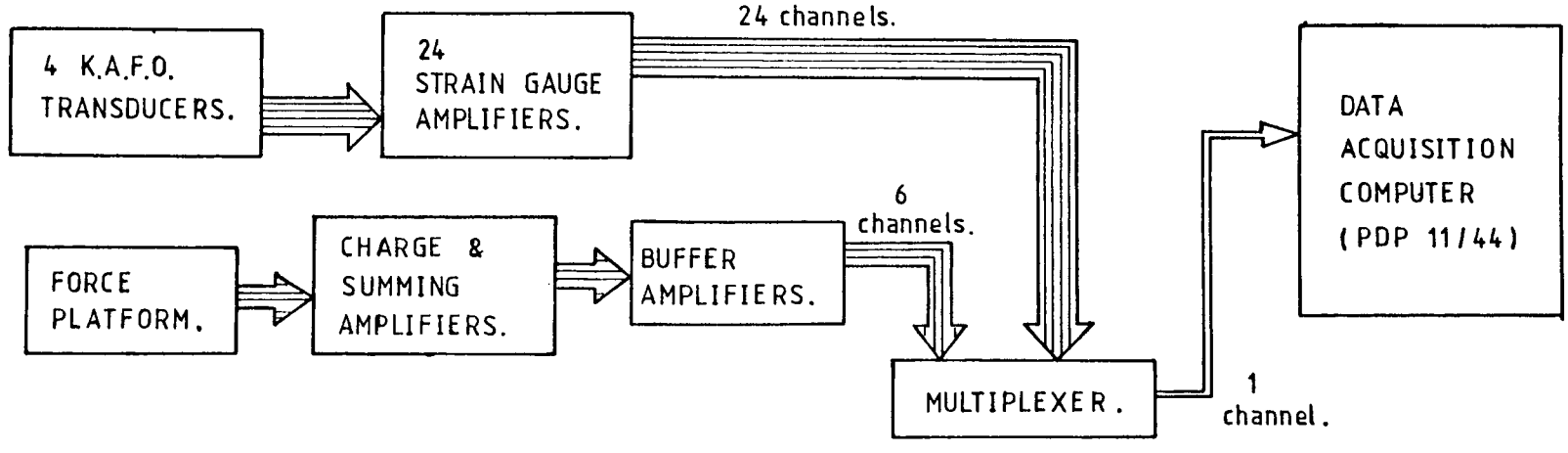


FIG 5·4	Block Diagram of The Signal Processing System.

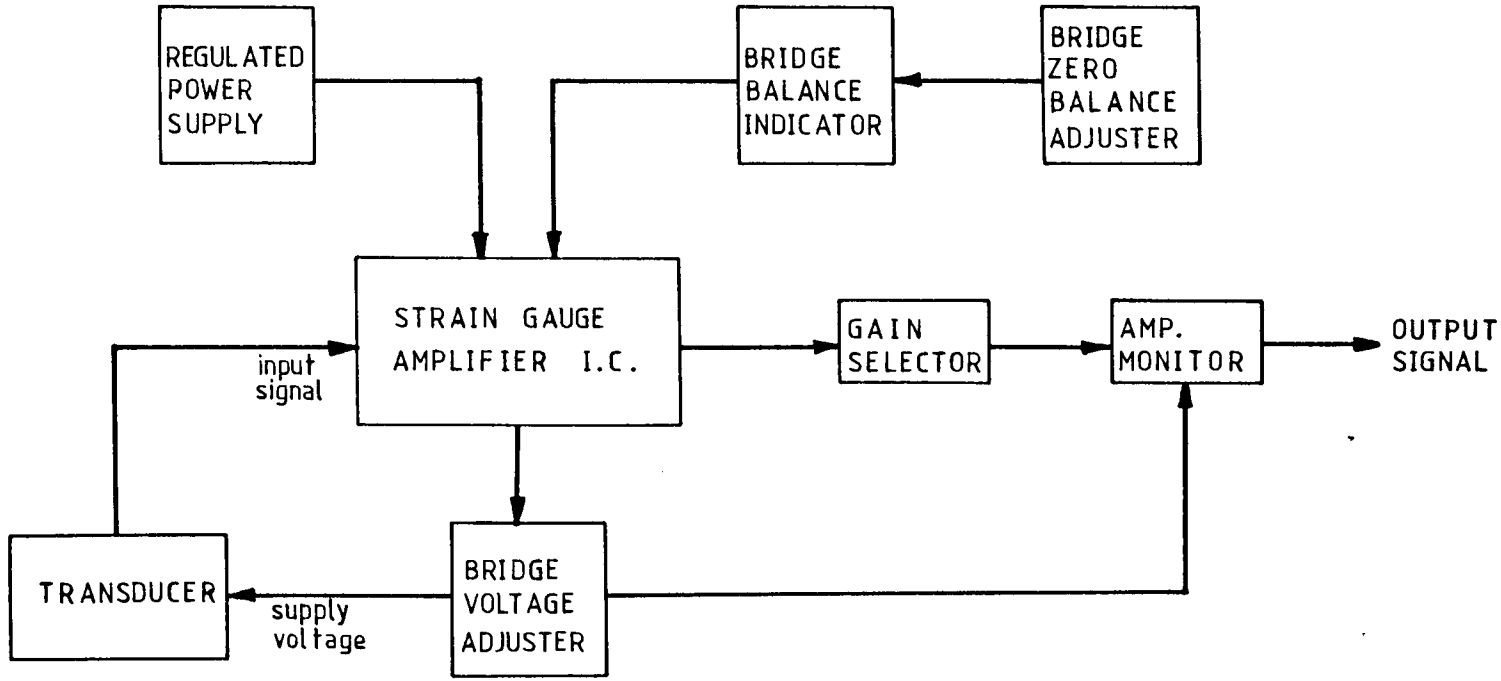


FIG 5·5 Block Diagram of the Strain Gauge Amplifier.

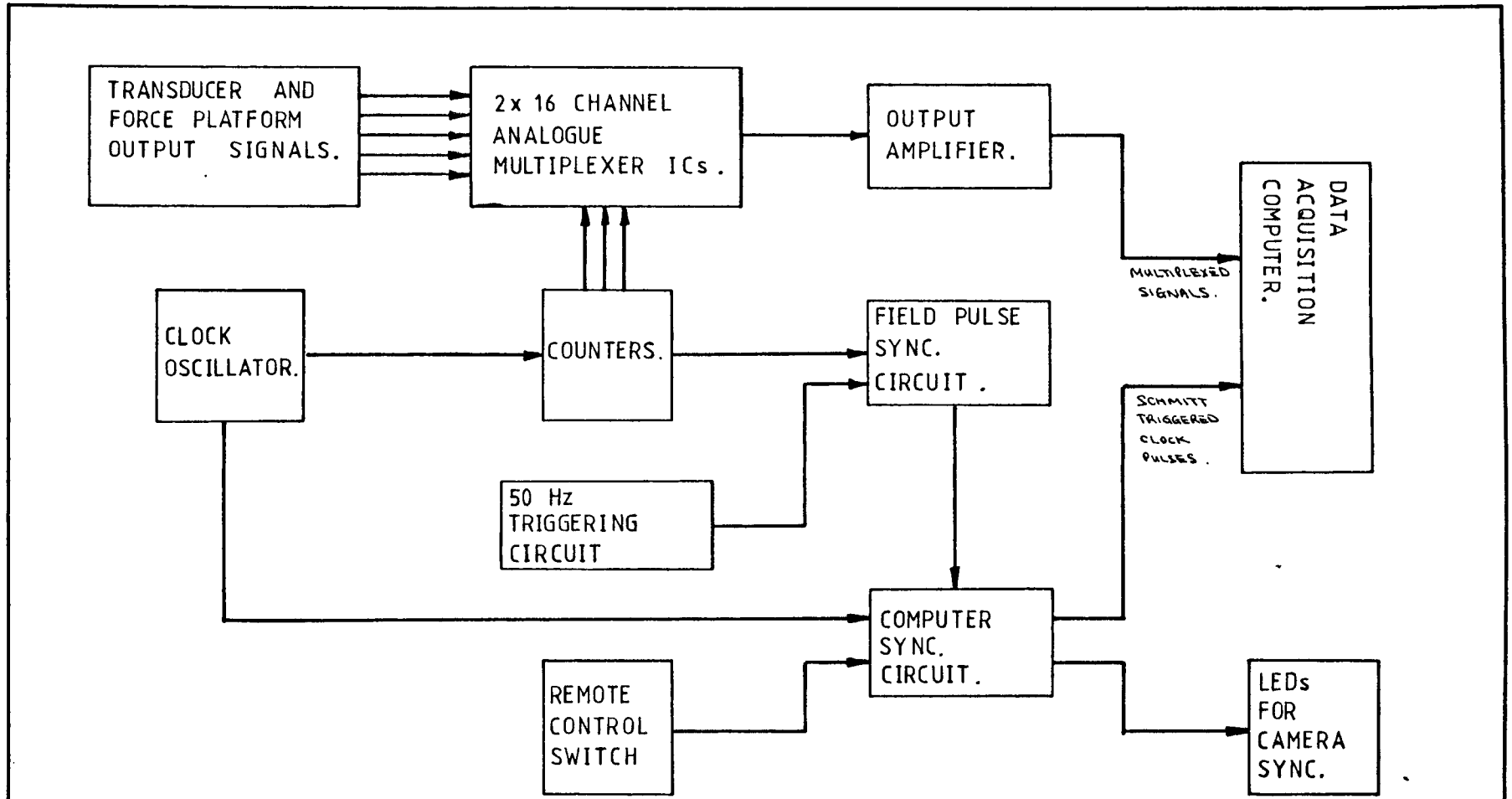


FIG 5·6 Schematic Diagram of the Multiplexer Unit.

Ltd Strain Gauge Amplifier Integrated Circuit Type SGA100, which provides a regulated positive and negative bridge voltage to the transducer, and at the same time amplifies the bridge signal with an adjustable gain factor (Fig 5.5). A 20 VA toroidal transformer isolates the d.c. circuit from the a.c. mains power supply. Incorporated into each power supply module are the amplifier monitoring unit and a bridge balance indicator. These monitoring devices can be switched to any of the amplifiers in the bank so that the working condition of that amplifier can be checked.

5.3.2.2 Force platform amplifiers: The signal processing system for the force platform consists of 8 charge and 2 summing amplifiers. The charge amplifiers convert the electrical charges produced by the piezoelectric force transducers of the platform into proportional voltages, while the summing amplifiers are essentially analogue computers which compute the charge amplifier outputs into the 6 orthogonal output components of the load. Before the signals are sampled by the data acquisition computer, buffer amplifiers attenuate the signals of each channel to a level acceptable to the computer.

5.3.2.3 The Multiplexer: A multiplexing operation is a process whereby a single electrical pathway is used for the simultaneous transmission of several signal channels without the loss of information to any of the individual signal channel. In time division multiplexing, fast acting solid-state analogue switches are used to scan the signal channels sequentially at a predetermined high sampling frequency.

The multiplexer employed for the test is capable of transmitting 32 separate signal channels, and is directly compatible to the data acquisition computer. It basically consists of two 16-channel analogue integrated circuit multiplexers (Hybrid System Type MUX204), and an output line amplifier to compensate for the possible voltage loss in the signal transmission line (Fig 5.6). A control circuit supplies digital coded address signals to the multiplexer I.C.s for channel switching, and at the same time provides control signals to the data acquisition computer and camera synchronising LEDs.

The multiplexer has a free running sampling cycle frequency of 50Hz, synchronised to the a.c. mains power supply. Two other synchronising circuits were also used, namely the field pulse synchroniser

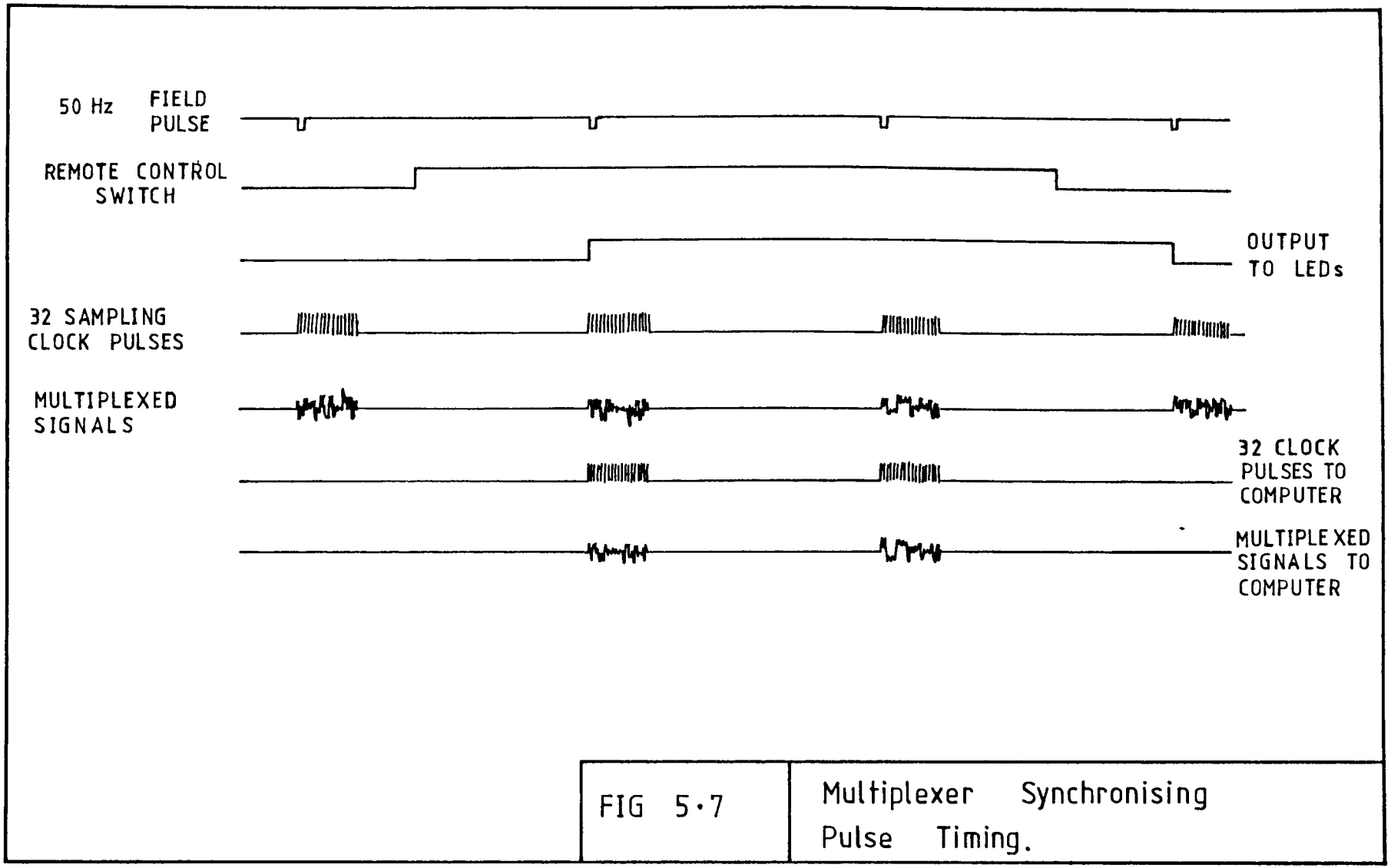


FIG 5·7 Multiplexer Synchronising Pulse Timing.

and the computer synchroniser. The field pulse synchroniser ensures that the field pulse produced from the 50Hz triggering circuit is correctly timed with the train of 32 Schmidt triggered clock pulses, which is further synchronised with the computer by controlling the transmission of these clock pulses, which activate the computer to sample the multiplexed data. When the remote control switch is switched on, the computer synchroniser allows the 32 clock pulses at the next field pulse to pass through to the computer, while at the same time it illuminates the LEDs. The multiplexer synchronising pulse timings are shown in Fig 5.7 . Details of the multiplexer and strain gauge amplifier designs can be found in Trappitt (1979) and Marlow (1982).

5.3.2.4 Data acquisition computer: The computer used for load data acquisition and also for data storage purposes is a Digital Equipment Company PDP 11/34 multi-user minicomputer system. A library of softwares, mostly developed by the Data Analysis section of the Bioengineering Unit, is stored in the system disk and can be utilized readily. The sampled data are collected and stored in real time onto a disk mounted onto another drive.

"MFRUN" is the name of the software used in the acquisition of KAFO loading data, and it controls the running of a number of programmes and interactively prompts the user for relevant information (Fig 5.8). "SCHM512" collects data via the "LPA11K", and the sampling action of this is controlled by the schmidt triggered clock pulses supplied by the multiplexer. On receiving each clock pulse, the computer samples the multiplexed data presented to it along with the clock pulse, and stores the signal voltage into its corresponding points in the computer units (1 volt = 2048 computer units). As the train of 32 channels are sampled, the data are sequentially collected into arrays of 512 sampled points.

Other programmes within the MFRUN include "SORMF" which arranges the unformatted sequentially stored data from SCHM512 into the appropriate channels, and "DISPMF" which allows the user to view and edit the data on a display unit.

5.3.3 Kinematic Data Recording System.

The function of a kinematic data recording system is to obtain the coordinates of important landmarks on the limb-orthosis complex

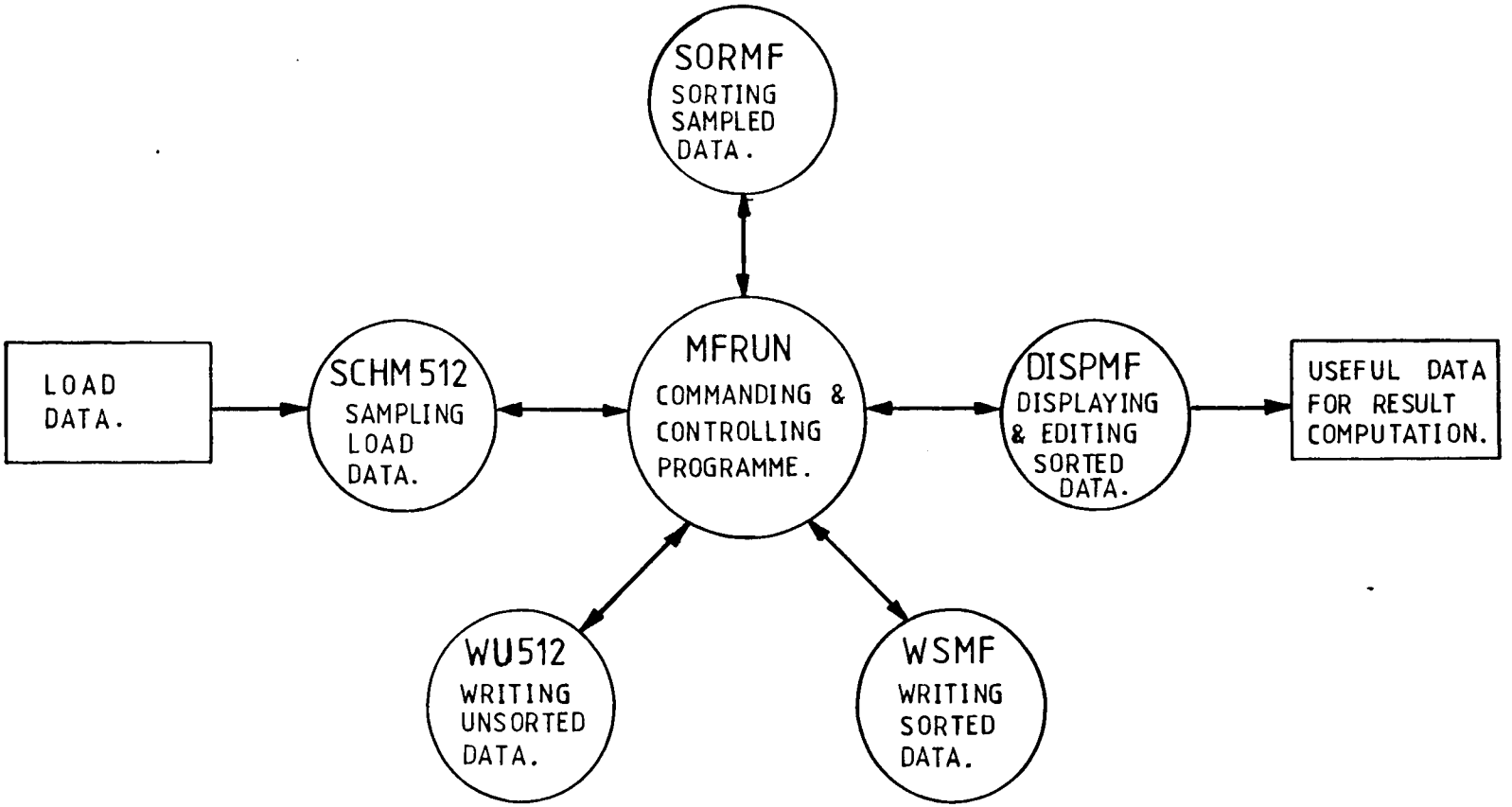


FIG 5·8 Block Diagram of the Load Data Acquisition System.

during a walking test. Since a camera can only provide a 2-dimensional image, a minimum of two cameras will be required for the 3-dimensional analytical system.

In the Bioengineering Unit, three kinematic recording systems are available. They are:

1. TV-computer system;
2. Selspot system; and
3. Cinephotographic system.

Both the tv-computer and Selspot systems provide rapid recovery of kinematic data when used on-line with a computer, and they have the additional advantage of a very low running cost. However, their major drawback is that only points of the landmarks are recorded, without the complete body image. The cinephotographic method's data recovery is tedious and time-consuming, and on top of that, it also has a high running cost. However, its greatest advantage over the other systems is the provision of complete visual image of the subject, which allows qualitative assessment of the pathological gait. It was mainly for this reason that the cinephotographic method was chosen for this study.

The cinephotographic system consists of three 16 mm "Bolex H16" reflex cine cameras, driven by synchronous motors through a gearing system to produce a frame rate equivalent to the mains frequency of 50 Hz. The cameras are orthogonally oriented one with respect to the other, with the optical axis of the front camera parallel to, and looking down the X-axis of the Ground Reference System (GRS). The other two cameras are placed on both sides of the walkway, with their optical axes parallel to the Z-axis of the GRS. The left camera looks along the Z-axis while the right camera looks down the same axis. The realtionship between the camera optical system and the origin of GRS is shown in Fig 5.9 . Definition of the GRS is given in Section 5.4.1 .

For this study, only two cameras are used at any one time, i.e. the front camera and either one of the side cameras, depending on the affected side of the patient. If the patient wears a KAFO on his/her right limb, then the front and right cameras will be used.

5.3.4 Equipment Synchronisation.

In order to perform the analysis accurately, the kinematic data, the force platform data, and the transducer data should all be sampled

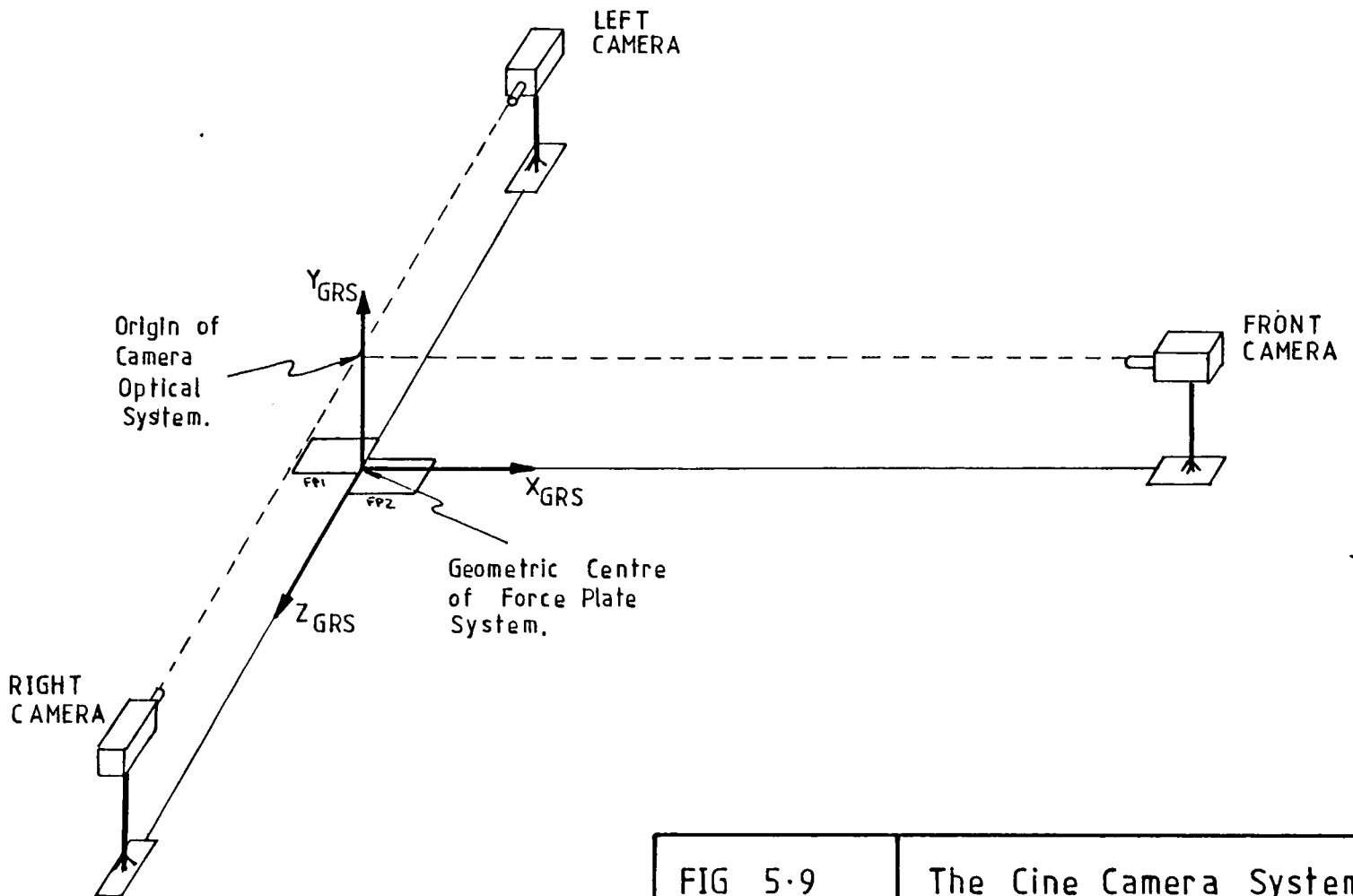


FIG 5·9 The Cine Camera System.

at the same instant of time, and at a constant time interval throughout the duration of the test.

For the data gathering system, the 3 pieces of equipment that require to be synchronised with each other are: the cine camera, the multiplexer, and the data acquisition computer. The sampling or field rate of the cameras and the multiplexer is synchronised with each other through the a.c. mains power supply. The multiplexer is in turn synchronised with the computer by clock pulses generated from the multiplexer. However, there is also a need to identify a particular cine film frame with the corresponding multiplexer sampling field. This is achieved by the use of high density LEDs, which are illuminated at the same instant as the first of the clock pulses are being sent to the computer when the sampling action is initiated. The frame on which the first LED illumination is seen will correspond to the first sampled load data field.

5.4 EXPERIMENTAL PROCEDURE.

The KAFO loading tests carried out are:

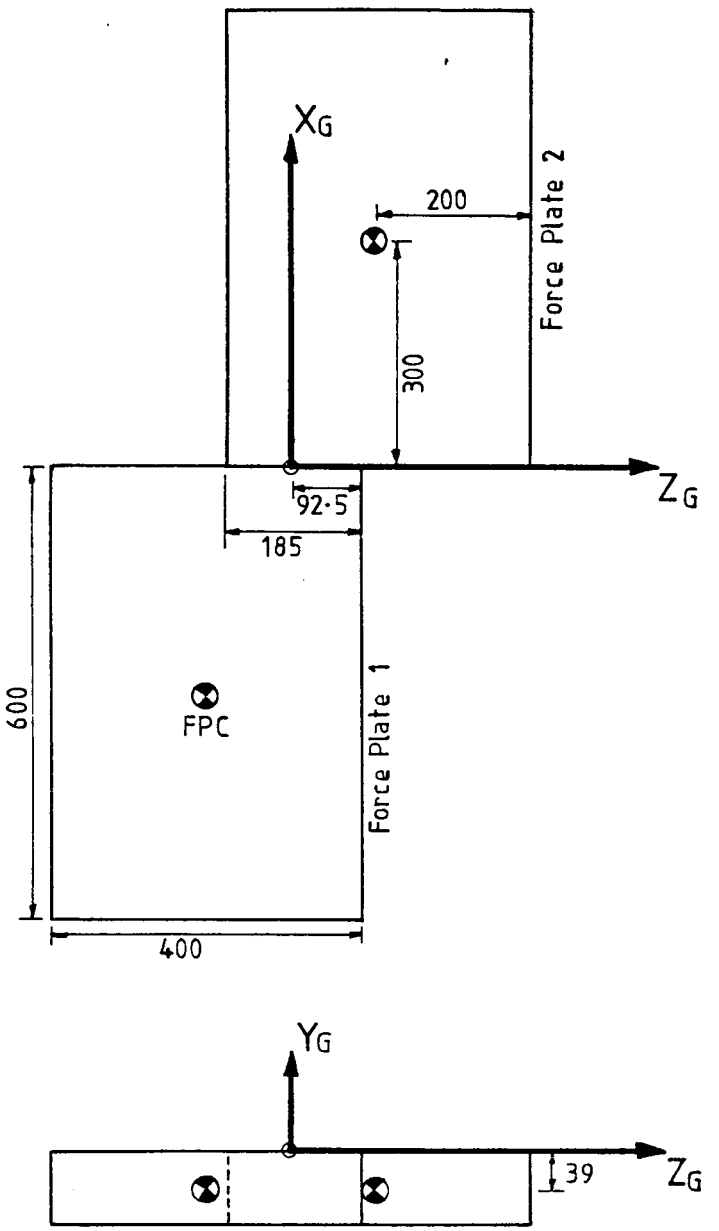
1. the "Normal Level Walking" (NLW) tests; and
2. the "Limited Range Free Walking" (LRFW) tests.

In the NLW tests, the patients are asked to walk along the walkway, in a straight line and at their own preferred speed. Cameras are used to collect the kinematic data of the limb-orthosis complex as it transverses over the force platform. However, only the data acquired over a single gait cycle when the patient stepped onto the force-plate will be used in the analysis.

The LRFW tests collect and analyse orthotic loading data over a number of steps, and for various types of ambulatory activities. These tests are very useful in showing the loading pattern experienced by the orthosis in different situations. However, they are unable to indicate the anatomical loading characteristics since the force platform and the cine cameras are not employed in the tests.

5.4.1 Normal Level Walking Tests.

Three sets of data are collected simultaneously for each "run" of the normal level walking (NLW) tests, and they are: the kinematic



(Dimension : mm)

FIG 5·10	The GRS Origin and The Force Plate Centres.

data from the 2 cine cameras, the orthotic loading data from the KAFO transducers, and the ground-reaction load data from the force platform. To inter-relate these data with one another, a common reference system is required. The method employed is to define a Ground Reference System (GRS) with respect to a fixed point on the walkway in the kinematic system such that the loading data may in turn be related to it.

5.4.1.1 The ground reference system: The origin of the GRS is taken to be at the geometric centre of the force platform system, which is a point at the ground level of the walkway immediately below the intersection of the optical axes of the cameras. This should be differentiated from the force plate origin which is a point 39 mm below the ground level of the walkway, at the centre of the individual platform.

The X-axis is defined as a line parallel to the optical axis of the front camera, and in the direction of progression of the walking tests. The Z-axis originates at the centre of the force platform system, parallel to the optical axis of the side cameras, and pointing into the right camera. The Y-axis is perpendicular to both the X and Z axes, and pointing from the ground level towards the ceiling of the laboratory. The relationship between the origin of GRS and the force plate centres is shown in Fig 5.10 .

5.4.1.2 Marker analysis: A total of 24 markers was used in the NLW tests. They can be generally divided into 3 groups according to their applications, i.e. the field and scaling markers, the markers that are sited on the orthosis, and the markers attached to the patient (Table 5.2). The markers are made from beads, of 10 mm diameter, painted in fluorescent yellow and connected to a matt-black base through a fluorescent orange light metal or wooden stick (Fig 5.11).

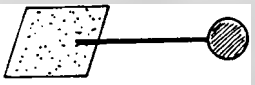
Field markers are employed to relate the image captured on the cine film with the GRS, and they are positioned around the force platform in such a way as to ensure constant vision by both cameras. Their positions with respect to the force platform are shown in Fig 5.12a, where SM1, SM2, and SM3 will be used with the right camera, while SM1', SM2', and SM3' are for the left camera.

The scaling calibration markers are placed on a grid board at a height of 860 mm above the ground level, i.e. the height of the camera

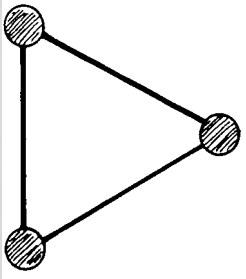
PL,
KL

KM

AL,
AM



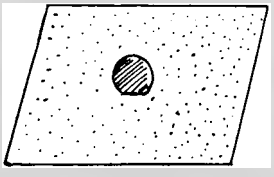
Transducer
Marker
Triangle.



HTR,
HTS,
HTT

BM

SM1,
SM2,
SM3



All
Other
Markers.

(not drawn to proportion)

FIG 5·11	Configuration of Marker Sticks.

Table 5.2 Marker Analysis.

Marker	Marker Utilised in: Static Calibration	Patient Tests
CALIBRATION MARKERS		
Field Markers: SM1,SM2,SM3	Yes	Yes
Calibration board:		
Horizontal: BX or BZ	Yes	No
Vertical: BY	Yes	No
ORTHOSIS MARKERS		
4 Transducer Marker Triangles	Yes	No
Lateral Knee Markers: KL,PL	Yes	Yes
Medial Knee Marker: KM	Yes	Yes
Lateral Ankle Marker: AL	Yes	Yes
Medial Ankle Marker: AM	Yes	Yes
PATIENT MARKERS		
Hip-Tail Marker Stick: HTR,HTS,HTT	No	Yes
Nearside Hip Marker: H1	No	Yes
Offside Hip Marker: H2	No	Yes
Knee Centre Markers: KC1,KC2	Yes	No
Ankle Centre Markers: AC1,AC2	Yes	No
Shank Markers: Proximal: BS	Yes	Yes
Distal: BI	Yes	Yes
Medial: BM	Yes	Yes

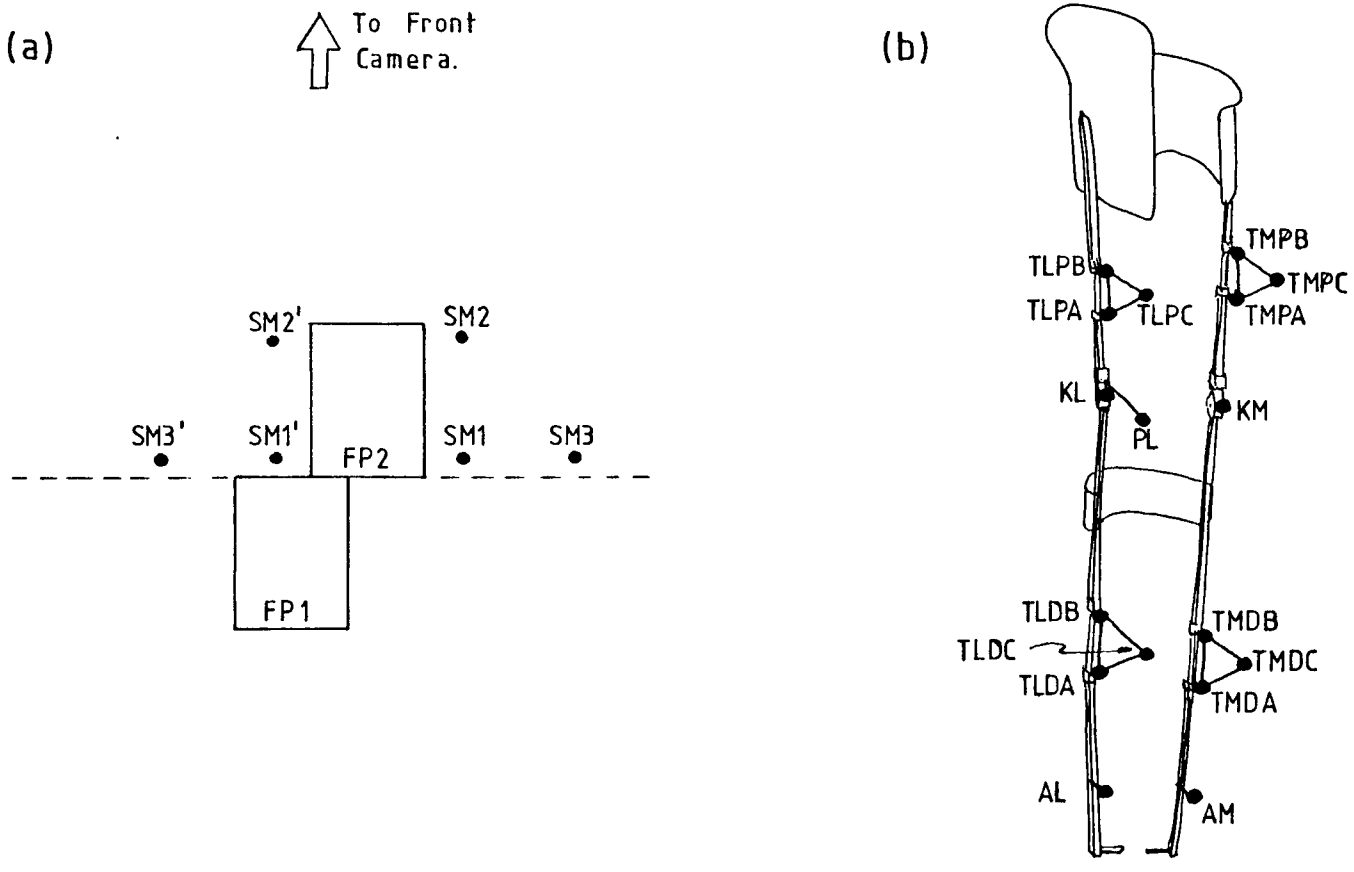


FIG 5·12 | Field and Orthotic Marker Positionings.

optical axes. They are for relating the dimensions of the projected cine film images with their corresponding actual physical sizes.

Positions of the orthotic and limb markers are shown in Fig 5.12b and Fig 5.13 respectively. Their significance will be discussed in Chapter 6.

5.4.1.3 Laboratory test procedure: The key to a successful patient test is the systematic approach in setting-up the necessary equipment and ensuring that they are in good operational condition, and in carrying-out the test according to a well-defined protocol. It is equally important to create a relaxed and confident atmosphere throughout the test.

Before the test, the KAFO load transducers are securely attached to the orthosis fabricated for that particular patient. It is imperative to handle the transducers with due care at all times, and to ensure that they are not over-stressed during the installation processes. The uprights of the instrumented orthosis are then sprayed matt black to reduce the glare effects of the metallic surfaces on the film records. Other reflective surfaces are also covered with black tapes. The signal processing and data acquisition equipment are checked to ensure that they are in good working order. This is particularly important since the large number of measuring channels will mean that the probability of a failure in the system increases correspondingly.

The transducer strain gauge amplifiers are set to their appropriate bridge supply voltage levels using a digital multimeter (DVM). With the instrumented KAFO leaning in an upright position against a chair, the amplifiers are zero-balanced at the highest gain setting. After zero-balancing, the amplifiers are reset to the expected gain levels, or to a gain of 500 for a new patient.

The cine cameras, i.e. the front and one of the side cameras (depending on the affected side of the patient) are loaded with the appropriate type of film (Kodakchrome ASA40), and the aperture set to the following f-stops: f1.5, f1.6, and f1.3 for the front, left and right cameras respectively. The light emitting diodes (LEDs) are positioned, and overhead lights checked to ensure that the force-plate area is evenly and sufficiently illuminated.

Just before the patient's arrival, all the equipment are switched

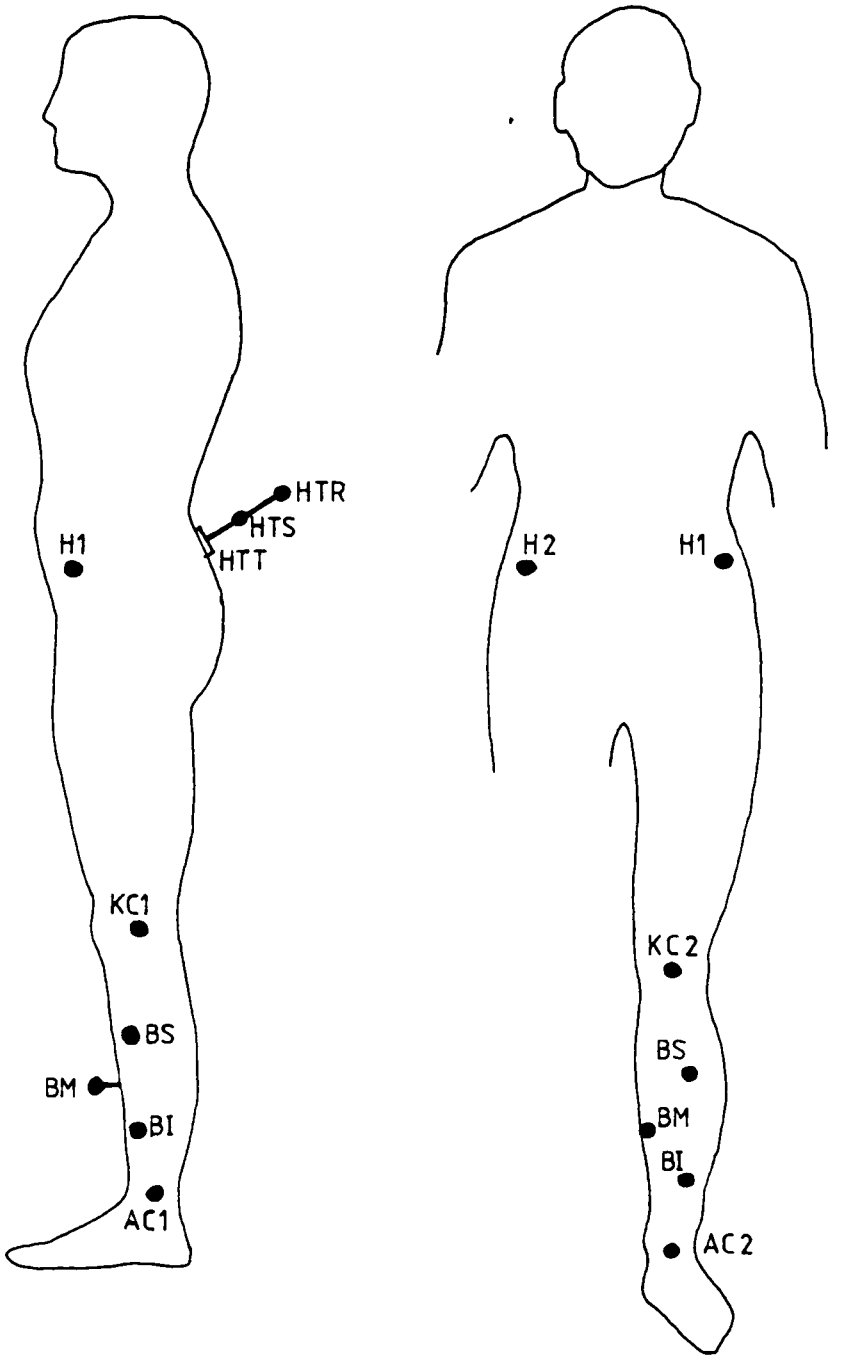


FIG 5·14 Position of Markers on a Left Limb Affected Patient.

on and allowed to warm up for at least 30 minutes. Field markers are attached with double-sided adhesive tapes onto the walkway, beside Force-Plate 2. Any other final preparation should also be completed before the patient arrived.

Initially, the patient, wearing the instrumented orthosis and with the markers securely attached, is asked to walk freely around the laboratory. The maximum and minimum magnitudes of the transducer output voltages attained by all the channels as recorded by the storage oscilloscope may then be used for adjustment of the amplifier gain settings to ensure that the output voltages would not exceed ±1.0V, which is the range acceptable by the computer.

The sequence of events in a patient tests are as follows:

1. static calibration of the patient with the instrumented KAFO;
2. the walking test proper;
3. static calibration of the instrumented orthosis; and
4. scaling calibration using the grid board.

In the KAFO-patient static calibration, the patient with the knee and ankle centre markers attached, is asked to stand on the force-plate facing the front camera. The cameras are switched on for approximately $\frac{1}{2}$ second to record the relationship of the markers on the orthosis with those on the patient's limb. The duration of this calibration is normally kept to the minimum because of the full "heating effects" of the cine lightings on the patient while standing stationary on the force platform.

The patient is then positioned at the starting point of the walkway, and asked to walk towards the front camera, with an assistant holding the data transmission cable behind the patient. If the patient does not step centrally on the force plate, the starting position is adjusted, and the procedure repeated. Normally, the location of the force platform is concealed from the patient.

When the patient is ready for a test run, the cine test reference numbering system is set to the required number, and the computer activated with the "MFRUN" programme. As soon as the computer is ready for sampling, the patient is instructed to start walking whenever he/she wishes to do so. At heel-strike of the affected limb prior to stepping on the force platform, the cameras are started, followed by initiattion of the sampling process at toe-off of the same stance phase. After the patient has cleared the force platform area, the

cameras should be stopped to conserve the film. For the first test run, the load channels are examined on the visual display unit of the computer terminal to ensure that the sampling has been satisfactory. In the subsequent runs, a check on the storage oscilloscope display is sufficient for the purpose of ensuring the amplifier outputs are within the allowable voltage range.

After the walking tests, the hip-tail marker distances are measured and recorded. The orthosis, together with its attached markers, is removed from the patient with care to prevent displacing any of the markers.

When the patient has left, the orthosis with the transducer marker triangles attached, is leaned against a stool positioned on the force platform. The markers are checked to ensure that they are all visible to both the front and side cameras. The cameras are then run for approximately $\frac{1}{2}$ second.

The scaling calibration grid board is placed immediately above the GRS origin, facing the front camera. Before running the camera, markers are placed on the board, at 860 mm above the ground level and 510 mm away from the central vertical line of the board. The same procedure is repeated for the side camera.

The x,y,z components of the distance vectors from the markers to various points on the orthosis are also measured and recorded. These measurements are taken in the respective orthosis upright coordinate systems.

A protocol outlining the procedure of the test is included in Appendix 6, while Appendix 7 contains samples of the test recording forms.

5.4.2 Limited Range Free Walking Tests.

These tests are designed to simulate some of the situations that a KAFO wearer may have to face in the day-to-day ambulatory activities, and to quantitatively monitor the loading performance of the orthosis while being used to carry out these activities. Ideally, the test should be carried out in an actual environment, such as while the patient is walking along a busy street. However, for practical and instrumentational constraints, it was decided that the tests should be carried out in the biomechanics laboratory.

Therefore, the range and types of activities will be restricted by the length of the data transmission cable from the orthosis to the amplifiers, the floor area of the laboratory, and also by the ability of the patients to perform these activities.

In these tests, only the orthotic loading data are collected. The ground reaction loads are not recorded because the results obtained from a single step will not be representative of the continuously changing events, such as walking round a 90° bend. Similarly, the equipment and analytical limitations also prevented rational kinematic results to be obtained from the cine cameras. If necessary, the essential direction cosine matrices can be obtained from the normal level walking tests, assuming that the patient uses the same instrumented orthosis under the same conditions.

Two general area of activities were investigated, namely, level walking, and manoeuvres to clear obstacles. These can be divided into six basic types of ambulatory activities as follow:

1. level walking in a straight line;
2. level walking, negotiating a set of double right-angled bends;
3. level walking, following a U-shaped path;
4. stepping up and down a low platform;
5. ascending and descending an inclined walkway; and
6. ascending and descending stairs.

5.4.2.1 Straight line level walking: This is to simulate normal walking at leisurely or preferred speed. The patients are asked to walk along a walkway for a distance of approximately 10 metres to allow for at least 6 full walking cycles to be sampled. A selected number of patients are also asked to walk at a speed as fast as they possibly can, and also at a very slow speed. For two of the patients, data were collected as they were starting to walk from a stationary position, as well as when they were slowing down to a halt. This would give an indication of the changes in the walking characteristics as compared to the steady state level walking.

5.4.2.2 Level walking round double bends: It is a simulation of KAFO wearers having to walk round a corner, or to avoid colliding into fellow pedestrians while walking along a crowded street. For this test, the patient is asked to walk along the walkway, then take a

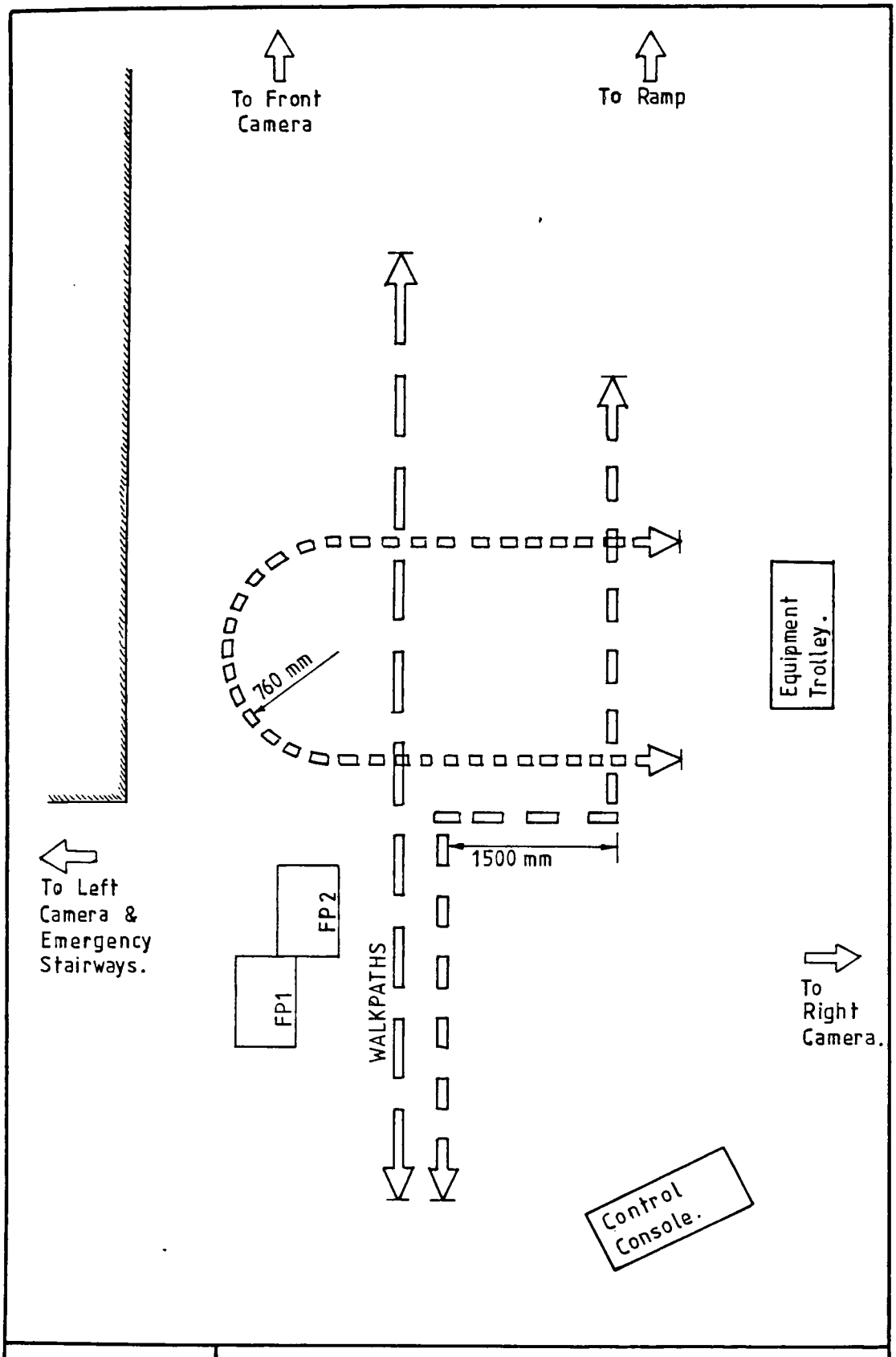


FIG 5·14 Arrangement of The Biomechanics Laboratory for LRFW Tests.

sharp 90° right turn, walk a further 2 to 3 steps, and take a right-angled left turn before continuing the walk. Short lengths of masking tapes are placed on the laboratory floor along the intended walk path. Two high round stools are also positioned at the inside of the right-angled turns to allow the patient to walk round the bends without having to fix the eyes on the masking tapes. The approximate positions of the heel-strikes on the KAFO side with respect to the bends are also noted.

5.4.2.3 Level walking round U-shaped path: To simulate walking round an obstacle, the patients are asked to follow a U-shaped walk path with a semi-circular radius of 760 mm. In the same way as in the previous test, a stool positioned inside the semi-circular path guides the patient round the intended route. For both these tests, the patients are asked to look in the forward direction as they normally would, and use the stool to guide them round the intended path. This would allow them to use their preferred way to negotiate the obstacles. A diagram showing the general laboratory arrangement for the level walking tests is given in Fig 5.14 .

5.4.2.4 Up/down low platform: This manoeuvre aims at simulating a situation where a curb, or an unlevelled floor is to be negotiated. A low wooden platform approximately 100 mm high, with an area of 760 mm by 900 mm was placed across the walkway. The patients are asked to walk along the walkway, step onto the platform, walk across, and then step down from it before completing the rest of the walkway.

5.4.2.5 Up/down ramp: A ramp with 1:7 incline (approximately 8° gradient) and 3 metres in length is employed in this part of the test. For patient's safety, handrails are built around the whole length of the ramp as well as at the top of the slope. In addition, the surface of the ramp is also covered with a non-slippery flooring material (Fig 5.15a). The patients are asked to ascend the ramp, turn round at the top before proceeding with the descent. Data are collected for both the ascending and descending walk.

5.4.2.6 Up/down stairs: A 6-step stair module is available in the laboratory which was designed by researchers working on the gait analysis of normal subjects (Morrison,1967). However, the construction

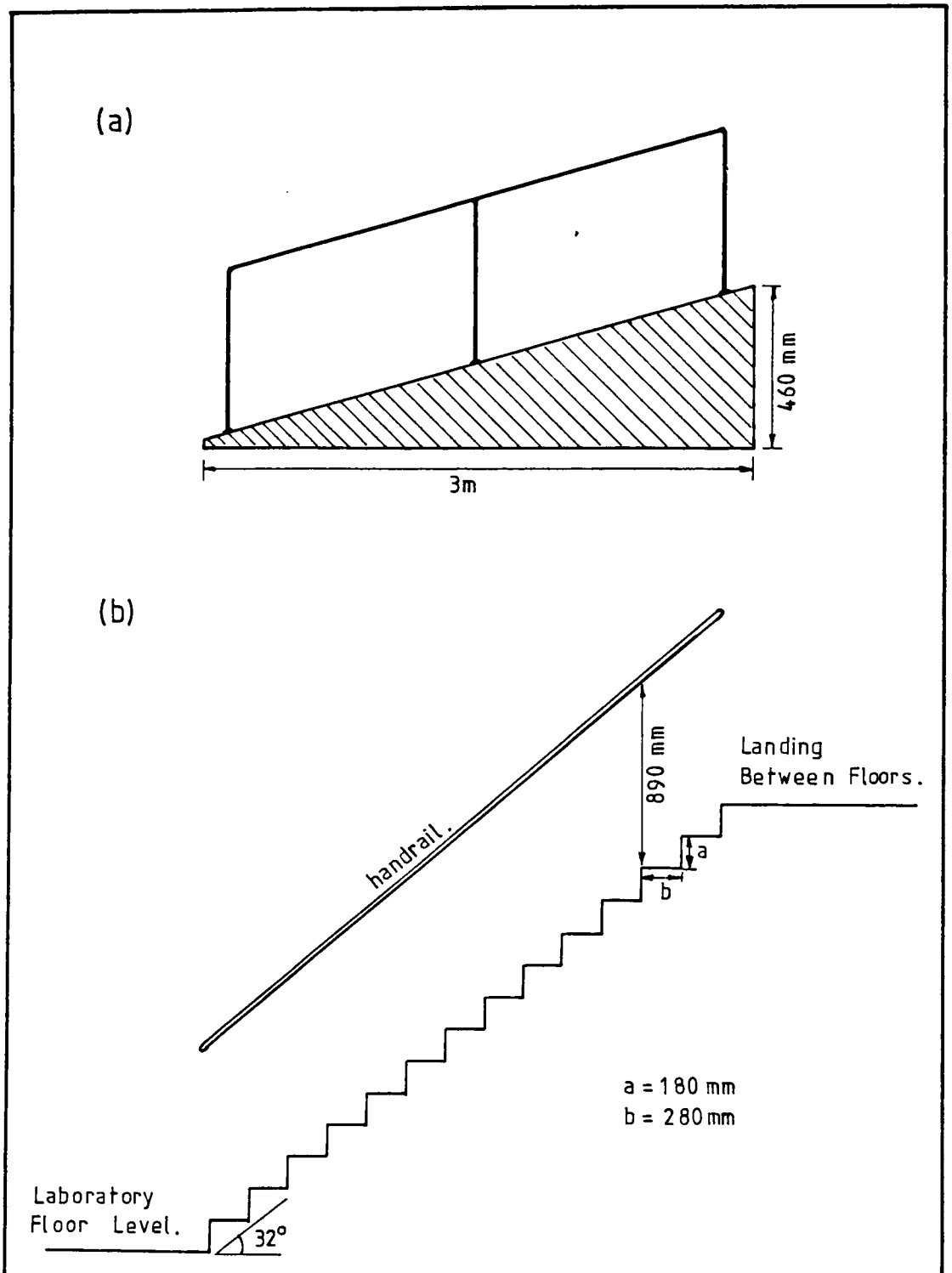


FIG 5·15	The Ramp and Stair for The LRFW Tests.

was based on a design for small dwelling houses, with very shallow depth of tread (180 mm), and a steep pitch of some 45° inclination to the horizontal. It was, therefore, considered to be unsuitable for tests on KAFO wearers.

The test is carried out using the "emergency staircase" by the side of the laboratory, which has a 280 mm depth of tread, a 180 mm riser, an unobstructed width of 900 mm, and a generous pitch of 32° (Fig 5.15b). The dimensions of this staircase are classified as optimal for 'semi-public' stairs (BS 5395:1977). In the tests, the patients are asked to ascend or descend 7 to 8 steps to allow for a 6-step cycle to be sampled. The patients are advised to use their preferred speed and given adequate rest between each climb.

5.5 DATA ANALYSIS.

5.5.1 Data Reduction and Preparation.

Due to the enormous amount of data collected in each test run, the useful data have to be identified and separated from the unwanted excess information. The amount of loading and kinematic data collected are directly proportional to the length of sampling and camera running times respectively.

The load data is edited by using "MFRUN" to identify the stance phase recorded by the force platform. It can be easily and precisely identified by the use of the Fx and Fy channels of the force-plate data, where they possess very distinctive loading and unloading characteristics at heel-strike and toe-off. By displaying the force-plate Fy channel on the visual display unit of a computer terminal, and placing the left cursor on the first sampled point of the stance phase and the right cursor at the toe-off point, the stance phase data between, and inclusive, of the cursor points can be transferred and stored in a designated file at the PDP computer. The positions may be verified, if necessary, with the force-plate Fx channel. All the other channels are then sequentially displayed at the terminal and recorded into the designated file.

Since the kinematic data are recorded as images on the cine films, they have to be converted into the digital form. This is realised by the employment of a trace analyser system, consisting of a 16 mm cine

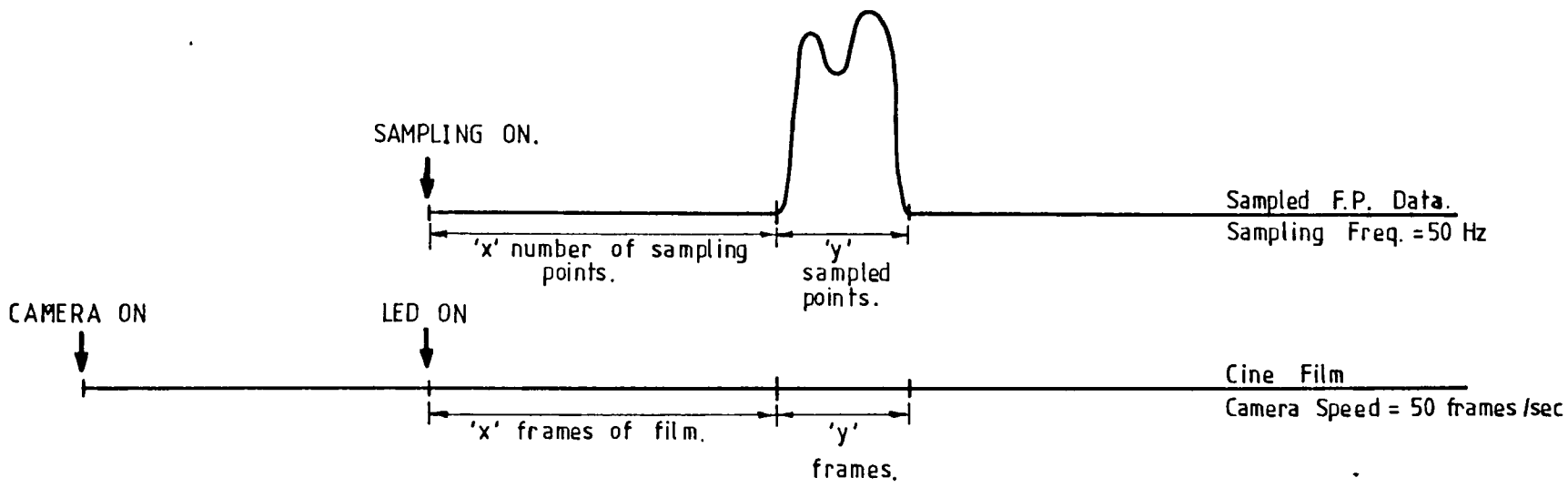


FIG 5·16 Synchronisation of Kinematic and Load Data.

projector and a "Calcomp 9000" digitiser. The image recorded on the film is projected onto a mirror mounted at 45° to the horizontal directly above the digitiser tablet so that the image falls squarely onto the tablet. A flying cursor is then positioned over the marker to be digitised, and by depressing a key on the cursor, the position of the marker on the digitiser tablet surface can be electromagnetically converted into the digital coordinate points with respect to an origin at the lower left corner of the tablet.

To synchronise the cine film with the load data, the film is advanced frame by frame until the first frame in which the illuminated LED is observed, and this will correspond to the first sampled point of the load data (Fig 5.16). The number of sampled points between the initiation of sampling and the on-set of force-plate Fy trace would be the same as the number of cine frames between the first LED illumination and the heel-strike.

Digitisation is then carried out on the film from the side camera, first on the scaling calibration grid board image, then on the KAFO static calibration, the patient-orthosis static calibration, and finally on the dynamic walking tests. The same procedure is repeated for the front camera.

The load and kinematic data are then transferred to the University's main frame computer (DEC type VAX 11/782) via a private exchange system between the department and the Computer Centre. Other data such as patient identification and measured or recorded data are then added onto the data files.

For the Limited Range Free Walking tests, force-plate data are not available for locating the precise instants of heel-strike and toe-off. Initially it was considered that foot switches might be needed, but this would inevitably increase the number of sampling channels. However, detailed studies of the Fy traces obtained from the transducers revealed that heel-strike and toe-off can be accurately established from the rather distinctive turning points. The accuracy was verified by comparing with the force-plate tracings, and the maximum error was less than 2 sampled points, i.e. better than 40 milliseconds.

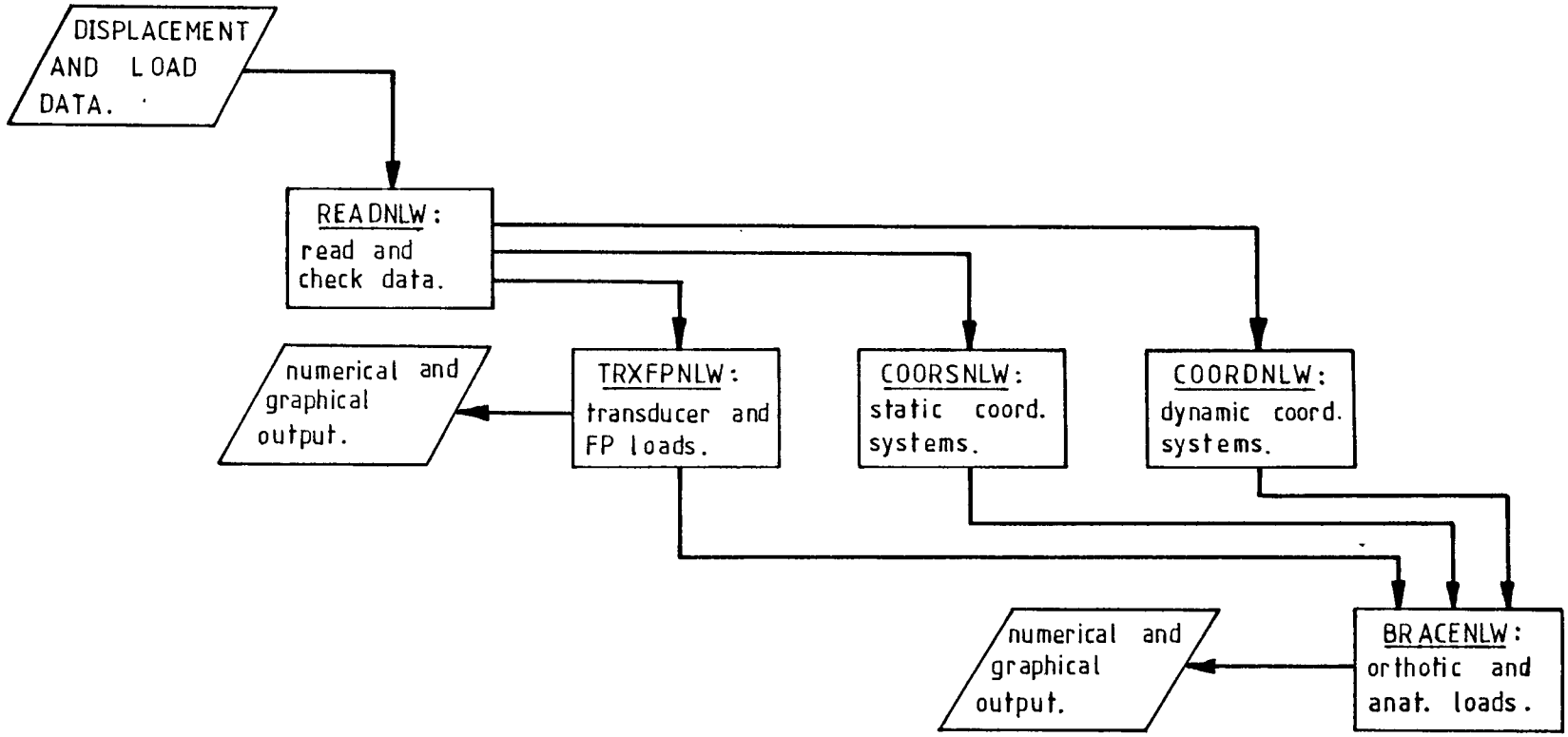


FIG 5·17 Computation of Results for Normal Level Walking Tests.

5.5.2 Computation of Results.

Main frame computers are used in the handling and calculation of results for the KAFO tests. At the initial stages of the study, the Computer Centre's ICL 1904S George III system was used. It was a system suitable for processing large amounts of data as 'background' jobs. However, this computer was later phased out, and was replaced with a DEC type VAX 11/782 system. In changing over to the new system, the computer programmes and file store designation had to be amended, and the execution commands rewritten according to the requirements of the new system. The following section describes the computation of results using the VAX system.

The computer programmes are written in the "FORTRAN 77" language, and stored in the executable binary form. For the NLW tests, the analysis is based on the programme named "AETPROG1" reported by Trappitt (1979). Instead of the large AETPROG1 programme, the new enhanced and improved version consists of 5 smaller programmes forming a 3-tier data processing system. With this, better accuracy, wider versatility, greater use of the primary input data, and quicker turn over of results can be achieved. The first stage of the system consists of the programme "READNLW" which checks data for 'out-of-range' and synchronization errors in the load and kinematic data. It then distributes the data into appropriate output files for the next stage of the analysis, which is formed by the 3 programmes: "COORSNLW", "COORDNLW", and "TRXFPNLW". Direction cosine matrices and distance vectors of the limb-orthosis complex are calculated by "COORSNLW" and "COORDNLW" for the static kinematic calibration and dynamic walking tests respectively. "TRXFPNLW" analyses the transducer and force-plate loadings, and presents the results both numerically and in the graphical forms. All these results are then used by "BRACENLW" to calculate the loadings on the orthosis and their effects on the anatomical joints (Fig 5.17).

For the LRFW tests, the results are analysed by "TRXLR" and expressed in the respective transducer coordinate systems. If results in the orthotic upright coordinate systems are required, an optional feature in the programme allows the calculated direction cosine matrices and distance vectors from the NLW tests to be transferred to "TRXLR". These D.C. matrices are assumed to be true if the same

orthosis and transducer configuration are used in both the NLW and LRFW tests for the same patient. Numerical and graphical outputs of at least 5 gait cycles are obtained from the analysis.

"KSTRAP" is the programme written for analysis of the knee strap forces of a KAFO. The results are expressed either with respect to the transducers' own coordinate systems or in the orthosis' reference system. Its purpose is mainly for comparison of the calculated results from "BRACENLW" and those actually being measured by the transducers.

CHAPTER 6.

THEORETICAL ANALYSIS.

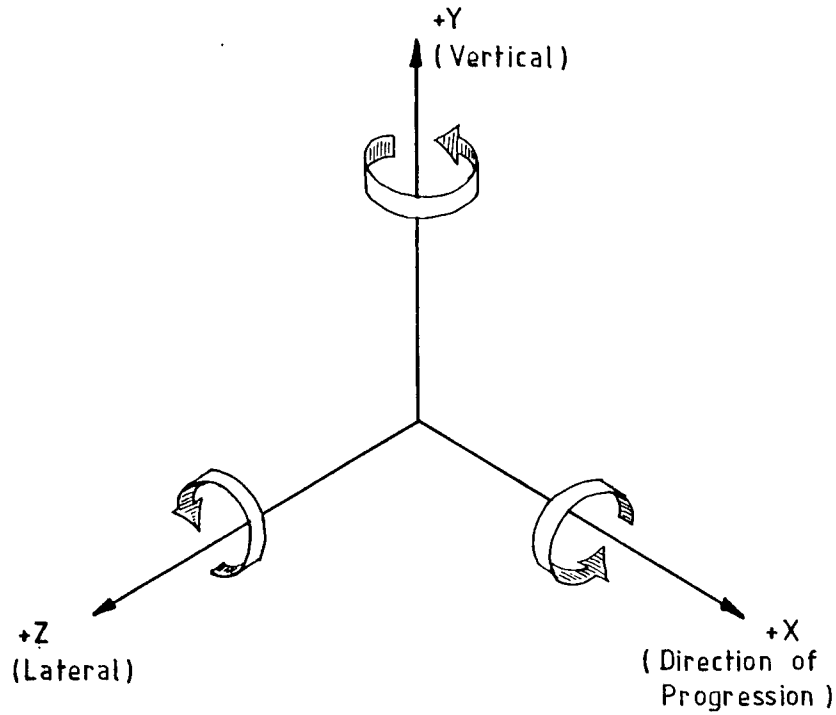


FIG 6·1	The Cartesian Coordinate System.

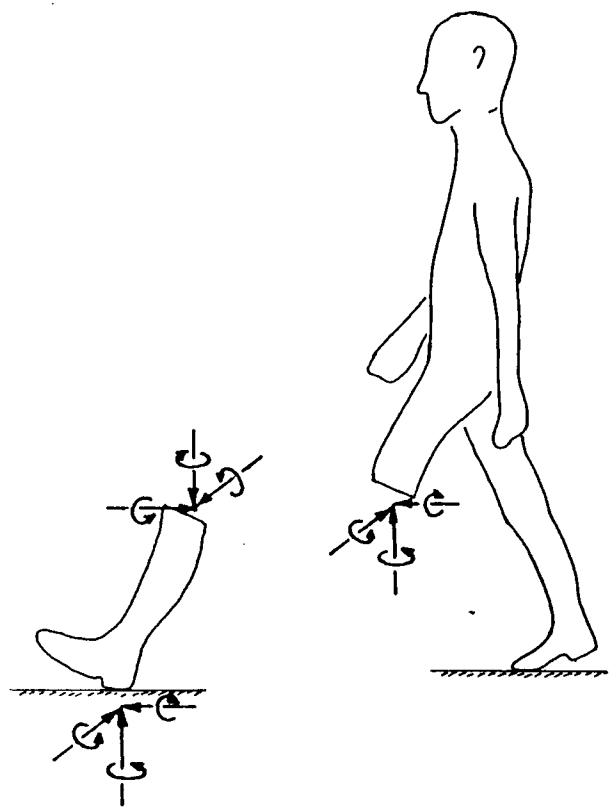


FIG 6·2	Positive Loading on a Body Segment.

6.1 INTRODUCTION.

This chapter discusses the background theories and concepts for the analysis of the experimental data. The various procedures in the processing of the 'raw' data are set-out under their appropriate headings, and in logical form, from the analysis of the spatial to the loading data. A basic knowledge of vector algebra and matrix operations is assumed, and detailed, step-by-step mathematical manipulations are omitted. In the text, only the equations in their final forms are given, and where appropriate, illustrations concerning how particular sections of the analysis are performed will be provided. The assumptions used in the analysis will also be explained under their appropriate headings.

Although the theoretical analysis provides the basic principles for the computation of results, the computer programmes do not follow chronologically the sections set-out in this chapter, but employ only the relevant segments to form the analysis required for that particular programme. Details of the computer programme listings can be found in Appendix 9 (microfiche).

6.2 SIGN CONVENTION AND NOMENCLATURE.

The Cartesian System, using the Right Hand Rule, is used throughout the analysis. The sense of the orthogonal system follows those recommended by a working party on the Standardization of Gait Analysis Parameters (1975), the directions of which are:

X-axis : horizontal and positive in the direction of progression;
Y-axis : vertical and positive in the upward direction;
Z-axis : horizontal and positive from left to right.

For angular measurements, a positive rotation is taken as a clockwise rotation viewed in the positive direction of an axis as according to the Cartesian System. The same argument is applied for both force and moment measurements, where a positive load is taken as one which tends to accelerate the body segment in the positive direction (Fig 6.1).

Applying the above sign convention to the biomechanical analysis of a body segment, a positive force (or moment) is one that is being applied by an external body, such as ground or orthosis, onto the body segment at the distal end (Fig 6.2). Similarly, the positive load

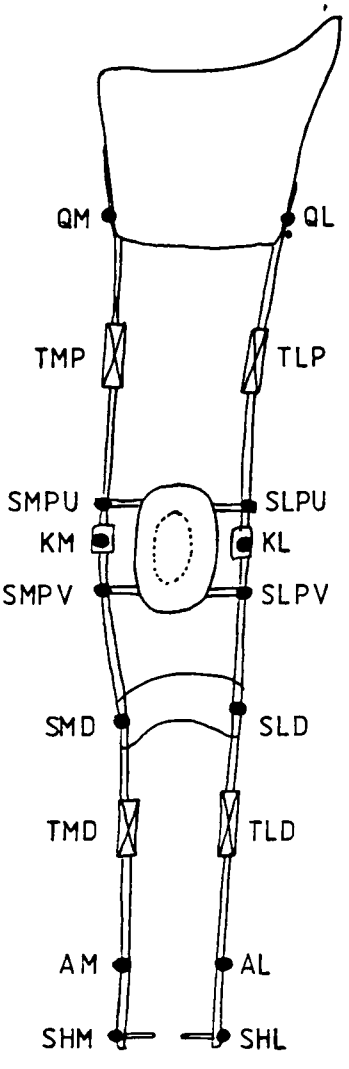


FIG 6·3	Designated Names of Points on Orthosis.

actions recorded by the transducers are assumed to be acting from an external source onto the distal end of the device.

The nomenclature for identification of various points on the instrumented orthosis is mainly based on the medial/lateral and superior/inferior positional relationship of the points to be considered. A transducer (abbreviated to 'T') that is mounted on the orthotic upright nearer to the mid-sagittal plane of the body is a medial (M) transducer while those on the outer upright are classified as lateral (L). For transducers on the same upright, the one that is superiorly placed with respect to the other is the proximal (P) transducer, while the inferiorly located one is the distal (D). Other points on the orthosis are:

Q = thigh band connection point;
K = orthotic knee joint centre;
A = orthotic ankle joint centre;
S = strap attachment point;
SH = shoe attachment point; and
ST = T-strap connection point.

These names are followed by a subscript 'L' or 'M' to denote points on the lateral or medial upright. Some strap attachment points have multiple subscripts (Fig 6.3). For anatomical joints, AC, KC, HC, and IS are used for denoting the ankle, knee, and hip centres, and ischial tuberosity respectively.

The various coordinate systems formed around the body or orthotic segments are represented by:

H = pelvic system;
OL = lateral orthotic upright system;
OM = medial orthotic upright system;
S = shank system;
T = shank triangle system; and
TH = thigh system.

These names are normally prefixed by the letter C to denote direction cosine matrices. Furthermore, the subscripts S or D are also added to the names to differentiate between those obtained from the static calibration and those from the dynamic locomotion data.

Vectors connecting two different points are denoted by combining the names of the two points, for example, HCH1 represents the vector

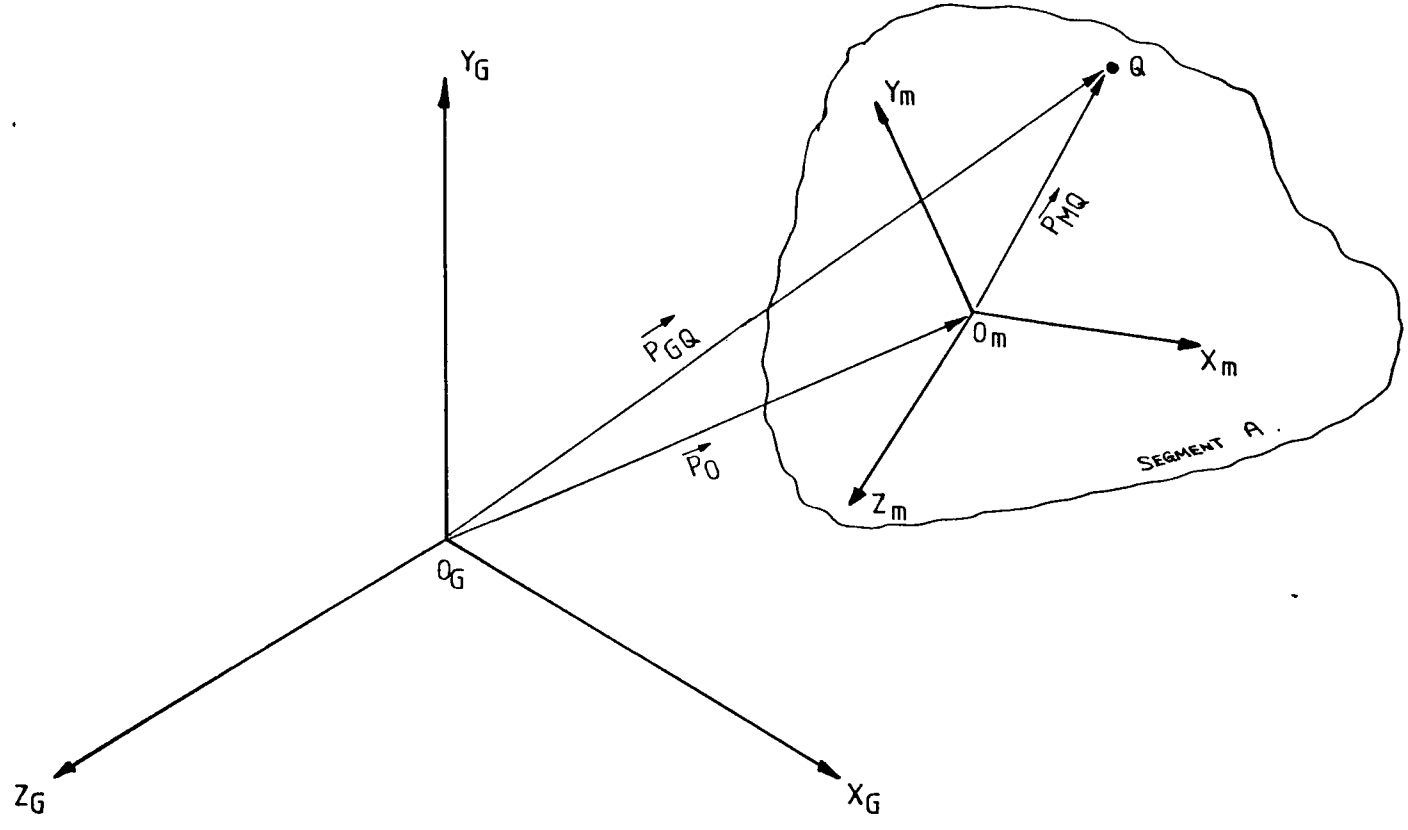


FIG 6·4	Relationships of Two Coordinate Systems.

from the hip centre HC to the hip marker H1.

6.3 SIGNIFICANCE OF COORDINATE SYSTEMS.

Generally, the coordinate systems may be divided into 2 broad classes, i.e. static and dynamic coordinate systems. The static coordinate systems are responsible for relating the orthotic components with the enclosed limb segments, and also to the Ground Reference System (GRS). As the name suggests, the dynamic or moving coordinate systems define the instantaneous positions of the components of the limb-orthosis complex in space. These coordinate systems are determined by the positions of the markers placed on the patient and on the orthosis during a locomotion test.

At various stages of the analysis, it may be necessary to relate data from one coordinate system with one another. This can be achieved by defining both the systems with respect to the common GRS, and by developing an inter-system relationship between the two sets of data. A number of assumptions are therefore required: firstly, the body segments and other orthotic structures are considered as rigid bodies, i.e. the positions of the markers relative to the body segment or orthosis remains the same throughout the test. Secondly, all the anatomical joints of the lower limb are assumed to be simple hinge joints.

Consider a rigid body A suspended in a 3-dimensional space defined by the GRS (Fig 6.4), where its origin with respect to the GRS may be represented by:

$$\overline{P_o} = \begin{bmatrix} x_o \\ y_o \\ z_o \end{bmatrix}$$

The orientation of the rigid body's orthogonal axes system (Xm,Ym,Zm) can be defined relative to the GRS by a direction coside (D.C.) matrix,

$$[B] = [DCMG] = \begin{bmatrix} B_{GXMX} & B_{GXMY} & B_{GXMZ} \\ B_{GYMX} & B_{GYMY} & B_{GYMZ} \\ B_{GZMX} & B_{GZMY} & B_{GZMZ} \end{bmatrix}$$

where the element B_{GiMj} is the cosine of the true angle between the positive directions of the fixed (G) and moving (M) axes. The matrix

also specifies the magnitude and direction of rotation from one frame of reference into the other.

To transform a set of measurands in the moving system to that of the GRS, the measurants are multipled with the DC matrix, thus,

$$\begin{bmatrix} X \\ Y \\ Z \end{bmatrix}_G = [DCMG] \begin{bmatrix} X \\ Y \\ Z \end{bmatrix}_M$$

This can be represented mathematically as:

$$X_G = (X_m * B_{GXMX}) + (Y_m * B_{GXMY}) + (Z_m * B_{GXMZ})$$

$$Y_G = (X_m * B_{GYMX}) + (Y_m * B_{GYMY}) + (Z_m * B_{GYMZ})$$

$$Z_G = (X_m * B_{GZMX}) + (Y_m * B_{GZMY}) + (Z_m * B_{GZMZ})$$

To obtain the coordinates of the moving system from a set of parameters expressed in the GRS, the inverse of the DC matrix is used, i.e.

$$[A]_M = [DCMG]^{-1} [A]_G$$

Because of its orthogonal properties, the inverse of the DC matrix is also equal to its transpose arrangement, hence,

$$[DCMG]^{-1} = \begin{bmatrix} B_{GXMX} & B_{GYMX} & B_{GZMX} \\ B_{GXMY} & B_{GYMY} & B_{GZMY} \\ B_{GXMZ} & B_{GYMZ} & B_{GZMZ} \end{bmatrix}$$

Therefore, by the appropriate multiplication of the DC matrices, it is possible to interrelate 2 moving systems through the GRS. An illustration is given below for the transformation of quantities from a moving system M to another moving system N.

$$[A]_G = [DCMG] [A]_M$$

$$[A]_N = [DCNG]^{-1} [A]_G$$

Combining the above,

$$[A]_N = [DCNG]^{-1} [DCMG] [A]_M$$

Referring to Fig 6.4, the coordinate of [C] of an arbitrary point Q on the moving body A may be defined by the position vector $\overline{P_{MQ}}$ in the moving reference system, and $\overline{P_{GQ}}$ in the GRS. That is:

$$\overline{P_{MQ}} = [C]_M$$

$$\overline{P_{GQ}} = [C]_G$$

Therefore the changing components of the vector $\overline{P_{GQ}}$ may be related by,

$$\overline{P_{GQ}} = \overline{P_o} + [DCMG]*\overline{P_{MQ}}$$

i.e. $$[C]_G = [C]_o + [DCMG]*[C]_M$$

where $[C]_o$ is the position vector of the origin of the moving system with respect to the GRS.

6.4 SPATIAL DATA RATIONALISATION.

The spatial data derived from the cine film contain a number of errors, and they may be grouped under the following headings:

1. displacement of GRS origin;
2. scaling (or magnification) and parallax errors;
3. digitisation errors;
4. lens distortion errors;
5. film distortion in processing;
6. phase difference in the multi-camera system; and
7. random errors.

The effects of lens and film distortion were determined experimentally by projecting the image of a calibration grid onto the digitising tablet. It was found that the errors were not significant when the digitised points are within the inner three-quarters of the total image area. Therefore, by ensuring that the test subject remains within the inner $\frac{3}{4}$ area of the camera's field of vision, these errors may be ignored.

Errors due to phase difference are caused by the unsynchronised shutter opening times of the multi-camera system. This is inevitable since there are no mechanical linkages between the synchronous cameras but through the 50 Hz mains supply. The maximum phase-difference will be less than one frame, i.e. 20 milliseconds, and this is considered to be non-critical since rapid movements are usually absent in the patient's walking cycles. However, linear interpolation may be used to synchronise the data, as used by Ishai (1975) and Goh (1982).

Other sources of errors include those due to 'random noise' in

the camera and digitisation systems. These are errors superimposed onto the displacement data, and are generally caused by imperfection of the sprockets and driving gears of the cameras and projectors, mains frequency variations, imperfections in the cine films, and other spontaneous system errors. Although this is a very serious source of error if the displacement data were to be double-differentiated for the derivation of acceleration and forces, the digital filtering process is not applied to the data in the analysis since only displacement information is needed.

6.4.1 Origin of the Ground Reference System.

The coordinates of points obtained from the digitiser are not referred to the ground reference system (GRS), but to the 'origin' of the digitising tablet. It is, therefore, necessary to redefine the coordinate systems to those of the GRS.

Throughout the test, at least one static ground marker, designated here as SM2, can be constantly seen by both the front and side cameras. Since the relationship between marker SM2 and the origin of the GRS is known from the grid board calibration test, all other digitised points can be subsequently transferred to the GRS through SM2. In addition, the digitisation linear displacement errors (see Section 6.4.4) are also being eliminated by the use of marker SM2.

From the grid board calibration test, the relationship between the GRS origin (PPC) and the static marker SM2C is:

$$[PPSM2C] = [SM2C] - [PPC]$$

where the subscript 'C' denotes coordinates obtained from the calibration test, and in the digitiser's frame of reference. Assuming that the vector PPSM2C remains the same throughout the course of the test, the GRS origin can then be defined by,

$$[PP] = [SM2] - [PPSM2C]$$

where SM2 being the coordinates of the static marker recorded in that particular frame of data being analysed. Therefore, the coordinates of a digitised point QQ can be defined in the GRS by,

$$[QQ]_{GRS} = [QQ]_{Digitiser} - [PP]$$

6.4.2 Scaling Correction.

Since the size of the image projected onto the digitising tablet represents only a proportion of the actual size of the test subject, it is important to convert all these apparent sizes onto their true dimensions. The method employed is by the calibration of 3 markers, with known inter-marker distances, on the grid board. The scaling factor (SF) can be defined as:

$$SF = \frac{\text{True Size}}{\text{Apparent Size}}$$

$$= \frac{\text{True distance between calibration markers}}{\text{Apparent distance between the same markers}}$$

The true coordinates of QQ is therefore,

$$[QQ]_{true} = [QQ]_{apparent} * SF$$

By incorporating the correction for GRS origin into the above equation,

$$[QQ]_{true,GRS} = \left[\frac{QQ - PP}{BB - PP}\right]_{digitiser} * SF$$

where BB is the scaling calibration marker.

6.4.3 Parallax Correction and Hidden Marker Determination.

Parralax error is caused by the perspective effect in the cine-photography, and it increases with:

1. decreasing camera-to-object distance;
2. increasing object-to-reference plane distance; and
3. increasing deviation of object from the camera axis.

Therefore, ideally, the camera should be at an infinite distance away from the object, with the object lying on the reference plane immediately in front of the camera. However, in the laboratory conditions, the cameras were placed at a distance of between 4.0 and 9.5 metres away from the force-plate geometric centre.

As a result of the orthogonal arrangement of cameras, and the placement of markers, some of the markers may be obscured from the view of either the front or the side camera. The hidden marker coordinate determination and the parallax error correction can be carried out

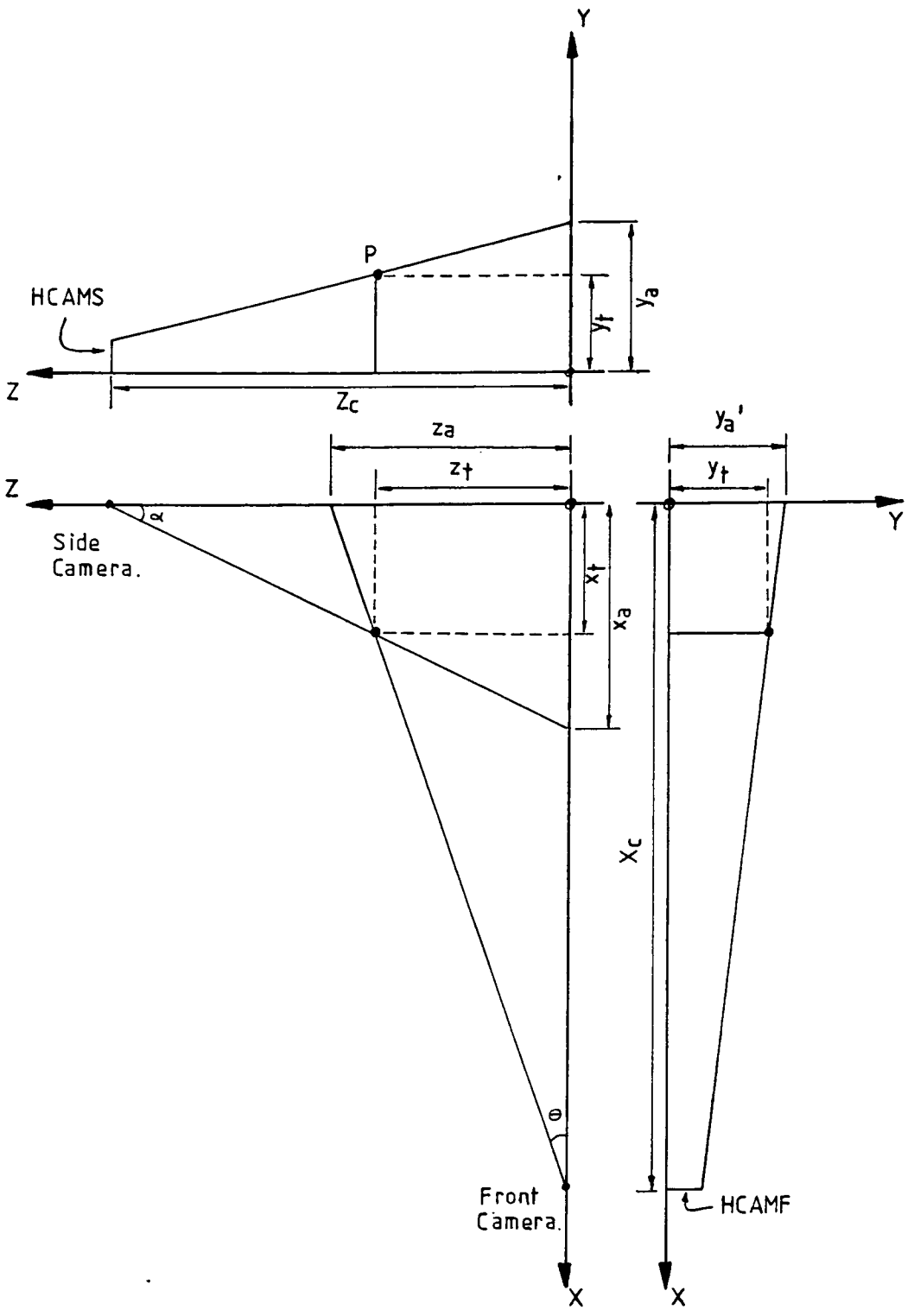


FIG 6·5	Parallax Errors.

simultaneously. The three possible cases are:

1. markers seen by both cameras (ALL-VISIBLE);
2. markers seen only by front camera (FRONT-VISIBLE); and
3. markers seen only by side camera (SIDE-VISIBLE).

6.4.3.1 ALL-VISIBLE case: Referring to Fig 6.5, the following parallax equations can be derived from similar triangle theorems:

$$x_t = [\, x_a(Zc - z_t)]/[K * Zc] \quad \cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots (6.1)$$

$$z_t = [\, z_a(Xc - x_t)]/Xc \quad \cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots\cdots (6.2)$$

$$y_t = y_a + [\, z_t\ (HCAMS - y_a)]/[K * Zc] \quad \cdots\cdots\cdots\cdots\cdots (6.3)$$

$$y_t = y_a' + [\, x_t\ (HCAMF - y_a')]/Xc \quad \cdots\cdots\cdots\cdots\cdots\cdots (6.4)$$

where:

x,y,z = coordinates of the marker under consideration;
Xc,Zc = distances of front and side camera from GRS origin;
HCAMS = height of side camera from ground level;
HCAMF = height of front camera from ground level;
t = subscript denoting true coordinates;
a = subscript denoting apparent coordinates;
y_a' = apparent y coordinates of a marker as recorded by the front camera;
K = 'side' correction factor: +1 for right side, -1 for the left.

From equations 6.1, 6.2, and 6.3, the following parallax error correcting equations are obtained:

$$x_t = [(Xc * x_a)(z_a - K*Zc)]/[(x_a * z_a) - (K*Zc * Xc)]$$

$$y_t = y_a + [\, z_t\ (HCAMS - y_a)]/[\, K*Zc]$$

$$z_t = [(K*Zc * z_a)(x_a - Xc)]/[(x_a * z_a) - (K*Zc * Xc)]$$

6.4.3.2 FRONT-VISIBLE case: During a test, markers positioned on the other side of the patient's body away from the camera will be obscured from the camera's view (Fig 6.6). To determined the true coordinates of these hidden markers, the following additional information are needed:

1. the true coordinates (x_m, y_m, z_m) of an adjacent marker in the same rigid body, and
2. the true distance (MAG) between the hidden marker and the selected adjacent marker.

The relationship between the distance MAG and the coordinates of the two markers can be expressed as,

$$MAG = SQRT\ [(x_t-x_m)^2 + (y_t-y_m)^2 + (z_t-z_m)^2] \quad \cdots\cdots\cdots (6.5)$$

By substituting equations 6.2 and 6.4 into 6.5, a quadratic solution for the missing x-coordinates for x_t can be derived:

$$x_t = -B/A \pm SQRT[(B/A)^2 - (C/A)]$$

$$y_t = y_a + [\ x_t\ (HCAMF - y_a)]/Xc$$

$$z_t = [z_a(Xc-x_t)]/Xc$$

where:

$$A = 1 + [z_a^2 + (HCAMF-y_a)^2]/Xc^2$$

$$B = -x_m + [(HCAMF-y_a)(y_a-y_m) + z_a(z_m-z_a)]/x_m$$

$$C = x_m^2 + (z_m-z_a)^2 + (y_m-y_a)^2 - MAG^2$$

The relevant results for x_t is selected by geometrical consideration between the hidden and reference markers.

6.4.3.3 SIDE-VISIBLE case: The solution for marker coordinates hidden from the side camera are obtained by solving the equations 6.1, 6.3, and 6.5 :

$$z_t = -B/A \pm SQRT[(B/A)^2 - (C/A)]$$

$$y_t = y_a + [z_t(HCAMS-y_a)]/(K*Zc)$$

$$x_t = [x_a(K*Zc - z_t)]/(K*Zc)$$

where:

$$A = 1 = [x_a^2 + (HCAMS - y_a)^2]/Zc^2$$

$$B = -z_m + [(HCAMS-y_a)(y_a-y_m) + x_a(x_m-x_a)]/z_m$$

$$C = z_m^2 + (x_m-x_a)^2 + (y_m-y_a)^2 - MAG^2$$

Again, the relevant solution is obtained by geometrical considerations of the markers involved.

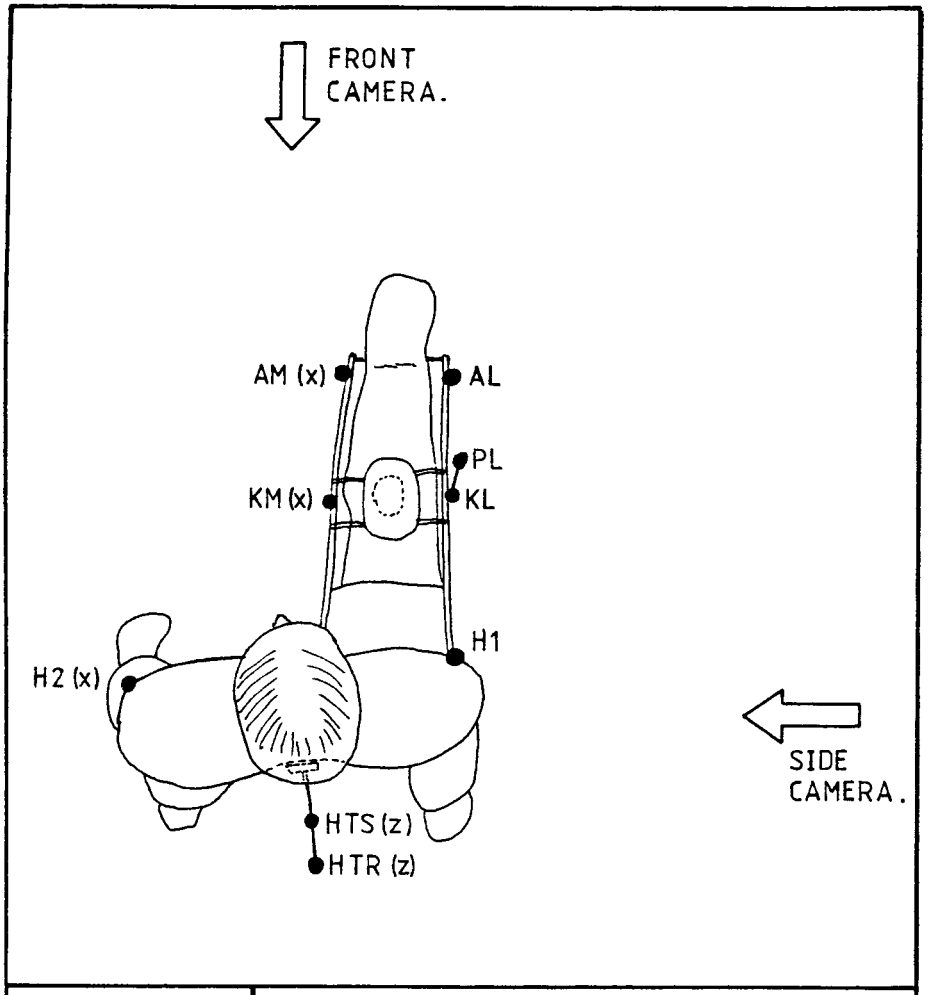


FIG 6·6 The Hidden Markers.

6.4.3.4 Selection of reference markers: To obtain a reliable and accurate solution for the missing coordinates, the reference marker should be carefully selected by considering its geometrical relationship with the hidden marker. One of the main selection criteria being that the component of the coordinates under review should always bear a constant relationship between the 2 markers throughout the stance phase.

The markers hidden from the front camera (i.e. z-coordinate missing) are the two tail markers HTR and HTS, while the off-side hip marker H2 and the knee and ankle markers on the medial orthotic upright, i.e. KM and AM, have missing x-coordinates (Fig 6.6).

The near-side hip marker H1 is used as the reference for calculating missing HTR(z) and HTS(z), since H1(z) is assumed to be always more positive or more negative than HTR(z) and HTS(z), depending on which side camera is being used in the test. A quick literature search (University of California,1947; Chapman & Kurokawa,1969) have shown that the maximum pelvic rotation of a normal subject is 15° in the transverse plane. This indicates that the assumption will be valid even if the maximum pelvic rotation of an orthosis wearer reaches twice that of the normal.

A similar argument is used for markers hidden from the side camera, where the reference marker HTT and PL are employed for the determination of H2(x) and KM(x) respectively. However, the solution for missing AM(x) is slightly more complex because there is a lack of markers that will satisfy both the conditions of having a constant distance from AM and at the same time possesses a fixed relative position with AM throughout the stance phase of the gait cycle. The solution adopted is to use the marker KL and AL on the lateral upright to determine the position of AM relative to the reference marker KM, assuming that the orthosis experiences no excessive torsional deformation and the uprights are parallel to each other.

A summary of the solution for the hidden marker is tabulated in Table 6.1 .

6.4.4 Digitisation Errors.

The errors associated with digitisation of spatial data are the 'rounding-off' or quantisation errors, the errors due to malalignment

Table 6.1 The Relevant Solutions for Hidden Markers.

CASE	CAMERA SET-UP	MISSING COORDINATE	RELEVANT * SOLUTION	REFERENCE MARKER	CONDITIONS
SIDE-VISIBLE	Front/Right	HTR(z) & HTS(z)	-M-SQRT(N)	H1	HTR(z) & HTS(z) < H1(z)
	Front/Left		-M+SQRT(N)	H1	HTR(z) & HTS(z) > H1(z)
FRONT-VISIBLE	Front/Right or Front/Left	H2(x)	-M+SQRT(N)	HTT	H2(x) > HTT(x)
		KM(x)	-M-SQRT(N)	PL	KM(x) < PL(x)
		AM(x)	-M+SQRT(N)	KM	KL(x) < AL(x)
			-M-SQRT(N)	KM	KL(x) > AL(x)
			AM(x)=KM(x)	KM	KL(x) = AL(x)

* $M = B/A$
$N = (B/A)^2-(C/A)$

of the projected image with the digitiser, and the human errors. Quantisation error is considered to be insignificant because the Calcomp 9000 digitiser used has a manufacturer's claimed resolution of 1000 lines per inch and an accuracy of ±0.0105 inch (Calcomp,1982).

The principal error is caused by the malalignment of the projected image with the digitising tablet, although care is taken to ensure proper alignment before the digitisation process. This is due to the fact that the projected image does not always fall squarely onto the same position at the digitising surface when the film is being advanced from frame to frame. To assess the magnitude of this error, a section of cine film containing the image of a calibration grid board was advanced through 150 consecutive frames, and the projected image traced onto a paper fixed to the tablet. Three types of uncertainties were discovered, namely, 2 linear translational errors along the horizontal and vertical axes of the digitiser, and a rotational error. The maximum errors were found to be less than 5 mm and 1° for the linear and rotational uncertainties respectively. Elimination of the linear errors is performed through the static marker SM2 when the digitiser data are being defined relative to the GRS (see Section 6.4.1). The rotational error is considered to be non-critical for the analysis.

6.5 ANALYSIS OF SPATIAL DATA.

This section discusses the formation of coordinate systems and distance vectors of the body segments and orthotic structures. The analysis can be generally divided into 2 areas, firstly, those coordinate systems derived from the static calibration, and secondly, those from the dynamic locomotion situations. Positional relationships within the limb-orthosis complex and the instantaneous orientation of the various systems can thus be obtained.

In the analysis, 3 basic assumptions are employed:

1. the orthosis experiences insignificant deformations before and after it has been applied to the patient;
2. there is insignificant orthotic deformation during walking or other ambulatory activities; and
3. the body segments and orthotic structures are considered to be rigid.

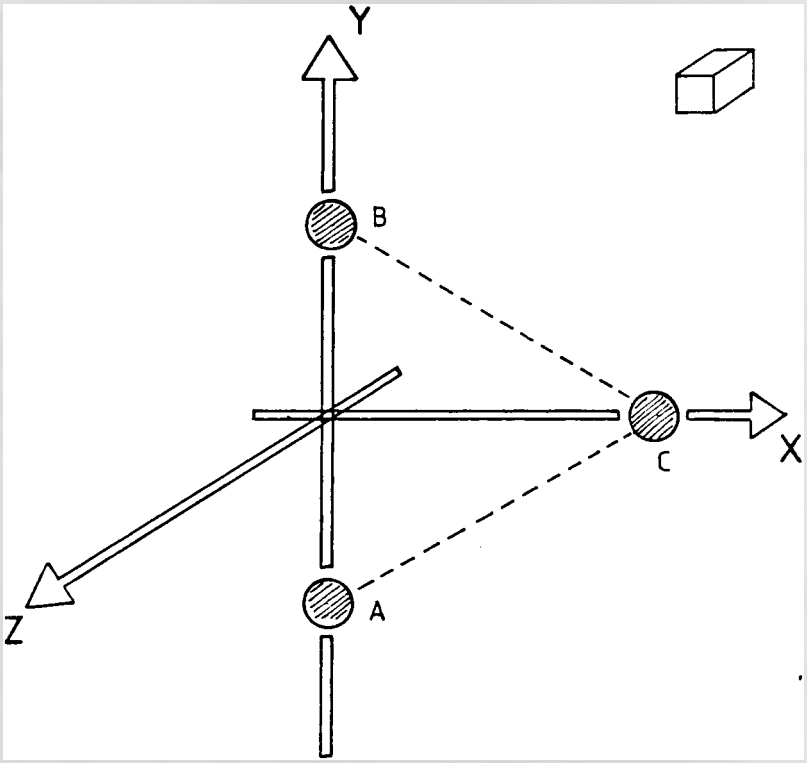


FIG 6·7	Formation of Coordinate System from Three Markers.

6.5.1 Static Coordinate Systems.

A knowledge of the positional relationships between the different components of the limb-orthosis complex is required for the transference of data from one segment of the complex to another. The coordinate systems are defined with respect to the GRS:

1. DC matrices of the transducers (CTLP,CTLD,CTMP,CTMD);
2. DC matrices of the orthotic uprights (COLS,COMS);
3. DC matrix of the shank triangle system (CTS); and
4. DC matrix of the shank (CSS).

In all but the shank system, 3 markers arranged in a triangle are used to define the directions of the axes. The markers are arranged in such a way that a line joining the centres of markers A and B will define the Y-axis (Fig 6.7). A line perpendicular to the Y-axis and passes through the marker C determines the direction of the X-axis, while the Z-axis is normal to the plane formed by the markers A, B, and C. The DC matrix thus obtained is expressed with respect to the GRS since the coordinates of the markers were measured in the ground system, and its formation is illustrated below.

Vector along the Z-axis,

$$Z(I) = AB(I) \wedge AC(I)$$

where AB and AC are the vectors joining the markers A,B and A,C respectively. The symbol '$\wedge$' denotes the vector cross-product while (I) may be 1, 2, or 3 to represent the x, y, and z coordinates respectively. The direction cosines of the axes are therefore:

Y-axis, $$CMT(I,2) = \frac{AB(I)}{MAG(AB)}$$

Z-axis, $$CMT(I,3) = \frac{Z(I)}{MAG(Z)}$$

X-axis, $$CMT(I,1) = CMT(I,2) \wedge CMT(I,3)$$

where,

CMT = DC matrix of the marker triangle,

MAG(x) = magnitude of vector x.

The formation of the individual transducer coordinates system is a direct application of the method described above, with the X-axis pointing towards the front camera while the Y-axis is directed

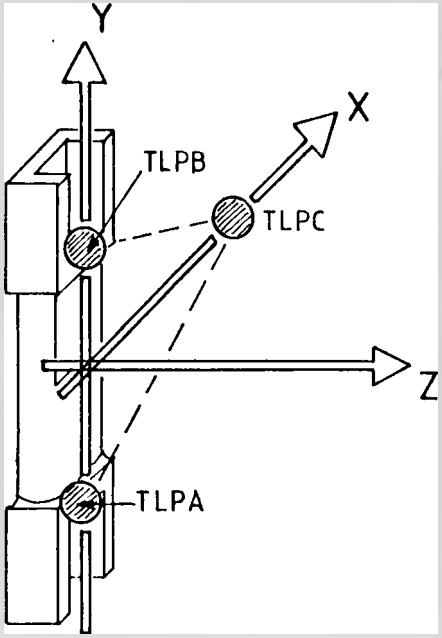


FIG 6·8	Coordinate System for Transducer TLP.

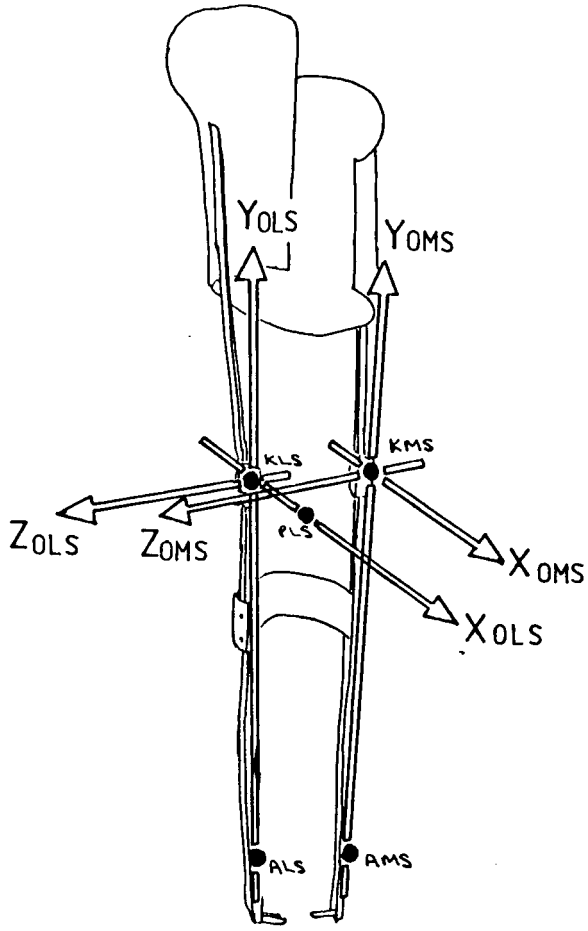


FIG 6·9	Coordinate Systems of Orthotic Uprights.

proximally (Fig 6.8). However, for the orthotic uprights, the situation is slightly more complex, and attempts to define the double uprights with a single coordinate system can lead to numerous practical difficulties. Therefore, two coordinate systems are used, one for each upright, with the medial system possessing a certain degree of dependency on the lateral system. The markers KLS, ALS, and PLS are employed in the formation of the lateral upright coordinate system (COLS), with its Y-axis (Y_{OLS}) defined by ALS and KLS (Fig 6.9). For the medial upright, only two markers are available to define its coordinate system (COMS). These two markers (KMS, AMS) define the Y-axis (Y_{OMS}), with its Z-axis (Z_{OMS}) taken to be perpendicular to a plane formed by Y_{OMS} and the X-axis of the lateral upright (X_{OLS}).

Two coordinate systems are employed to define the 3-dimensional position of shank in space. They are the shank triangle system (CTS) and the shank system (CSS). The CSS system provides information of the mechanical axis of the shank between the knee and the ankle joint centres with respect to the GRS, while the CTS system relates the positions of 3 markers attached to the tibia and fibula to the GRS. This is necessary because in the dynamic walking tests, the anatomical knee and ankle joint centre markers are removed to avoid interfering with the motion of the orthotic uprights, and only the instantaneous positions of the CTS markers are defined.

The shank triangle system is specified by the triangle comprising markers BSS, BIS, and BMS, as shown in Fig 6.10 . However, for the shank system, only the anatomical knee and ankle joint centres are specified by the markers KC1, KC2, and AC1, AC2 respectively. This defines the Y-axis (Y_{SS}), with the Z-axis (Z_{SS}) perpendicular to the plane formed by the Y-axis of this system and the X-axis of the GRS.

At the anatomical joint centres, 2 markers are employed to locate a single point. The markers laterally offset from the joint centre (AC1 & KC1) define the x and y coordinates while the z coordinate is derived from the anteriorly offset markers (AC2 & KC2). Errors associated with such a system are considered to be insignificant because the offset is very much smaller than the magnitude of the camera distances Xc and Zc. Another possible source of error is due to the rotation of the leg about its long axis, and this can be reduced by asking the subject to stand, facing the front camera, in a comfortable posture during the static calibration test.

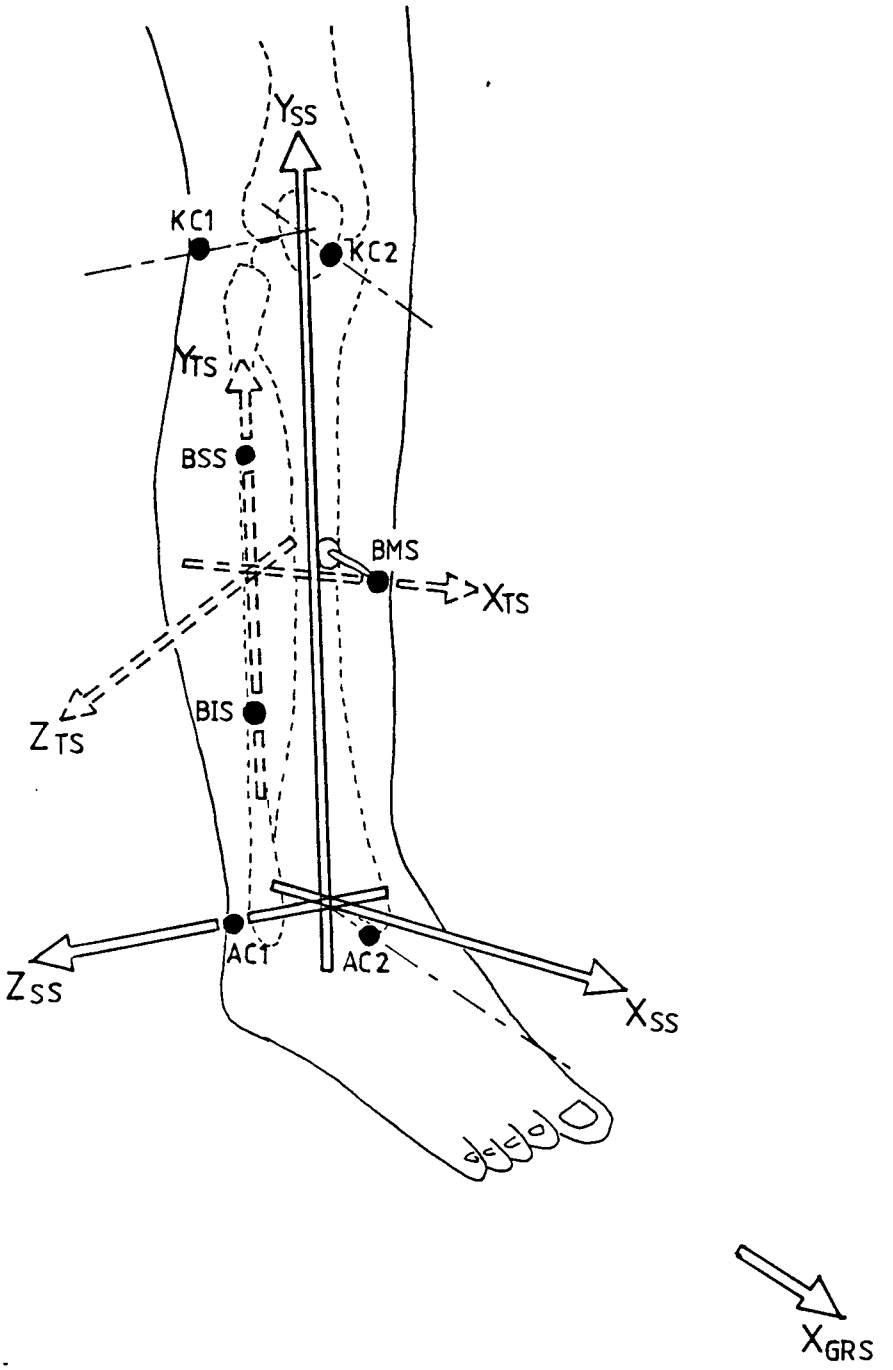


FIG 6·10	The Shank and Shank Triangle Coordinate Systems.

A number of distance vectors between important points on the limb-orthosis complex are also obtained from the static calibration spatial data. These include the vectors:

1. between strap connection points;
2. strap connection points to transducer centres;
3. orthotic knee and ankle joint centres to transducer centres;
4. quadrilateral socket connection points to transducer centres; and
5. orthotic knee joint centres to strap connection points.

The 'transducer centre' or 'transducer origin' (see Section 4.4.2.2) is denoted by point D, relates to marker 'A' of the transducer marker triangle through the distances MAGAM and MAGMD (Fig 6.11). The orthotic joint centre markers are offset from the actual mechanical joint centres by the amount represented by the vector MC.

In the analysis, the vectors required are obtained through simple vector algebra, using those with known values to deduce the unknown, and taking into consideration the different reference frames of the vectors. An example of the operation is illustrated below, where the vector under consideration is KLTLP, i.e. the vector from the lateral orthotic knee centre to the lateral proximal transducer centre (Fig 6.12).

$$[TLDTLPA]_G = [TLPA]_G - [TLDA]_G$$

$$[TLDTLPA] = [COLS]^{-1} * [TLDTLPA]_G$$

where [TLDTLPA] is the vector from marker 'A' of transducer TLD to marker 'A' of transducer TLP, in the lateral upright coordinate system;

[TLDA], [TLPA] are the position vectors of marker 'A' of transducers TLD and TLP respectively;

G is the subscript denoting GRS;

$[COLS]^{-1}$ is the inverse DC matrix of the lateral upright.

Also,

$$[TLDAKLS]_G = [KLS]_G - [TLDA]_G$$

$$[TLDAKLS] = [COLS]^{-1} * [TLDAKLS]_G$$

where [TLDAKLS] is the vector from marker A of transducer TLD to the lateral knee marker, and [KLS] being the position vector of the lateral

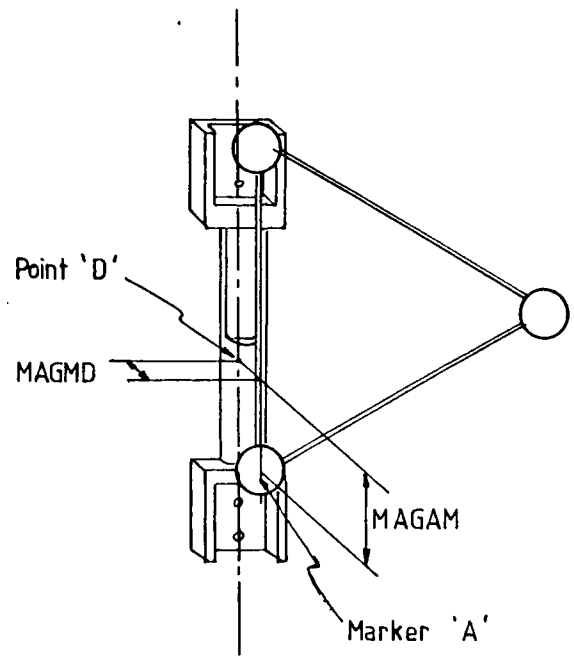


FIG 6·11	Position of Transducer Centre.

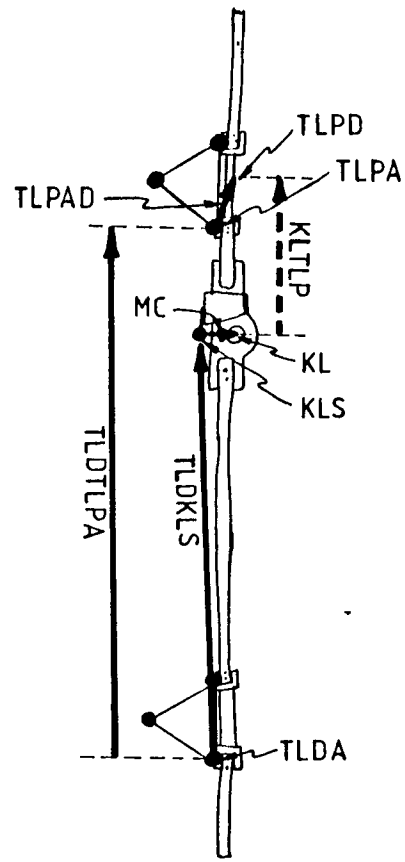


FIG 6·12	Derivation of Vector KLTLP.

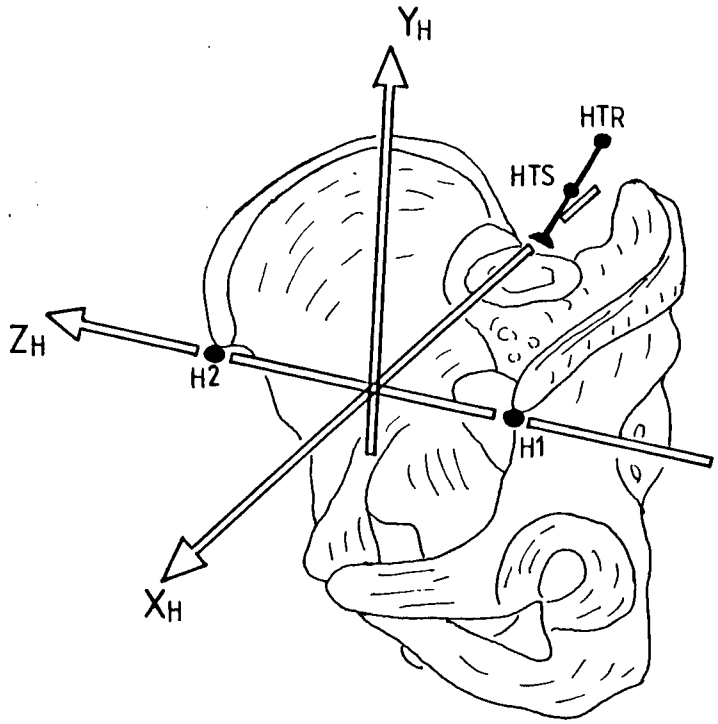


FIG 6·13	The Pelvic Coordinate System.

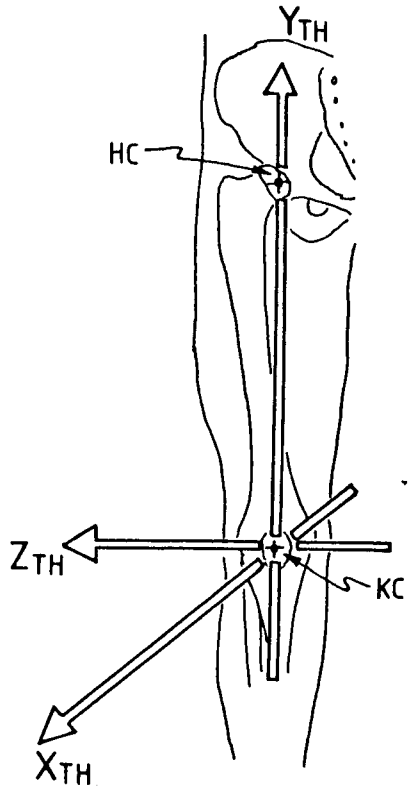


FIG 6·14	The Thigh Coordinate System.

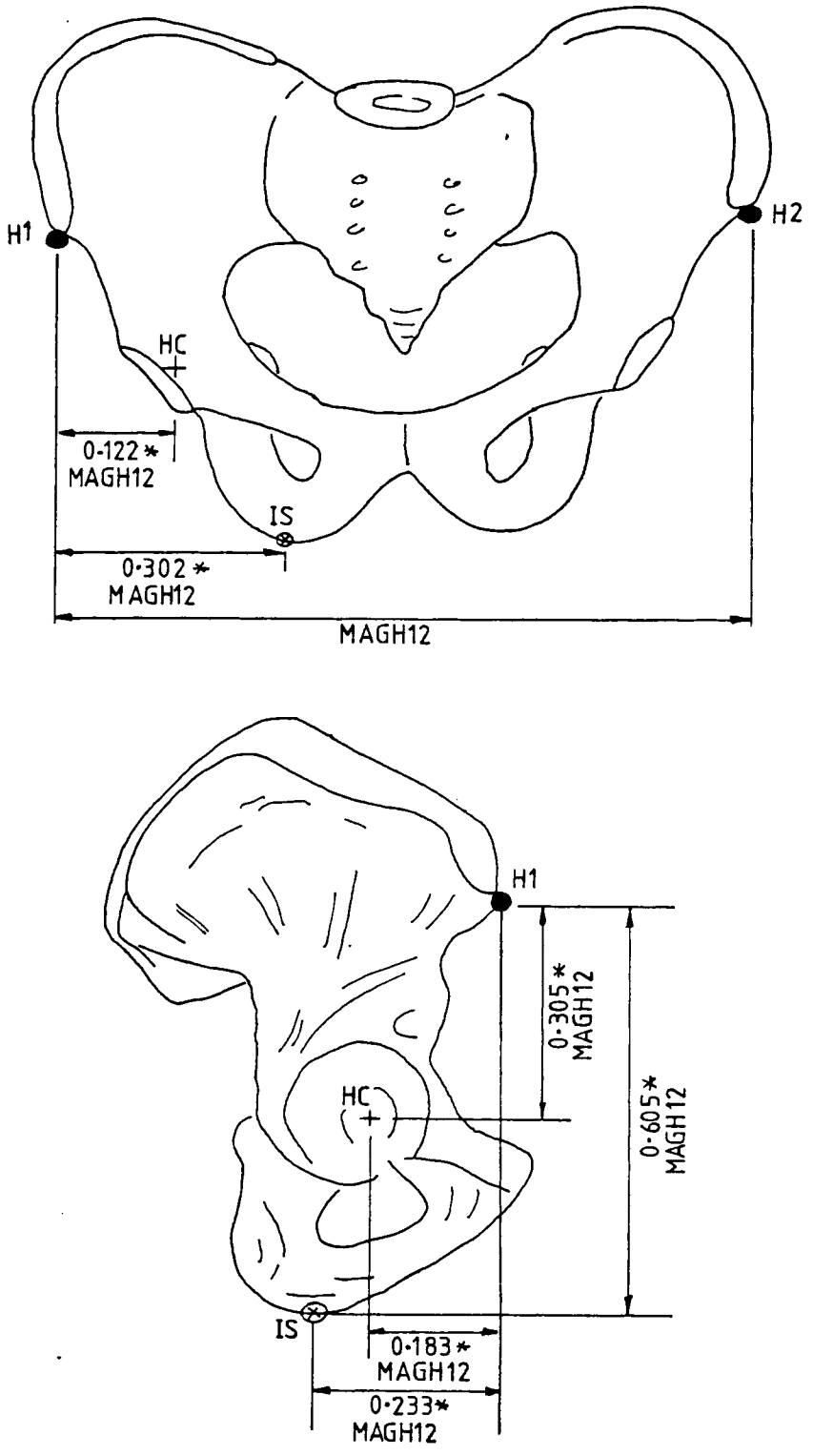


FIG 6·15	The Hip Scaling Factors.

knee marker. Therefore,

$$[KLTLP] = [TLPAD] - [TLDAKLS] + [TLDTLPA] - [MC]$$

where [TLPAD] is the vector joining the transducer marker A to the transducer centre of TLP.

6.5.2 Dynamic Coordinate Systems.

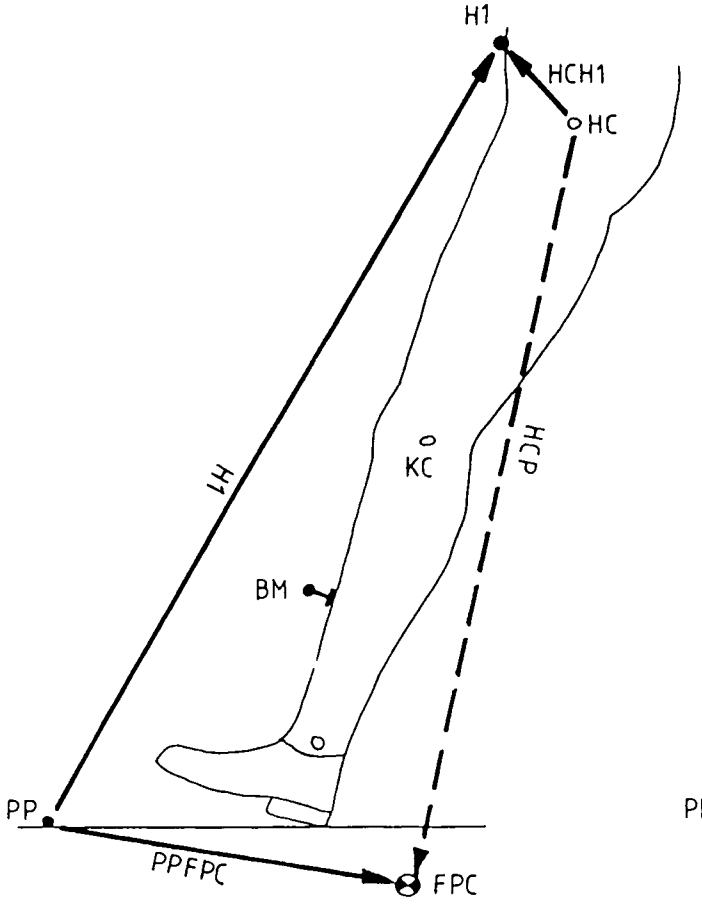
This section discusses the formation of direction vectors and instantaneous coordinate systems of the limb-orthosis complex. These are defined in every frame of the sampled data, i.e. at an interval of 20 milliseconds (50 Hz). The DC matrices calculated are:

1. orthotic upright systems (COLD,COMD);
2. shank triangle system (CTD); and
3. pelvic and thigh systems (CH,CTH).

The direction cosine matrices of the lateral and medial orthotic uprights, and that of the shank triangle system, denoted by COLD, COMD, and CTD respectively, are defined in a similar manner as their static calibration counterparts (i.e. COLS, COMS, and CTS). The subscripts 'S' and 'D' identify the coordinate systems' nature or origin.

For the pelvic girdle, its direction cosine is determined by the 2 anterior superior iliac spine (ASIS) markers H1 and H2, and the base of the sacral marker HTT (Fig 6.13). The markers H1 and H2 determine the Z-axis (X_H) of the system, while the direction of its X-axis (X_H) passes through HTT, in the plane formed by the 3 markers and perpendicular to Z_H. A line normal to both X_H and Z_H forms the Y-axis, Y_H. It should be noted that a certain degree of uncertainty may have been introduced into the formation of this axis system because 2 of its 3 markers were hidden from the view of the cameras.

The formation of the thigh system (CTH) is rather complex because neither the anatomical hip joint centre nor the knee joint centre are defined by markers (Fig 6.14). Scaling factors are therefore used to locate the anatomical hip joint centre from measurements taken on the test subject. Harrington (1974) reported that from measurements of a skeleton and X-rays of normal subjects, the following relationships between the hip joint centre (HC) and the ASIS (H1) may be assumed for an average subject (i.e. average values for both male and female subjects):



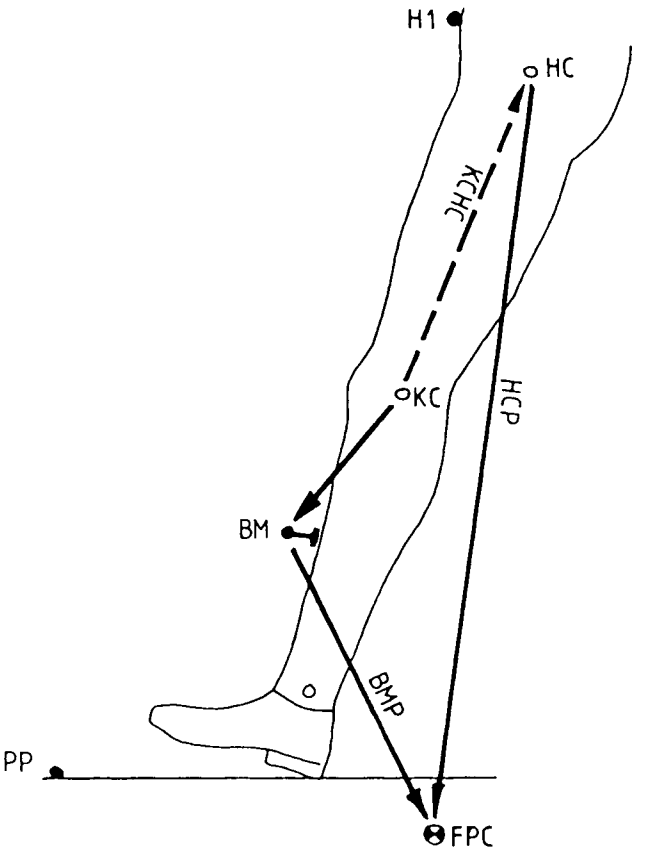


FIG 6·16 | Formation of Vector KCHC.

HCH1 (x) = 0.183 * MAGH12

HCH1 (y) = 0.305 * MAGH12

HCH1 (z) = 0.122 * MAGH12 * ISIDE

where MAGH12 is the measured distance between markers H1 and H2 (Fig 6.15). The coordinates of HC can then be calculated from the vectors HCH1 and H1.

The instantaneous location of the knee joint centre (KC) is also derived from the spatial relationship of other markers with the knee centre. The raised shank marker BM acts as an intermediate point relating the knee centre to the force plate centre (FPC). If the hip centre is also referred vectorially to FPC, the vector KCHC which forms the Y-component of the thigh coordinate system can be calculated, as shown in Fig 6.16. Therefore,

[HCP] = [HCH1]-[H1]-[PPFPC]

[KCHC] = [KCBM]+[BMP]-[HCP]

Y_{TH} = [KCHC]/MAGH12

where [PPFPC] is a fixed vector, from the GRS origin to FPC;

[H1] is the position vector of marker H1; and

[KCBM] is the vector from KC to BM, obtained from static calibration test.

Its Z-axis (Z_{TH}) is perpendicular to the plane formed by Y_{TH} and X-axis of the shank system, X_{SS}. The X-axis is orthogonal to both Y_{TH} and Z_{TH} (Fig 6.14).

The ischial load bearing point is another landmark that requires scaling factors to approximate its location. The scaling factors utilised were taken from the measurement by Trappitt (1979) on a skeleton and 6 normal subjects (3 males and 3 females). The average values of these factors are:

ISH1 (x) = 0.233 * MAGH12

ISH1 (y) = 0.605 * MAGH12

ISH1 (z) = 0.302 * MAGH12 * ISIDE.

Vectors relating the anatomical joint centres to the corresponding points on the orthosis are also being calculated. When performing the vector algebra, it is important to ensure that all vectors are referred to the same coordinate system before they are being added to

or substracted from one another. For example, to convert the vector BMP from GRS to the shank system, it has to be triple multipled with the inverse of dynamic shank triangle system ($[CTD]^{-1}$), the static shank triangle system ([CTS]), and then the inverse of the shank system ($[CSS]^{-1}$),

$$\text{i.e. } [BMPB] = [BMP]*[CTD]^{-1}*[CTS]*[CSS]^{-1}$$

where [BMPB] is vector BMP in the shank system.

6.6 FORCE-PLATE AND TRANSDUCER DATA.

The ground reaction and orthotic loadings were sampled at a frequency of 50Hz, and the acquired analogue signals were subsequently converted into the digital form by the data acquisition computer. In the digital form, a numerical number of 4096, corresponding to 2^{12} is allocated for an analogue signal of +1.0 V, while the minimum allowable negative voltage of -1.0 V is represented by the number 0000. For a zero voltage input, the number 2048 (i.e. 2^{11}) will be registered.

These digital numerical number have to be converted back into the force and moment units of newtons and newton-metres respectively. This is achieved by multiplying the magnitude of the registered number with an appropriate calibration factor. For the force-plate data, ground-to-body loadings are obtained from:

$$\text{Force (N)} = \frac{(N_i - N_o) * CA * I}{204.03 * BMF}$$

$$\text{Moment (Nm)} = \frac{(N_i - N_o) * CA * I * 0.264}{204.03 * BMF}$$

where N_i = output of frame i from computer;
N_o = output of final frame of data;
CA = charge amplifier setting;
BMF = buffer amplifier setting; and
I = sign correction factor, the value of which depends on the force-plate used and also on the measurement channel.

In the above equations, the value of N_o is used as a datum instead of the value 2048. This in effect eliminates any errors due to the offsets in the force-plate channels.

For the KAFO transducer data, the following equation is used:

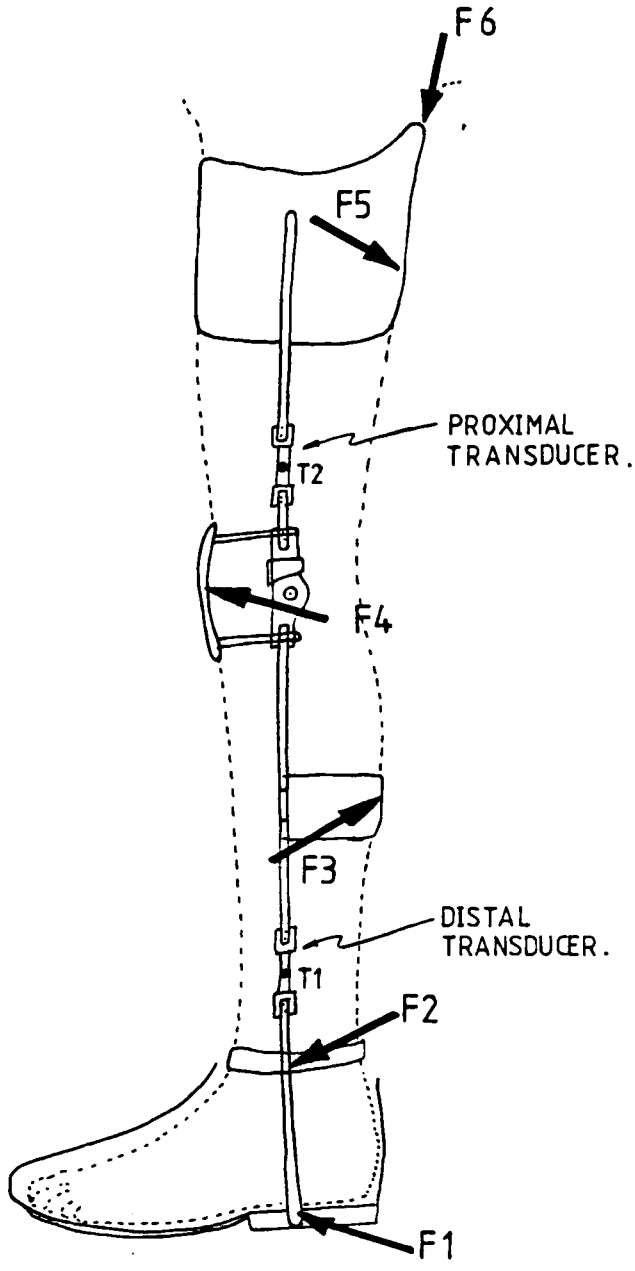


FIG 6·17	External Loads Experienced by a KAFO.

Force (N) or moment (Nm) of the orthotic upright is,

$$= [(N_i/2048.0)-1.0] * [CAL]^{-1}/[GAIN*VOLTS]$$

where N_i = output for frame i at the computer;

$[CAL]^{-1}$ = inverse calibration matrix of the transducer concerned;

GAIN = strain gauge amplifier gain setting; and

VOLTS = bridge supply voltage.

The load actions calculated above are expressed in the respective transducer coordinate systems, and are acting on the transducer from distal to proximal end through the centre of the transducer.

6.7 LOADINGS ON ORTHOSIS.

Since the load measuring transducers are placed strategically on the orthotic uprights, the loadings at various points on the orthosis can then be deduced from the transducer readings and the geometrical relationships of the transducer with the point to which the loading is required. The points of interest are: the shoe attachment points, orthotic knee and ankle joint centres, and the strap connection points, which include the thigh and calf bands, T-strap, knee straps, and the knee apron. Loadings are calculated for both the lateral and medial uprights. It is generally assumed that the orthosis experiences negligible amount of distortions under the loading conditions.

Fig 6.17 shows the arbitrarily applied external loadings, F1 to F6, that may be experienced by an orthosis in a walking condition. The transducers replace sections of the uprights at T1 and T2, and thus monitor the 'sectional loads' at these levels, which are the resultants of all the loads more proximally than T2 for the proximal transducer, and more distally than T1 for the distal device. For analytical purposes, each individual upright can be taken as three separate 'free bodies', as shown in Fig 6.18. The unknown forces and moments can then be determined by considering the equilibrium of forces and moments at each individual free body.

The first step in the analysis is to transform the loadings measured in the transducer frame of reference to their respective orthotic upright systems. Thus,

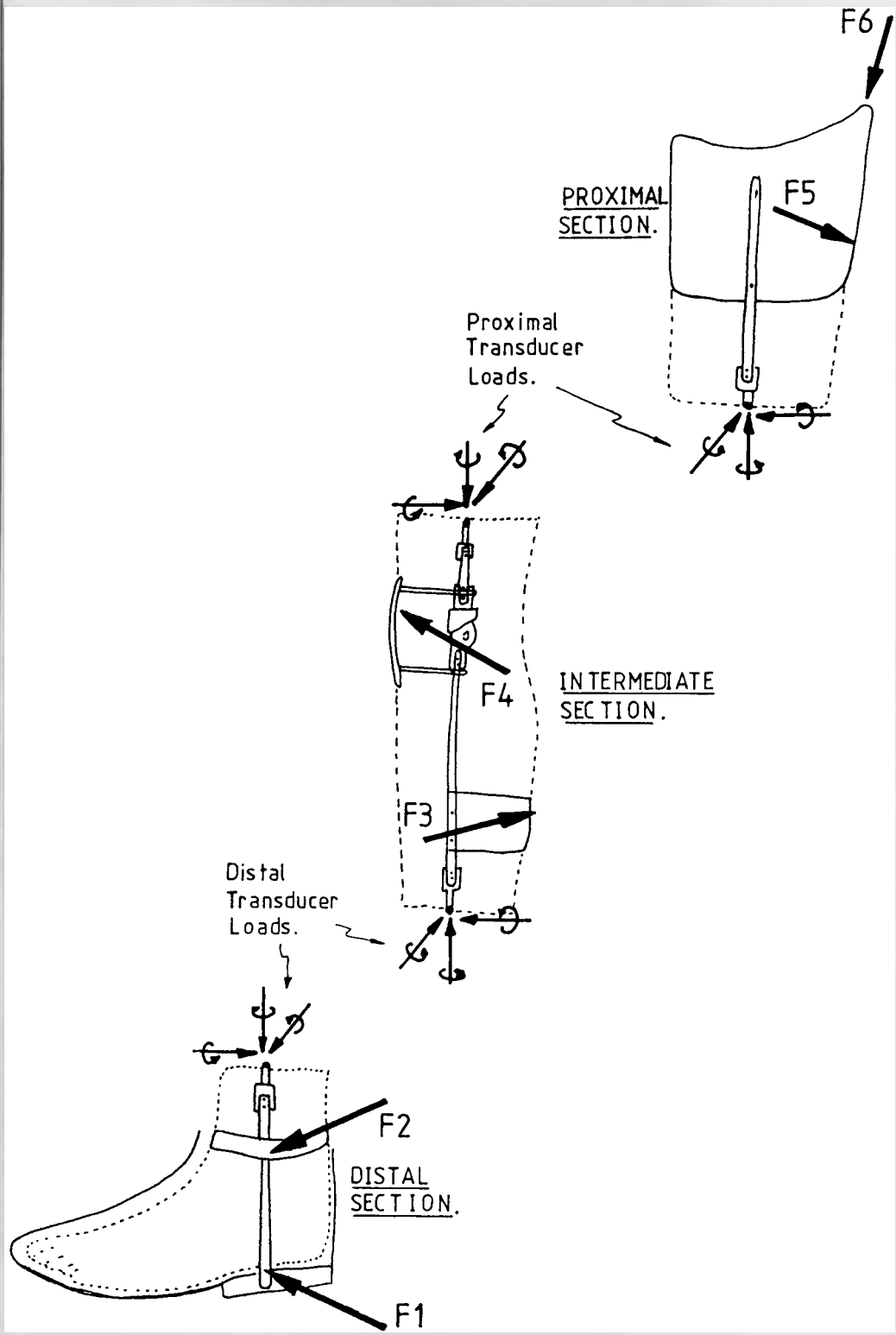


FIG 6·18	Free Body Diagrams Used in The Analysis.

$$\begin{bmatrix}\text{transducer load}\\ \text{in orthotic}\\ \text{upright system}\end{bmatrix} = \begin{bmatrix}\text{transducer load}\\ \text{in transducer}\\ \text{system}\end{bmatrix} * \begin{bmatrix}\text{DC matrix}\\ \text{of}\\ \text{transducer}\end{bmatrix} * \begin{bmatrix}\text{DC matrix}\\ \text{of orthotic}\\ \text{upright}\end{bmatrix}^{-1}$$

Applying to the lateral proximal transducer (TLP),

$$[FTLPO] = [FTLP]*[CTLP]*[COLS]^{-1}$$
$$[MTLPO] = [MTLP]*[CTLP]*[COLS]^{-1}$$

where [FTLPO],[MTLPO] = arrays containing the forces and moments of transducer TLP in lateral upright system;

[FTLP],[MTLP] = arrays of forces and moments of TLP in transducer TLP reference system;

[CTLP] = DC matrix of transducer TLP with reference to GRS;

[COLS] = DC matrix of lateral upright system, with reference to GRS, and obtained from static calibration tests.

From the free body diagram, the forces FLR at any point 'LR' on the intermediate section of the orthosis (Fig 6.18) can be found from the force equilibrium equations, i.e.

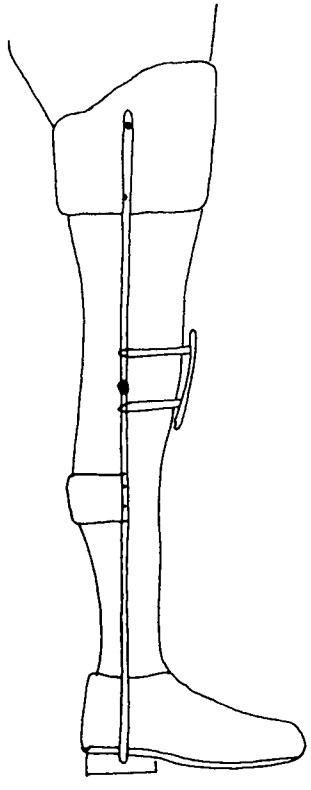
$$[FLR] = [FTLP]-[FTLD]-\Sigma(\text{all other forces})$$

To calculate the moment MLR at the point LR, the distance vector LRS, from the point LR to an adjacent point LS, and in the appropriate upright coordinate system is needed. The point LS is normally the centre of an adjacent transducer, but may also be an adjacent point where the forces and moments are fully known. MLR is then equal to the sum of the moment MLS at LS and the cross-product of the distance vector LRS and the forces FLS, i.e.

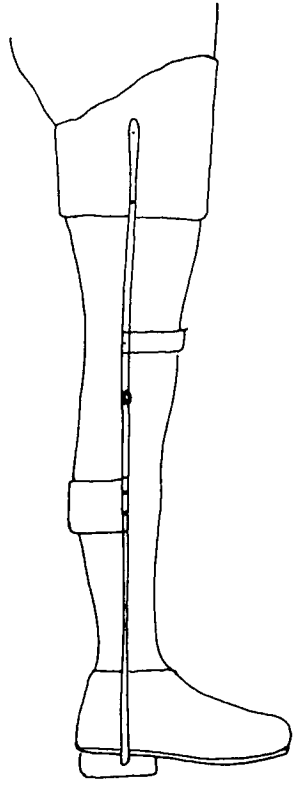
$$[MLR] = [MLS] + [LRS]\wedge[FLS]$$

where the symbol $\wedge$ denotes vector cross-product.

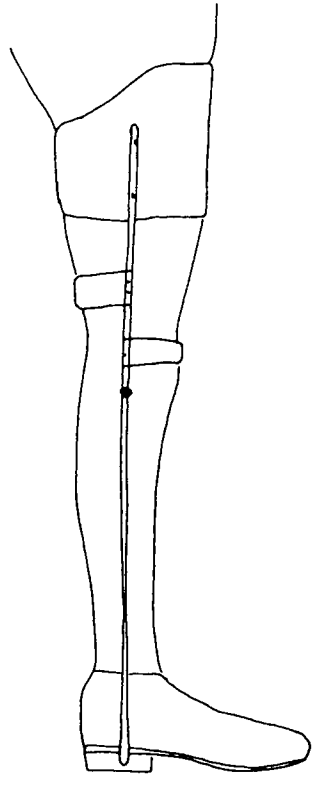
The loadings at the knee restraining straps are influenced by the strap configuration used. Since the strap configuration in the instrumented KAFO during a test follows closely to that prescribed to the patient, a situation arises whereby a number of different options are needed to cover all the normal types of strap configurations. A two-tier system of choices is therefore devised, consisting of options SOPT and SFOPT. The option SOPT provides for the variation in the strap configurations (Fig 6.19), while SFOPT suppliments SOPT by accommodating the strap axial force assumptions. Details of these options are given below:



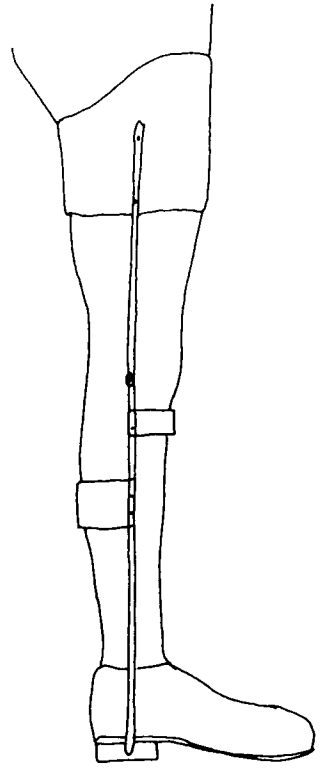
SOPT = 0
Knee Apron
and
Calf Band.



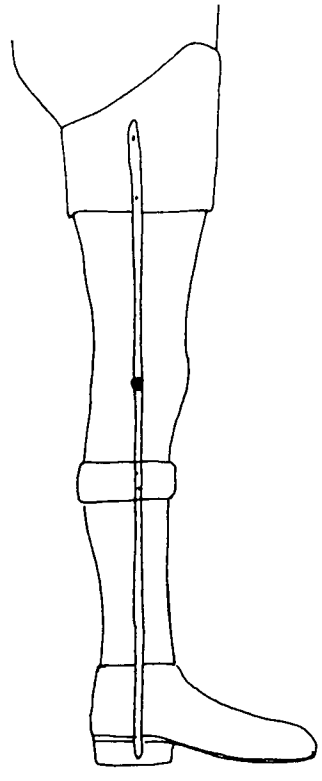
SOPT = 1
AK Band
and
Calf Band.



SOPT = 2
Double
AK Bands.



SOPT = 3
BK Band
and
Calf Band.



SOPT = 4
Calf Band
Closure only.

FIG 6·19

Knee Restraining Strap Option SOPT.

SOPT = 0, knee apron and calf band used;

SOPT = 1, above-knee (AK) restraining strap and calf band used;

SOPT = 2, double AK restraining straps, no calf band;

SOPT = 3, below-knee (BK) restraining strap and calf band used;

SOPT = 4, calf band only, no other knee straps;

and,

SFOPT = 1, only proximal strap is weight-bearing;

SFOPT = 2, only distal strap is weight-bearing;

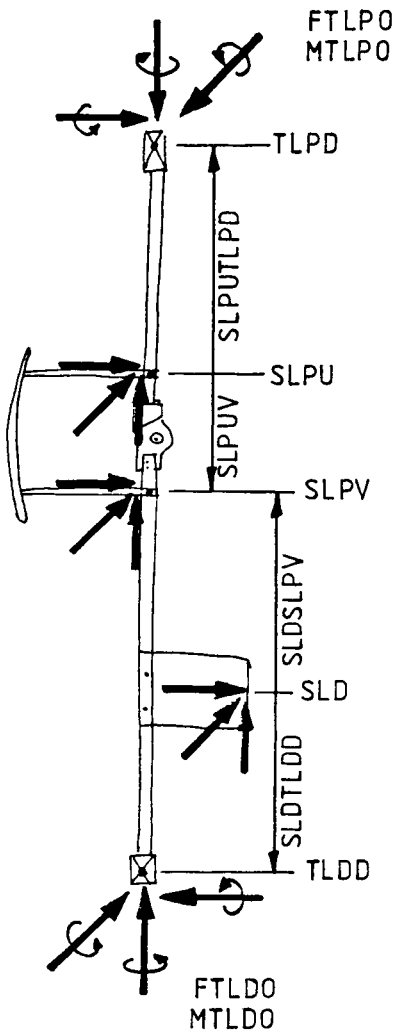
SFOPT = 3, only calf band is weight-bearing, used in conjunction with SOPT =4;

SFOPT = 0, weight-bearing is shared by both strap, used in conjunction with SOPT=0, where the axial load is shared in a 7:3 ratio between the apron and the calf band.

Considering the 'knee apron/calf band' complex, where the knee apron exerts 2 separate loading points on each upright while a somewhat non-uniformly distributed load is applied by the limb on the calf band. This situation can be simplified with the following assumptions: the two leather straps of the apron are assumed to carry equal loads, and the distributed load on the calf band is assumed to be a concentrated load passing through the centre of the band. It is further assumed that these straps do not produce any moment at the load application points on the upright. After discussion with orthotists and other workers in the field, the axial load in the complex is assumed to share between the two straps in a ratio of 7:3, with the knee apron taking the larger load. Therefore, by considering the upright between the 2 transducers as a free body, the forces applied by the straps and calf band can be fully defined (Fig 6.20).

For the situation where a combination of 2 straps or bands is placed between the proximal and distal transducers (i.e. SOPT=1, 2, or 3), an assumption in the weight-bearing of the straps/bands is required (i.e. SFOPT options). Either of the straps can be assumed to carry most of the axial load in the strap combination. Again the unknown can be found from the force and moment equilibrium of the free body.

A facility for calculating the loadings at the T-strap and the shoe attachment points is also included in the analysis, where options are available for the cases where the straps is either around the medial or the lateral upright, or when it is not being used at all



(lateral upright)

FIG 6·20	Load Analysis of The Knee Apron and Calf Band Complex.

(option IST). For this part of the analysis, the free body is taken to be the section of the upright more distal than the distal transducer centre, including the T-strap and shoe-attachment points on the upright (Fig 6.21). However, this free body possesses too many unknowns, and the following assumptions are taken to simplify the situation. Firstly, the T-strap is assumed to be situated more superiorly than the ankle joint, and no external forces are applied at the orthotic joint centre. It is further assumed that the strap neither carries any moments nor any axial load, i.e. Mx=My=Mz=0. No anterio-posterior moment (Mz) is applied at the shoe attachment point, since round spurs are used in the orthoses. These assumptions reduce the analysis to a statically indeterminate situation with a single redundancy, and it can therefore be analysed using the energy method of solution.

6.8 ANATOMICAL LOADINGS.

Under this heading contains the analysis for the determination of the intersegmental loads at the anatomical joints of the affected limb. The anatomical loads are obtained from the difference between the ground reaction loads and the loadings on the upright at the joint level, i.e.

$$\text{Load}_{(\text{anatomical})} = \text{Load}_{(\text{FP})} - \text{Load}_{(\text{KAFO})}$$

The sign convention for the loading on orthosis, as discussed in Section 6.2, will also apply for the anatomical forces and moments. However, at the limb-orthosis interface, the same loading changes sign across the interface, i.e. the same loading expressed as external load acting on the limb is equal and opposite to the same loading when it is expressed as an external load on orthosis by limb. To avoid this confusion at the interfaces, the limb-orthosis complex is taken as a single unit, and sections are taken away from the interfaces. The anatomical loads are then calculated at the ankle, knee, and hip joint levels, as well as at the ischial loading area.

The forces (FAC) and moments (MAC) at the anatomical ankle joint can be calculated from:

$$[\text{FAC}] = [\text{FP}]-[\text{FAM}]-[\text{FAL}]$$
$$[\text{MAC}] = [\text{MP}]-[\text{MAM}]-[\text{MAL}]$$

where FP and MP are the forces and moments from the force-plate results,

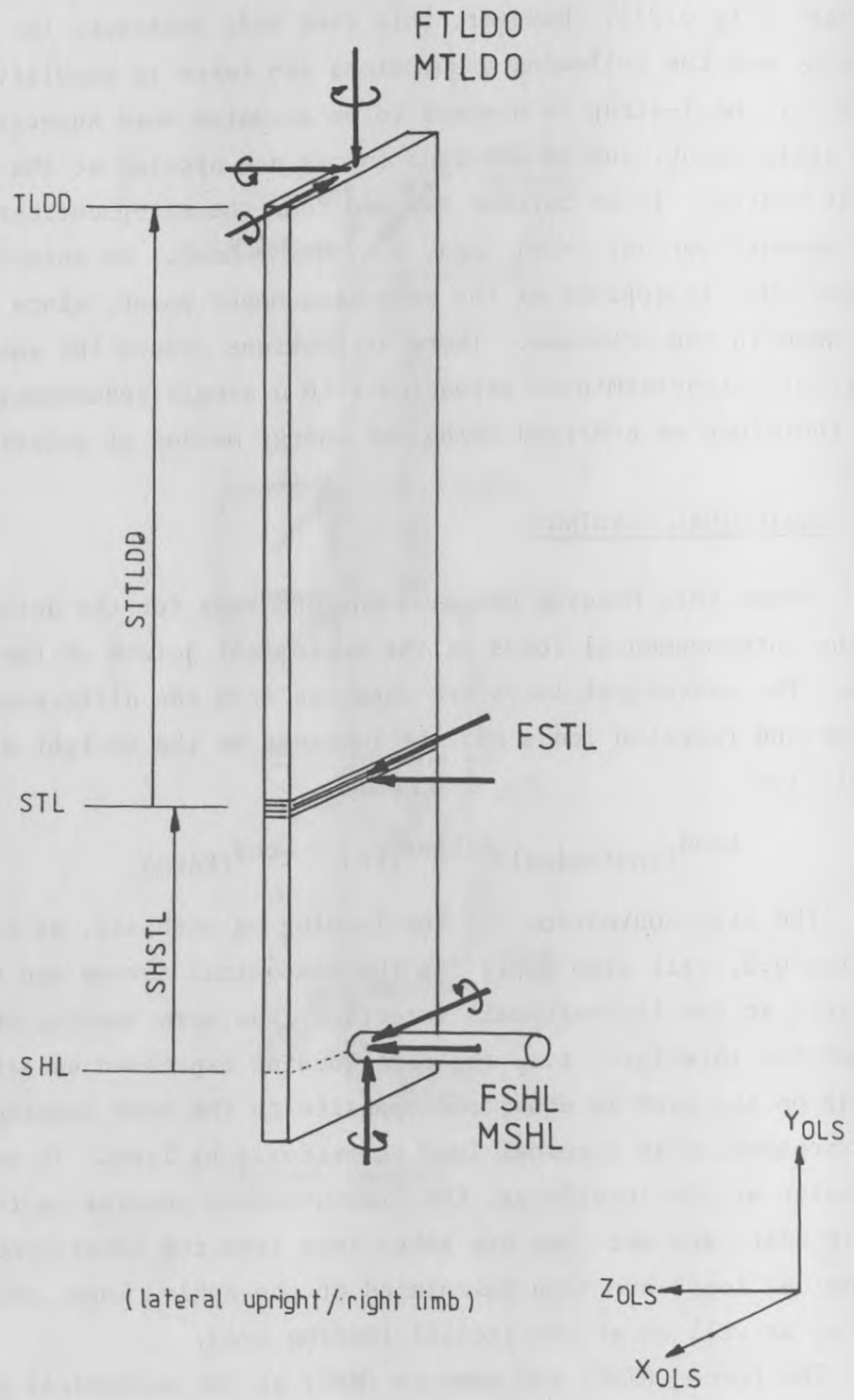


FIG 6·21	Load Analysis of the T-Strap and Shoe Attachment Point.

and FAM,MAM and FAL,MAL are the forces and moments of the medial and lateral uprights at the ankle joint level respectively. A similar analysis can be used to obtain the knee loadings.

For the ischial forces, it is assumed that the axial load of the orthotic uprights above the knee level is fully supported by the ischial seat. Therefore the ischial load is equivalent to the sum of the two proximal transducer forces. This in turn determines the hip force which is defined as the difference between the force-plate and the ischial forces.

6.9 KNEE STRAP TRANSDUCER LOADINGS.

As with the KAFO Mark I and II transducers, the knee strap transducers are also designated by their positional relationships with each other. However, the strap transducer names are prefixed with 'TX' instead of 'T' to avoid causing confusion with the other transducers. The 'full names' of the transducers are therefore, TXLP, TXLD, TXMP, and TXMD for the lateral proximal and distal transducers, and medial proximal and distal transducers respectively.

The Y-axis of the TXLP and TXLD coordinate systems are assumed to be parallel to the line defined by markers TXLA and TXLB (Fig 6.22). Their Z-axes are normal to a plane formed by the Y-axes of the transducer systems and the X-axis of the lateral upright system. The X-axes are orthogonal to both the Y and Z axes of the transducer systems (Fig 6.23).

The coordinate axes system for the medial proximal strap transducer TXMP is formed in a similar manner as those for the lateral transducers, but with the Y-axis defined by markers TXMA and TXMB, and with its Z-axis normal to the X-axis of the medial upright system. However, at the distal end of the medial orthotic knee joint assembly, the upright attachment slot is bent laterally to follow the contour of the anatomical knee joint structures, and the Y-axis of the transducer TXMD has to be formed at an angle inclined to the Y-axis of TXMP. This angle was determined experimentally by measuring 14 pairs of "Otto Bock" orthotic knee joints, and found to be inclined at 16° laterally. The X and Z axes formations are similar to TXMP.

Loadings from all the transducers are subsequently expressed in the lateral upright coordinate system such that data from any of the

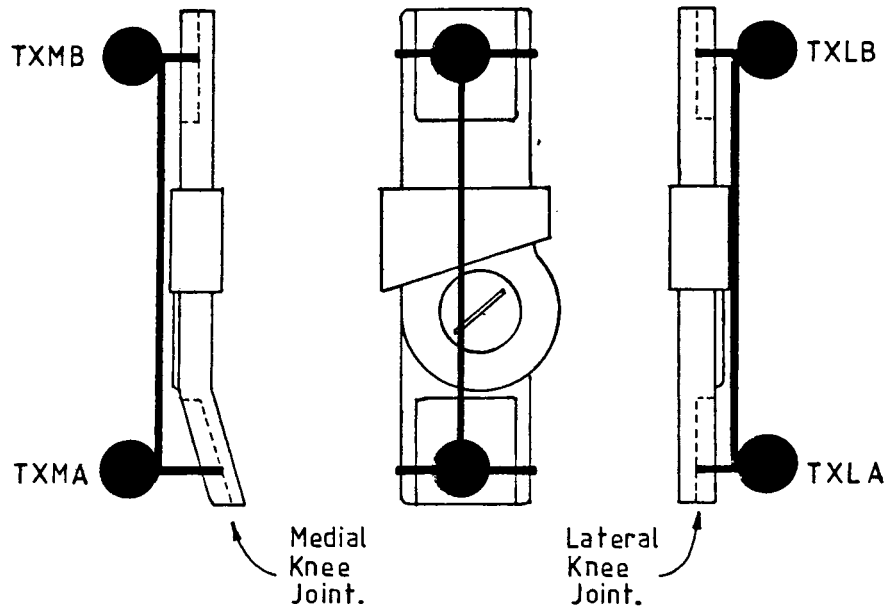


FIG 6·22 Markers Attached to the Orthotic Knee Joints.

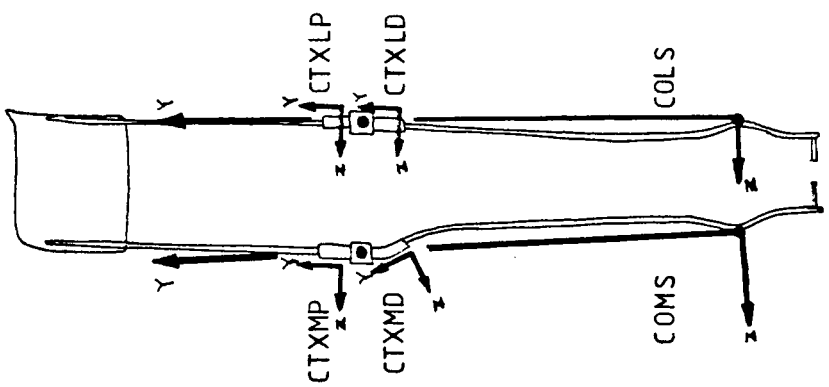


FIG 6·23 Coordinate Systems of Strap Transducers.

transducers may be compared or combined with each other. The axes conversion process from one frame of reference into another is similar to that described in Section 6.7 . Calculations were performed to determine the magnitude and direction of the resultant load for each transducer, as well as for the lateral, medial, proximal, and distal pairs of the transducers. In addition, the resultant knee strap loading of the 4 transducers, which includes the point of application of the force, were also calcualted.

As an illustration, the resultant load at the TXLP transducer is:

$$RTXLP = SQRT[\ FTXLPO(x)^2 + FTXLPO(z)^2\ ]$$

where FTXLPO is the forces at transducer TXLP, in the lateral upright reference system;

x,z = x and z components of the force FTXLPO.

CHAPTER 7.

RESULTS AND DISCUSSIONS.

7.1 INTRODUCTION.

This chapter presents and discusses the results of the various tests carried out on a group of subjects which comprises a normal subject and 8 post-poliomyelitis patients. Due to the limitations of the patients' ability to carry out the prescribed ambulatory activities, and also to keep the disruption of the patients' normal work to a minimum, it is inevitable that some of the subjects will not be able to participate in all the tests.

The results that will be presented include the findings of a KAFO Survey on the participating patients, the orthotic loading patterns recorded with the different ambulatory activities performed by the patients, and last but not least, the anatomical and strap loadings. Only a single patient's results will be presented and discussed in details in each of the tests. Any significant differences in the results of any other patients will also be discussed, and the findings compared with the results obtained by other researchers.

7.2 RESULTS OF KAFO SURVEY.

Six of the post-poliomyelitis patients (3 males and 3 females) participating in the KAFO study were requested to complete the KAFO Survey questionnaires, a copy of which has been included in Appendix 7. The results of the survey are summarised below.

All the patients were prescribed with conventional double-upright KAFOs by their clinicians, except for patient 'MN' who wears a cosmetic plastic KAFO, and they are able to put on the orthoses with relative ease. The orthoses were considered to be uncomfortable for walking and standing by two patients, while the rest indicated that the appliances were either "tolerable" or "O.K.". The thigh band and the weight-bearing top were the main sources of discomfort for patients with conventional KAFOs. It was unanimously agreed that the orthoses are on the heavy side, however, most of them "have got used to" the weight of the device.

A surprisingly high proportion of the orthoses (67%) were altered or repaired by the patients in a "do-it-yourself" manner. It ranges from minor repairs such as replacing worn-out nuts and bolts at the knee joints, to bending the uprights to relieve pressures on the knee,

and also the welding of the ankle joint. A patient even went to the extent of redesigning the knee apron straps to make it more "comfortable and effective".

Ten incidents of orthotic failures were reported, with failures of the 'spur' being the commonest (40%). A patient had her orthosis broken twice at the medial proximal section. The first incident occurred at her place of work while she was coming down from stairs, and the other failure happened when she tried to get off a bus. Another patient also described how his orthosis failed at the medial distal section when he was getting down from a bus. The most unfortunate failure was experienced by a patient while holidaying abroard. One of the spurs managed to work itself out from the socket and the orthosis was very badly twisted. In another incident, the knee joint of the same patient's orthosis failed, and she sustained a minor injury to the limb.

Two-thirds of the patients are very confident with their orthoses, while the others are not too happy because of the orthotic failures they had experienced. All the patients can manage most of the common ambulatory activities without too much difficulties. A particular patient pointed out that her sound leg normally gets tired first before the affected one.

Most of the patients indicated that they would like to have lighter and more close-fitting cosmetic KAFOs for any future prescriptions. One patient suggested that telescopic proximal sections should be fitted to the orthosis so that the weight-bearing tops may be adjusted to suit the needs according to the circumstances.

7.3 ORTHOTIC LOADINGS IN AMBULATORY ACTIVITIES.

The loading patterns experienced by the orthoses in 6 different types of ambulatory activities will be discussed in the following scetions. The results are presented as loadings on the orthosis, with their directions defined according to the sign convention described in Section 6.2 . In addition, the results are also reported in the respective transducer reference systems, and assumed to act at the centres of the transducers. The transducers have been pre-arranged to be at approximately the mid-point along the length of each section of the uprights.

Only the results of a typical patient will be described in detail, and where appropriate, the results obtained from other patients will be discussed with reference to that patient. The orthotic loadings over 5 gait cycle periods are presented in graphical form, and they are arranged in groups at the end of this section so that comparison between the different activities may be carried out with relative ease.

7.3.1 Level Walking in a Straight Line.

The results obtained from a typical test for a right-side affected active male patient 'MR' are shown in Fig 7.1 to 7.8, where the solid and broken lines represent loadings at the proximal and distal sections respectively. A very good repeatability over 5 gait cycles can be seen in Fig 7.1 and 7.2 which illustrate the axial loads (FY) of the lateral and medial uprights respectively. During the stance phase, a higher proportion of the body weight is being carried through the lateral than the medial upright. The results also revealed a triple-peak loading pattern for the uprights, where the first and the highest peak occurred just after heel-strike (HS) upon full weight-bearing by the affected side. The second and also the lowest peak occurs around mid-stance while the last peak reaches its maximum just before heel-off (HO). For the medial upright, the distal section always carry a higher axial load than the proximal. However, at the lateral upright, after the distal section has reached the peak value of some 300N (i.e. 0.44 x body weight), the proximal section becomes more heavily loaded. This loading condition remains for the rest of the gait cycle.

Another interesting characteristic of the patient's axial loading pattern is that the maximum FY recorded fluctuates in a distinctive way with every step taken by the patient. A similar loading level is obtained in every alternative step, with a difference of approximately 50N between any two consecutive steps, while the durations of the swing and stance phase remain the same throughout the gait cycles. Although this can be attributed to the patient's particular style of walking, the actual causes responsible for this unusual phenomenon have not been fully understood.

At the end of the stance phase and the beginning of swing, a small peak of about 15N in magnitude can be seen in the lateral upright, and it corresponds to a trough of similar magnitude on the medial side. When this is compared with the loadings in other channels, i.e. Fig 7.3 to 7.8, the phenomenon was found to exist in all the graphs. This is probably due to a slight plantar flexion of the foot-shoe complex after the end of the stance phase. The underlying cause of this delayed plantar-flexing action can be attributed to the non-alignment of the anatomical ankle joint and the 'spur-and-socket' arrangement at the shoe. This consequently causes frictional effects at the spur and socket interface and also a slight pistoning action of the orthosis with respect to the enclosed limb.

During the swing phase, the orthosis does not become fully unloaded, but remains at approximately 10N of axial load at the medial upright and 20N at the lateral proximal section of the upright. For the distal section of the lateral upright, it becomes tensile (FY= -20N) immediately after toe-off. This is caused by the pulling action of the T-strap on the lateral distal upright.

For both the uprights, the anterioposterior force (FX) acts anteriorly on the proximal sections and posteriorly on the distal throughout the gait cycles. These forces are higher by some 40% on the medial than on the lateral upright (Fig 7.3 & 7.4). The results indicate that the limb applied considerable forces on the orthosis just after HS, trying to flex the orthosis about the knee. This, in turn, can be interpreted as that extension forces are being applied by the orthosis onto the enclosed limb. It therefore fulfills the function of stabilizing the affected limb. These stabilizing forces are maintained throughout the stance phase, with another peak of activities at HO, also to prevent flexion of the enclosed knee.

A mediolateral instability was found to occur just after HS, as shown in the FZ traces of Fig 7.5, where all 4 sections of the uprights are acting laterally to prevent a collapse of the knee in the medial direction. Thereafter, these forces remain at low values for the rest of the stance phase.

The maximum mediolateral moment (MX) was found to be at the medial distal section, where it reaches a magnitude of 11 Nm (Fig 7.6). All other section experienced MX moments of about half of that value.

Another observation is that the MX moment acts in the positive direction on the lateral upright but negatively on the medial, i.e. the uprights are acting to keep the enclosed limb in mediolateral stability.

The medial distal section was also found to be under the maximum torsional moment MY of 6 Nm (Fig 7.7). The lateral distal section is loaded to 5 Nm, just lower than its medial counterpart. However, they are being stressed in the opposite direction to each other. The MY moments in the proximal sections are much smaller, and are acting in the same direction as the medial distal. The correcting moment of the T-strap on the lateral distal section may have caused the large moment that acts in the opposite direction as the other sections. Although this would normally be expected to transmit to other sections of the orthotic uprights, the shoe-foot and the knee-spron/calf-band complexes may have absorbed the excessive moment and prevent the other transducers from recording higher MY values.

Unlike the MX and MY moments, the highest anterioposterior moment (MZ) occurs at the lateral proximal section, with a maximum load of 25 Nm acting in the negative direction (Fig 7.8). This is some 5 Nm higher than those in the other sections, with all 4 of them acting in the same direction throughout the gait cycle. It thus verifies the previous finding that the orthosis is preventing the bending of the knee at all times.

To illustrate the good repeatability of the results, the FY and MZ loadings on the lateral proximal section of 6 different runs of the same patient 'MR' performing the straight line level walking tests were plotted (Fig 7.9 & 7.10). A 100N axial load variation (i.e. 30% of the maximum recorded) and a 10 Nm MZ moment (i.e. 20% of maximum MZ) differences were recorded in the maximum values in these channels. When the fluctuation of maximum loadings in alternative steps of this patient (as explained earlier) has been taken into considerations, the FY and MZ load variations becomes 15% and 10% of the maximum values respectively.

The results of patient 'MR' were then compared with those obtained from the other patients and the normal subject participating in the tests. Some of the more important fingings are discussed below.

The factors that may affect the amount of axial load experienced by the orthosis are: the shape of the weight-bearing top, the fitness of the orthosis, and to a certain degree, the gait pattern of the

patient. However, for this study, all these factors except the individual gait pattern of the patient may be considered as constant since the orthoses were all measured, manufactured, and fitted by the same orthotist. Therefore, it is assumed that the principal variable will be the gait patterns.

For an active and 'aggressive' walker like patient 'MR' described above, large axial forces of up to 300N (0.44 x body weight) were recorded. Other active but not so 'aggressive' walkers (patient MC, MN, and WI) yielded a maximum of 200N, and for very 'careful' walkers (patient DC, DR, and EM), the axial loads recorded were less than 100N. The normal subject produced an axial loading of around 120N. A wide variety of loading patterns was recorded for the FY channels, varying from a single peak loading characteristic to one with triple peak. The load distribution between the two uprights ranges from an equally distributed situation to one that is heavily depended on the lateral upright for weight bearing, as for patients WI and DR (Fig 7.11), where the medial uprights were in tension. This rather unusual loading pattern can be attributed to the excessive lateral trunk bending motion of the patients during the stance phase. Another interesting point is that while all the post-poliomyelitis patients have higher loadings on the lateral upright, the normal subject 'CR' produced a result whereby the medial upright is twice as highly loaded as the lateral (Fig 7.12).

The FY loading patterns discussed above, where the medial upright is in tension for some patients, correlates very well with the results obtained by Scott and coworkers (1971) on their research on the UCLA Functional KAFO. They discovered that the medial bar can be in tension throughout the stance phase while the lateral bar is under compression. Lippert (1971) who employed a normal subject for his study on lower limb orthotics reported that the FY loadings on the medial uprights were very much greater than the lateral side. His results are therefore in good agreement with the loading pattern obtained for normal subject CR. Anderson (1974) claimed that for polio patients standing in an upright posture, the lateral upright always carries much less loads than the medial member. This was found to be in contrary to the results of the patients (MR and DR) who were involved in the standing tests. The lateral upright was twice as highly loaded as the medial.

It is normally assumed that the distal sections of the uprights carry higher axial load FY than the proximal sections since some of the axial load would have been absorbed by the anatomical knee structures through the knee straps. However, the results show that this may not always be true. Some patients have produced results where the proximal sections were loaded to either marginally above or considerably more than the distal sections. This phenomenon may be due to the strapping actions on the uprights which redistributed the axial loads on the orthosis.

The results obtained for the FY loadings during swing phase indicate that the uprights do not fully unload as assumed by Kirkpatrick, Day and Lehmann (1969) (see Section 3.4.5). Since the orthotic upright can either be in compression or tension prior to heel-strike, their results will therefore be over-, or under-estimated. Although they used the "zero-force" datum, they reported that the orthosis did not return to zero loading at the end of stance because of the test subject's muscle power. However, the results obtained by the author of this thesis shows that the effects is more likely to be caused by the combination of orthotic fitting and strap actions than by muscle actions.

Although the majority of the patients produced an anterioposterior shear force (FX) pattern similar to that of MR, three other variations were also recorded. EM has the FX forces acting in the opposite direction as MR, i.e. forces at proximal sections acting posteriorly and distal forces anteriorly (Fig 7.13). The patient also produced a peak of activity after the end of stance phase, and this is considered to be caused by the delayed plantar flexion of the foot-shoe complex as discussed previously for MR. For patient DR, the FX forces at the proximal sections act anteriorly at the beginning of the stance phase. At foot-flat, FX changes to a posteriorly acting force before acting anteriorly again at heel-off. Thereafter, it remains anteriorly throughout the swing phase. The distal section FX forces are in the opposite direction as the proximal (Fig 7.14). The normal subject CR has the FX forces at the medial upright acting anteriorly, and lateral posteriorly, with the medial FX forces more than twice that of the lateral.

The mediolateral shear force (FZ) varies considerably with patients and there is a total lack of similarity between any two patients. This is probably caused by the combined effects of the differences in the gait patterns and of the strap configurations. In general, FZ has a small magnitude of less than 40N, except in MC and WI where the lateral proximal section reaches 90N and 60N respectively. Since the mediolateral bending moment (MX) is closely related to the FZ forces, the MX obtained from the tests are also expectantly to be small.

Similar to the FZ and MX loadings, the torsional moments (MY) about the long axis of the upright also varies from patient to patient. The maximum MY moment recorded were less than 6 Nm.

For the anterioposterior moments (MZ), all but two patients (DC and EM) applied the largest loadings on the lateral proximal section. The maximum MZ obtained was 26 Nm (patient MN), marginally above that of MR, and most of the patients have the proximal sections more highly loaded than the distals. For patient WI, the lateral proximal has a much higher loading than the other 3 sections, as shown in Fig 7.15 (c.f. Fig 7.8). The most severely loaded section for patients DC and EM are the lateral distal and the medial distal , where they reached 16 Nm and 13 Nm respectively.

Comparing with the results of Scott et al (1971), where they reported that the MZ moments on the distal section ("ankle bar") was approximately twice that of the proximal section ("knee bar"), this loading condition is true only on two patients of the Free Walking Tests. However, Trappitt (1979) who used a similat set-up as those used by the author, reported the maximum loadings measured being at the medial proximal section for both of his patients. The MX and MZ loadings recorded were 9 Nm and 23 Nm respectively. The MZ loading characteristics for some of the patients also agree with the assumption suggested by Lippert (1971) that the greatest bending moment occurs when the axial loads are at their highest.

A few selected patients were also tested on different days so that any variation in the orthotic loading patterns can be recorded. The differences obtained for patient DC was minimal, except in the axial load where the loading and unloading process is much steeper in one occasion that the other. There was also a slight change in the

Table 7.1 Table Showing the Highest Orthotic Loadings for Patient 'MR'.

Orthotic * Uprights.	Normal Walking		Other Activities		Overall Highest Loadings	
	max	min	max	min	max	min
Lateral Proximal						
FX	70	6	68	9	70	6
FY	325	15	314	12	325	12
FZ	-2	-19	17	-28	17	-28
MX	6	-1	6	-1	6	-1
MY	1	-3	2	-4	2	-4
MZ	0	-30	0	-29	0	-30
Lateral Distal						
FX	7	-56	16	-53	16	-56
FY	379	-26	379	-185	379	-185
FZ	21	-5	33	-15	33	-15
MX	5	-1	5	-2	5	-2
MY	6	0	8	-2	8	-2
MZ	1	-23	4	-23	4	-23
Medial Proximal						
FX	89	10	90	7	90	7
FY	238	-2	240	-34	240	-34
FZ	4	-21	17	-23	17	-23
MX	0	-7	0	-7	0	-7
MY	0	-4	1	-5	1	-5
MZ	-1	-22	-1	-22	-1	-22
Medial Distal						
FX	-3	-74	40	-69	40	-74
FY	274	-3	404	-66	404	-66
FZ	10	-32	32	-35	32	-35
MX	-1	-11	2	-15	2	-15
MY	0	-6	1	-7	1	-7
MZ	0	-23	15	-23	15	-23

* The units of loadings for FX, FY, and FZ is in newtons (N); and for MX, MY, and MZ is newton-metres (Nm).

axial load distribution between the two uprights. However, for patient DR, some substantially different loading patterns were obtained. Fig 7.16 shows the MZ moments obtained on three occasions, they range from a negatively loaded -6 Nm to some 20 Nm acting positively for the lateral proximal section. The variations in the axial load are illustrated in Fig 7.17 & 7.18, where the principal difference is in the loading patterns of the medial upright.

Two of the patients were also monitored during the process of starting to walk from a stationary position, and while slowing down to a halt. Fig 7.19 & 7.20 show the axial load FY of patient MR for the initial 4 walking cycles from a stationary position. It can be seen that FY progressively increases through the first 3 gait cycles before the normal FY values are attained. The results show a proportionally higher peak at HS prior to reaching the 'steady-state' walking process. It also revealed that the lateral upright carries as much as twice the medial upright axial load in the standing position.

Results of patient MR stopping from a steady walking situation are shown in Fig 7.21 & 7.22, where the loadings are gradually reduced through the final few gait cycles. A similar pattern is seen in all other transducer channels, and the MZ loading for both the above activities is presented in Fig 7.23 . This phenomenon of the reduced loadings may be due to the fact that the patient is more careful at 'stop-start', and subconsciously place more weight onto the anatomical structures than through the orthosis. Results obtained from patient DR for the 'stop-start' activities corresponds well with that of MR. This finding thus disproves the common belief that the orthotic loadings are higher in the 'stop-start' process than in normal steady state walking.

In addition to the above activities, patient DR was also requested to walk faster than her normal perferred speed, and also at a much slower pace. She was also asked to stop abruptly at mid-walk. No significant changes in the loading patterns were noted. When asked to stop suddenly at mid-steady state walking, DR used the sound leg to decelerate the body, which effectively avoided any significant increase in the orthotic loadings.

Table 7.2 Table Showing the Highest Orthotic Loadings for Patient 'MN'.

Orthotic * Uprights.	Normal Walking		Other Activities		Overall Highest Loadings.	
	max	min	max	min	max	min
Lateral Proximal						
FX	125	23	147	27	147	23
FY	112	-24	138	-19	138	-24
FZ	12	-9	15	-11	15	-11
MX	2	-2	2	-2	2	-2
MY	2	0	3	0	3	0
MZ	-2	-28	-3	-29	-2	-29
Lateral Distal						
FX	-34	-118	126	-137	126	-137
FY	243	26	264	-266	264	-266
FZ	8	-35	36	-17	36	-35
MX	1	-3	1	-3	1	-3
MY	5	-1	6	0	6	-1
MZ	-1	-21	1	-22	1	-22
Medial Proximal						
FX	93	3	186	4	186	3
FY	210	-5	254	-9	254	-9
FZ	-3	-55	52	-37	52	-55
MX	0	-5	1	-8	1	-8
MY	4	-1	6	-2	6	-2
MZ	-1	-26	22	-33	22	-33
Medial Distal						
FX	7	-69	92	-66	92	-69
FY	134	-70	105	-145	134	-145
FZ	28	-6	24	-17	28	-17
MX	2	-3	2	-3	2	-3
MY	2	-2	2	-3	2	-3
MZ	2	-15	4	-28	4	-28

* The units of loadings for FX, FY, and FZ is newton (N);
and for MX, MY, and MZ is newton-metres (Nm).

7.3.2 Walking Round a U-Shaped Path.

The detailed results from the right-side affected active male patient MR performing both the 'right-handed' (RH) and 'left-handed' (LH) U-turns will be discussed. Each of the tests were repeated to verify their reproducibility of the results.

The FX and FZ loading characteristics for both the activities are essentially similar to those of the previous case, i.e. walking in a straight line. However, there is a slight decrease in the magnitude of FX in all the sections as the patient negotiates the semi-circular portion of the walk path. For the distal section, the medial FX forces decrease at a higher proportion than the lateral during the RH U-turn. The opposite is true for LH U-turns.

There is also an overall reduction in the MX and MZ moments. The loading patterns for the proximal sections in the anterioposterior directions are shown in Fig 7.24, and the differences can be readily identified when it is compared with Fig 7.8 for the normal straight line walking test. Fig 7.25 & 7.26 show the changes in the MY loading patterns during the U-turns, where the loading in the lateral distal section increases by 40% to 7.5 Nm and at the same time the medial distal decreases by 30% in the RH U-turn. For LH U-turn, the reverse is true.

The most distinctive differences between the loadings of a straight line and a U-turn walk is found in the axial loads experienced by the orthosis. For RH U-turn, the lateral FY forces are somewhat similar to the normal walk. However, on the medial upright, the FY loads have a very much reduced peak just after HS, thereafter, the upright is almost totally unloaded. At the top of the U-turn, the medial upright becomes tensile (Fig 7.27 & 7.28).

For LH U-turn, FY reduces gradually from the straight section of the U-path to the top of the semi-circular section, and thereafter, the axial load increases again. There is also a momentary tensile loading at the lateral distal section (Fig 7.29). At the medial upright, the FY load pattern obtained shows an increasing peak value at heel-off as the patient walks into the semi-circular walkpath. It reaches a new height of 300N, some 80N higher than for normal walking. A similar pattern is obtained for the other right-side affected patients (EM and MC).

Table 7.3 Table Showing the Highest Orthotic Loadings for Patient 'MC'.

Orthotic * Uprights.	Normal Walking		Other Activities		Highest Overall Loadings.	
	max	min	max	min	max	min
Lateral Proximal						
FX	41	-5	51	-15	51	-15
FY	208	-2	210	-5	210	-5
FZ	-6	-95	-4	-99	-4	-99
MX	1	-2	2	-4	2	-4
MY	3	0	5	-1	5	-1
MZ	0	-21	1	-21	1	-21
Lateral Distal						
FX	8	-37	17	-44	17	-44
FY	197	-13	272	-56	272	-56
FZ	45	2	52	2	52	2
MX	3	-1	4	-1	4	-1
MY	4	0	6	-3	6	-3
MZ	1	-7	2	-7	2	-7
Medial Proximal						
FX	80	8	75	4	80	4
FY	141	-13	191	-16	191	-16
FZ	7	-22	9	-22	9	-22
MX	2	-5	3	-7	3	-7
MY	1	-3	1	-3	1	-3
MZ	-1	-18	-1	-18	-1	-18
Medial Distal						
FX	-11	-72	-12	-75	-11	-75
FY	180	-20	198	-82	198	-82
FZ	20	-7	32	-15	32	-15
MX	3	0	4	-1	4	-1
MY	3	-2	4	-2	4	-2
MZ	-1	-9	-2	-11	-1	-11

* The units of loadings for FX, FY, and FZ is newton (N);
and for MX, MY, and MZ is newton-metres (Nm).

The results of FY loadings for a left-side affected patient are the reverse of a right-sided patient, i.e. the lateral upright's FY increases instead of decreases for a LH U-turn (Fig 7.31 & 7.32).

From the results, it is deduced that the upright nearer to the centre of the semi-circle has an increased FY and MY, as compared with the upright further away from the origin of the radius. This applies for both the LH and RH U-turns, and also for both left-side and right-side affected patients. This effect may be due to the patient's leaning more toward the inside of the circle, and hence reducing the loading on the upright further away from the centre of the circle.

7.3.3 Walking Round a Double-Bend.

This activity involves the patient walking along the walkpath, take a right 90° turn, walk a further short distance, and take a left 90° turn before proceeding to the end of the walkway. As previouly, the results of patient MR will be illustrated as a typical example.

At the 90° right turn, the axial load on the medial upright becomes tensile (-70N), and to compensate for this abnormality, the lateral upright has to maintain a higher proportion of the body weight (340N). However, when the patient makes a left 90° turn, the lateral distal section goes into a -120N tensile while the lateral proximal section remains marginally compressive, i.e. carrying body weight. At this instant, the loading at the medial distal section rises to 400N, which is an increase of some 180N, i.e. approximately twice the normal level (Fig 7.33 & 7.34). This magnitude is much higher than the 50% increase anticipated by Henshaw (1976).

The FX and FZ shear forces measured are somewhat similar to those obtained for the U-turn tests, except that any deviation from the normal loading pattern becomes more distinctive. However, the overall maximum loading levels are still below those for normal walking.

The mediolateral bending moment (MX) reaches a new peak at 90° left turn, where the maximum loading level recorded at the medial distal section was 15 Nm, some 4 Nm higher than the 'norm'. Twice the normal level of torsional moment (MY) was also recorded in this test, where the lateral distal section was increased to 9 Nm. There is an overall reduction in the maximum MZ moments recorded. A similar pattern of results has also been recorded in the other 2 right-side affected patients.

Table 7.4 Table Showing the Highest Orthotic Loadings for Patient 'EM'.

Orthotic * Uprights	Normal Walking		Other Activities		Overall Highest Loadings	
	max	min	max	min	max	min
Lateral Proximal						
FX	6	-23	10	-22	10	-23
FY	76	-13	105	-15	105	-15
FZ	24	-7	36	-9	36	-9
MX	1	-1	2	-1	2	-1
MY	2	0	2	0	2	0
MZ	7	-1	9	-1	9	-1
Lateral Distal						
FX	31	-2	43	-2	43	-2
FY	94	-32	144	-48	144	-48
FZ	5	-22	9	-30	9	-30
MX	3	-1	3	-1	3	-1
MY	1	-3	1	-4	1	-4
MZ	6	-3	8	-5	8	-5
Medial Proximal						
FX	0	-33	3	-37	3	-37
FY	78	18	88	-4	88	-4
FZ	20	-3	13	-5	20	-5
MX	4	1	5	0	5	0
MY	2	-1	2	-1	2	-1
MZ	9	1	10	0	10	0
Medial Distal						
FX	50	13	63	10	63	10
FY	107	3	132	-24	132	-24
FZ	20	-7	24	-8	24	-8
MX	2	-2	2	-2	2	-2
MY	3	-1	3	-1	3	-1
MZ	14	1	16	1	16	1

* The units of laodings for FX, FY, and FZ is newton (N);
and for MX, MY, and MZ is newton-metres (Nm).

For the left-side affected patients, the changes in the loading patterns for the two uprights are interchanged as compared with the right-side affected patients, i.e. an increased in the loading level on the lateral upright for a right-sided patient will correspond to a similar increase at the medial upright for a left-sided patient.

A comparison of the results discussed above with those presented in the previous section shows that the severity of loading on some sections of the uprights increases with the turning angle undertaken by the patient.

7.3.4 The Low-Platform Test.

This involves the patient in walking along the walkway, stepping onto a low-platform, and transversing it before stepping down to continue the walk. All the subjects participating in the test used the procedure which is generally considered to be the safest for negotiating this obstacle. The 'sound leg' was used to step on and the 'KAFO leg' to step down from the low-platform.

For patient MR, the lateral upright axial load (FY) was reduced to about half its normal value just prior to stepping onto the platform. As the patient stepped down from the platform, the FY loading of the same upright increased to 325N, i.e. marginally higher than the level straight line walking test. The proportion of weight distribution between the uprights did not differ significantly from its normal ratio of 6:4, with the lateral upright carrying the higher load (Fig 7.35 & 7.36). The loadings in other directions are generally lower than for normal level walking, especially at stance phase of the affected side just prior to stepping onto the platform, where the loadings are approximately half of the normal values. This reduction of loading, as against the generally held assumption of higher loadings, is probably due to the reason that MR is not a 'careless walker'. It was observed that he descended from the platform in a swift and smooth motion, thus avoiding the 'dynamic loading' characteristics as seen in some patients, such as DR, whose results will be briefly discussed below.

The MZ moment loading level for patient DR was more than twice the normal levels just prior to stepping on and just after stepping down from the platform. The maximum MZ moment recorded is 12 Nm. DR

Table 7.5 Table Showing the Highest Orthotic Loadings for Patient 'DR'.

Orthotic * Uprights.	Normal Walking		Other Activities		Overall Highest Loadings	
	max	min	max	min	max	min
Lateral Proximal						
FX	36	-24	35	-39	36	-39
FY	95	-99	190	-22	190	-99
FZ	27	-2	51	3	51	-2
MX	3	-1	3	-1	3	-1
MY	0	-2	1	-3	1	-3
MZ	21	-8	5	-10	21	-10
Lateral Distal						
FX	20	-18	44	-25	44	-25
FY	124	-27	206	-19	206	-27
FZ	16	-27	-1	-47	16	-47
MX	4	-2	3	-3	4	-3
MY	1	-2	1	-4	1	-4
MZ	7	-3	12	-4	12	-4
Medial Proximal						
FX	47	-9	51	-34	51	-34
FY	71	-44	40	-122	71	-122
FZ	30	-14	47	-12	47	-14
MX	4	-1	3	-4	4	-4
MY	3	-2	5	-2	5	-2
MZ	6	-7	11	-7	11	-7
Medial Distal						
FX	25	-23	48	-25	48	-25
FY	140	-89	85	-125	140	-125
FZ	2	-57	2	-41	2	-57
MX	1	-1	1	-2	1	-2
MY	2	-2	4	-1	4	-2
MZ	6	-5	10	-5	10	-5

* The untis of loadings for FX, FY, and FZ is Newton (N);
and for MX, MY, and MZ is newton-metres (Nm).

tends to step down from the platform relatively hard and allows the body weight to 'drop' onto the orthosis, thus producing the large spike of twice the normal value, as shown in Fig 7.37 & 7.38 .

The anterioposterior bending moments of patients EM and DC also experienced an increase in the loading at stepping down from the platform. The magnitude of the increase are 50% and 45% of the normal MZ values for EM and DC respectively. For patient DC, the medial upright is fully unloaded while the lateral side has its FY loading marginally increased when he steps down from the platform. This, therefore, changes the weight distribution on the orthosis.

7.3.5 Ascending and Descending Slopes.

The finding of this test on all the participating patients have indicated unanimously that the loadings on the uprights are much higher while descending than ascending the slopes.

The results obtained for patient MR show a decrease in the FY loading on both the uprights, as compared to the level straight line walking tests, while ascending the ramp. The lateral FY reduction is approximately 50% (Fig 7.39 & 7.40). The first and the third peaks on the loading characteristics have also been reduced to the same level as the much smaller second peak, at approximately 150N. On the medial side, the third peak has been greatly decreased to well below the second peak.

On descending the slope, the first peak of the lateral upright FY loading increases to 380N, which is an increase of some 30% of the normal level walking tests (Fig 7.41 & 7.42). However, the subsequent peaks are reduced into a continuous downward slope of the FY loading curve. Similarly, the first peak of the medial side has also been increased to 270N.

In the MZ channels, the loadings on the lateral proximal section, which is the most highly loaded one, has been reduced by some 40% in the upslope run (Fig 7.43). The 2 peaks are reduced considerably to form a single rounded peak at -12 Nm. When coming down the slope, MZ recorded a new maximum of -28 Nm, an increased of 15% from the normal value (Fig 7.44). Unlike the upward run, the first peak is greatly amplified to -28 Nm, thereafter the curve slopes away to a plateau of some -17 Nm at foot-flat. The second peak has completely disappeared.

Table 7.6 Table Showing the Highest Orthotic Loadings for Patient 'DC'.

Orthotic * Uprights.	Normal Walking		Other Activities		Overall Highest Loadings	
	max	min	max	min	max	min
Lateral Proximal						
FX	91	0	108	15	108	0
FY	94	-29	103	-12	103	-29
FZ	37	0	47	0	47	0
MX	3	-3	1	-3	3	-4
MY	3	-2	3	-1	3	-2
MZ	2	-17	1	-19	2	-19
Lateral Distal						
FX	21	-43	7	-57	21	-57
FY	138	-31	176	-36	176	-36
FZ	9	-10	10	-10	10	-10
MX	1	-4	0	-4	1	-4
MY	3	0	3	0	3	0
MZ	3	-19	0	-15	3	-19
Medial Proximal						
FX	70	7	81	10	81	7
FY	126	-20	126	-29	126	-29
FZ	4	-30	3	-29	4	-30
MX	7	-1	6	-1	7	-1
MY	0	-3	0	-2	0	-3
MZ	10	-13	10	-11	10	-13
Medial Distal						
FX	-5	-53	-3	-51	-3	-53
FY	131	-35	143	-44	143	-44
FZ	-2	-20	-1	-22	-1	-22
MX	4	0	4	0	4	0
MY	2	-3	2	-4	2	-4
MZ	-1	-16	0	-16	0	-16

* The units of loadings for FX, FY, and FZ is newton (N); and for MX, MY, and MZ is newton-metres (Nm).

7.3.6 Ascending and Descending Stairs.

With much regret, this test was carried out only on two patients (EM and DR), due to difficulties in arranging the patients for the test, and also in the suitability of the patients to perform the activity.

The results obtained from patient EM on the ascending and descending stairs show two set of loading patterns which are very similar in shape and magnitude to those obtained from normal level walking tests. Moreover, EM consistently shows very low orthotic loadings for all the other activities. Thus no definite trend in the results can be drawn.

A similar situation also occurs for patient DR as with EM, where no definite pattern of loading characteristics can be drawn. However, for the axial load, considerable dynamic impact loadings were observed in the descending run, which causes the lateral distal section to rise to a peak of 200N, i.e. twice the normal value.

Table 7.1 to 7.6 shows the highest and lowest orthotic loading levels recorded in each section of the uprights, for 6 of the patients tested. The tables aimed to provide a comparison for the loadings obtained in the normal straight line walking with those measured in other ambulatory activities. The overall maximum and minimum loadings recorded are also tabulated.

Note: Section 7.4 follows after Fig 7.1 to 7.44 .

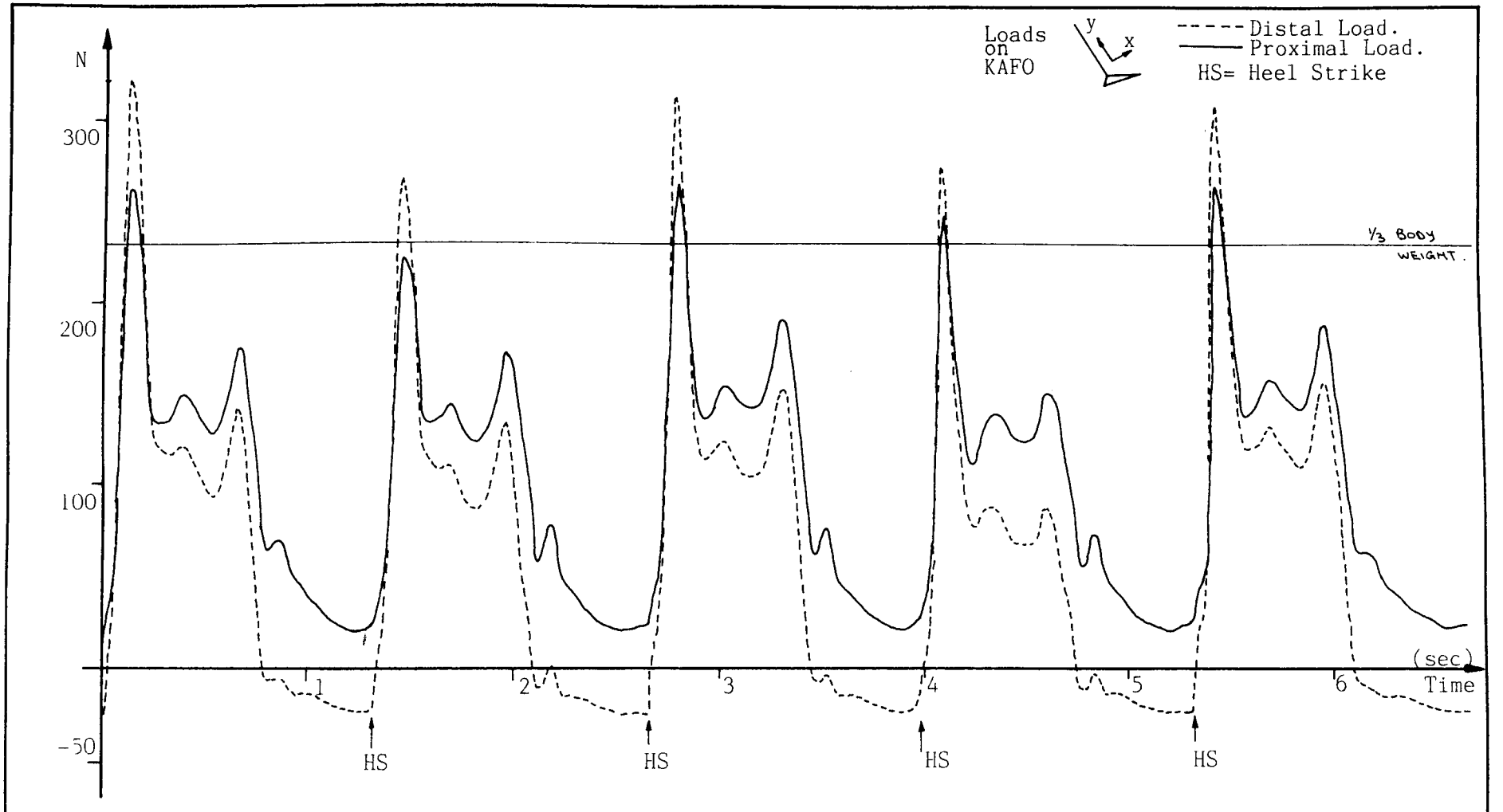


FIG. 7.1 FY Loadings on Lateral Upright of Patient MR.

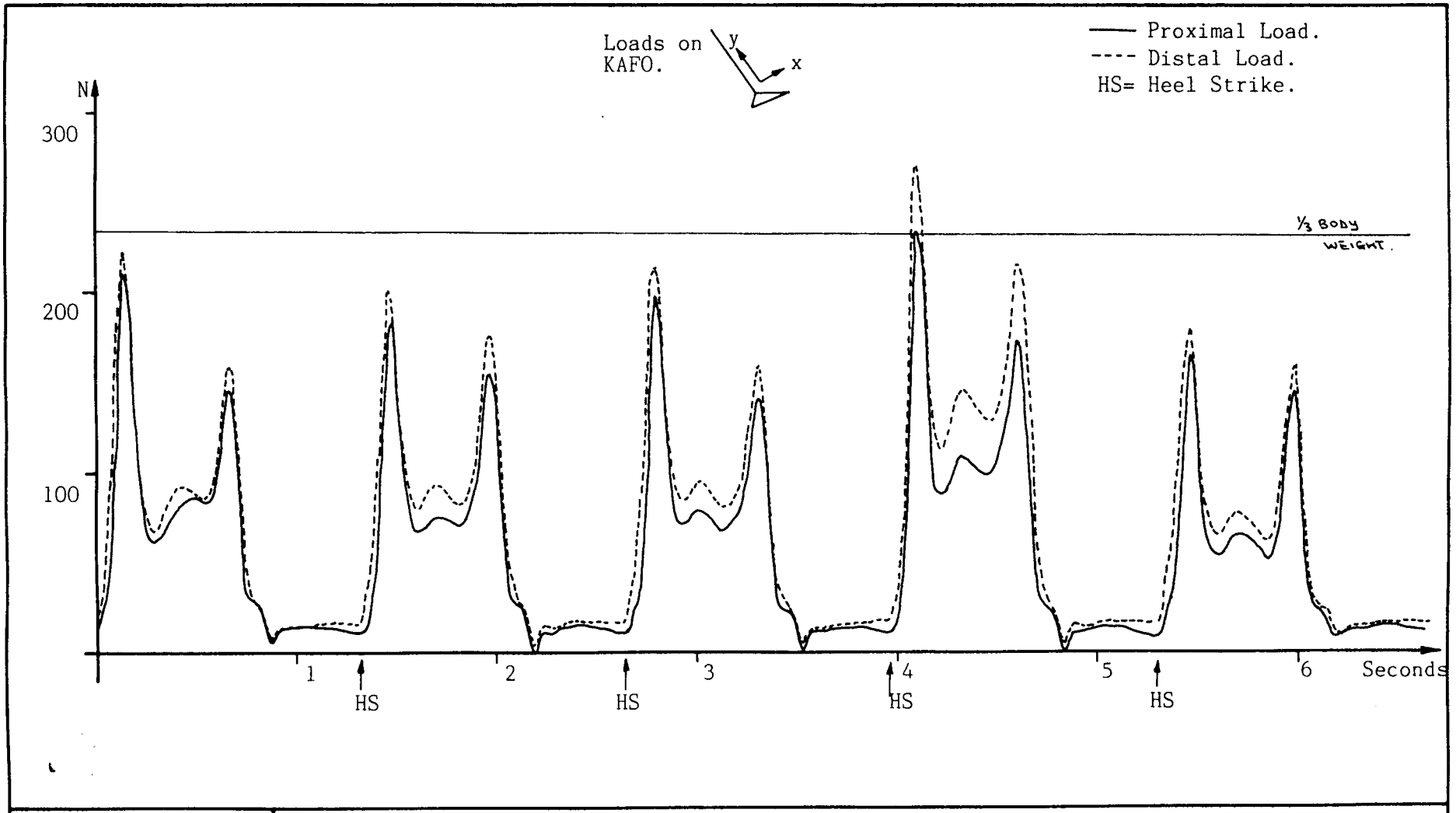


FIG 7.2 | FY Loadings on Medial Upright of Patient MR.

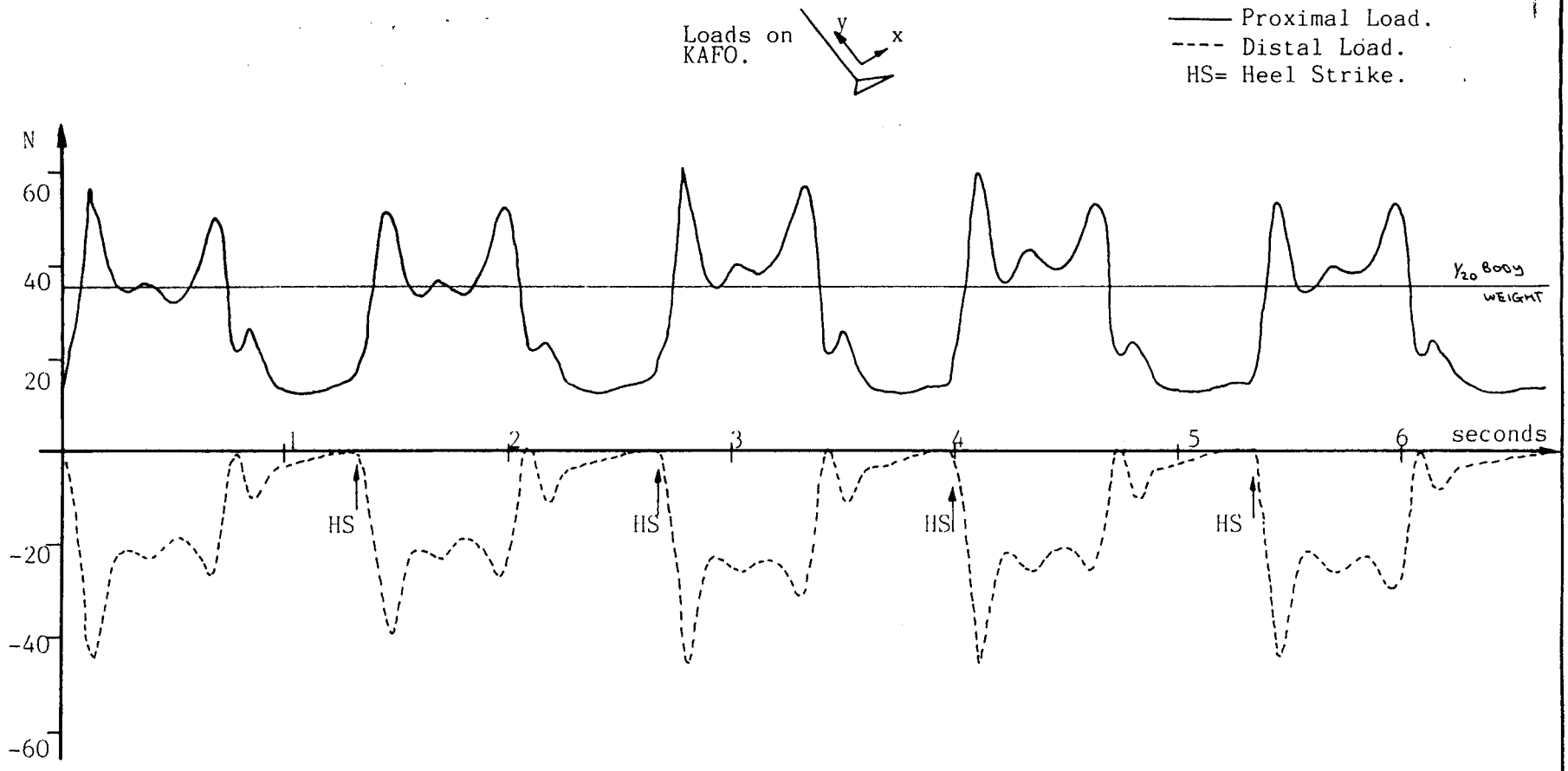


FIG. 7.3 | FX Loadings on Lateral Upright of Patient MR.

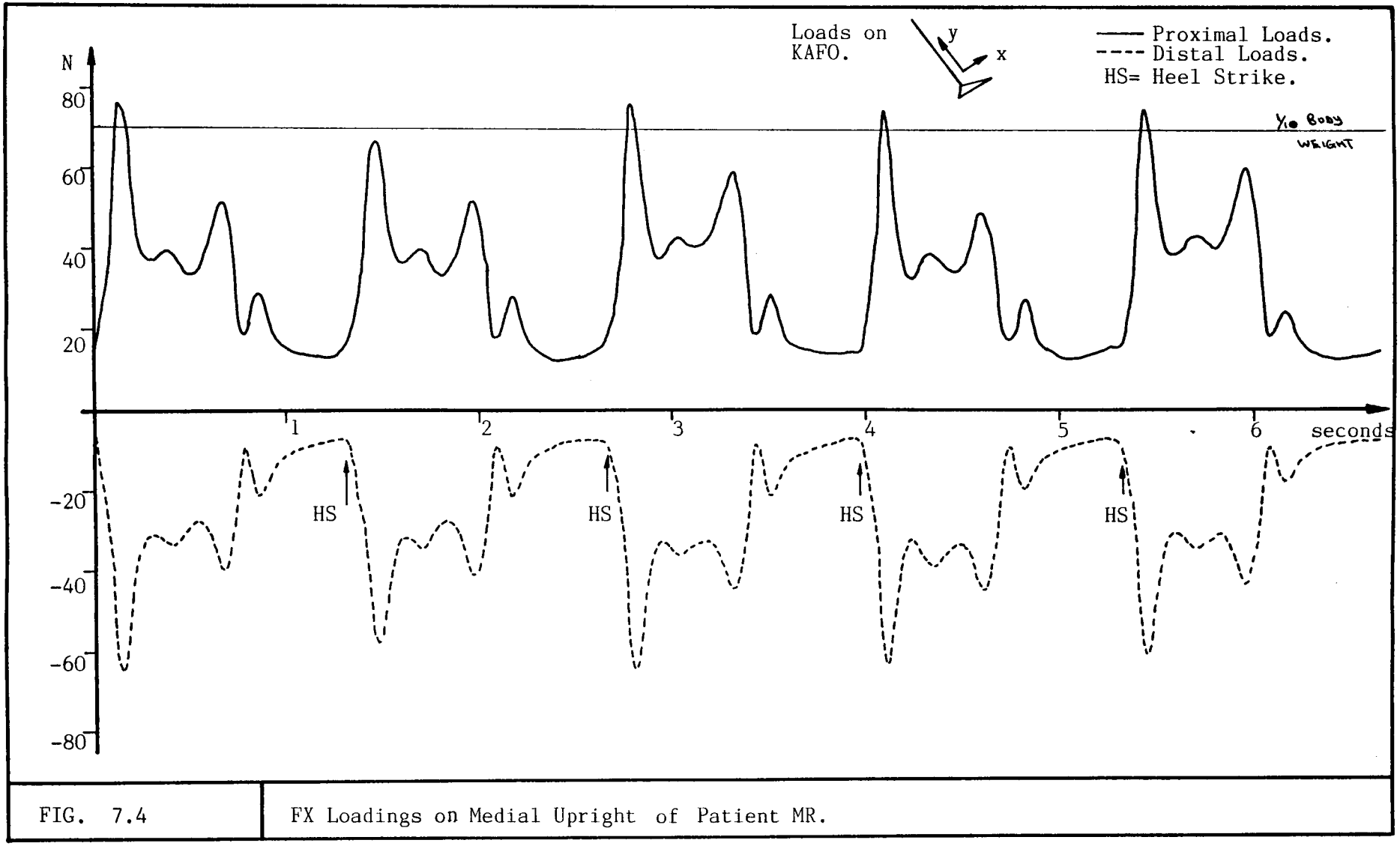


FIG. 7.4 FX Loadings on Medial Upright of Patient MR.

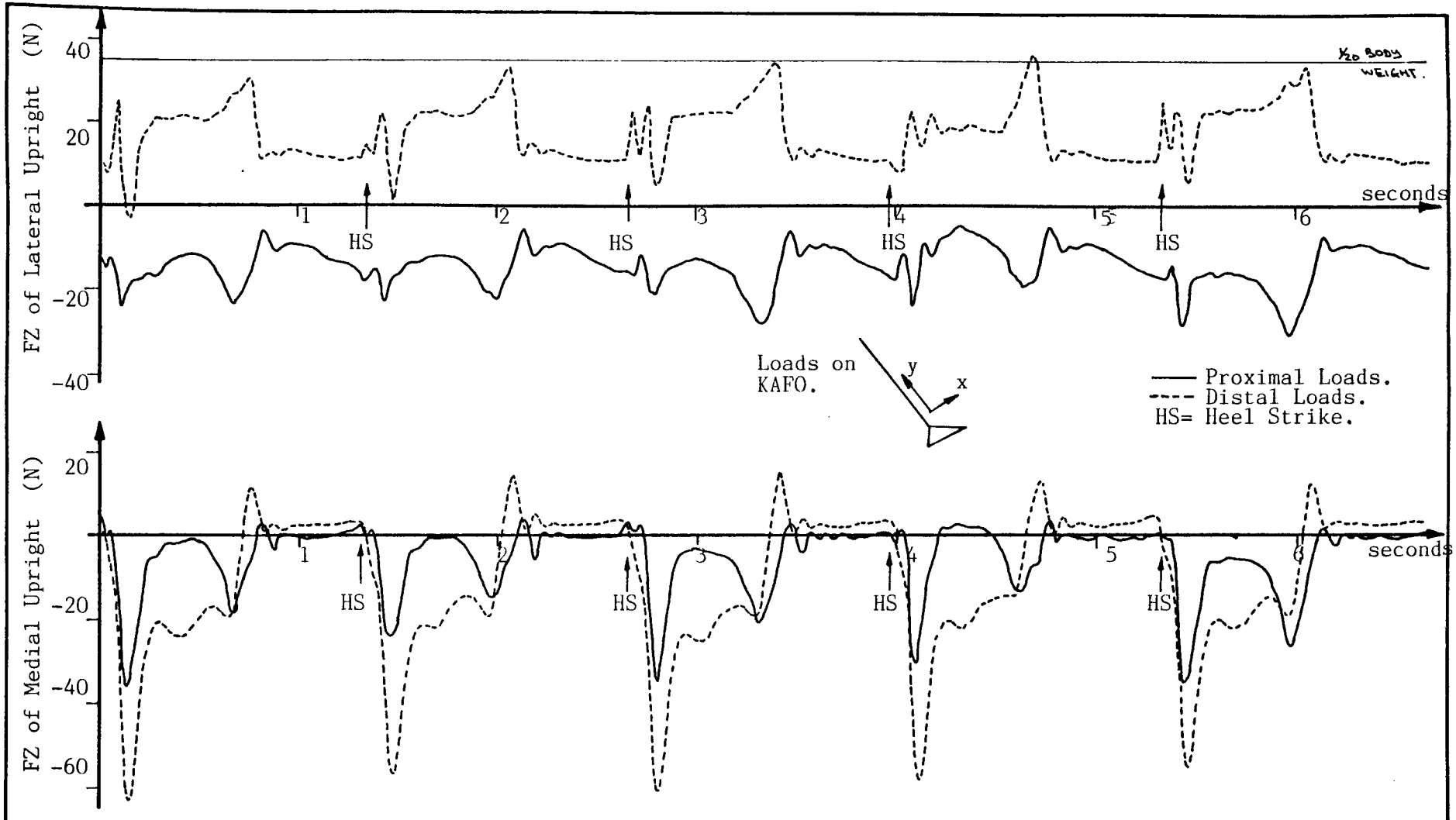


FIG. 7.5 FZ loadings on Lateral and Medial Uprights of Patient MR.

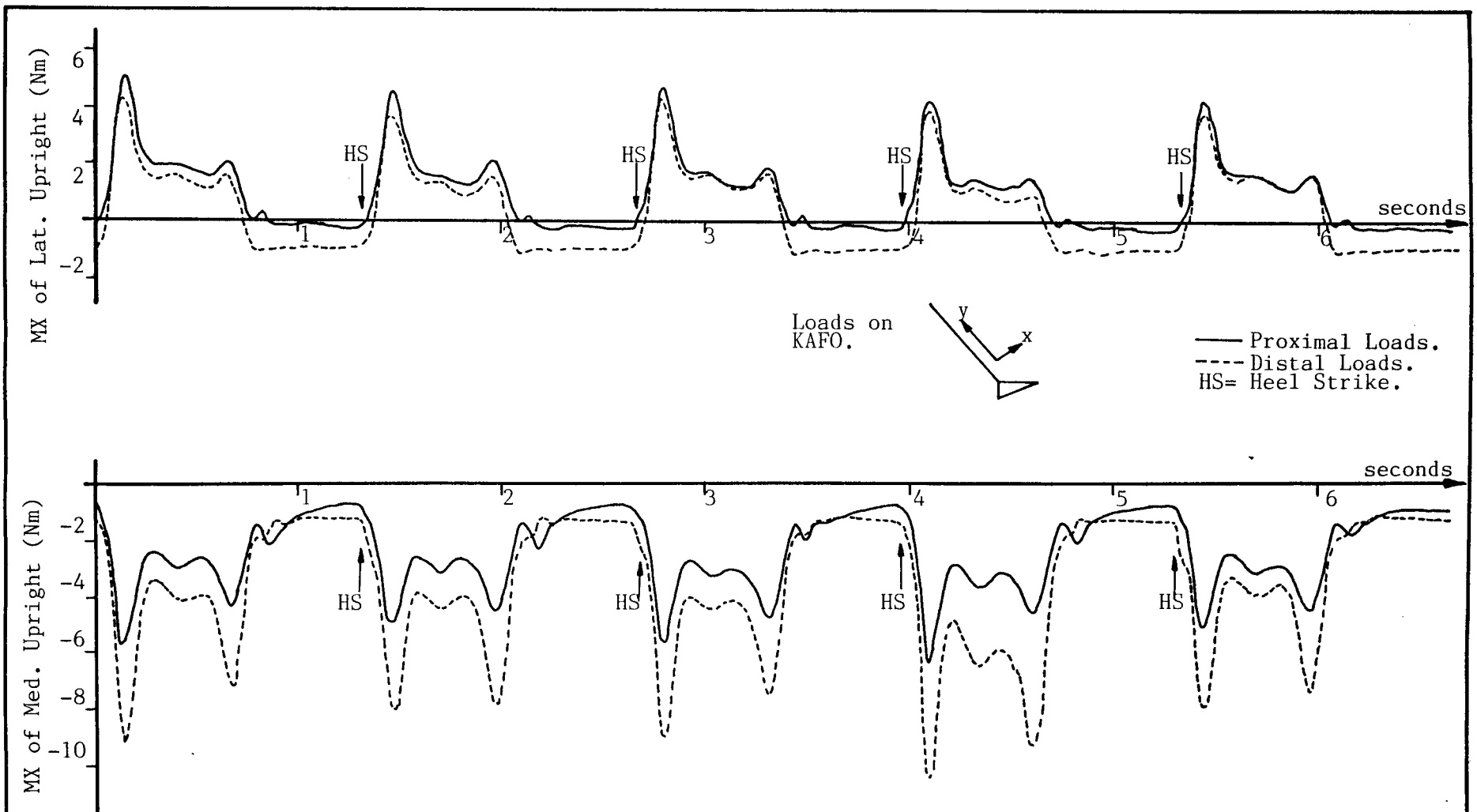


FIG. 7.6 | MX Loadings on Lateral and Medial Uprights of Patient MR.

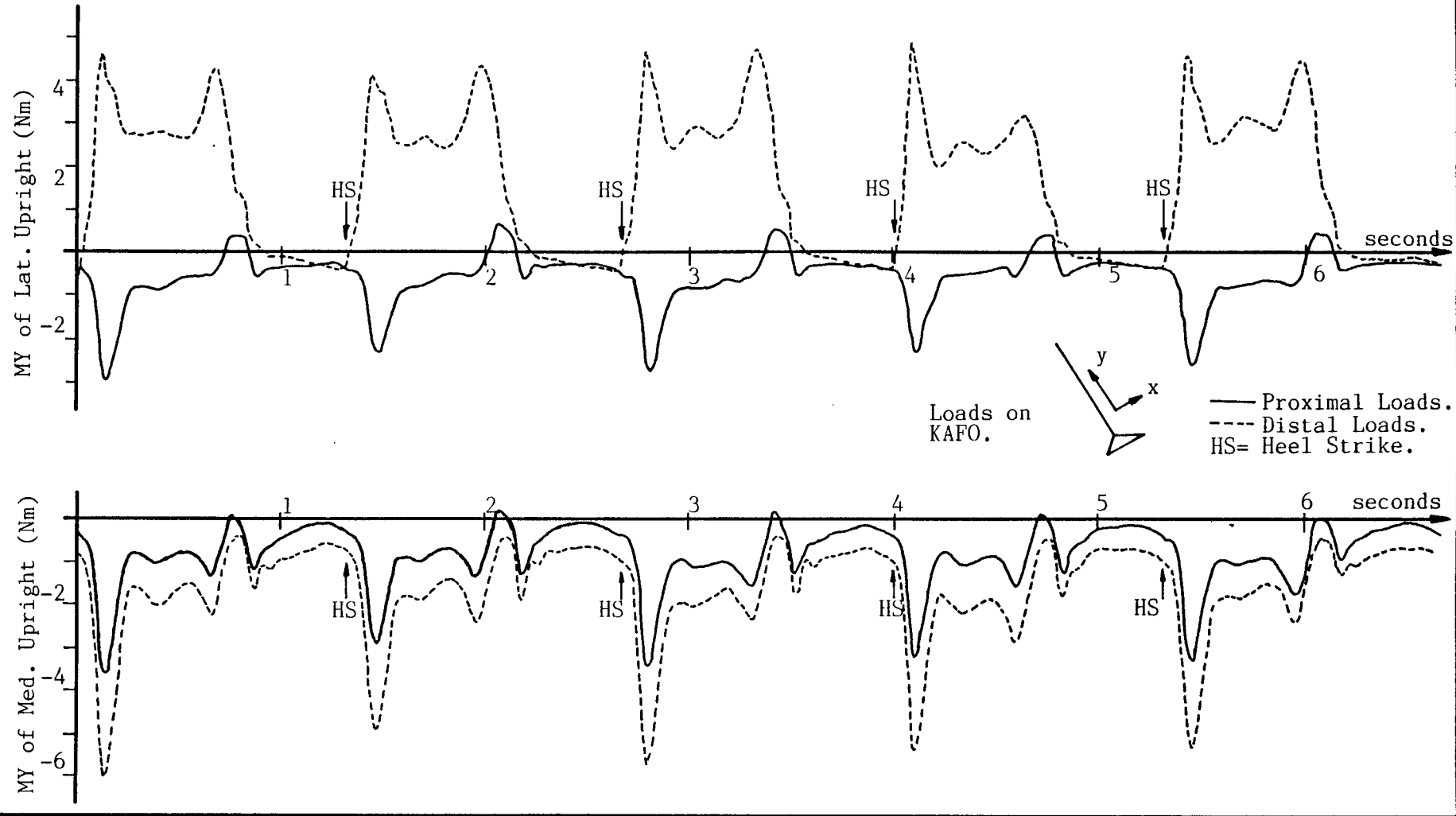


FIG. 7.7 MY Loadings on Lateral and Medial Uprights of Patient MR.

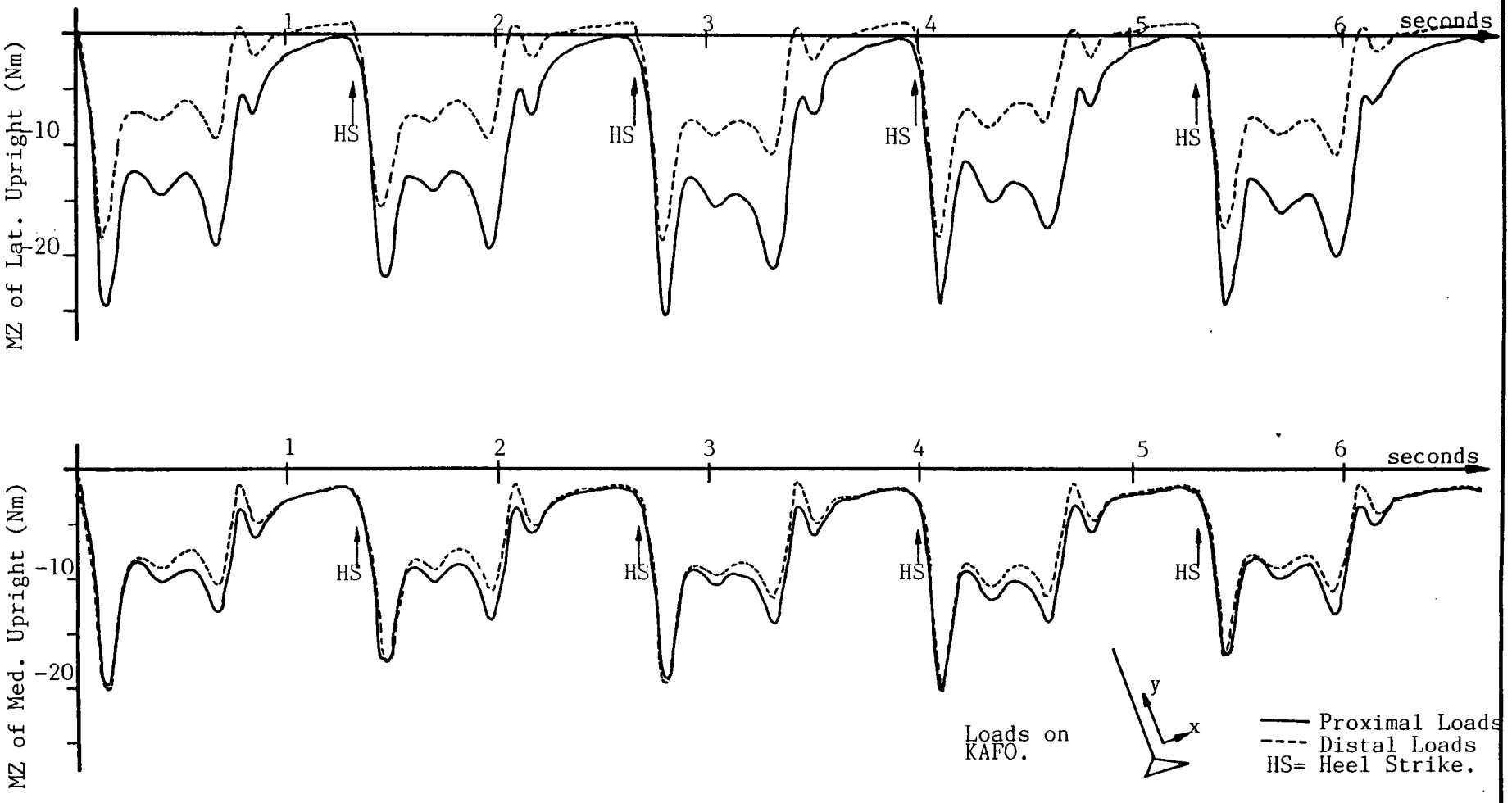


FIG. 7.8 MZ Loadings on Lateral and Medial Uprights of Patient MR.

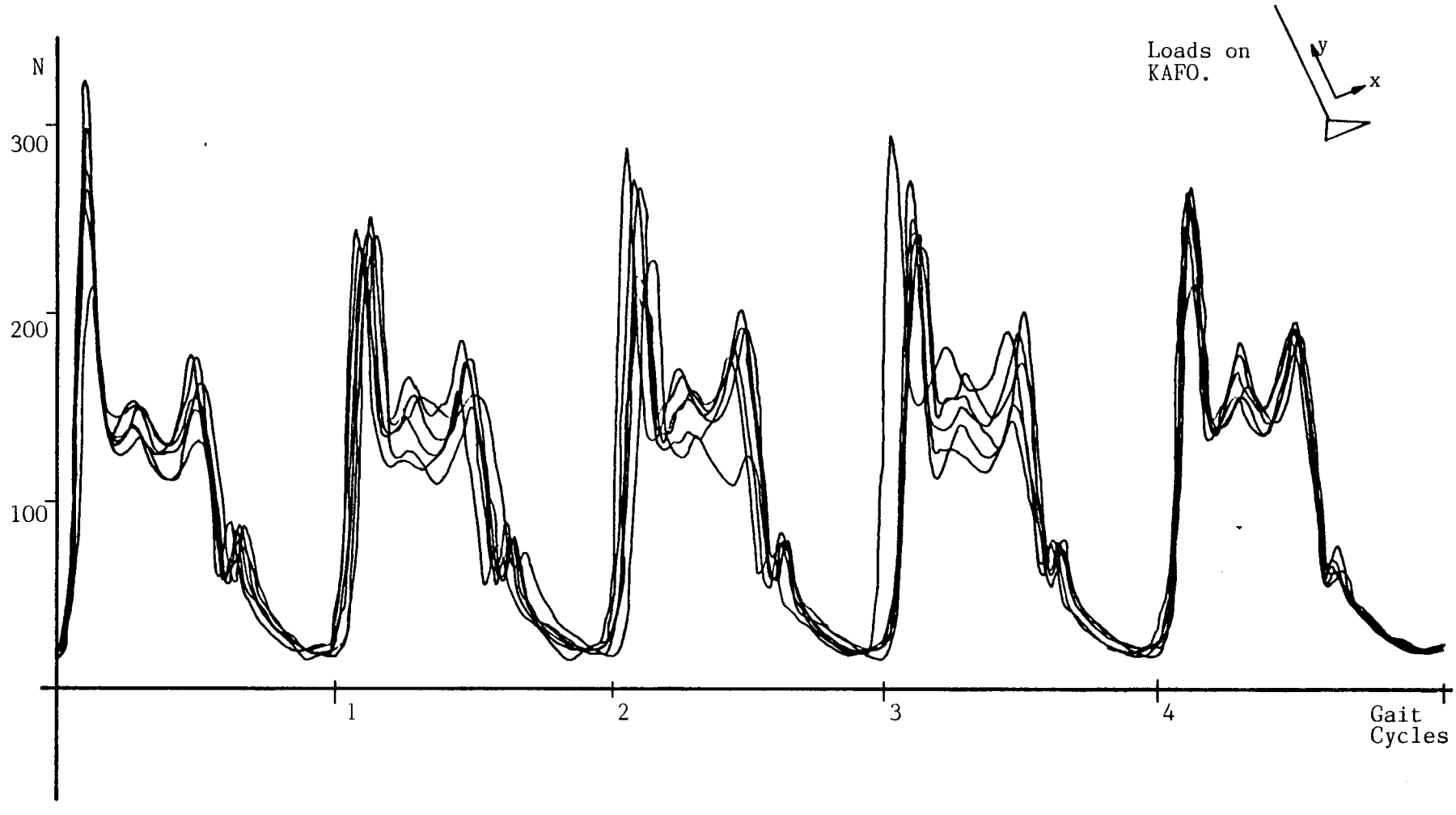


FIG. 7.9 Results of Lateral Proximal FY Loading from 6 Tests (Patient MR).

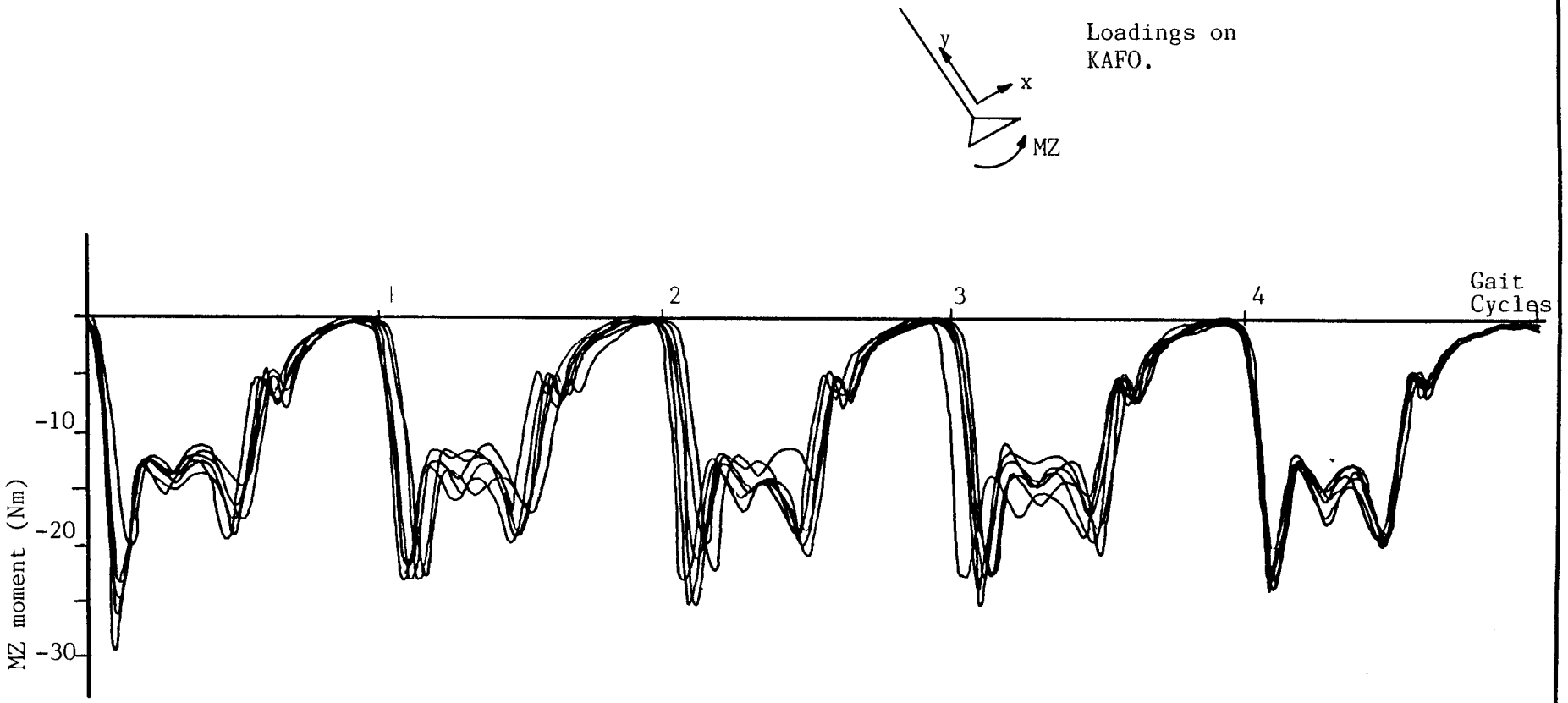


FIG. 7.10 Results of Lateral Proximal MZ loading from 6 Tests (Patient MR).

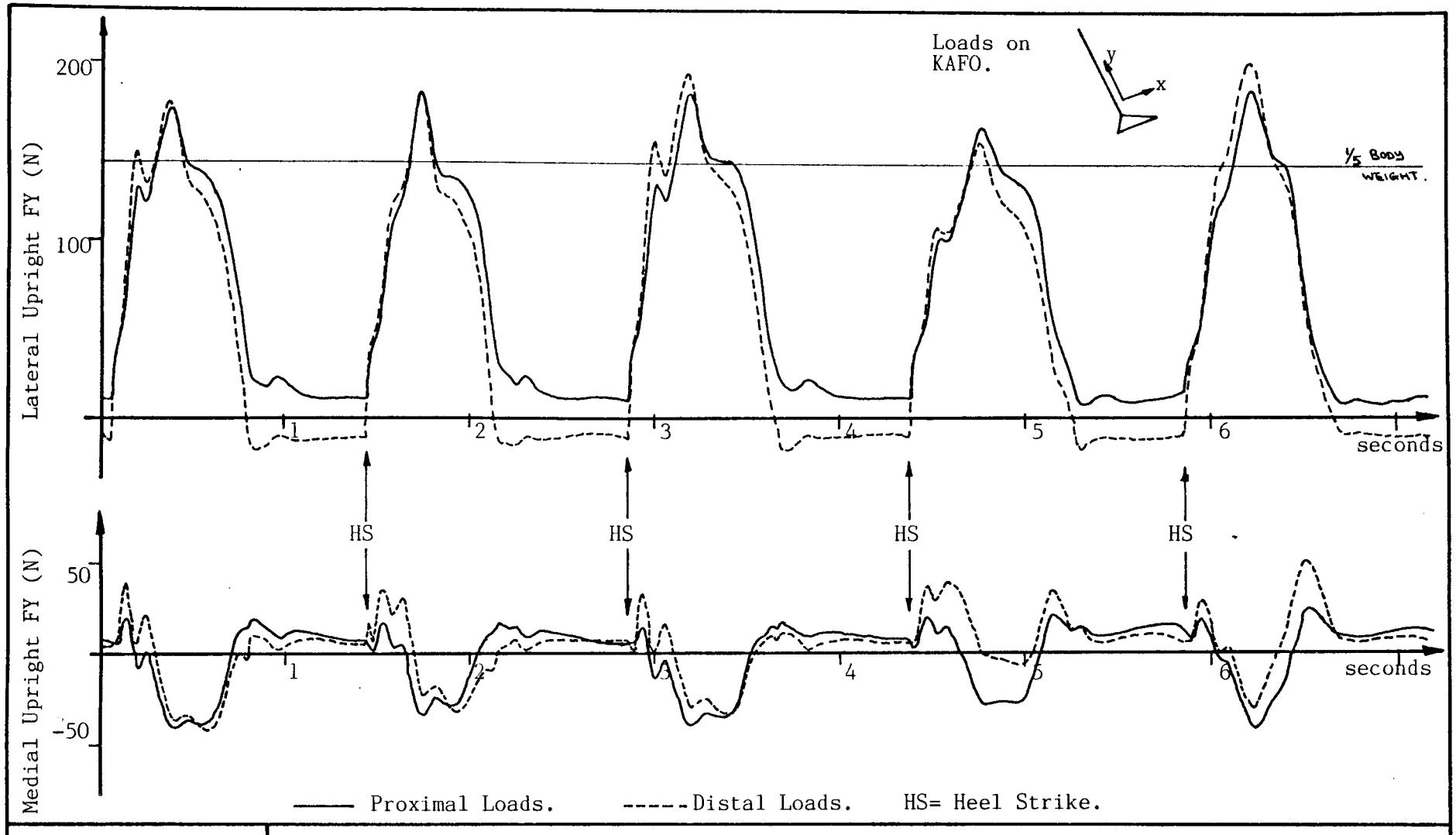


FIG. 7.11 FY Loadings on Lateral And Medial Uprights of Patient WI.

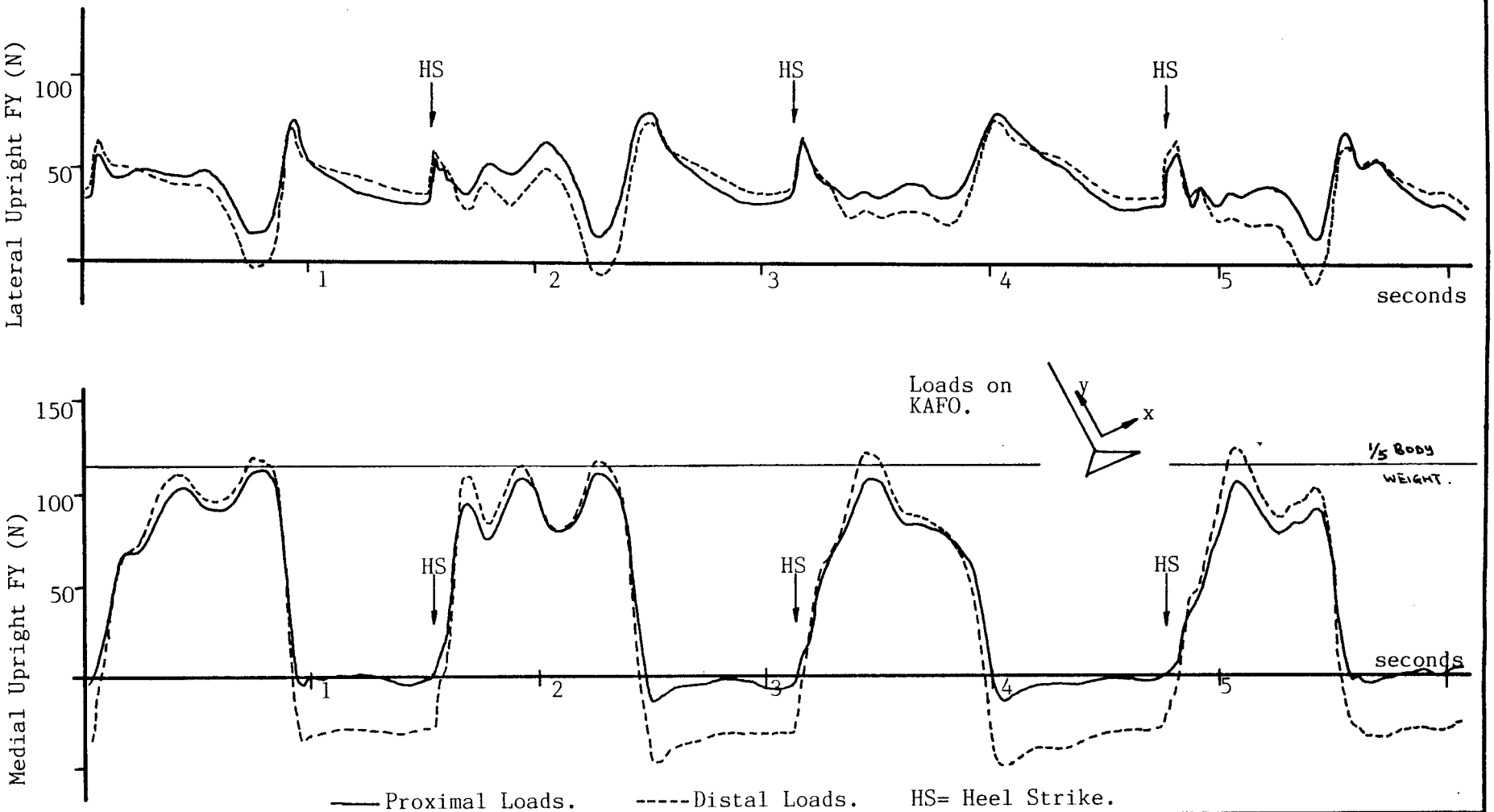


FIG. 7.12 FY Loadings on the Lateral and Medial Uprights of Normal Subject CR.

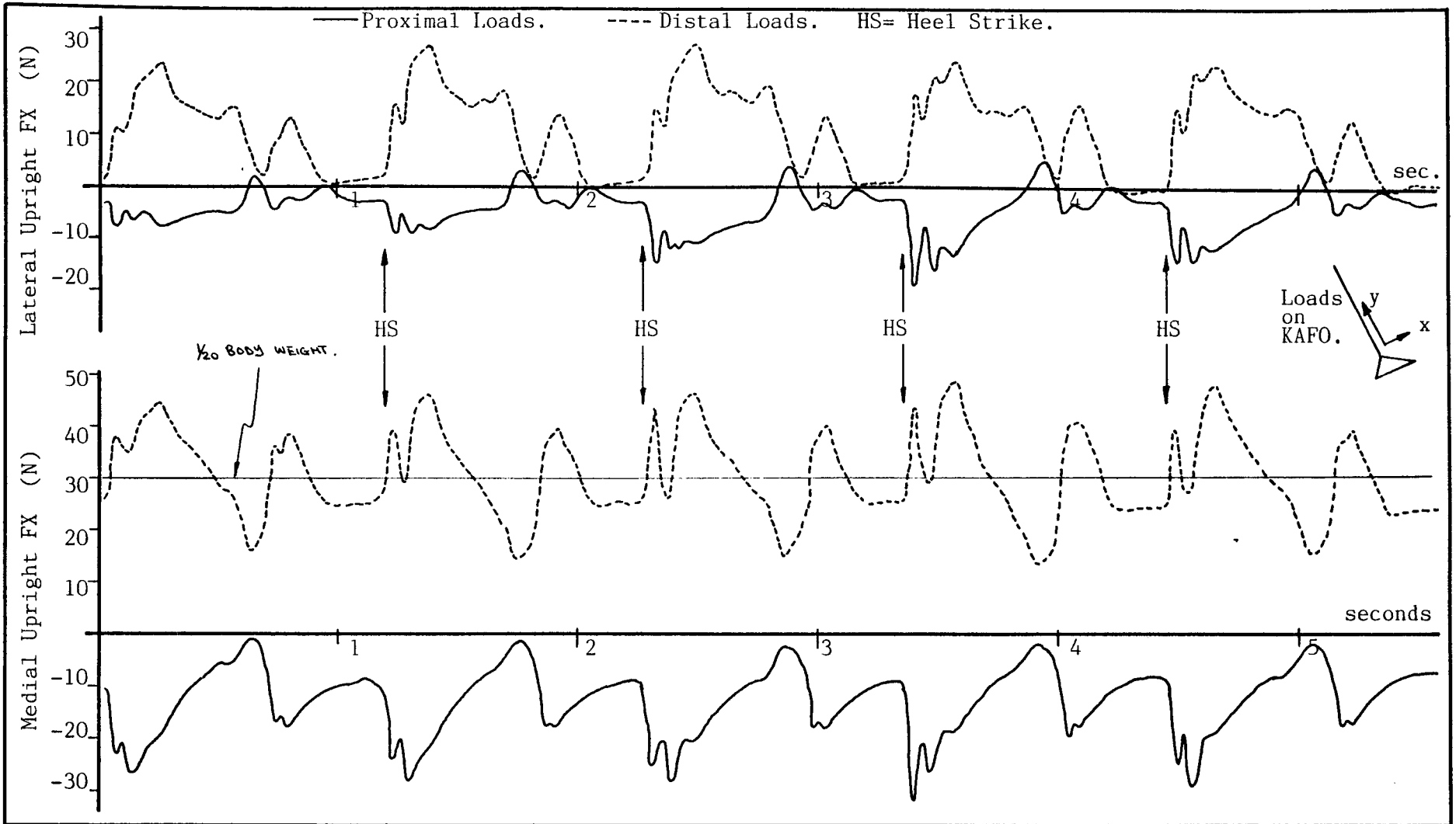


FIG. 7.13 FX Loadings on the Lateral and Medial Uprights of Patient EM.

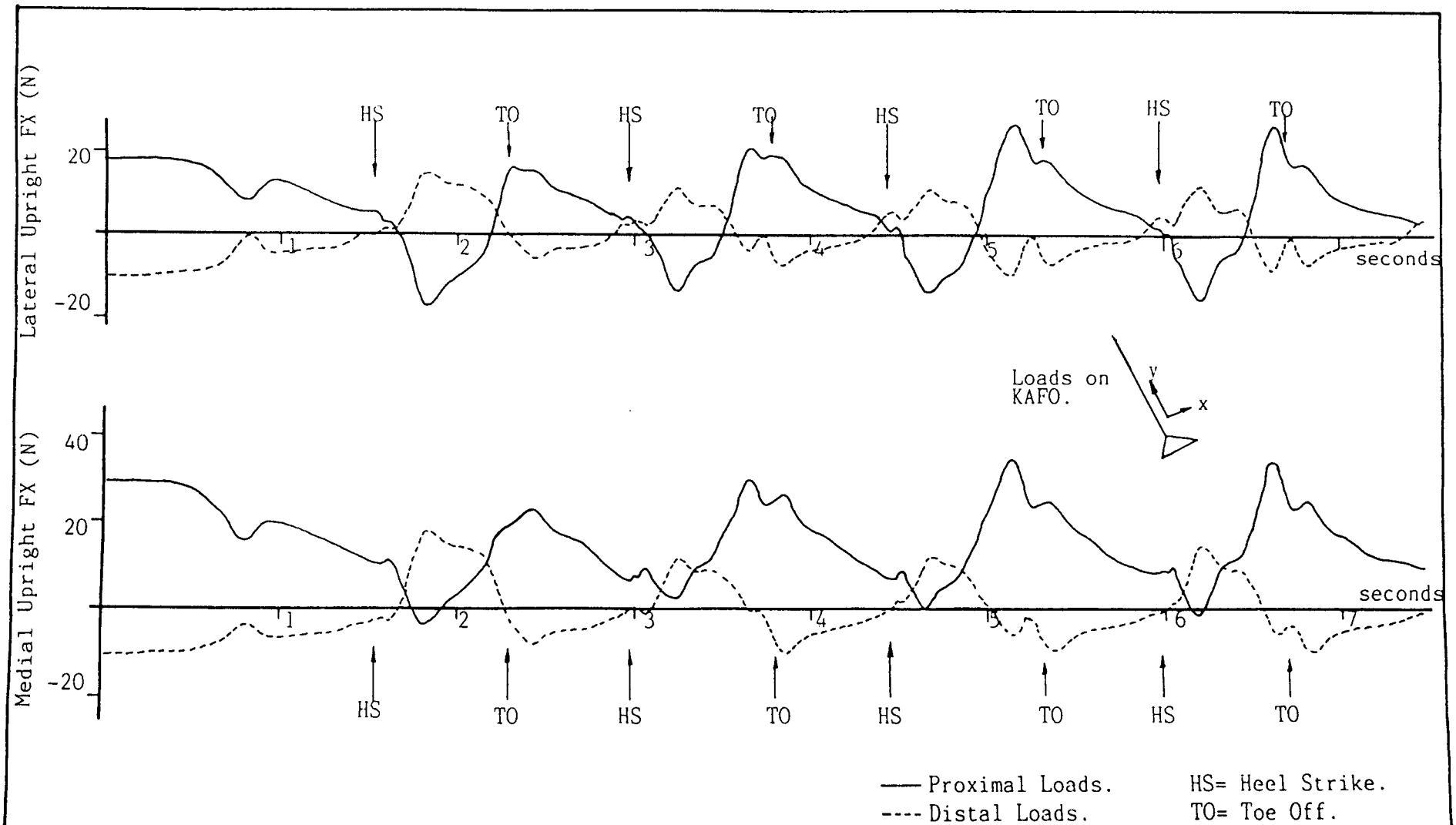


FIG. 7.14 FX Loadings on the Lateral and Medial Uprights of Patient DR. ("Start Walking")

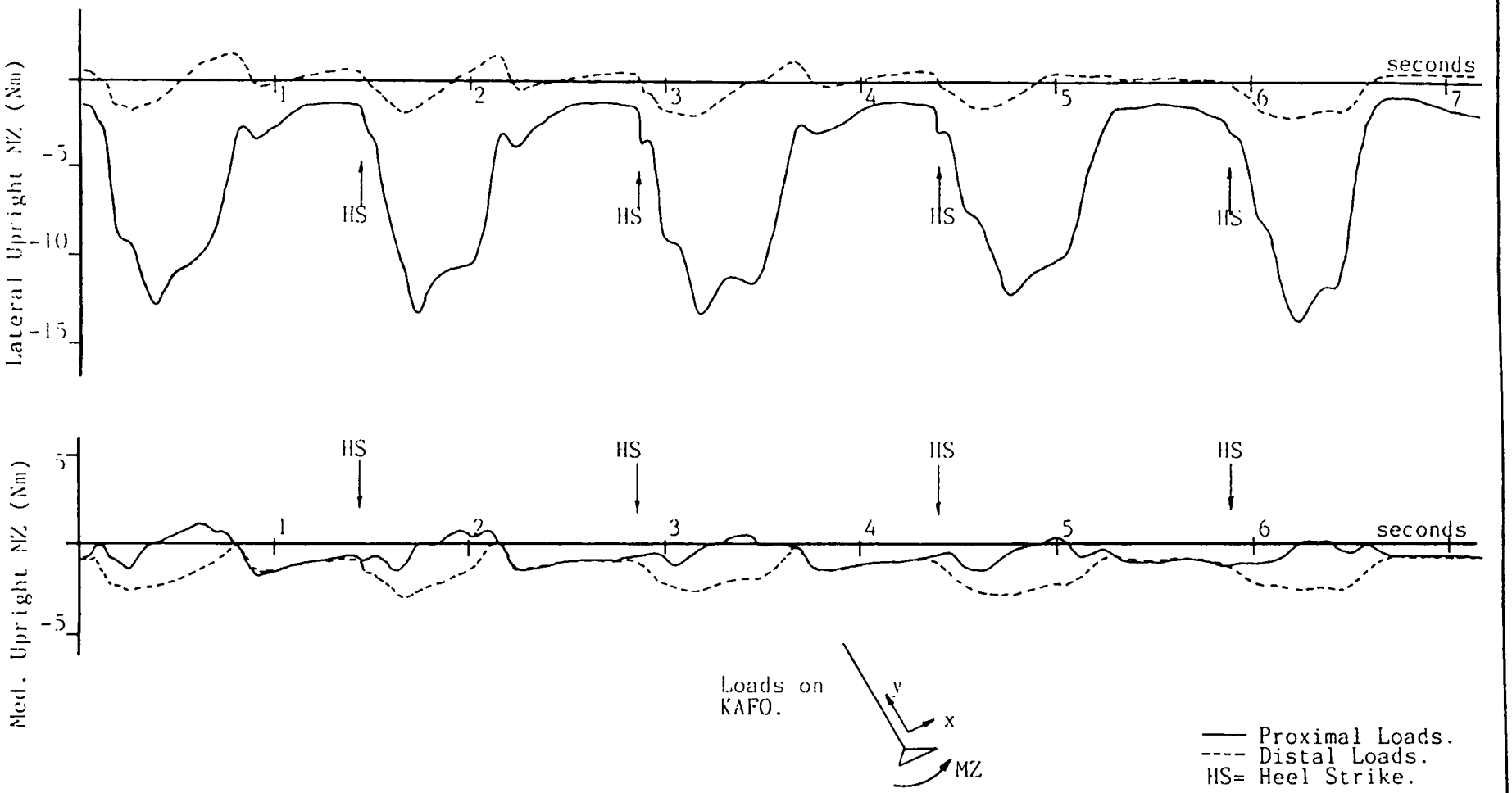


FIG. 7.15 MZ Loadings on the Lateral and Medial Uprights of Patient WI.

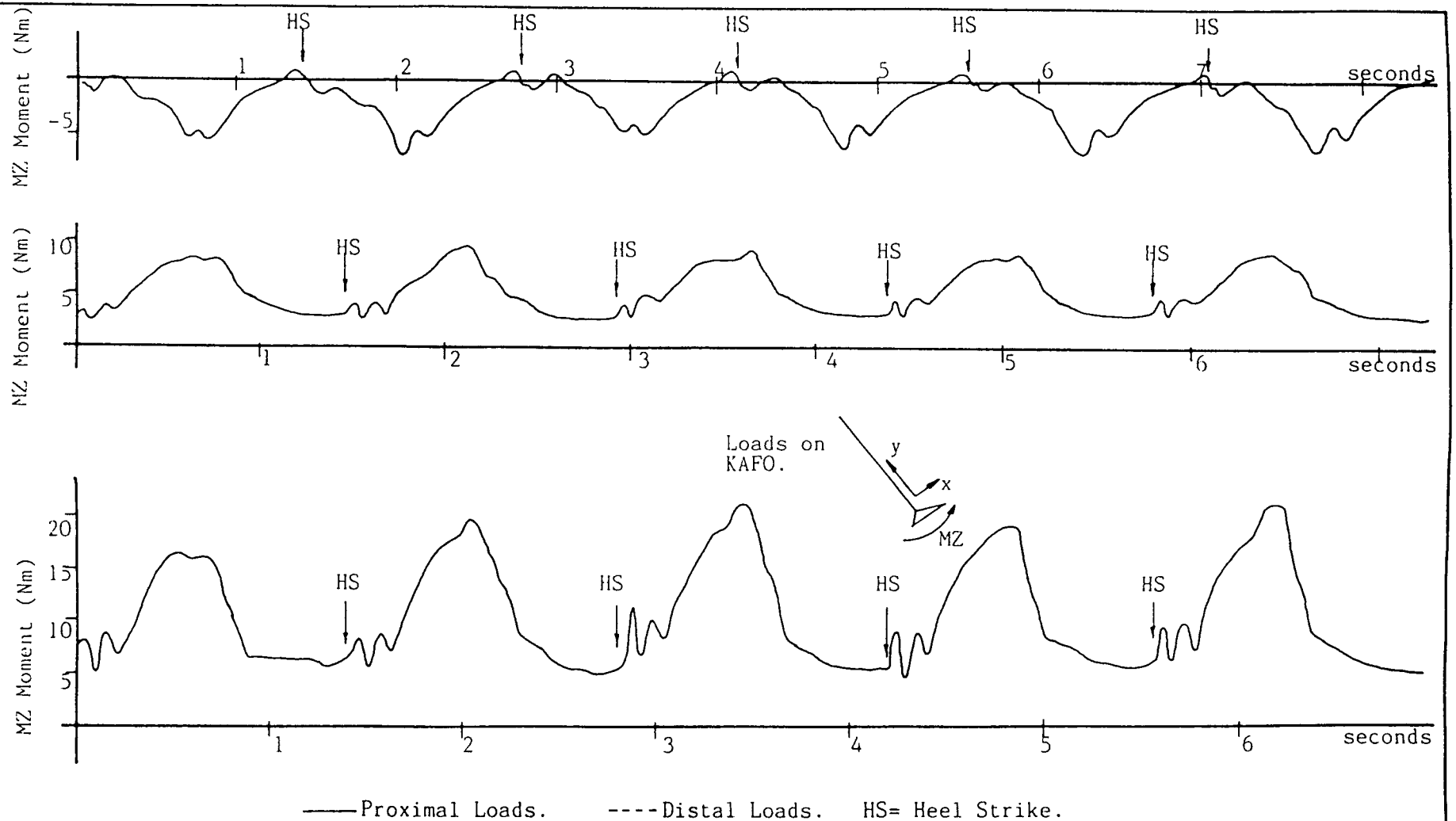
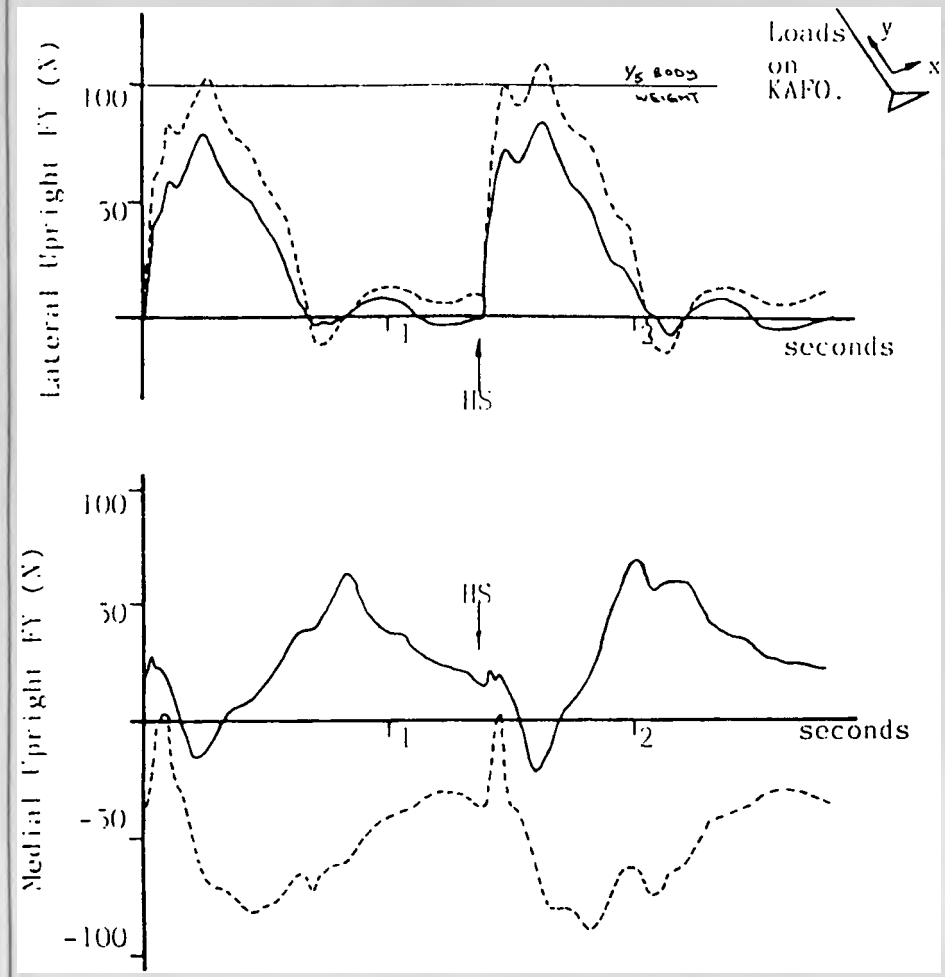


FIG. 7.16 Results of Lateral Proximal MZ Loadings for Patient DR Obtained from 3 Different Occasions.



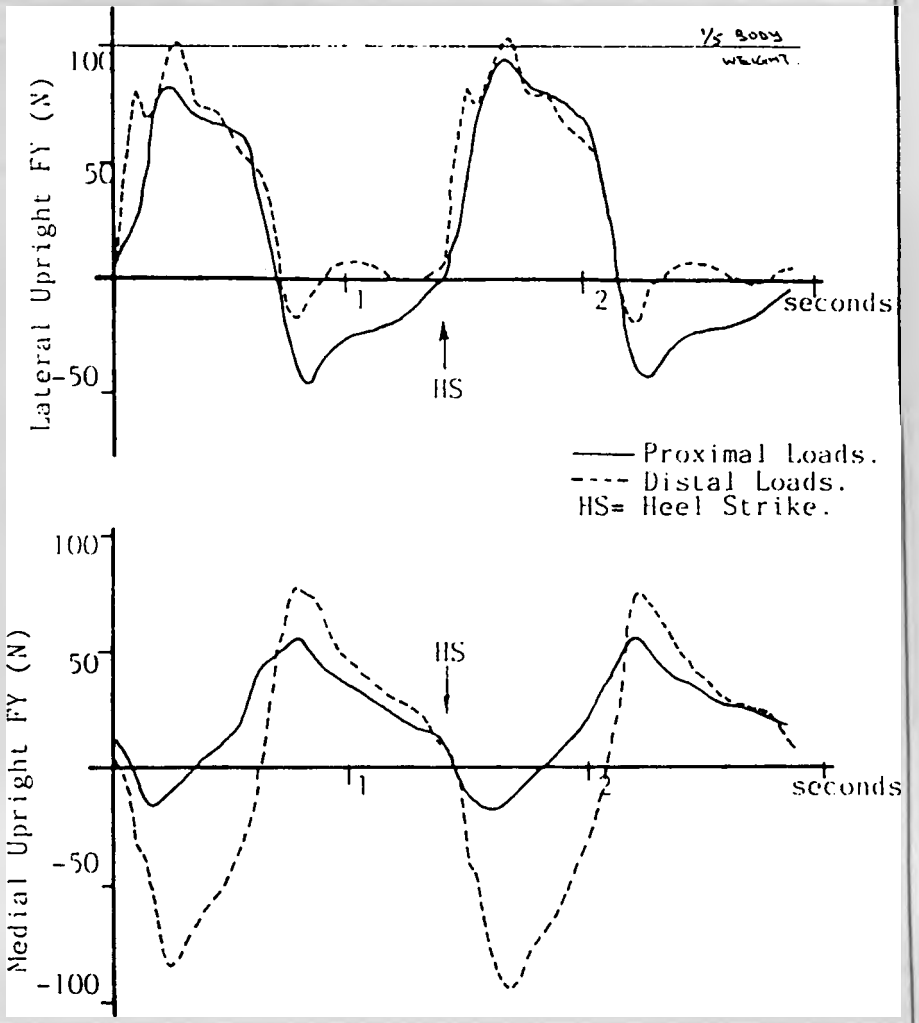


FIG. 7.17 Results of FY Loadings for Patient DR Obtained from 2 Different Occasions.

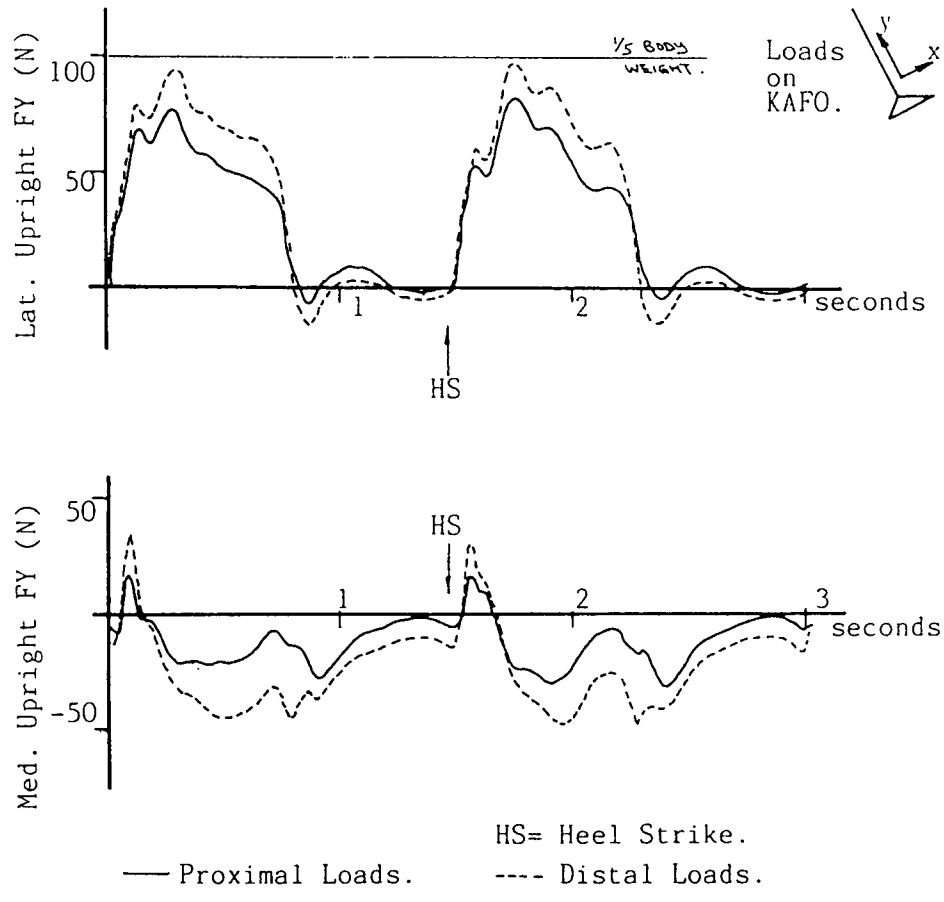


FIG. 7.18	Results of FY Loadings for Patient DR. (c.f. FIG. 7.17)

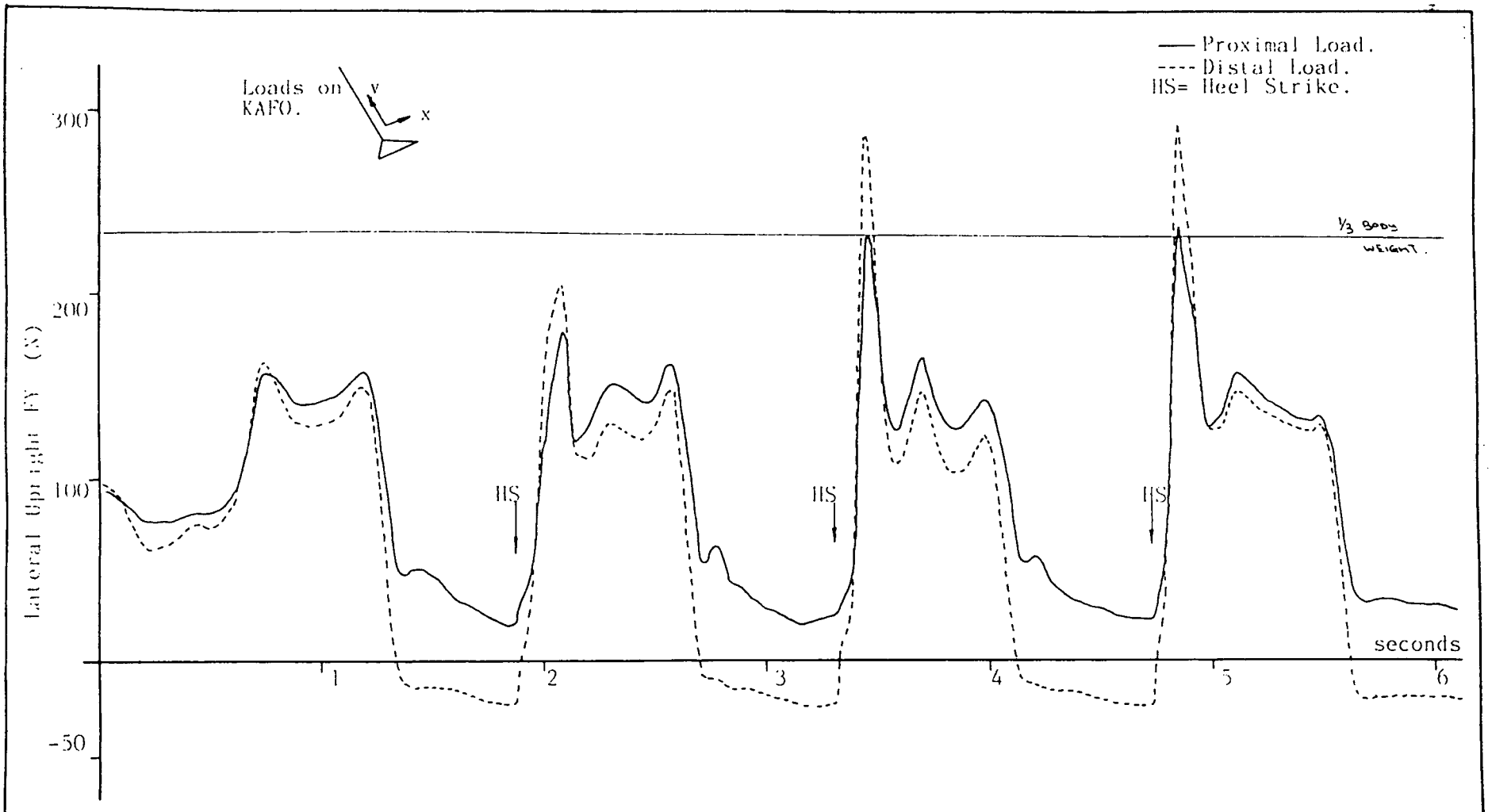


FIG. 7.19 FY Loadings on the Lateral Upright of Patient MR. ("Start Walking")

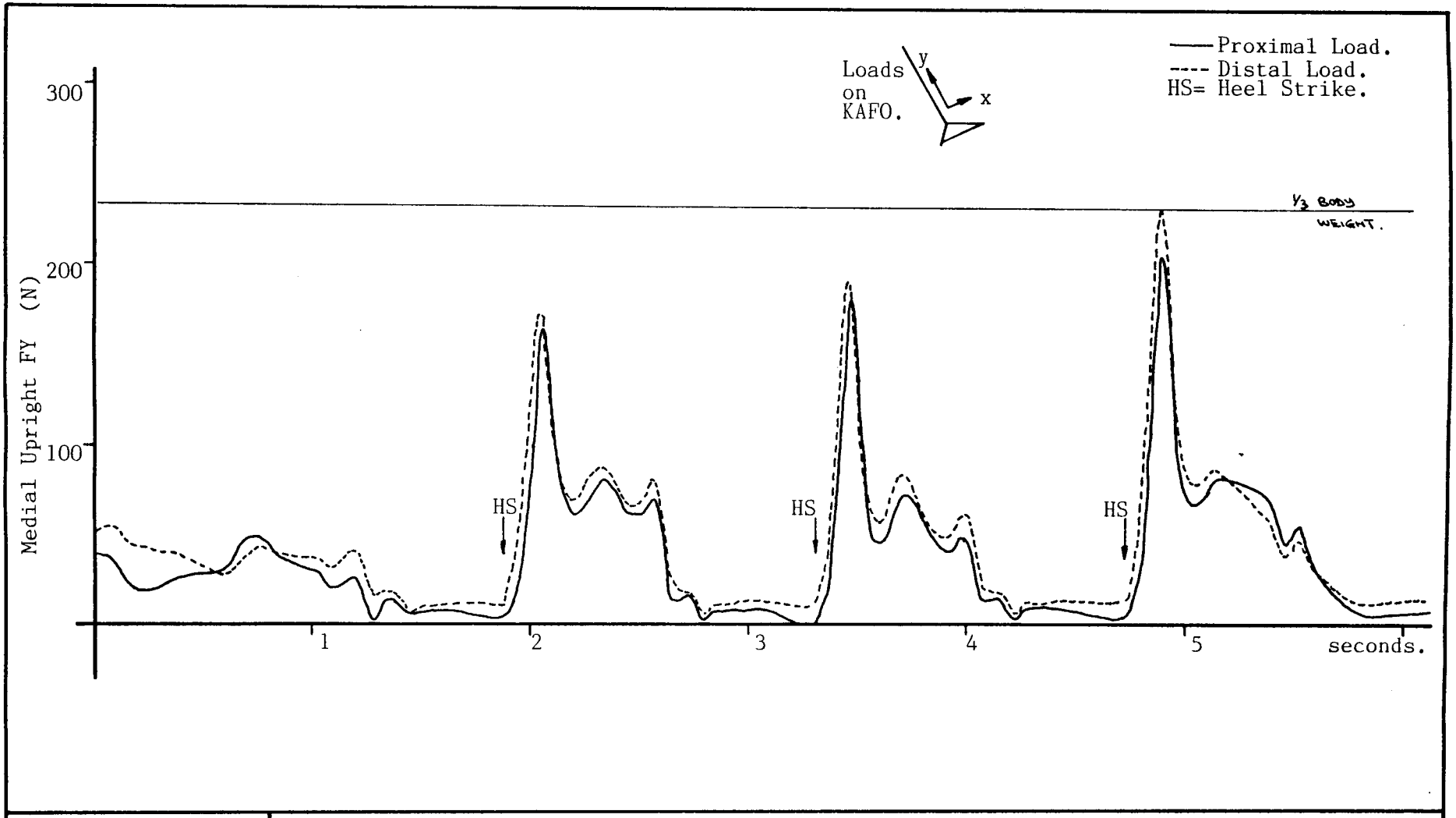


FIG. 7.20 FY Loadings on the Medial Upright of Patient MR. ("Start Walking")

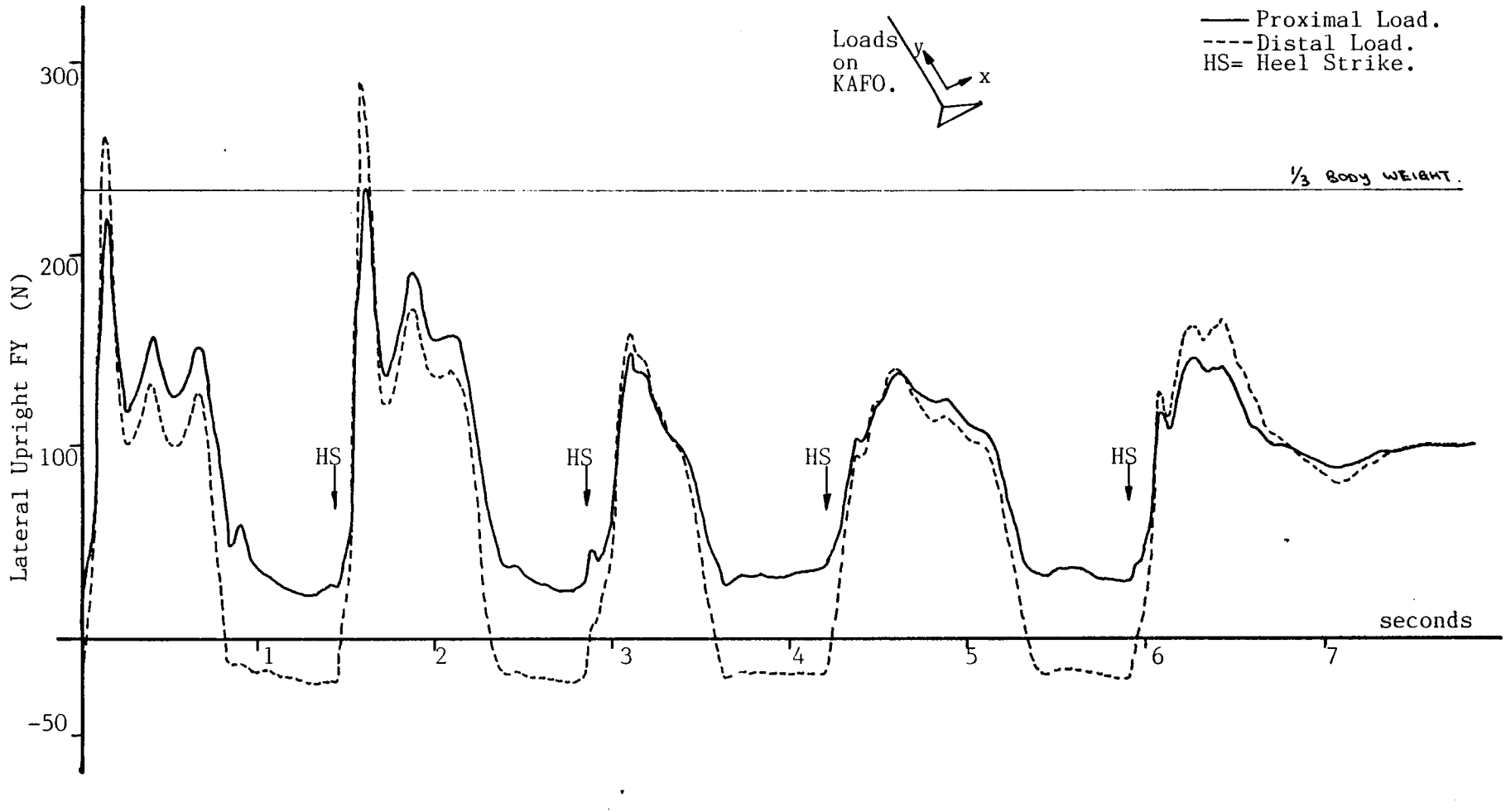


FIG. 7.21 FY Loadings on the Lateral Upright of Patient MR. ("Stop Walking")

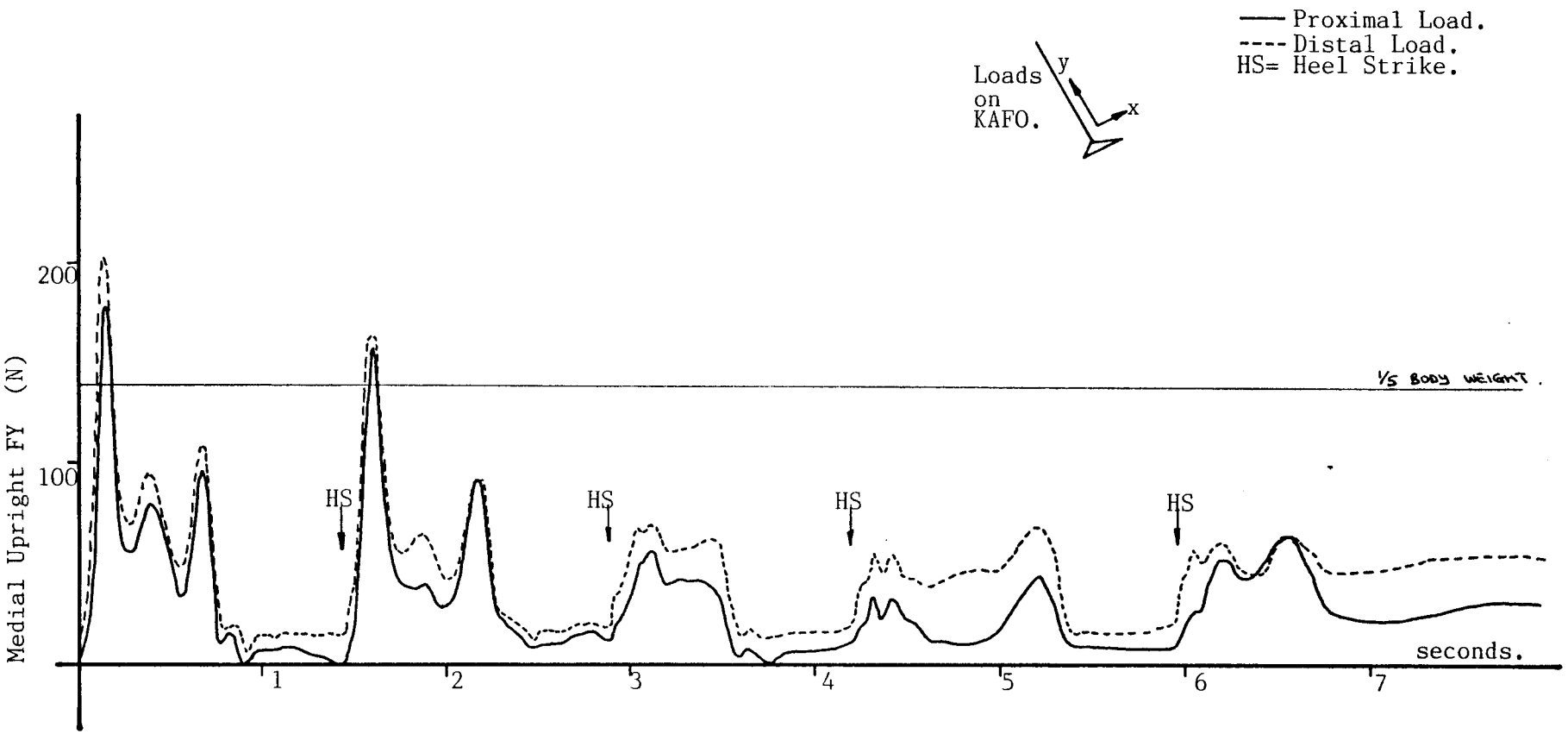


FIG. 7.22 | FY Loadings on the Medial Upright of Patient MR. ("Stop Walking")

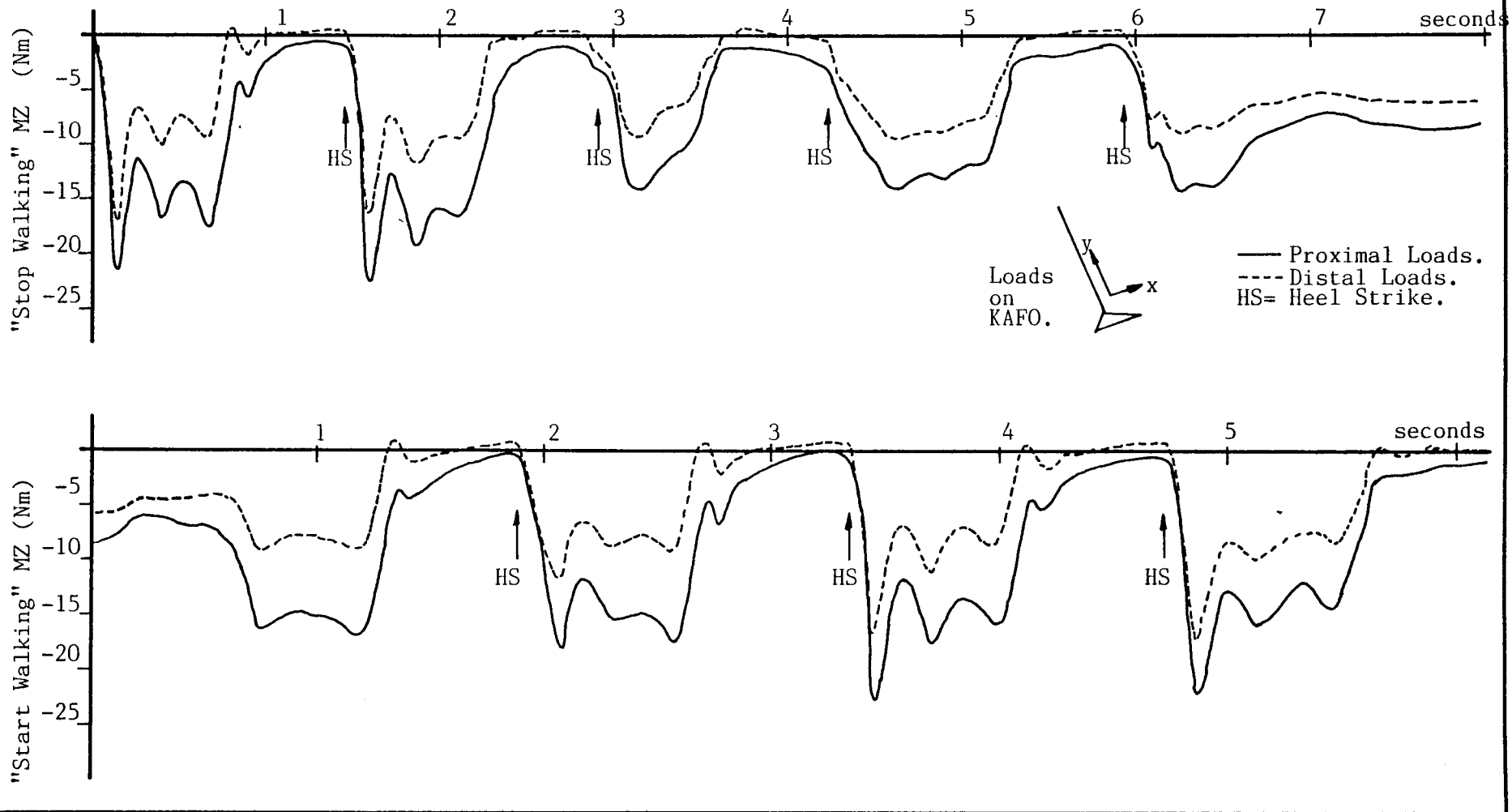


FIG. 7.23 | Results of Lateral Upright MZ Loadings for patient MR. ("Start Walking" & "Stop Walking").

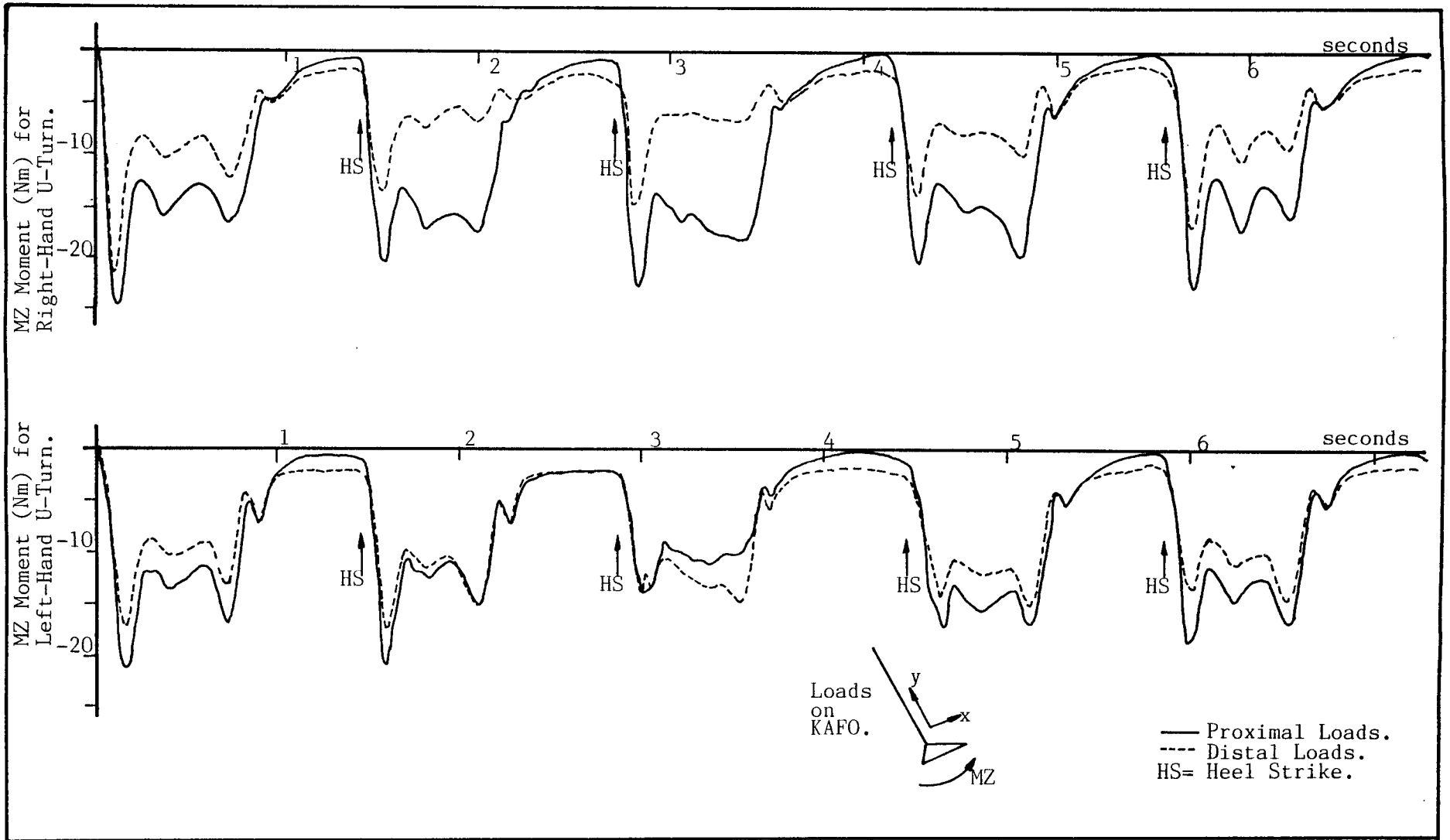


FIG. 7.24	Results of Lateral Upright MZ Loadings for Patient MR. (Left-Hand & Right-Hand U-Turns)

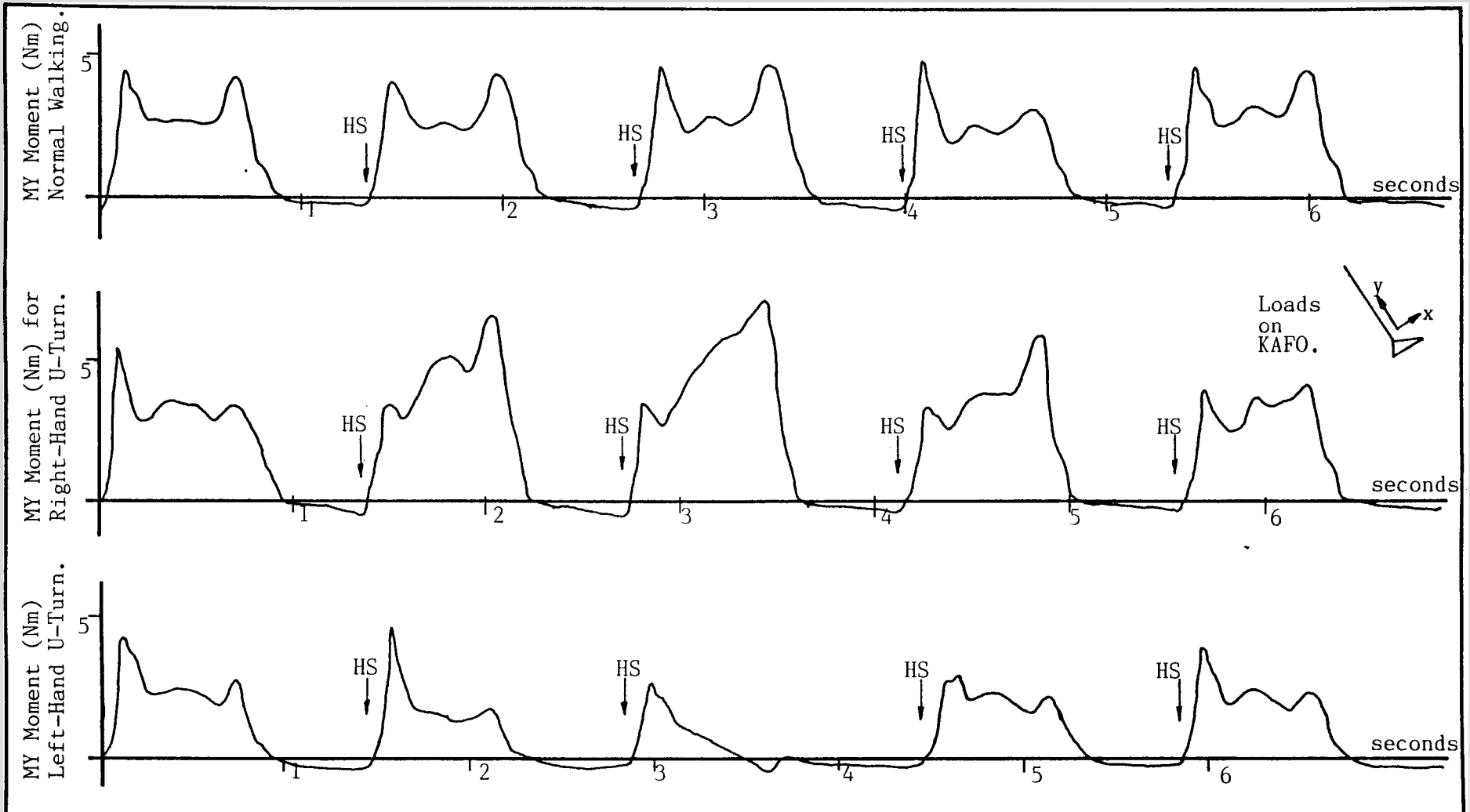


FIG. 7.25 Results of Lateral Distal MY Loadings for Patient MR. (Normal Walking & U-Turns)

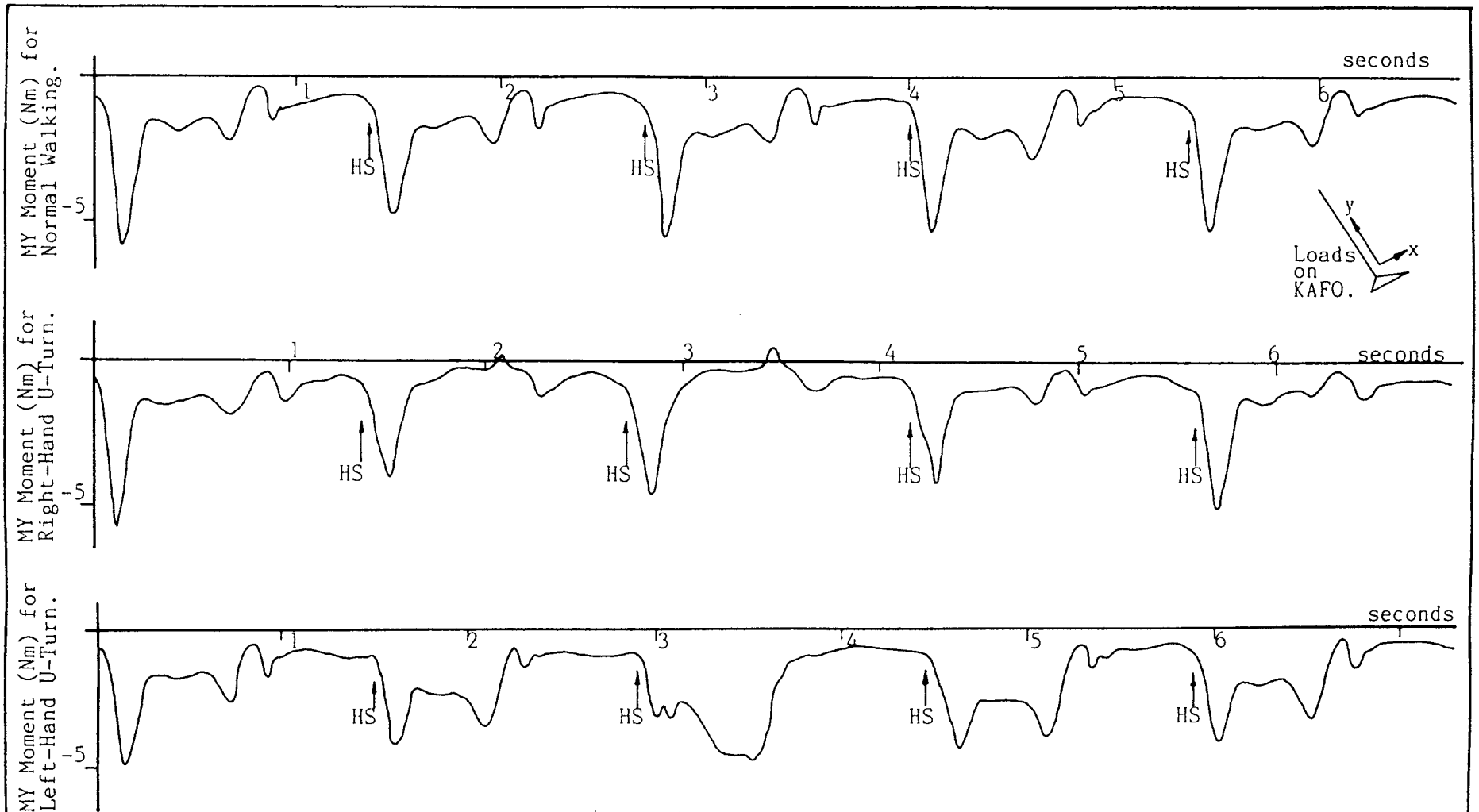


FIG 7.26 Results of Medial Distal MY Loadings for Patient MR. (Normal Walking & U-Turns)

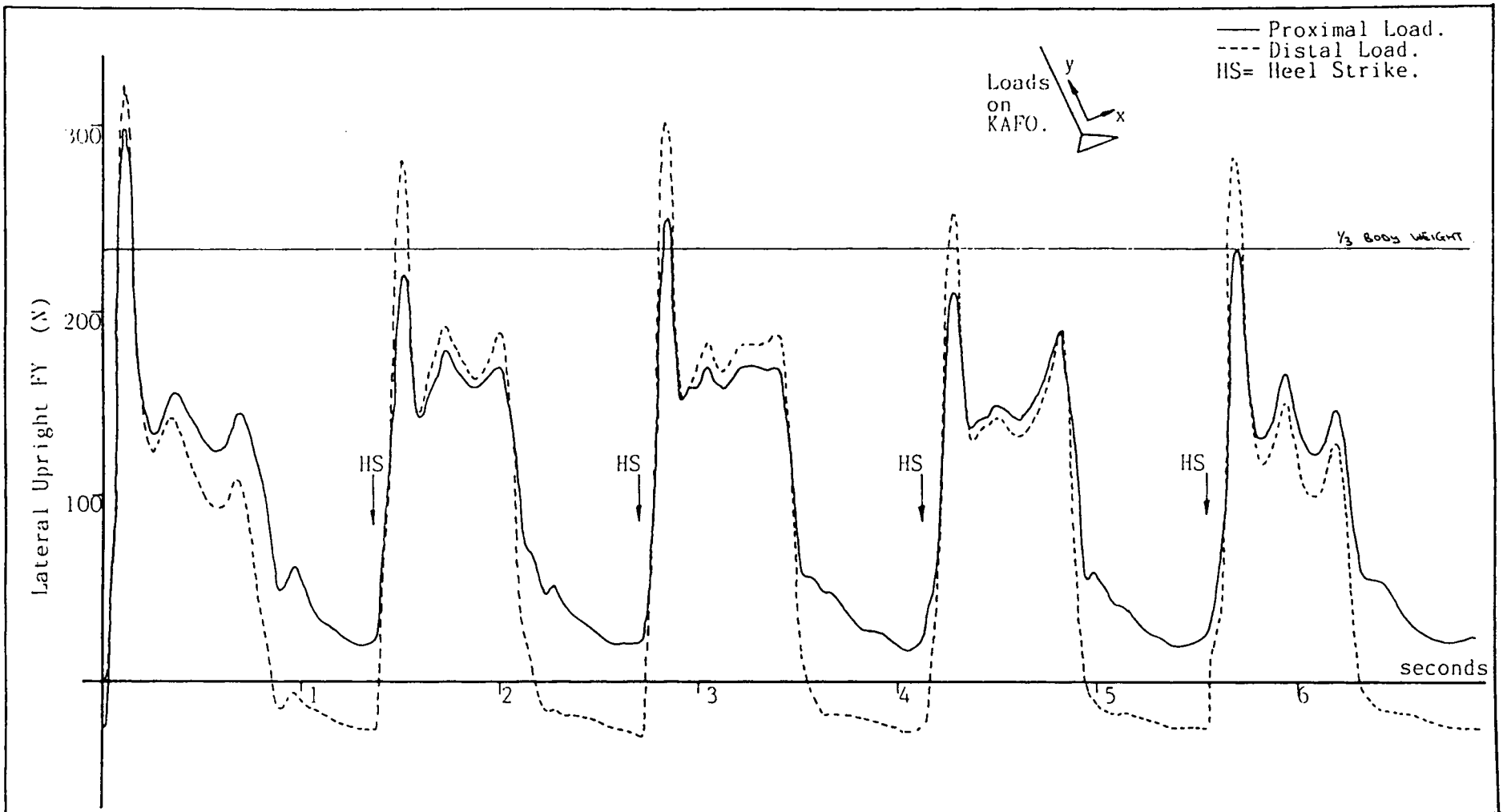


FIG. 7.27	FY Loadings on the Lateral Upright of Patient MR. (Right-Hand U-Turn).

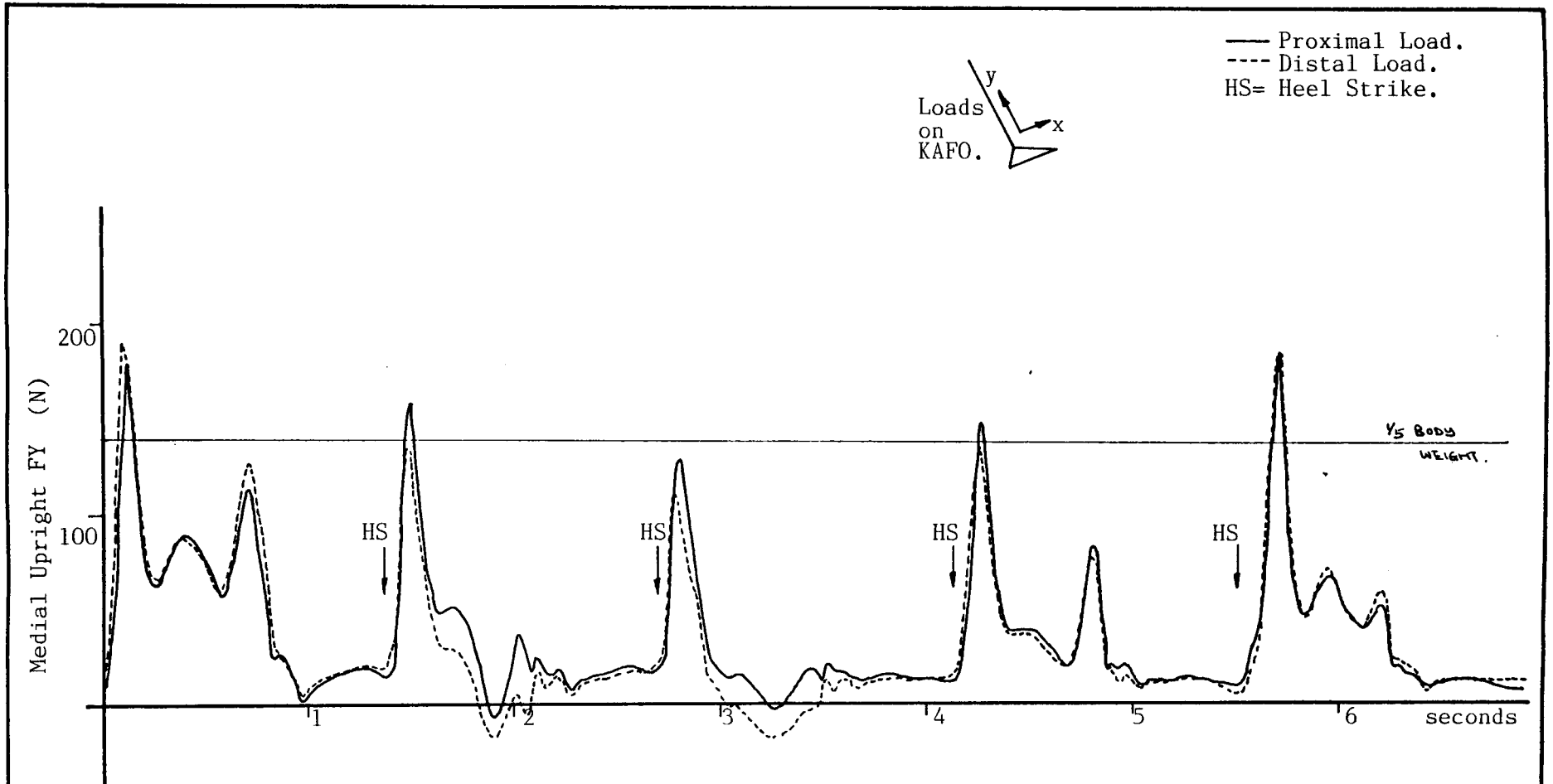


FIG. 7.28 FY Loadings on the Medial Upright of Patient MR. (Right-Hand U-Turn)

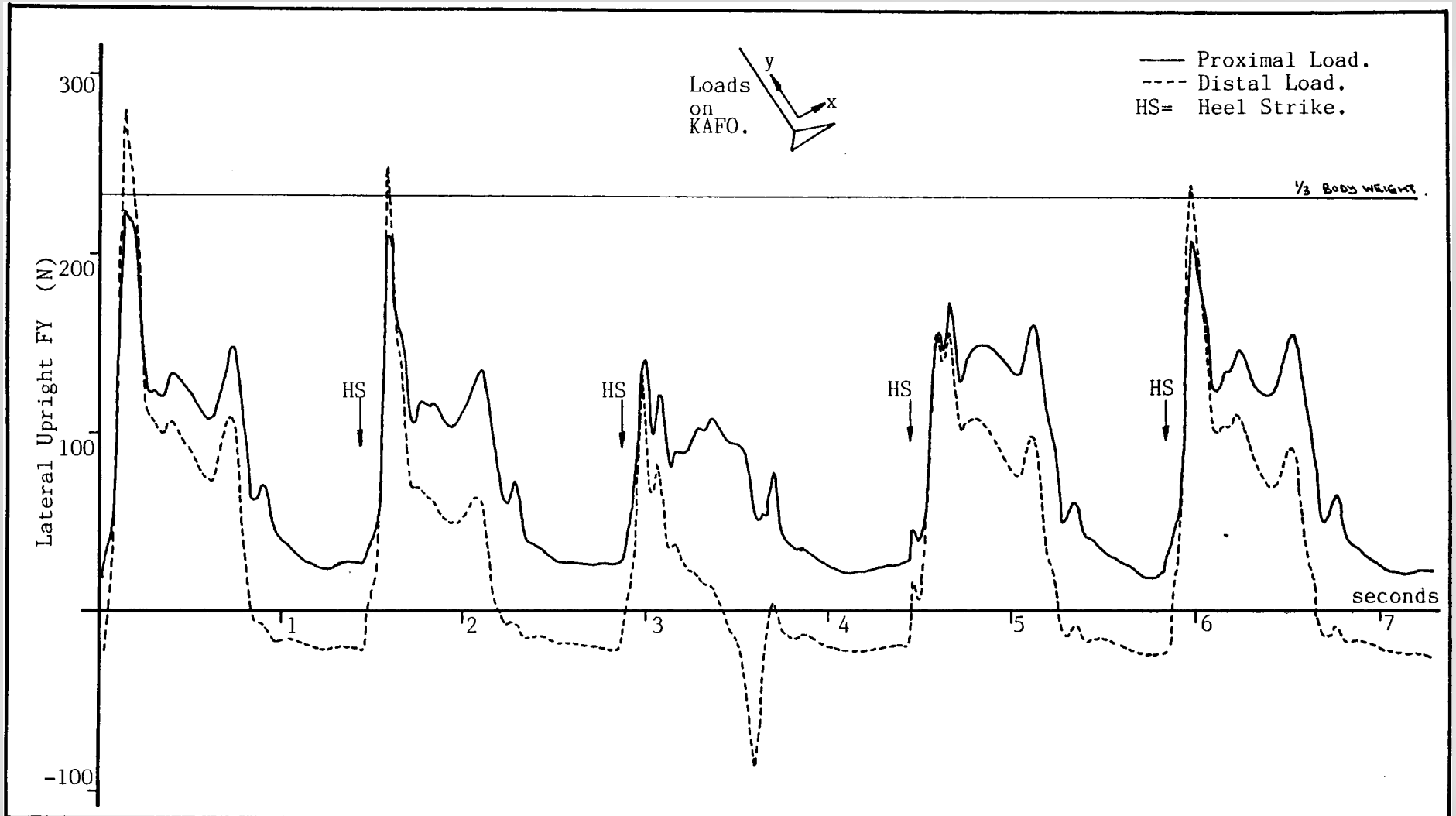


FIG. 7.29 FY Loadings on the Lateral Upright of Patient MR. (Left-Hand U-Turn)

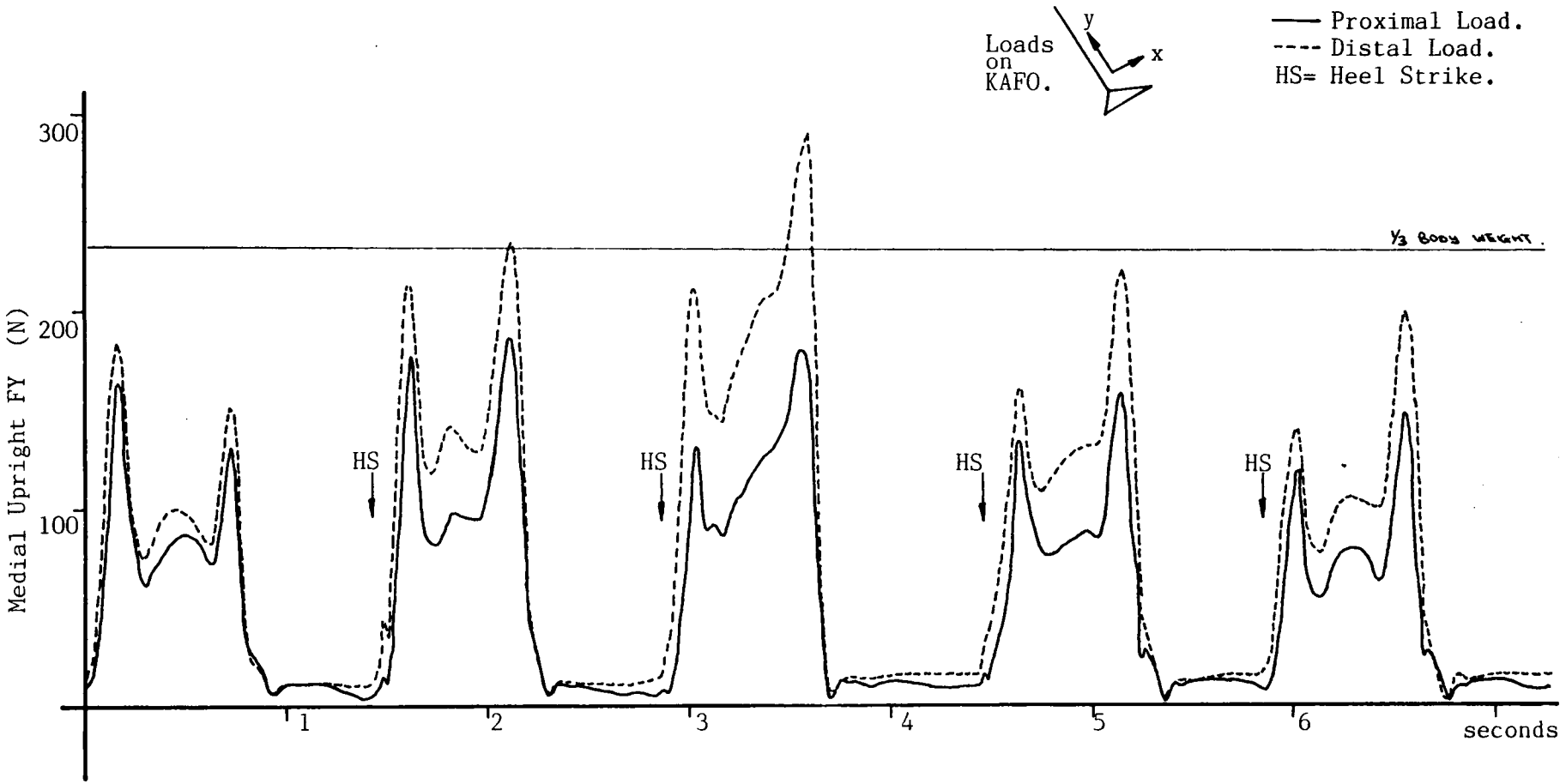


FIG. 7.30 FY Loadings on the Medial Upright of Patient MR. (Left-Hand U-Turn)

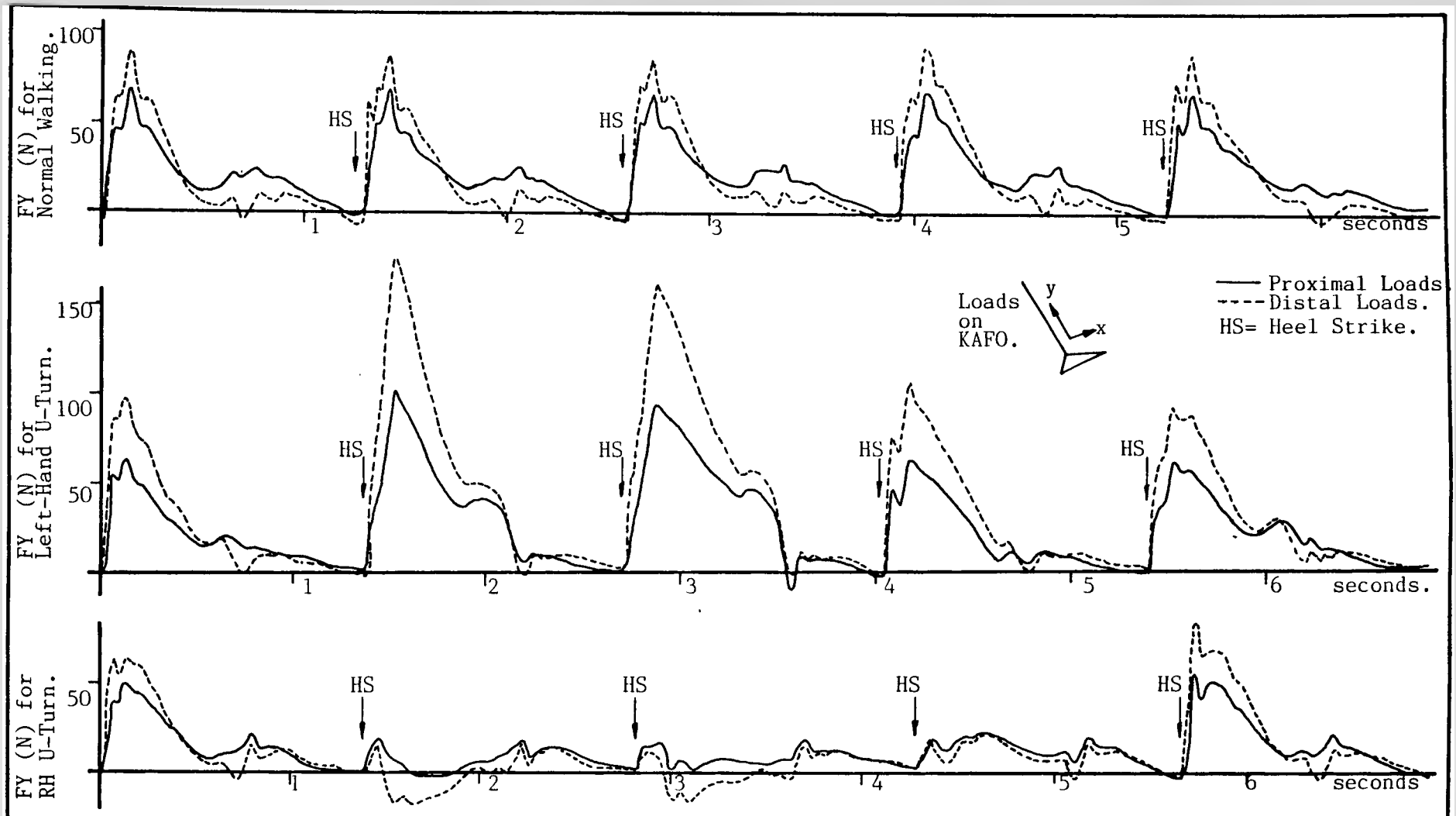


FIG. 7.31 Results of Lateral Upright FY Loadings for Patient DC. (Normal Walking & U-Turns).

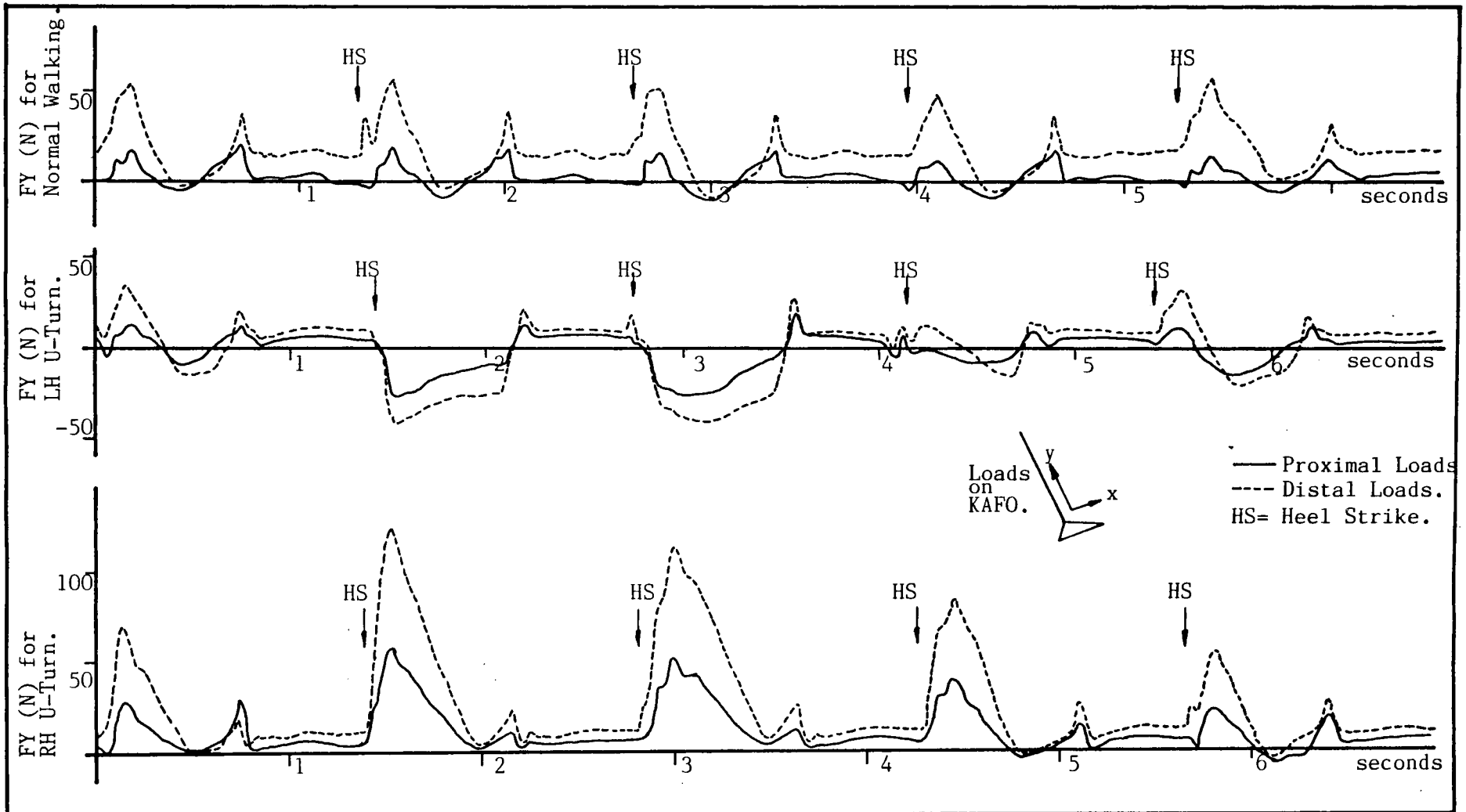


FIG. 7.32 Results of Medial Upright FY Loadings for Patient DC. (Normal Walking & U-Turns).

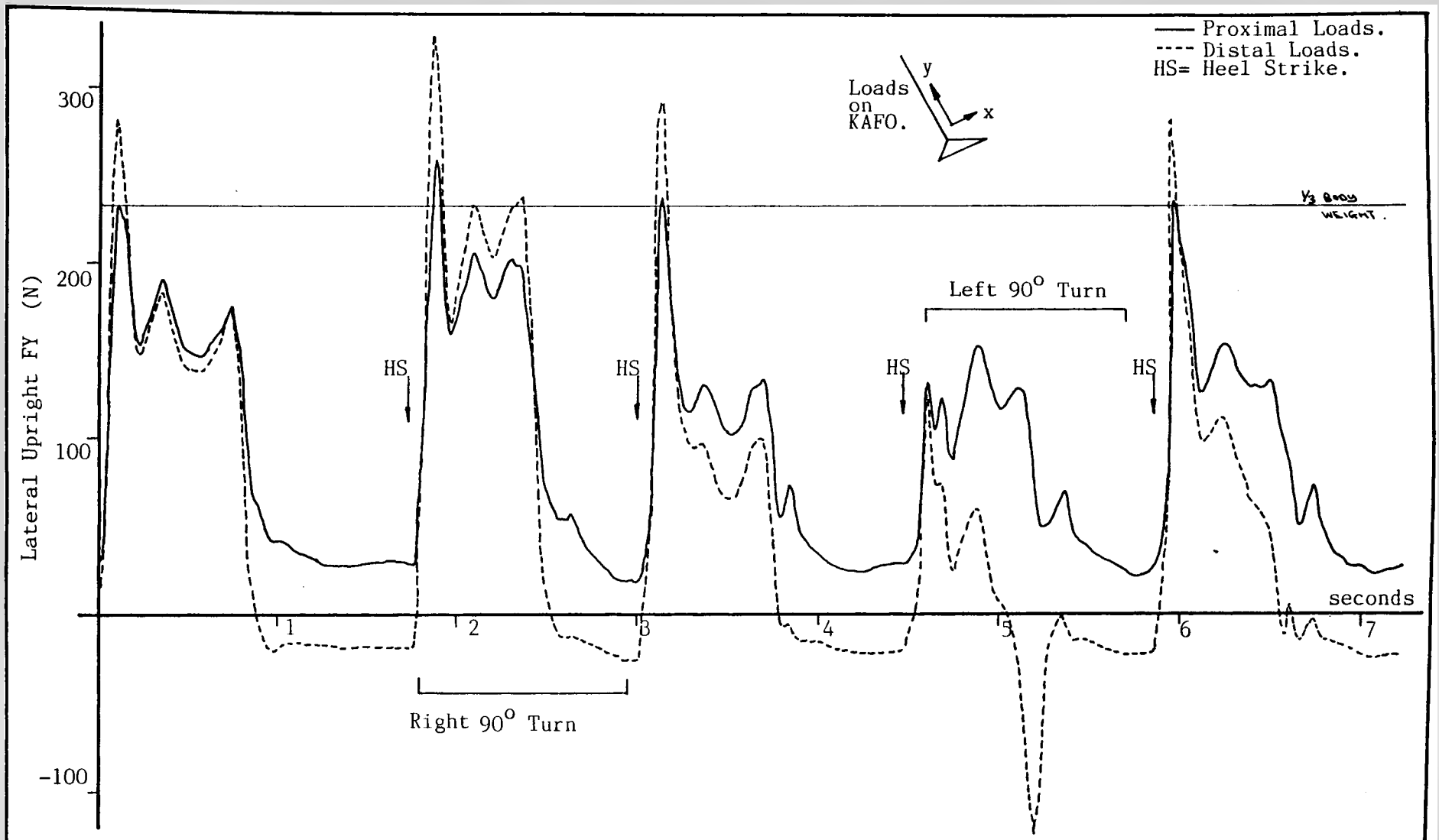


FIG. 7.33 FY Loadings on the Lateral Upright of Patient MR. (Double-Bend Test).

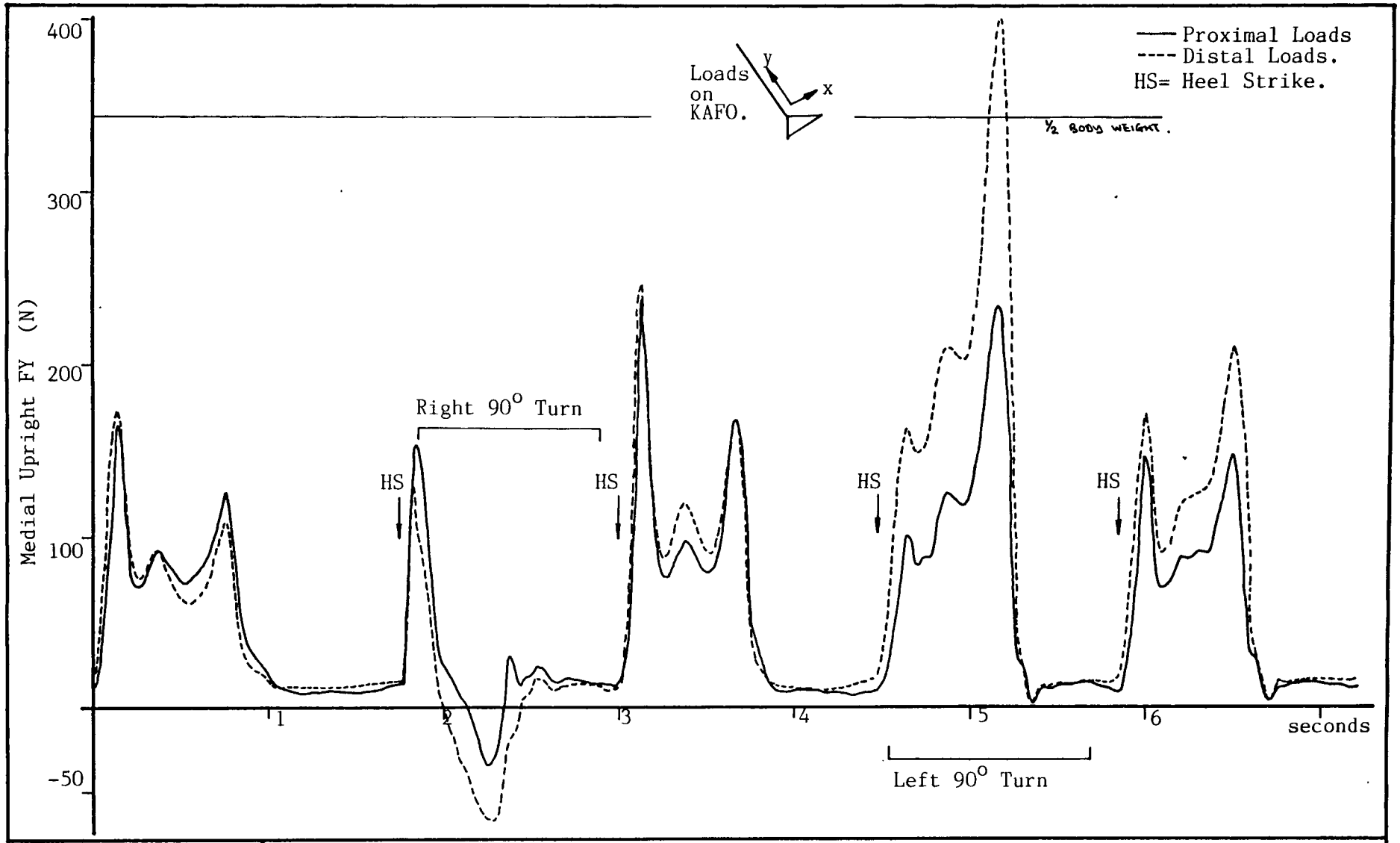


FIG. 7.34 FY Loadings on the Medial Upright of Patient MR. (Double-Bend Test).

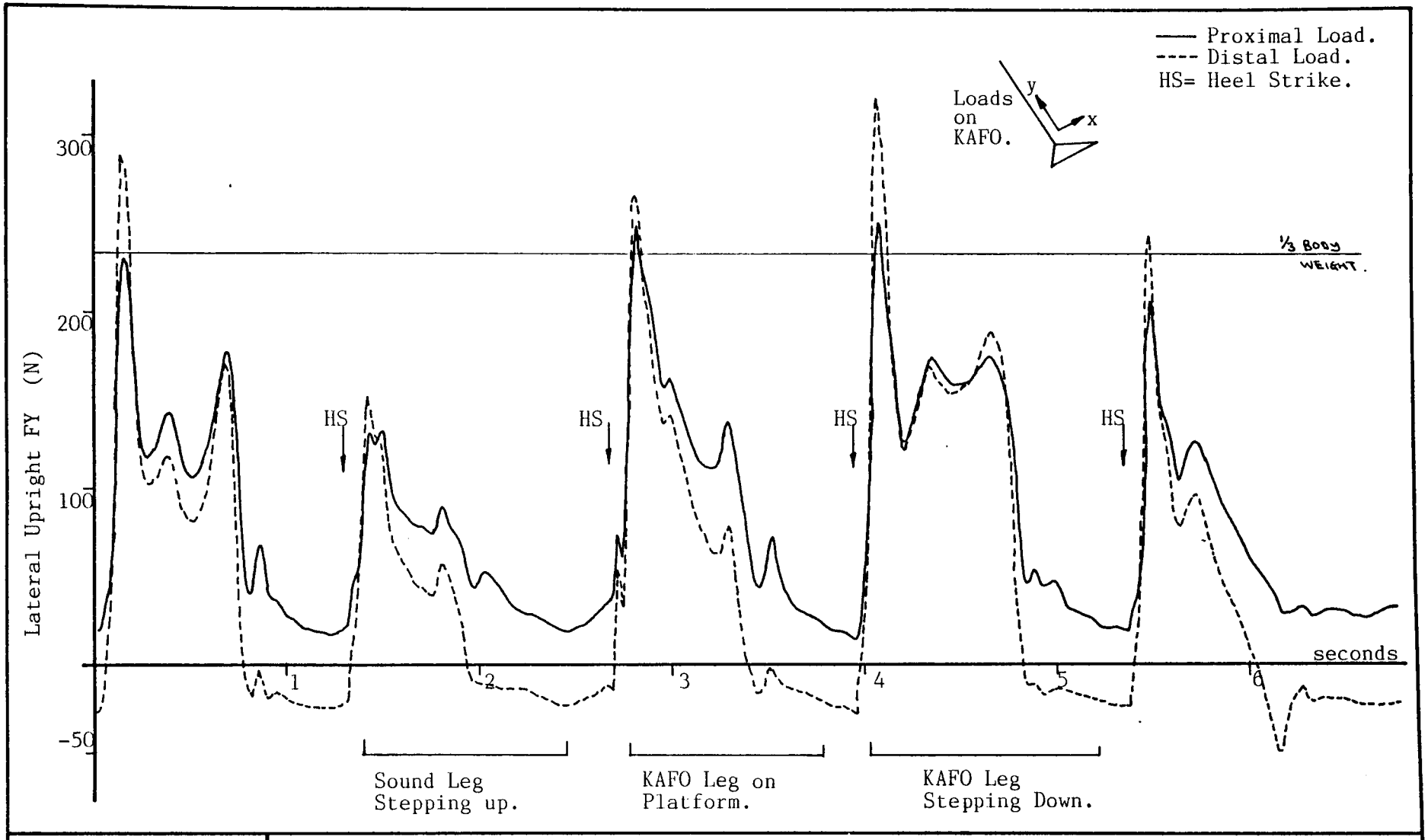


FIG. 7.35 FY Loadings on the Lateral Upright of Patient MR. (Low-Platform Test).

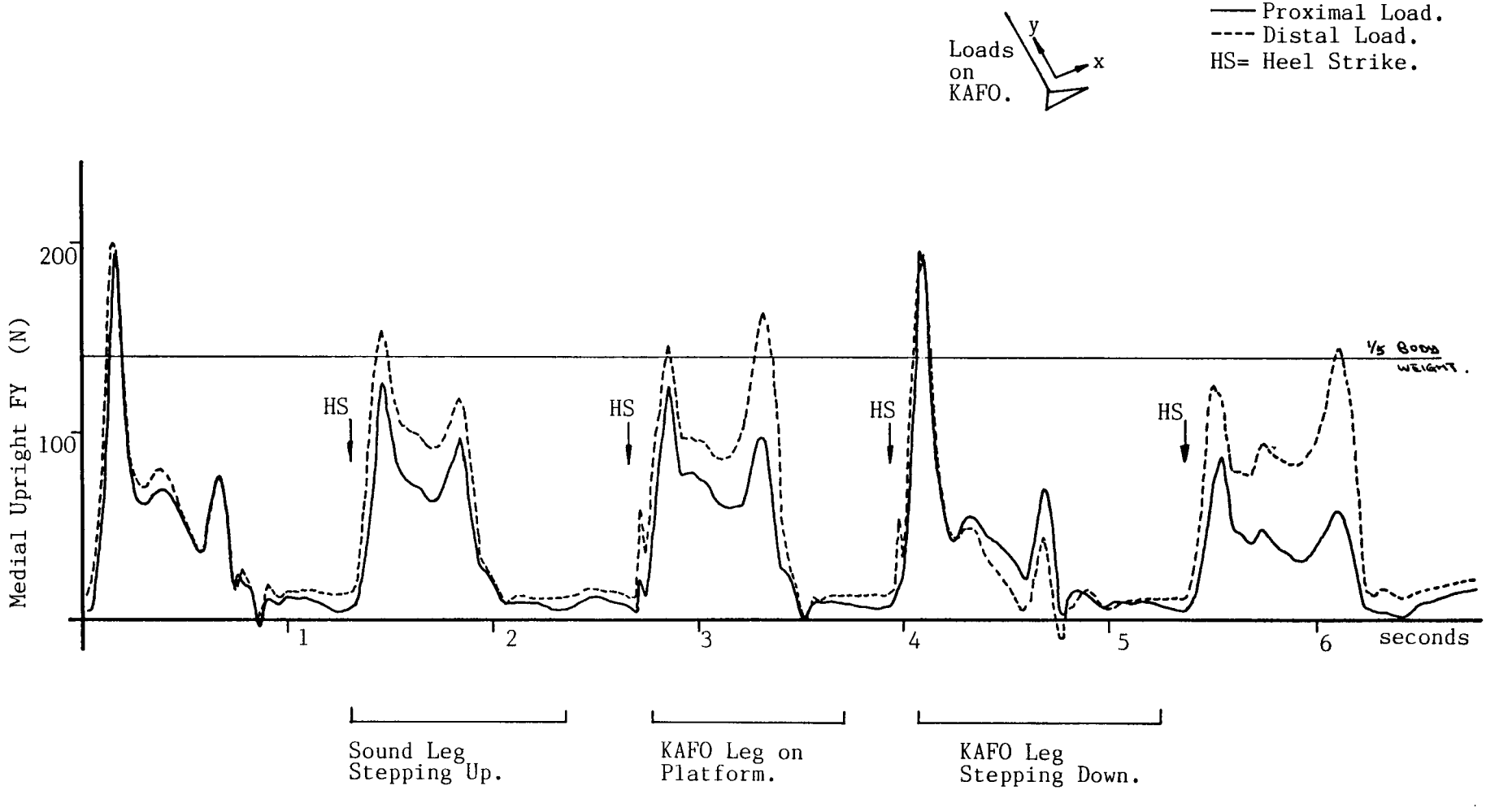


FIG. 7.36 FY Loadings on the Medial Upright of Patient MR. (Low-Platform Test).

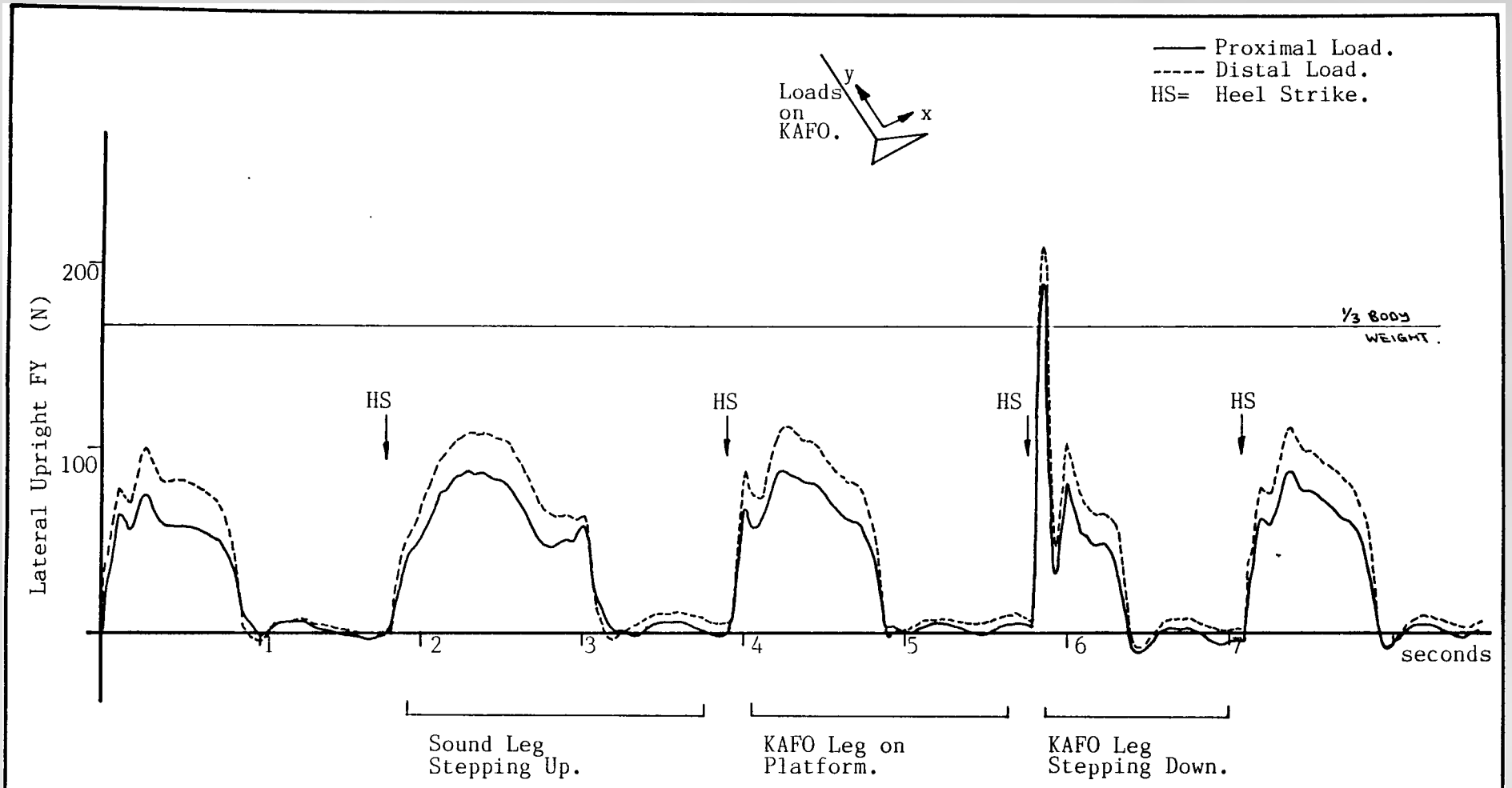


FIG. 7.37 FY Loadings on the Lateral Upright of Patient DR. (Low-Platform Test).

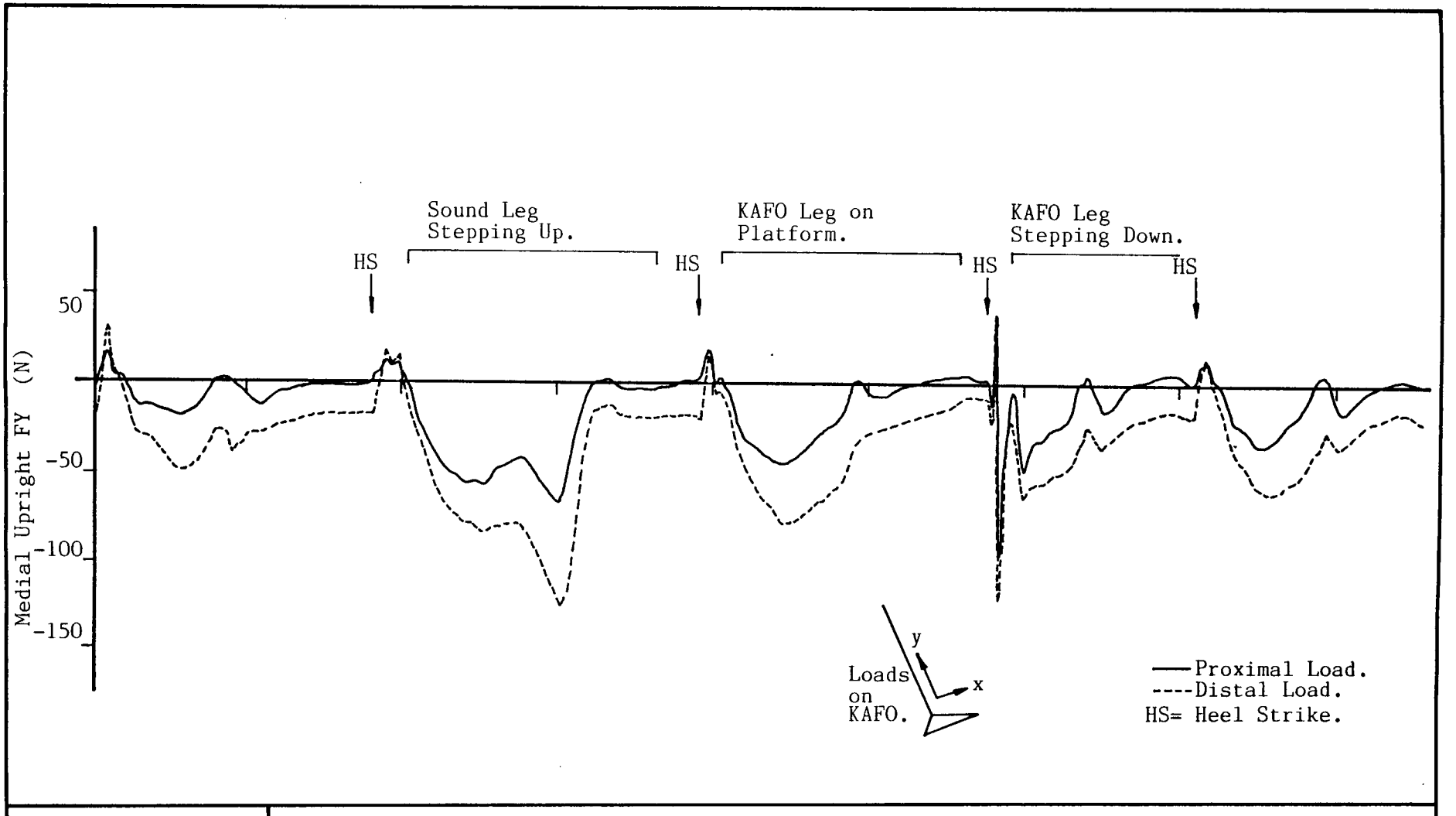


FIG. 7.38 | FY Loadings on the Medial Upright of Patient DR. (Low-Platform Test).

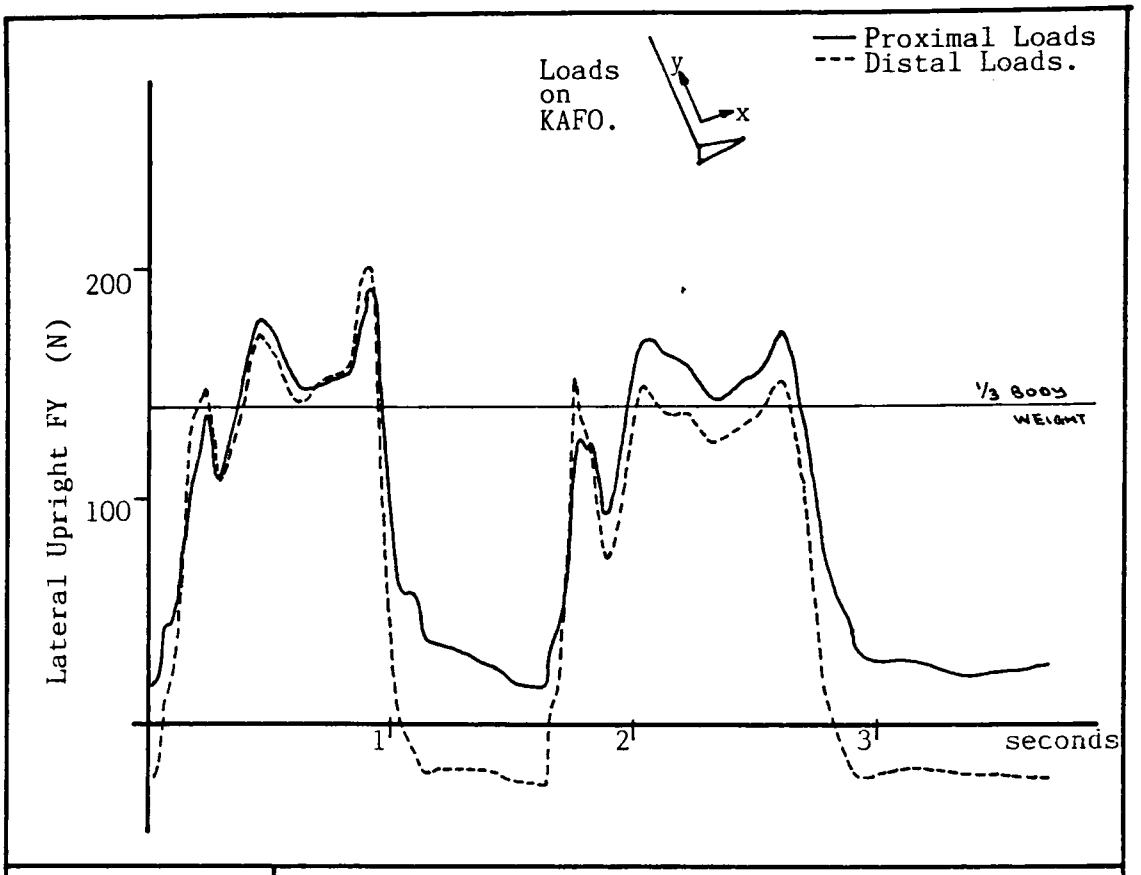


FIG 7.39	FY Loadings on the Lateral Upright of Patient MR. (Ascending Slope).

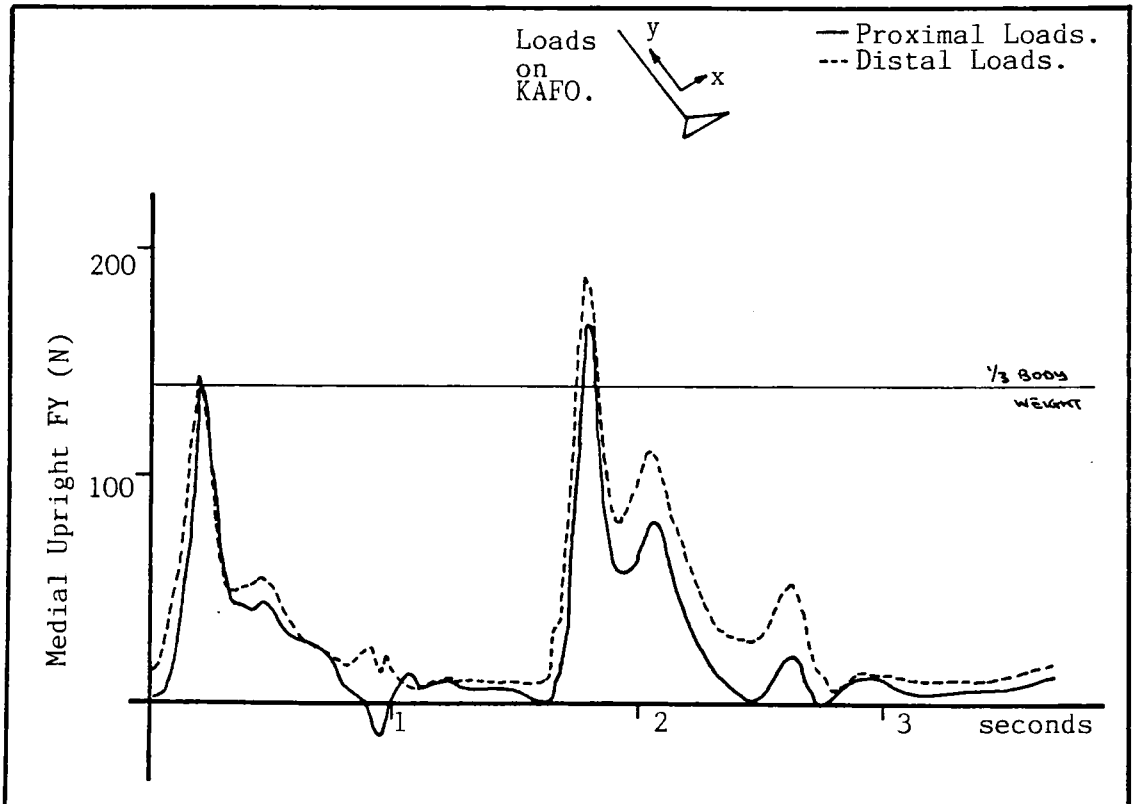


FIG 7.40	FY Loadings on the Medial Upright of Patient MR. (Ascending Slope).

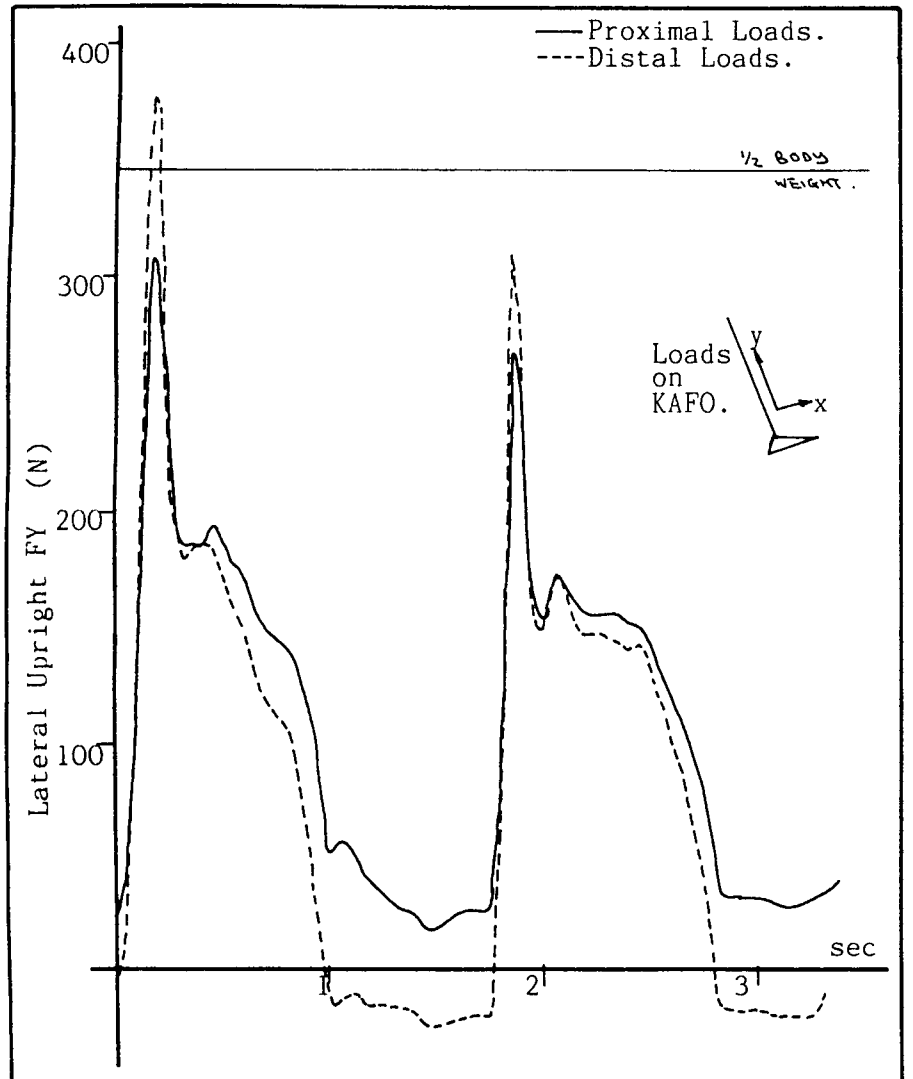


FIG 7.41	FY Loadings on Lateral Upright of Patient MR. (Descending Slope).

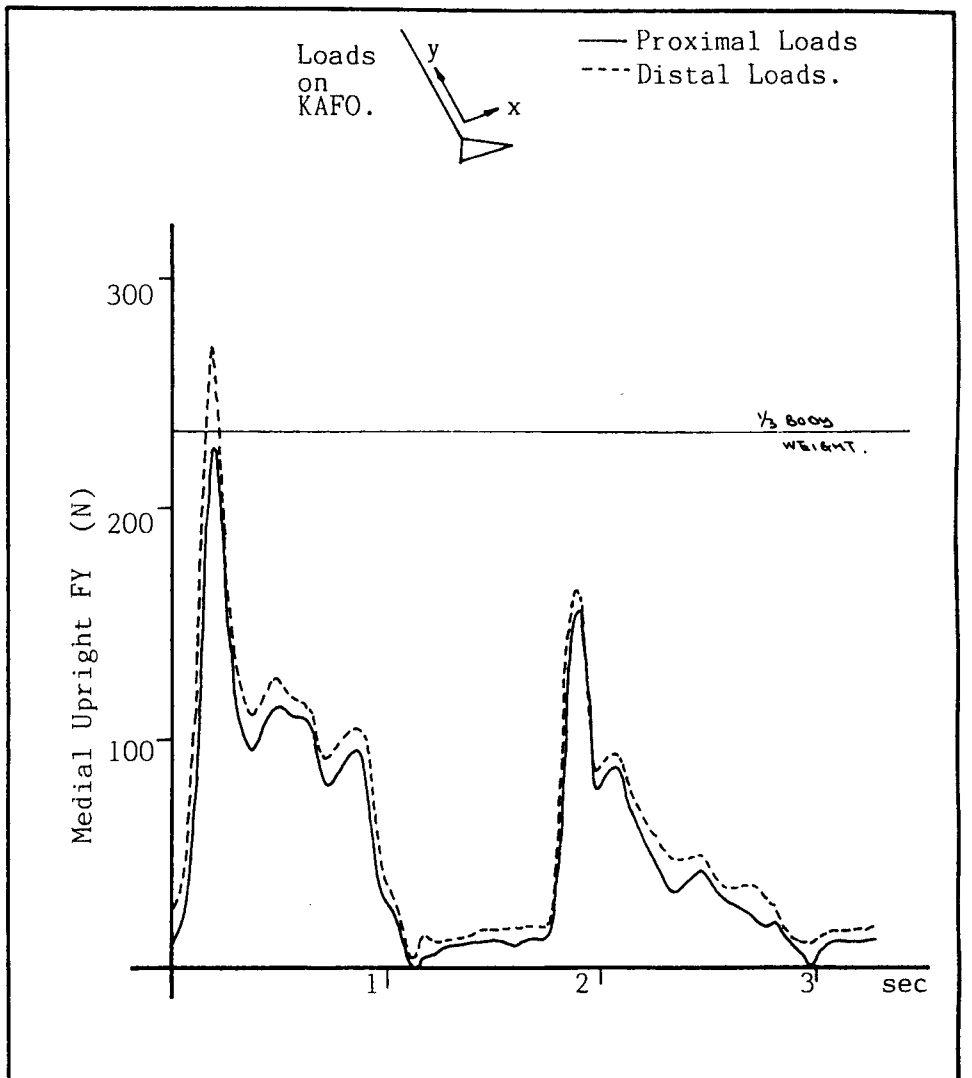


FIG 7.42	FY Loadings on Medial Upright of Patient MR. (Descending Slope)

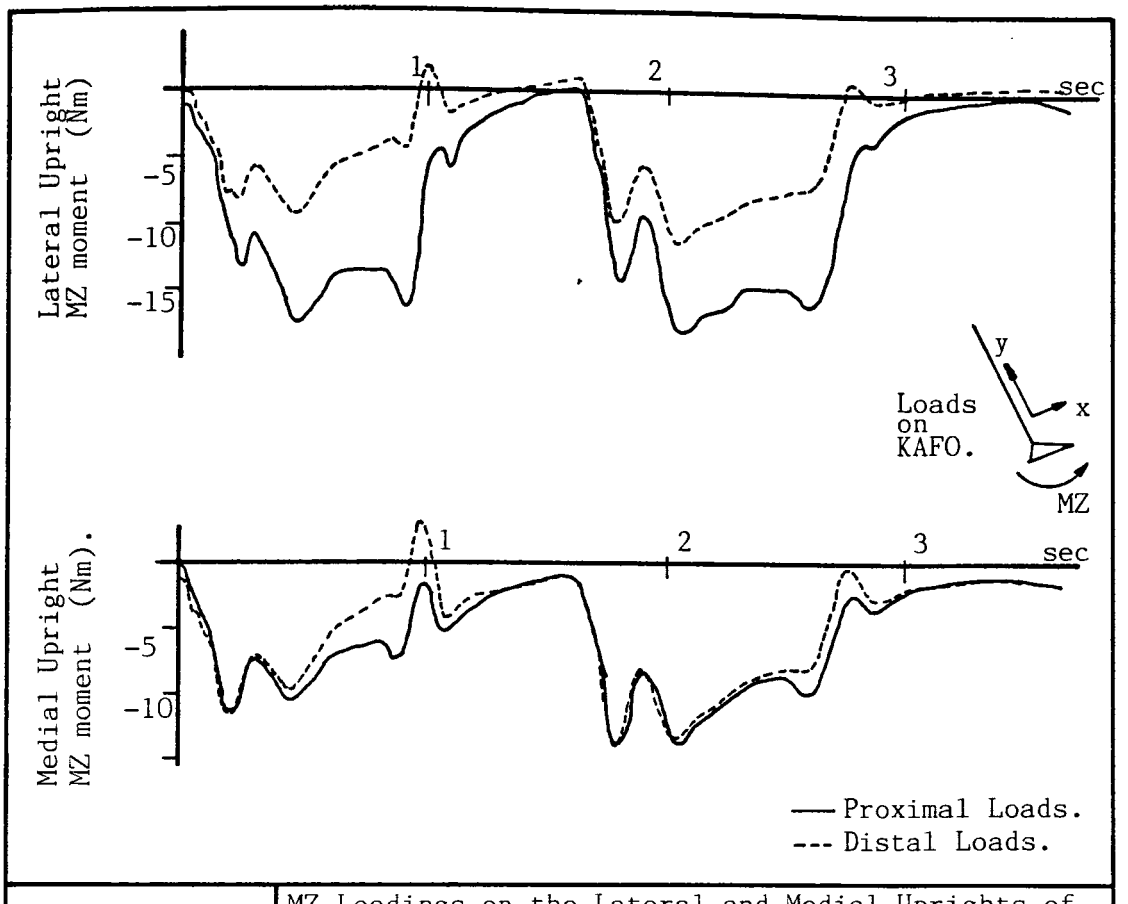


FIG 7.43 MZ Loadings on the Lateral and Medial Uprights of Patient MR. (Ascending Slope).

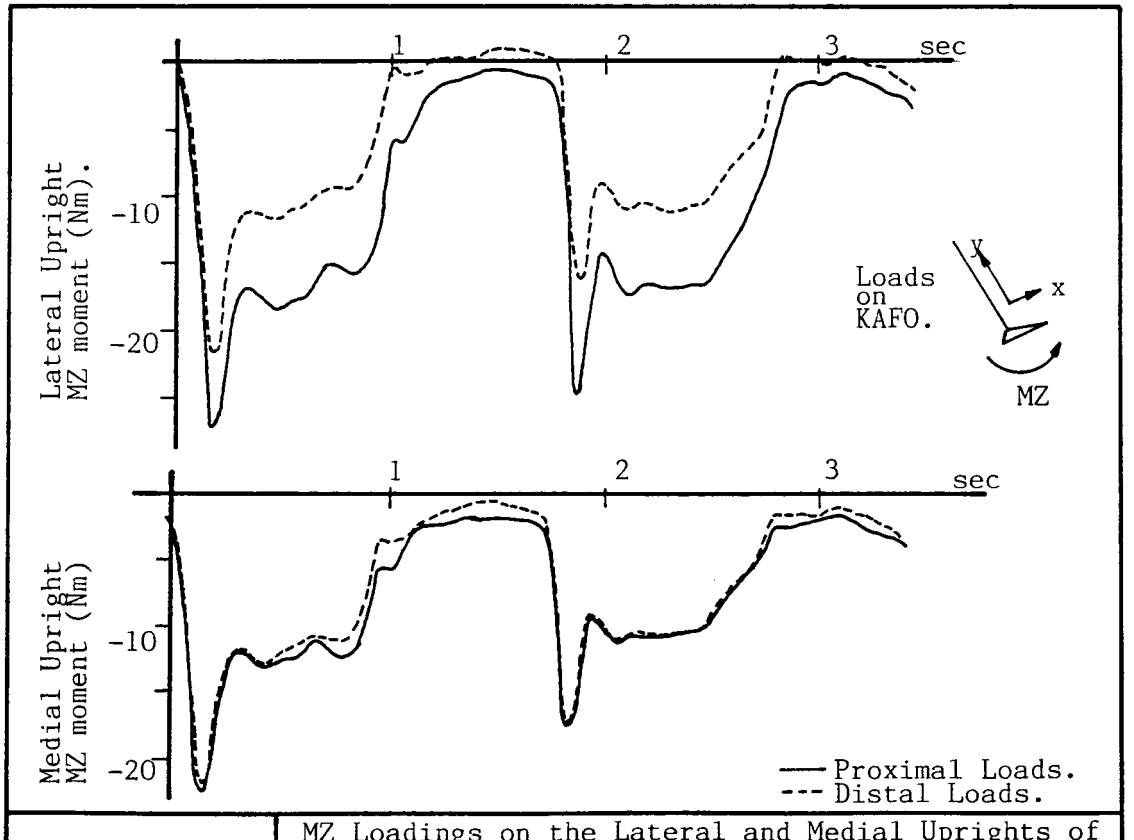


FIG 7.44 MZ Loadings on the Lateral and Medial Uprights of Patient MR. (Descending Slope).

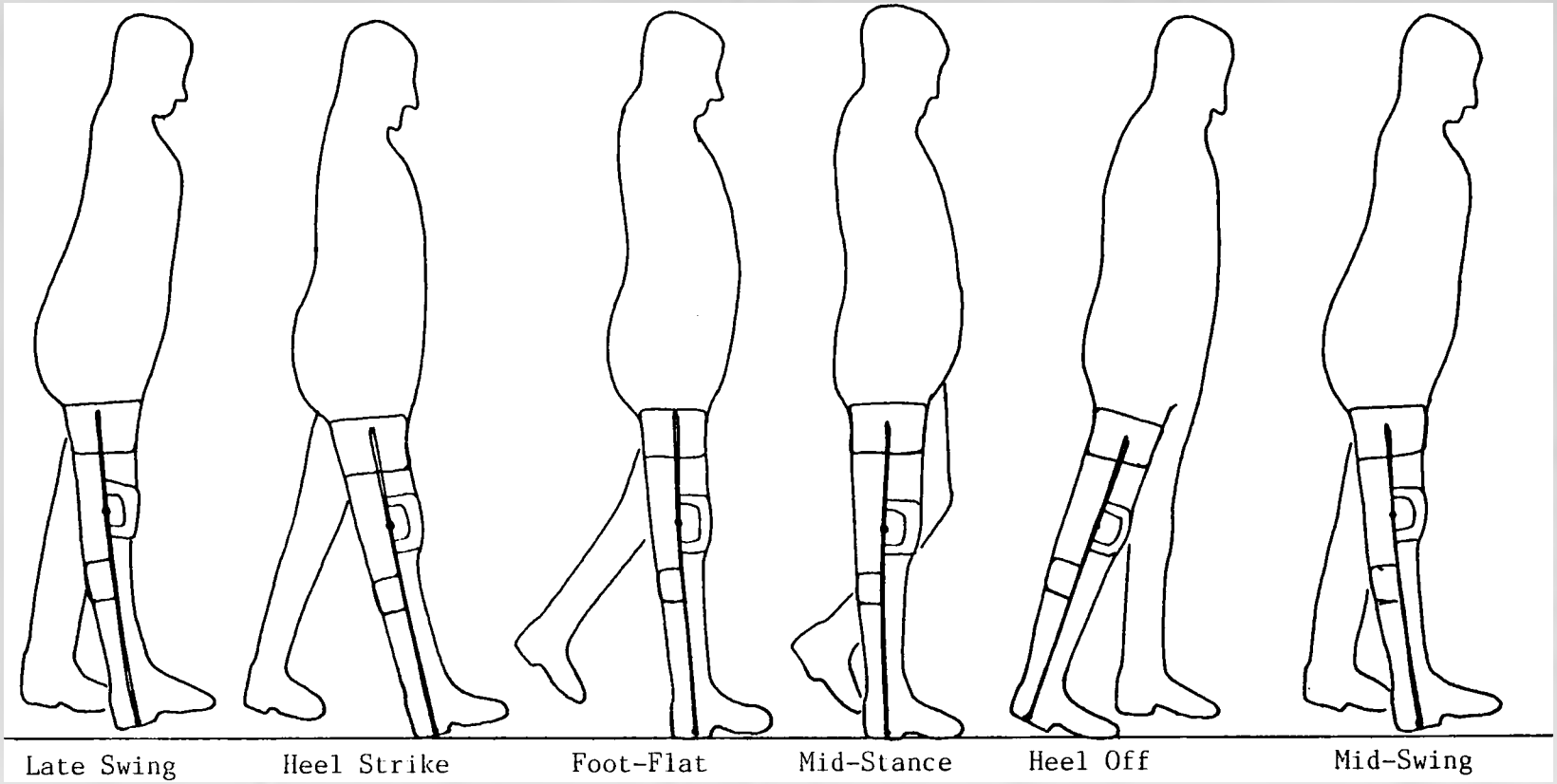


FIG. 7.45	The Lateral View of Patient DR's Typical Gait Cycle.

7.4 ANATOMICAL AND STRAP LOADINGS.

A total of five patients participated in the Normal Level Walking tests, i.e. tests where the kinematic data, and the orthotic and ground reaction loadings were recorded simultaneously throughout the walking cycles.

The kinematic data recorded were analysed both qualitatively and quantitatively. Fig 7.45 shows the gait pattern of patient DR traced from the cine film containing the 'side-views' of the patient's walking pattern, and the fully extended locked-knee type of gait can be very clearly seen from the diagram. The magnitude of lateral body sway for patient DR is illustrated in Fig 7.46, where the maximum lateral sway recorded was 140 mm at the head. This value is considerably larger than those for normal and amputee subjects, which were reported by Cappozzo (1982) as 55 mm and 95 mm respectively. The solid-lined image on Fig 7.46 shows the patient at mid-stance while the broken-lined image represents the mid-swing of the affected side. It gives a good indication of how the patient ensures that the affected limb clears the ground at swing through excessive lateral sway of the trunk. Due to a lack of abduction and adduction muscle power, the trunk moves over to the KAFO side during stance in order that the centre of gravity can be safely supported.

Fig 7.47 shows the amount of weight-bearing by the orthosis during the stance phase for patient DC. It indicates that the orthosis was most effective at the early stages of the stance phase when it carried approximately a quarter of the vertical FY load exerted onto the orthosis-limb complex. At the later stages of the stance, the vertical load on the orthosis-limb complex increases to slightly above the body weight. However, the orthosis carried only a fraction of the total FY loads, implying that the majority of the vertical ground reaction force was borne by the anatomical structure. This shows that the orthosis was being used by the patient primarily as a knee-stabilizing device, and not as a weight-bearing orthosis. The amount of weight-bearing of the orthosis varies with the requirement of the patients, ranging from 25% of body weight for patient DC and 75% for MR at early stance, to a fraction of DC's body weight and 50% of MR's weight at late stance phase.

The intersegmental forces and moments calculated at the anatomical

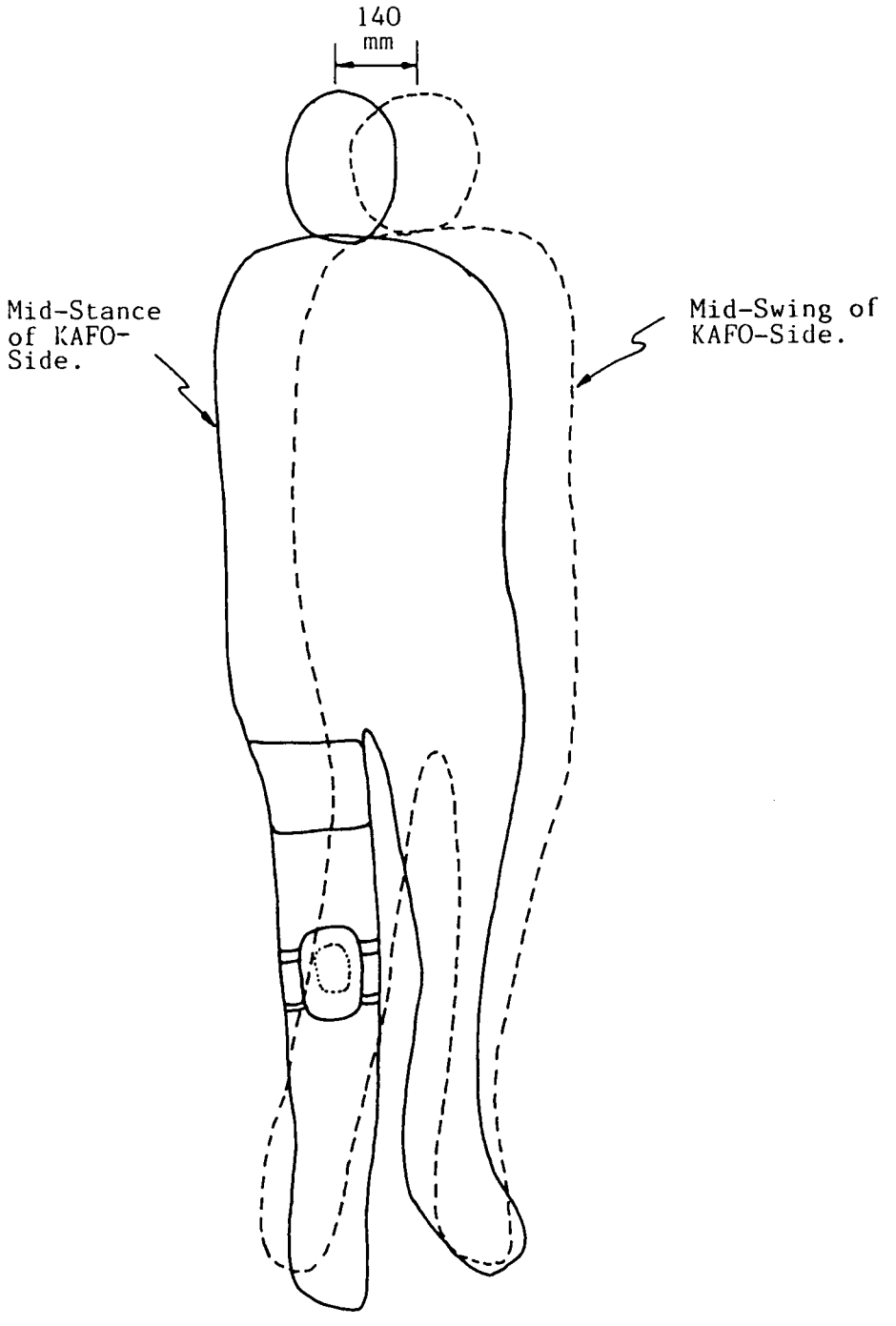


FIG. 7.46 Lateral Body Sway of Patient DR.

ankle, knee, and hip levels are shown in Fig 7.48 & 7.49 . A large posteriorly directed shear force FX of up to 500N was found to be acting across the anatomical knee joint centre. This unusually large shear force at the knee is due to the forces generated by the orthosis in response to its function for stabilizing the enclosed limb. However, Lehmann and Warren (1976) reported a maximum FX shear force of some 230N at the knee. This represents only 50% of the maximum value obtained for patient DC, and this large discrepancy may be due to the inaccuracies of the assumptions in the calculations for the strap forces, which will be discussed later in this chapter. When these anatomical loads are compared with the values for normal subjects reported by Bresler and Frankel (1950), considerable dissimilarities between the two sets of results can be seen (Fig 7.50). These differences may be due to the 'locked-knee' gait pattern of the KAFO patient, and may also caused by the inaccurate assumptions in the analysis for calculating FX and FZ shear forces at the anatomical joints.

Fig 7.48 also illustrates the amount of weight-bearing at the anatomical ankle, knee, and hip joints. By comparison with Fig 7.47, it can be seen that the majority of the body weight during stance phase was borne by the anatomical structure. The mediolateral shear forces (FZ) were less than 60N at the knee level. For the intersegmental bending moments, the knee is again having to sustain the highest loads, with the mediolateral bending moments (MX) and anterioposterior moments (MZ) reaching a magnitude of 90 Nm and 60 Nm respectively.

In order to acquire a better understanding of the effects of the knee apron strap forces on the orthotic uprights, custom built knee strap transducers were attached to the uprights for the measurement of the strap forces. The results obtained from this test for patient DC are shown in Fig 7.51 & 7.52, where the strap loadings are in the transducer reference systems. The transducer results show that the proximal straps are more highly loaded than the distal, with up to 3 times the magnitude of FZ forces on the lateral upright, and $2\frac{1}{2}$ times for FX of medial upright. The FX forces are acting anteriorly, indicating that the knee apron is performing its duty in controlling the knee flexion. The maximum knee apron FX forces measured was 140N, with the proximal straps contributing up to 65% of the load action. Although the knee apron for patient DC is designed primarily for the prevention of knee flexion during stance, the results obtained indicated that a

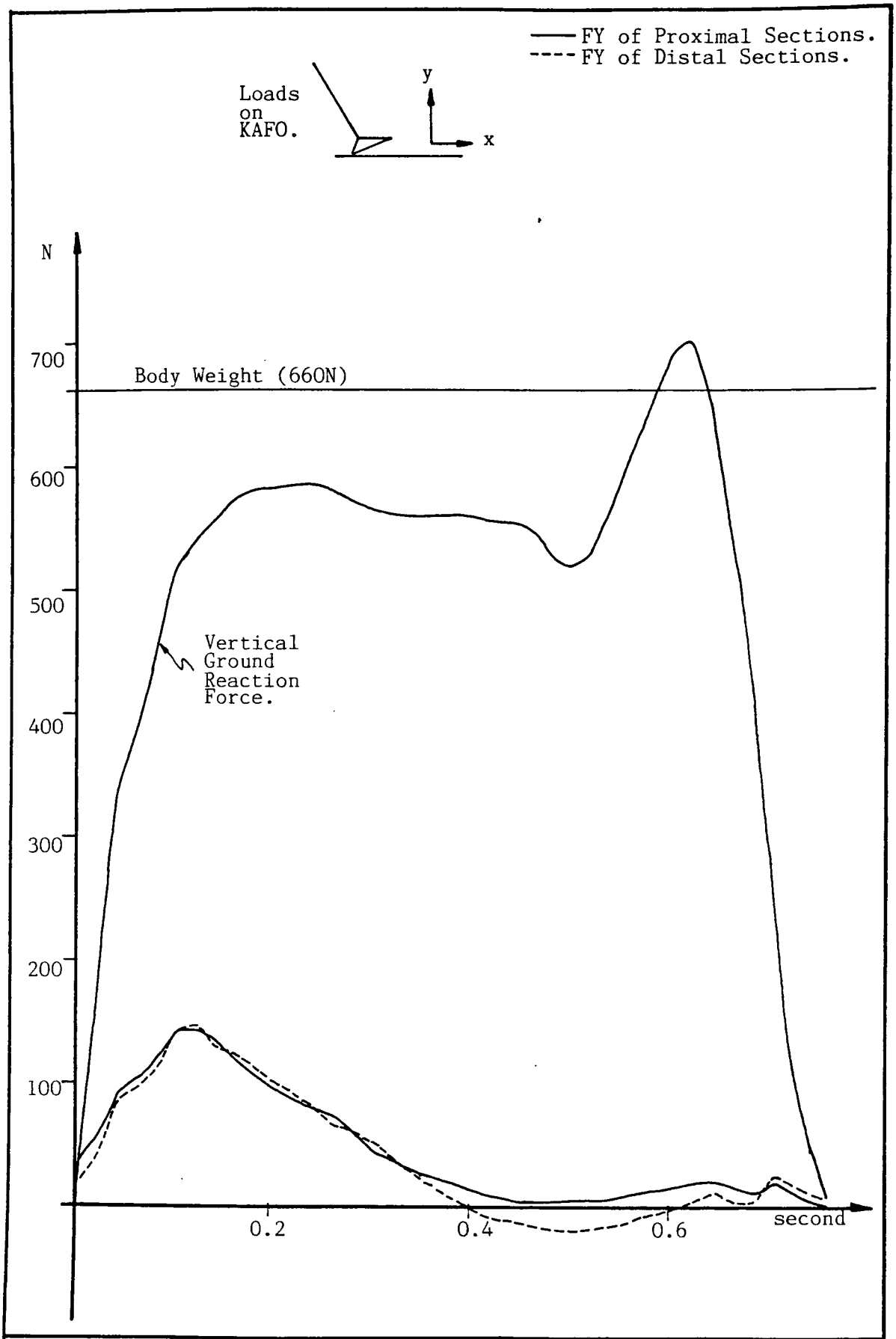


FIG. 7.47	Weight-Bearing Provided by Patient DC's Orthosis.

certain degree of mediolateral stabilization may also be achieved with the use of such a strap. The mediolateral knee strap forces on the lateral upright amounts to 80N, some 30N higher than those on the medial side. This additional 30N was most probably introduced by the patient to provide for some mediolateral support to the knee while putting on the orthosis.

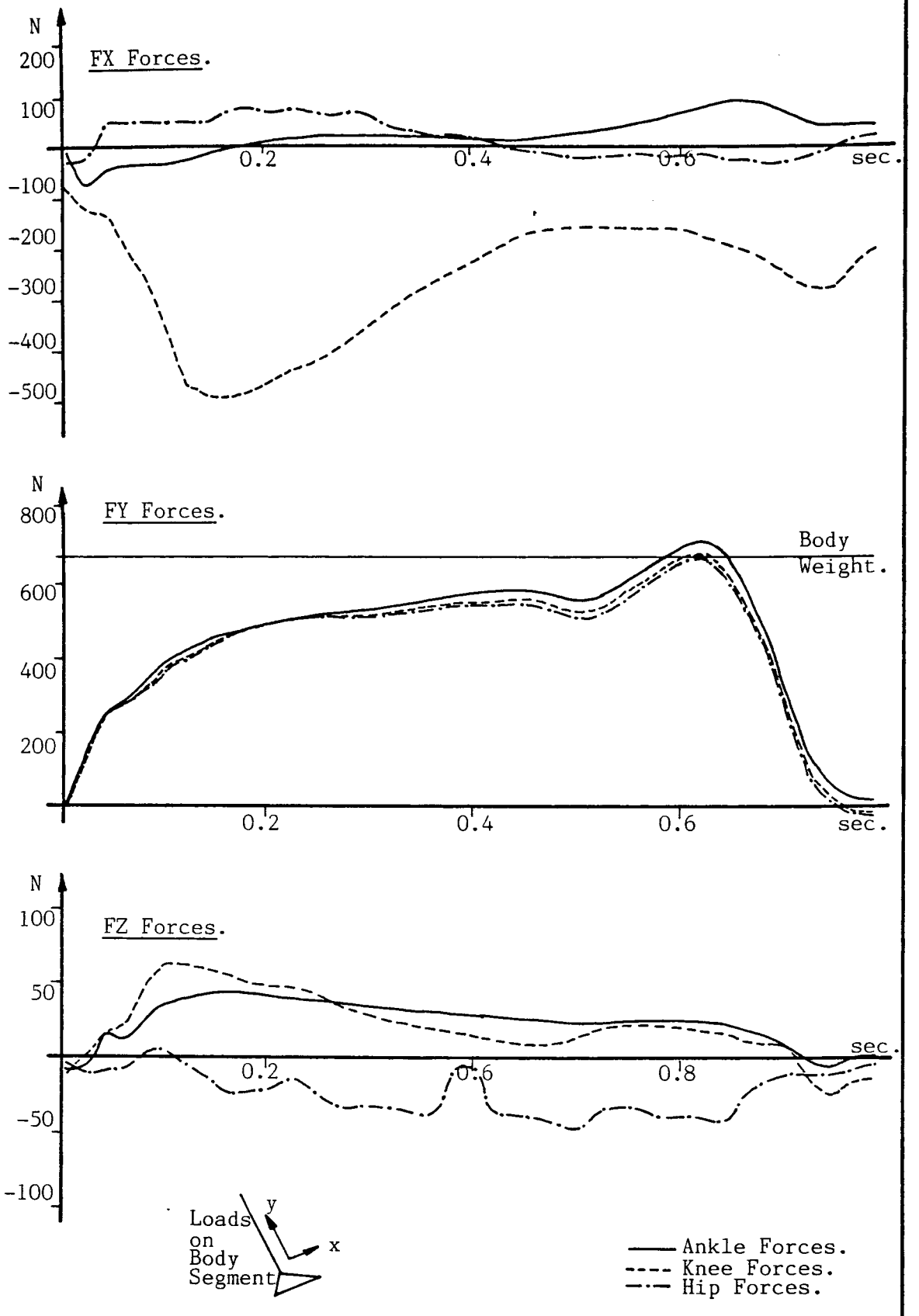
When the results of the knee strap forces obtained through the use of the strap transducers were compared with those estimated from the calculation, the short coming of the assumptions used in the orthosis load analysis becomes apparent. In the analysis, it was assumed that all the leather straps of the apron carry equal amount of forces. This was found to be incorrect in the actual loading situation (Fig 7.51 & 7.52). Although the calculated results show a similar loading pattern to those obtained from the transducers, they have been overestimated by about twice the magnitude in the FX channels, and underestimated by some 10-40N in the FZ direction. To obtain more realistic results, the strap transducers will have to be incorporated into the instrumented orthosis so that both the strap and upright loadings can be sampled simultaneously throughout the walking cycles.

The shearing forces and bending moemnts at various points on the orthosis were also calculated. In order to illustrated the anterioposterior and mediolateral shearing forces throughtout the length of the orthotic uprights, the results of the strap transducer forces were interpolated so that their values may be used in conjunction with those obtained from the calculation to form the shear force diagrams of Fig 7.53, 7.54 & 7.57 . Their corresponding MZ bending moment diagrams are shown in Fig 7.55 & 7.56, while Fig 7.58 provides the MX moment diagram for the early stance of patient DC. It can be seen that the loadings on the lateral upright is consistently higher than its medial counterpart, and that the orthosis has to sustain much higher moments at the early stages of the stance phase than the rest of the walking cycle.

The maximum MX bending moment was obtained at the orthotic knee assembly of the medial upright, when MX reached a value of 8 Nm. For the MZ moment, the maximum was calculated to be at the lateral distal section of the orthotic knee joint assembly, i.e. at the distal strap of the knee apron on the lateral upright, where the loading reaches a magnitude of -21 Nm, some 4 Nm or 24% above that measured at the lateral distal transducer of the orthotic upright. This shows that

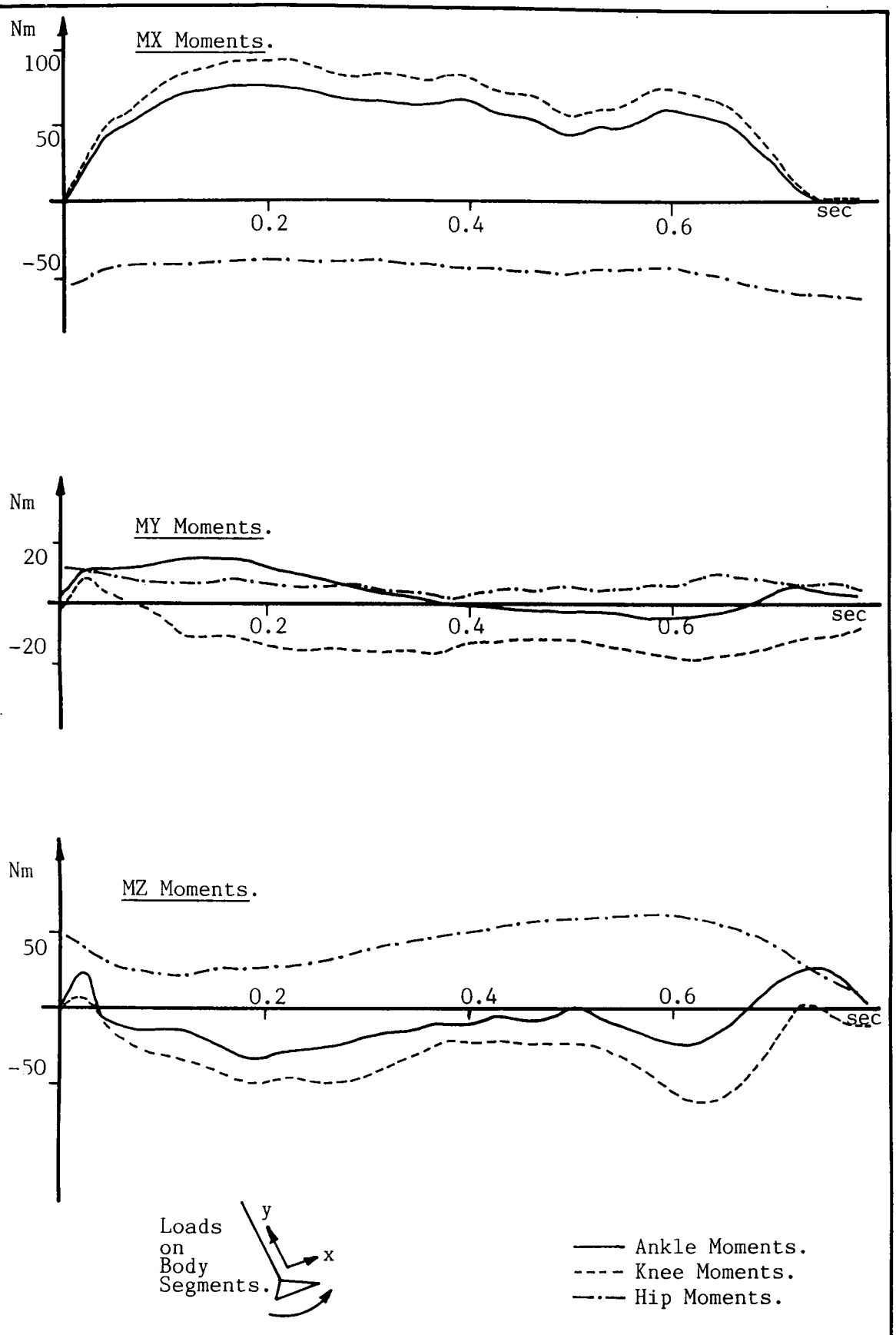
the transducer loading recorded provides a good indication for the approximate maximum loadings experienced by the orthosis.

The results of the transducer loadings expressed in terms of the respective transducer reference systems were also compared with those in the orthotic upright systems. It was found that the results correlate very well with each other, with an error of less than 5% for all sections except the lateral proximal, where the discrepancy is slightly larger. This is due to the fact that the lateral proximal section is normally bent laterally and posteriorly to follow the contour of the enclosed limb.



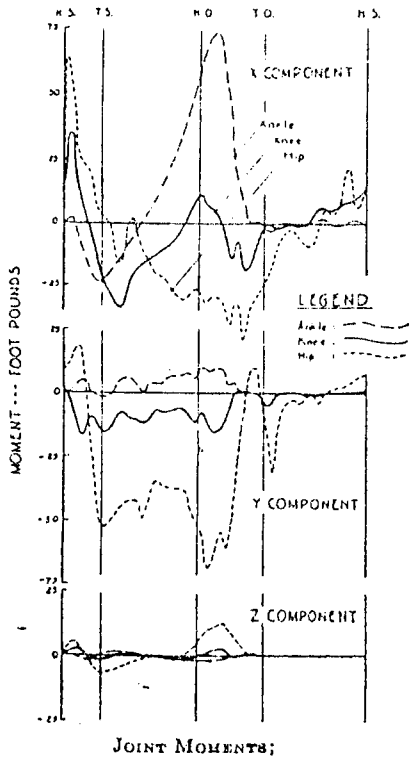
(Results of Patient DC)

FIG. 7.48	Intersegmental Anatomical Joint Forces.

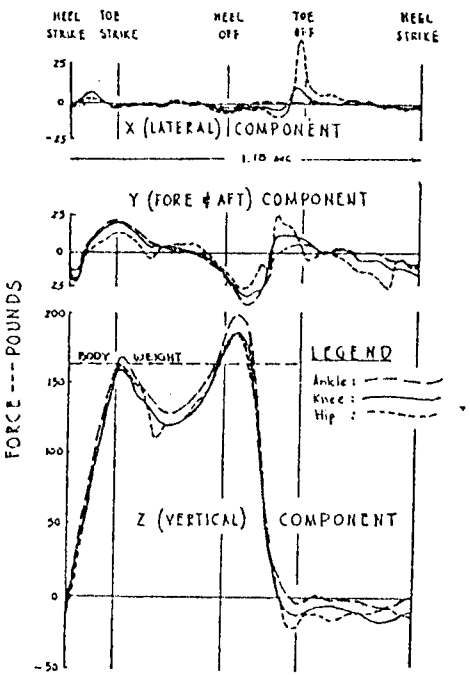


(Results of Patient DC)

FIG. 7.49	Intersegmental Anatomical Joint Moments.



JOINT MOMENTS;



JOINT FORCES;
(Time for one complete stride, 1.18 sec.)

(from Bresler and Frankel, 1950)

FIG 7.50	The forces and Moments (Intersegmental Loadings) During Level Walking.

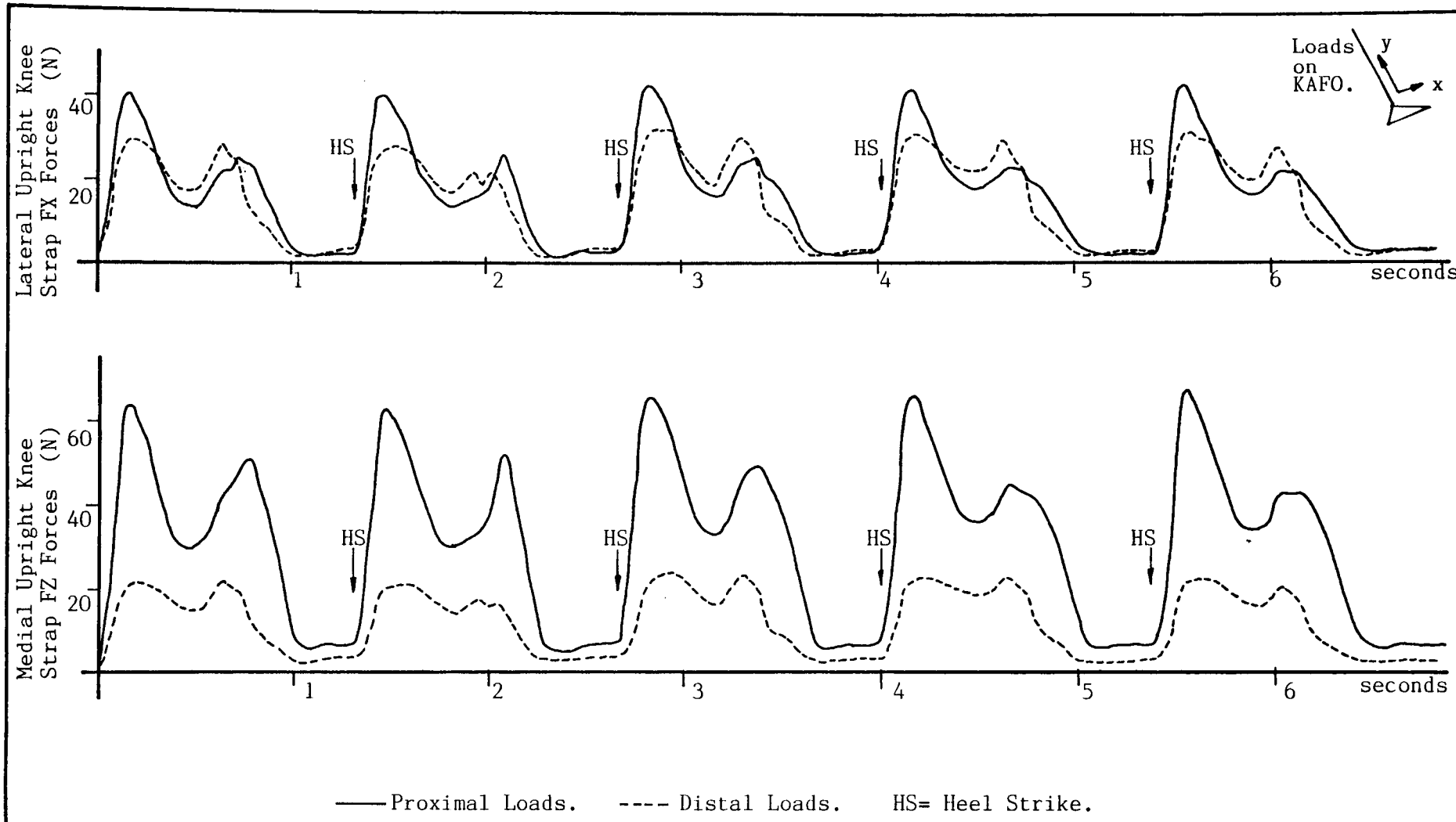


FIG. 7.51 Results of Knee Strap Transducer Forces on the Lateral Upright of Patient DC.

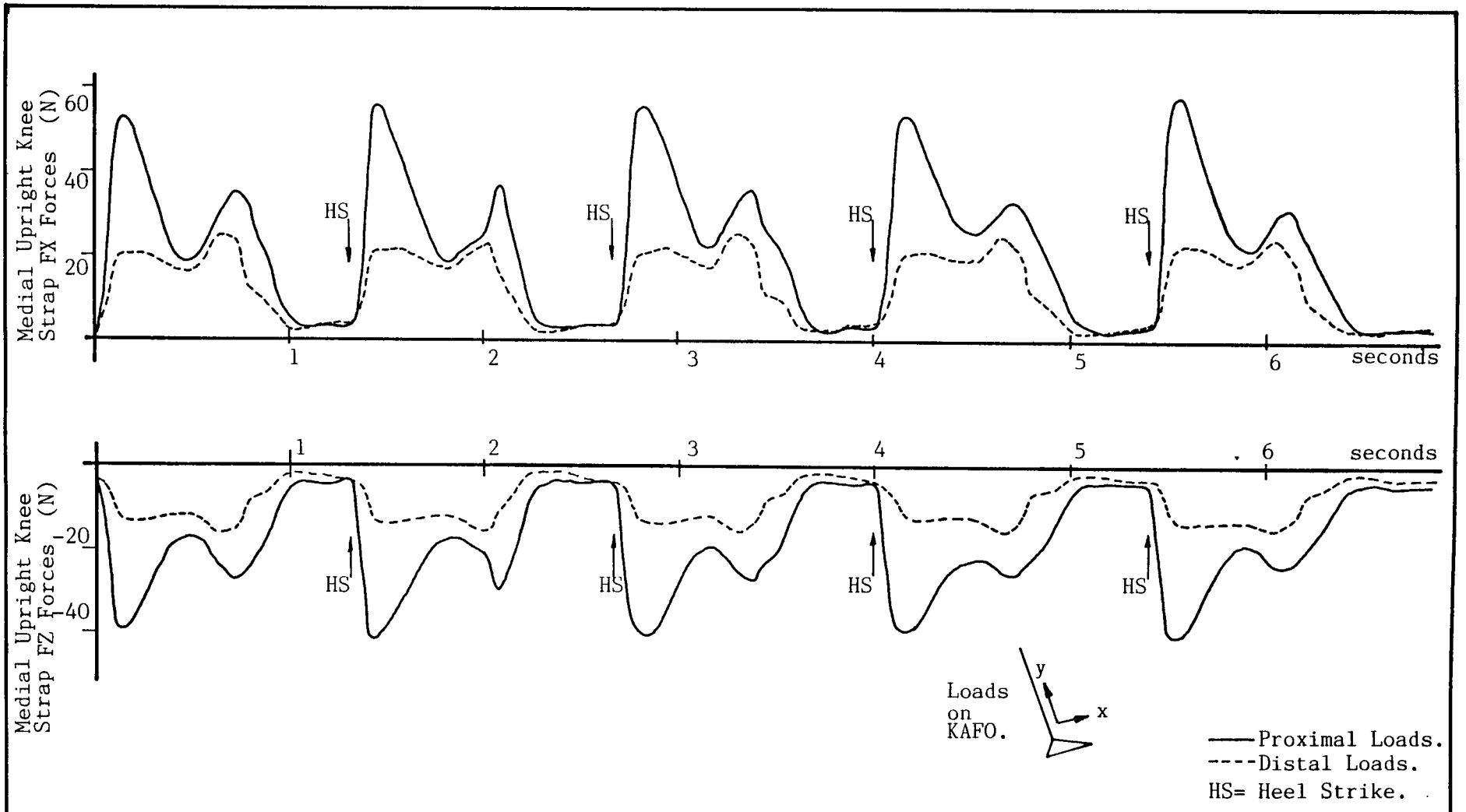
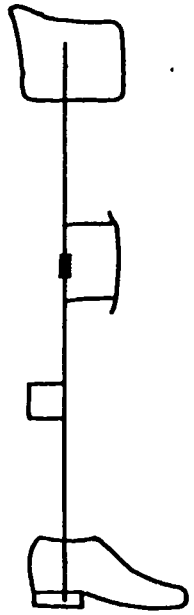
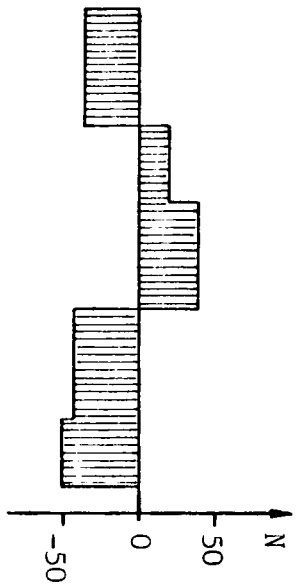
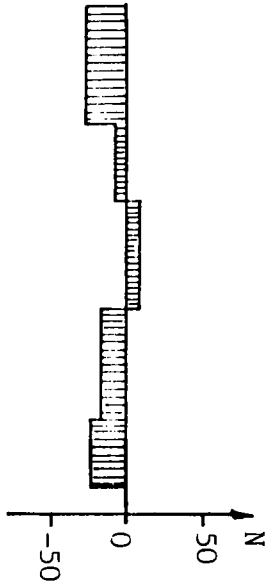


FIG. 7.52 Results of Knee Strap Transducer Forces on the Medial Upright of Patient DC.

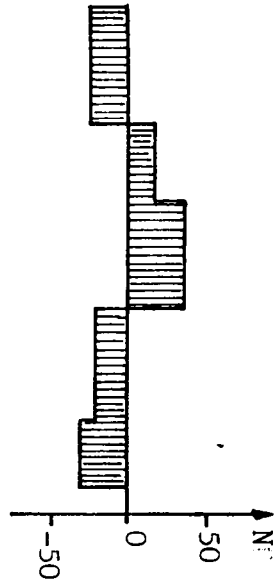




(Early Stance)

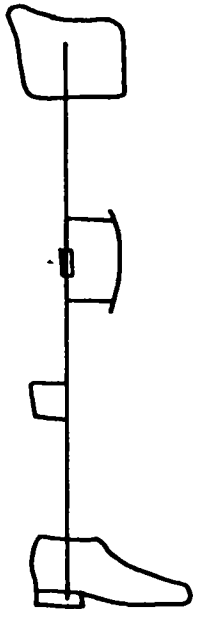


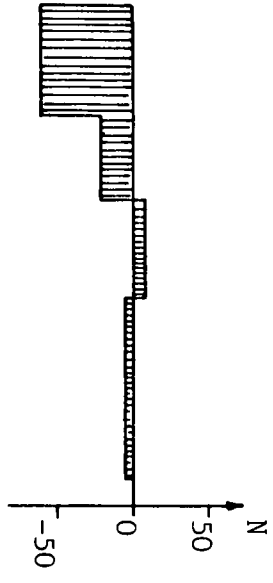
(Mid-Stance)



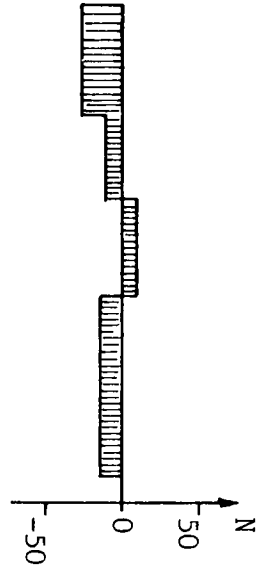
(Late Stance)

FIG 7.53	FX Shear Force Diagrams of the Medial Upright of Patient DC.

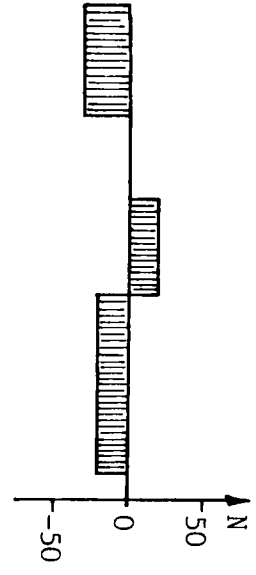




(Early Stance)



(Mid-Stance)



(Late-Stance)

FIG 7.54	FX Shear Force Diagrams of the Lateral Upright of Patient DC.

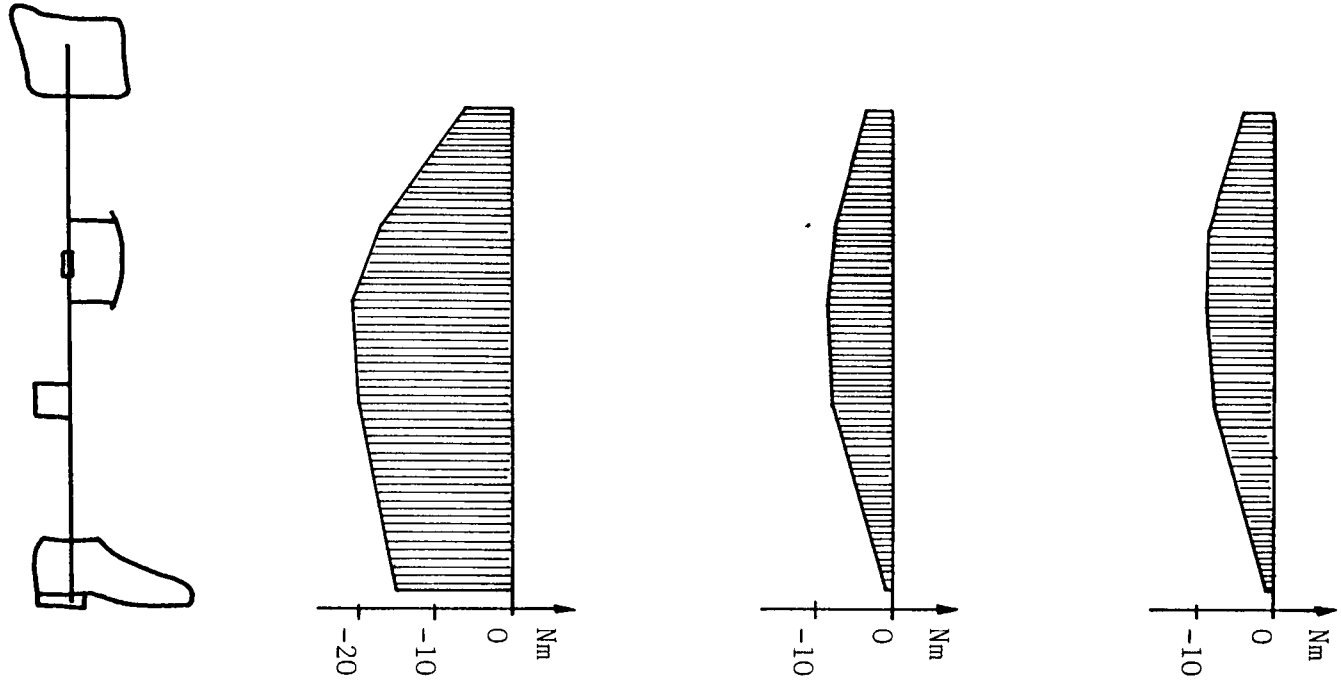
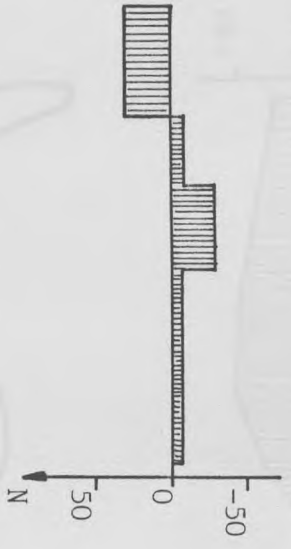
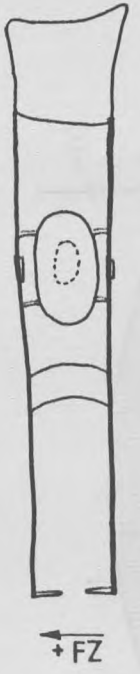


FIG 7.56 MZ Bending Moment Diagrams of the Lateral Upright of Patient DC.





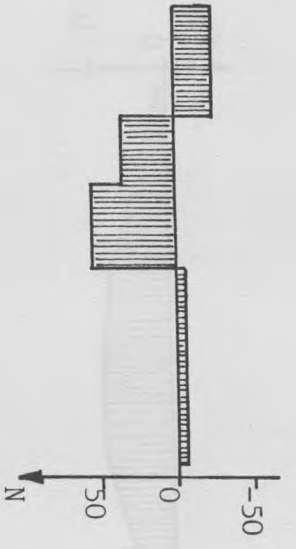


FIG 7.57	FZ Shear Force Diagrams of the Uprights for Patient DC at Early Stance.

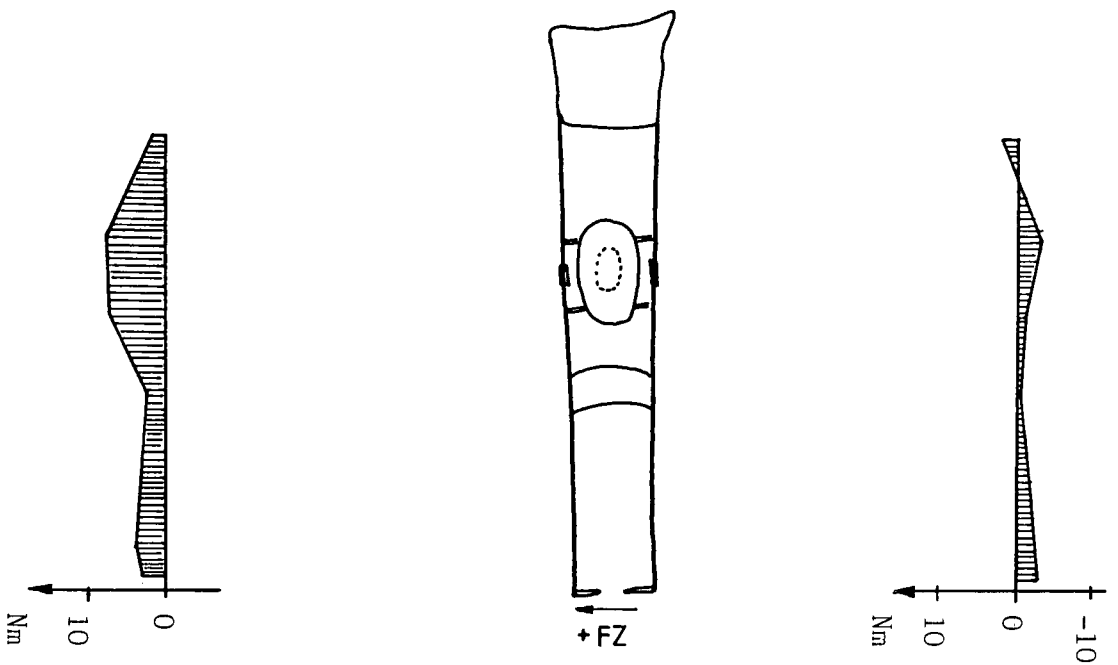
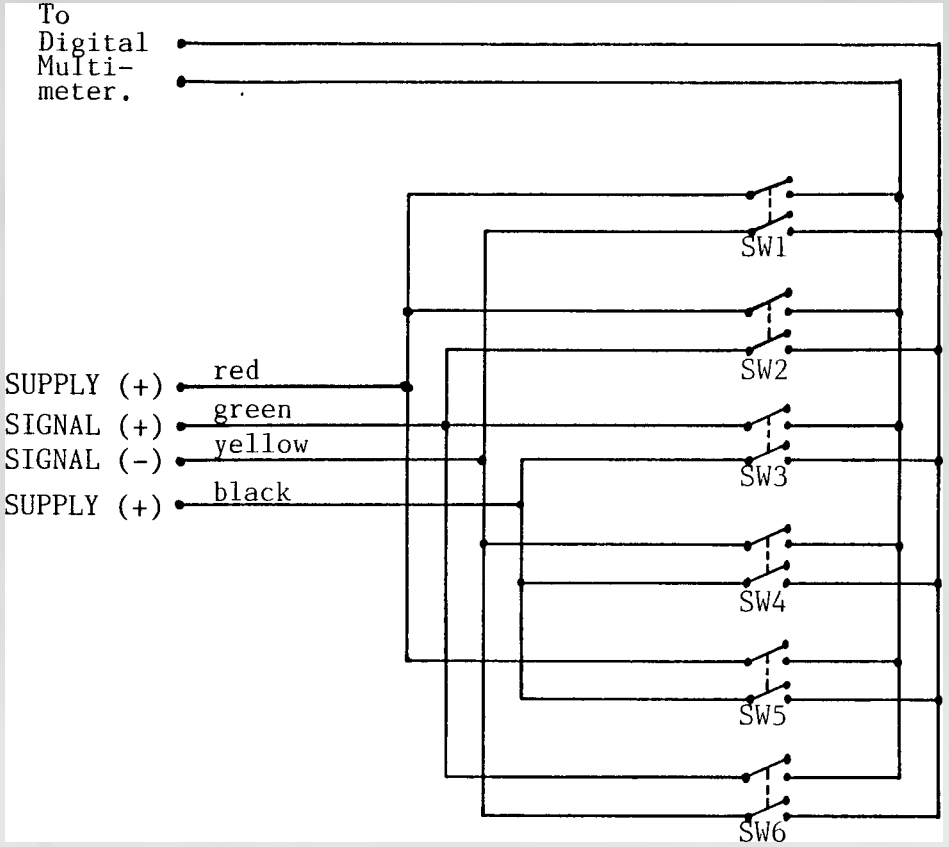


FIG 7.58	MX Bending Moment Diagrams of the Uprights for Patient DC at Early Stance.

CHAPTER 8.

CONCLUSIONS AND RECOMMENDATIONS.



SW1: SUPPLY (+) to SIGNAL (-)
SW2: SUPPLY (+) to SIGNAL (+)
SW3: SUPPLY (-) to SIGNAL (+)
SW4: SUPPLY (-) to SIGNAL (-)
SW5: SUPPLY (+) to SUPPLY (-)
SW6: SIGNAL (+) to SIGNAL (-)

FIG. 8.1	Bridge Continuity Tester.

8.1 INTRODUCTION.

This chapter presents a final discussion on the participating patients and the instrumentation used in the KAFO study. The problems encountered in the tests will be briefly described. Conclusions are drawn from the results of the orthotic loadings presented previously in Chapter 7.

A short analysis of the experimental accuracies is also included in this chapter, and finally, it concludes with some thoughts and ideas for any future research in this area of the lower limb orthotics.

8.2 CONCLUSIONS ON INSTRUMENTATION AND PATIENT SELECTION.

8.2.1 Instrumentation.

In general, the data gathering system performed very well throughout the study after a few initial teething problems have been overcomed. One of these was the intermittent occurrance of an open-circuit condition between the transducers and the amplifiers, the cause of which could either be due to a bad soldering point or a loose connection in the circuit. A "Bridge Continuity Tester" was therefore designed and constructed by the author to trace the source of the open-circuit (Fig 8.1). With the aid of the tester, the faults were able to be identified and corrected with relative ease.

All the transducers, namely, the Mark I and II KAFO transducers, and the knee strap transducers have also performed satisfactory in the tests. With the experienced gained, some comments on the design aspects may be made. The Mark I KAFO transducers were designed by Trappitt (1979) with sensitivity as the main criterion, of which he stated that they were designed to "as near to yield strain as possible with the expected maximum normal loading". This inevitably imposes a number of limitations on the usage of the device. For Mark II Transducer, the design criteria have moved more towards strength than sensitivity. This allows the transducers to be used in almost any type of activities that can be undertaken by the patients. However, with some hind-sight, a slightly too large reduction on sensitivity may have been sacrified for the improved strength.

The strain gauge amplifiers and the multiplexer have also

performed satisfactorily throughout the tests. The maximum drift and peak-to-peak noise level of the amplifier were monitored over a period of more than 6 hours to be at 5 μV and 1 μV respectively. For an amplifier output of less than 1V, the maximum drift will be better than 3% of the output level.

In the definition of the Ground Reference System (GRS), field markers were used instead of the more generally adopted "superimposed calibration grid board" method. The field marker system has been very successful, and if used properly, less human and random errors would be introduced since it was not necessary to rewind the film for re-exposure to the grid board.

A number of patients were sensitive to the noise of the synchronous motors which drive the cine-cameras. Hesitations were observed in their normal walking patterns as the cameras were being switched on just prior to the force platform heel-strike. Some patients have also shown sign of anticipation for the sudden onset of the noise during the tests. This unnecessary psychological burden, which in addition to the awareness of walking semi-clothed in an unfamiliar environment would not help to bring out the normal gait patterns of the patients. Moreover, the knowledge that one is being filmed would be an impediment to the normal behaviour of the subjects. Therefore, the kinematic data recording system should be improved such that filming can be carried out without causing undue anxiety to the patients.

Some problems were encountered in the selection of marker sites for the "raised shank marker BM". The marker tends to vibrate (up to ±10 mm) during the early stages of the stance phase, which is caused by a lack of muscles and excessive soft tissues at the affected limb. A compromise was reached by selecting a site with the minimum vibrational problem, and by accepting a fractional increase in the inaccuracy of the results. Any future work that demands a high level of accuracy would require either a new method of defining the shank system, or to filter the data digitally.

8.2.2 Selection of Patients.

There have been considerable difficulties in obtaining suitable patients for the tests. Lists of names together with brief case histories were supplied by collaborating clinicians from the local

Table 8.1 Table Showing the Highest Values of Loadings (in both Positive and Negative Directions) Experienced by a KAFO.

Loading Channels	Loadings on Orthotic Uprights.							
	Lateral Proximal		Lateral Distal		Medial Proximal		Medial Distal	
	Maximum	Minimum	Maximum	Minimum	Maximum	Minimum	Maximum	Minimum
FX (N)	147	-39	127	-137	186	-37	92	-75
FY (N)	325	-99	379	-266	254	-122	404	-145
FZ (N)	67	-99	52	-48	52	-55	32	-57
MX (Nm)	6	-4	5	-8	7	-8	4	-15
MY (Nm)	5	-4	8	-4	6	-5	5	-7
MZ (Nm)	21	-30	12	-24	22	-33	17	-28

hospitals. But the success rate in choosing the patients that satisfy the selection criteria outlined in Section 5.1 has been very low. For a particular case, where a clinician supplied a list of 100 cases of lower limb orthoses prescribed over a period of 14 years, a success rate as low as 3% was obtained. Of the 100 cases, 42 were classed under "below knee iron", 46 as "above knee iron" or "long leg caliper", and 12 were in other categories. The 46 patients assumed to have been prescribed with some form of KAFOs were contacted through the mail, and of the 22 who replied, only 3 were suitable.

This therefore indicates either over-demanding patient selection criteria, or a genuine lack of suitable patients. However, since lower limb orthoses are designed to aid patients in their attempt to resume normal ambulatory activities, it would not be too unrealistic to select patients who use them for normal walking purposes. There is also a group of patients who would only use the orthoses as a means of support in the upright posture, or for the transference to and from a wheelchair. In future works, the tests should also include this group of patients.

From the experience gained from patient selection, it has also become apparent that large proportion of the KAFO wearers are of the 'transient' type as against the 'permanent' type for post-poliomyelitis patients. The 'transient' KAFO wearers are normally the results of fractures or other injuries. With a minimum alteration to the testing procedures, these patients may also be included in any future studies. Another category of patients that should also be considered is the multiple sclerosis and spina bifida patients.

8.3 CONCLUSIONS ON RESULTS.

Table 8.1 shows the maximum loadings that had been experienced by a KAFO while being used for the 6 activities as described previously in Section 5.4.2 . From the tabulated results, it can be seen that the important loading channels are the axial load FY, and the MX and MZ bending moments in the mediolateral (ML) and anterioposterior (AP) directions respectively. The overall maximum AP moment obtained was -33 Nm in the medial proximal section of the upright, followed closely by -30 Nm at the lateral proximal upright. The maximum MZ values at the distal sections were -28 Nm and -24 Nm at the medial and lateral

sections respectively. In the ML direction, the overall maximum MX moment occurred at the medial distal section with a magnitude of -15 Nm. However, it should be noted that the relatively high -33 Nm MZ moment at the medial proximal section was obtained from patient 'MN' while performing the 'descending slope' and 'semi-circular path' ambulatory tests. For normal straight line level walking tests, the highest loaded section in the AP direction was the lateral proximal, at a magnitude of -30 Nm (by patient MR).

Assuming that the orthotic uprights were manufactured according to the recommendations laid down in BS 2932 and BS 2574, the minimum tensile and yield strength of a steel upright would be 700MPa and 355MPa respectively. For an "Otto Bock" (16mm x 4mm) steel upright, the -33 Nm MZ moment and the -15 Nm MX moment would produce bending stresses of 195MPa and 350MPa in the AP and ML directions respectively. The bending stresses due to the MX moment is thus above the fatigue strength of ±300MPa (at 10×10^6 cycles) for a smooth specimen at room temperature (Smithells & Braudes,1976). However, this fatigue strength has not taken any stress concentrations such as notches, irregularities, and drilled holes into its considerations, which could further reduce the fatigue strength. It therefore shows that although the MZ moment has a magnitude of more than twice that of MX, AP stresses are much lower than in the ML directions, due to the sectional properties of the uprights. Additionally, it also shows that with such values of MX moments, the uprights would be susceptible to fatigue failure. However, these loadings were the overall maxima obtained from all the tests. In normal circumstances, they would probably occur one in every few hundred walking cycles.

For the more representative results obtained from the normal level straight line walking tests, maximum repetitive loadings of approximately -30 Nm and -11 Nm were obtained for MZ and MX respectively. This would induce a stress level of 176MPa in the AP direction, and 258MPa in the ML direction. Although the stresses produced are below the fatigue strength of the material, any imperfection in the upright would, however, produce a stress concentration leading to a failure. Moreover, these loadings are being repeated with every step taken by the patient, and a level of 1,000,000 cycles a year can easily be reached even if the patient walks for an average of only a mile a day.

It should be noted that the results discussed above were the loadings measured at the transducer levels on the orthotic uprights. Since the ambulatory activities were monitored in the Limited Range Free Walking (LRFW) tests where kinematic data were not recorded, it is therefore not possible to calculate accurately the loadings on the orthotic knee assemblies. However, it has been shown in the previous chapter (Section 7.4) that the maximum loadings at the orthotic knee mechanism for patient DC are some 24% higher than its adjacent transducer readings. By considering that this factor also applies to other patients, the maximum bending moments at the orthotic knee mechanisms will be -40 Nm and -20 Nm in the AP and ML directions respectively.

Scothern (1982) reported on the results of mechanical tests carried out on a number of different types of orthotic knee joints supplied by 4 manufacturers. The failure loads in the AP direction ranges from 50 Nm to 290 Nm, depending on the make and type of the joint assembly. For the "Otto Bock" 16 mm knee joint used in this study, the joint mechanism and the upright were reported by Scothern to fail in the AP direction at 220 Nm and 160 Nm respectively. This indicates that there is a safety factor of 5.5 against a sudden failure of the knee joint. For the orthotic uprights, it is reduced to a factor of 4.8 . However, these safety margins will be further reduced when other stress concentration factors are taken into the considerations.

The test results do not provide a confident correlation between the orthotic loading patterns monitored and the residual muscle power of the patients. The patient-to-patient variations in the loading patterns can be caused by one or more of the following factors:

1. the patient's individual walking style or habit, which includes the "aggressiveness" of the walk, and the amount of lateral trunk bending during the stance phase;
2. the fit and alignment of the orthosis for the patient;
3. the design of the orthosis, including the shape of the weight-bearing top, and the type and placement of the bands and straps;
4. the amount of the body weight the patient is confident enough to exert on the orthosis;
5. the types of activity undertaken by the patient; and

6. the physiological and psychological conditions of the patient at that time.

It should be noted that the results reported in this thesis should not be taken as the established "norm" for the loading patterns of a KAFO, but as an indication towards the type of loadings that may be experienced by an orthosis in normal ambulatory activities. The maximum magnitudes of the orthotic loads may be used as the minimum strength requirement of an orthosis in any future design work.

8.4 EXPERIMENTAL ACCURACIES.

This section contains a brief analysis of the experimental accuracies of the results. The possible sources of errors are:

1. instrumentation errors;
2. errors introduced during the measurement process; and
3. errors caused by the use of simplifications and approximations.

The instrumentation errors emcompass those due to drift, noise, and resolution of the equipment used in the experiment. Additionaly, it also includes the calibration errors of the transducers. Some of these have already been discussed in the previous chapters.

The amplifiers and multiplexer have a maximum error of 3%, obtained from a long term monitoring of drift and peak-to-peak noise level. A total error of 2.5% is attributed to the transducer calibration process, where the dead weight accuracy was better than 0.15%. Since the PDP 11/34 computer converts a 1V signal with an accuracy of 1 in 2048, i.e. better than 0.05%, the errors of the transducer loads measured would be less than 5.5%. For Limited Range Free Walking tests, the results were presented in the transducer reference systems. This has an uncertainty of less than 5% as compared with those expressed in the orthotic upright systems. Therefore, the results of the LRFW tests will have an accuracy of better than 10%, i.e. errors of less than 30N in the FY forces, and 3 Nm in the MZ moments of patient MR.

For Normal Level Walking (NLW) tests, the greatest potential source of error will be from the kinematic data. If the patient stepped centrally on the force platform (i.e. within 440mm x 264mm of platform surface area), the manufacturer of the device claimed an accuracy of 2%, with a cross-sensitivity of better than 1.5% between the channels.

For the displacements, human estimation error of 2mm may occur in locating the centre of the markers with the flying curser of the digitising tablet, the accuracy of which is better than 0.3mm. Other errors include the laterally offset knee and ankle joint centre markers (see Section 6.5.1) of 5mm, and a 5mm vibration of the raised shank marker BM. An estimated maximum approximation error of 10mm in the location of hip joint centre may occur if the patient has a deformed pelvis. Another possible source of approximation error arises from the assumption that there is negligible deformation of the orthosis throughout the gait cycle. Although no upright deformation was visually detectable, a simple beam analysis has shown that the deformation may be up to 5mm, depending on the loading conditions.

By considering all the above factors, the errors due to the orthotic forces and moments will be within a maximum uncertainty of 15%. However, for the anatomical loads, the errors may be as high as 50% due to the inaccuracies of the assumptions made in the calculation of the knee strap forces (sss Section 7.4). This error reduces to less than 25% when the results from the strap transducers were incorporated into the analysis through interpolation.

8.5 RECOMMENDATIONS FOR FUTURE WORK.

As a result of the experience gained from this study, the following recommendations for future work are suggested:

1. make a wider survey of KAFO patients;
2. perform more patient tests;
3. incorporate the knee strap transducers into the normal level walking (NLW) tests;
4. improve the kinematic data recording system; and
5. conduct outdoor tests for KAFO patients.

A survey of KAFO patients in the local health board areas would be a logical approach in response to the difficulties faced in obtaining suitable patients for the tests. The survey should include all the orthopaedic departments of hospital in and around the city. The number and types of patients requiring KAFO for ambulation, the types of disability involved, and the number of KAFOs prescribed as compared with AFOs and other orthoses should be included in the survey. In addition

to identifying suitable patients for the test, it may also indicate areas of interest for future orthotic research.

In order that statistically acceptable results may be obtained, more patient tests will be needed, both in the number of participating patients and in the number of tests carried out for each activity. Each patient should also be tested on different occasions, preferably with an interval of several weeks, so that any changes in the loading patterns due to physiological and psychlogical factors can also be accounted for.

In future experiments, the knee strap transducers should be incorporated into the NLW tests. This would eliminate the requirement for the assumptions used in the calculation of the knee strap forces. However, this requires additional multiplexer channels to accommodate the 8 extra channels from the strap transducers. A redesigning of the amplifier-multiplexer system might be needed to improve the efficiency and functions of the orthotic load data acquisition system.

The noise generated by the cine camera system should be reduced or muffled to prevent it from distracting the patients in the tests. This can be achieved by either changing to a new set of driving motors, or more simply, by enclosing the driving mechanism with sound insulating materials.

Another recommendation for future work concerns the monitoring of the orthotic loadings in an out-of-the-laboratory situation where the tests would be carried out in an actual daily living environment. This inevitably will require the use of multi-channel data recording system, such as the tape-recording system by Lovely (1981). In this case, the number of channels recorded may be reduced by monitoring only the more critical loads such as the FY, MX and MZ channels.

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APPENDIX 1.

DISABILITY AND REHABILITATION IN DEVELOPING COUNTRIES.

A1.1 INTRODUCTION.

It has been estimated by the United Nations that out of the 4300 million world population, some 450 million have a physical or mental impairment, increasing by 15 million each year (UN,1983). In other words, one person in ten is disabled, and tragically, over two-thirds of these live in developing countries, mostly without any rehabilitative services. A complete breakdown of the types of disability and the population affected was not available at the time of writing this thesis. However, an estimated 30% of the total number of people having a disability were due to disease or accidents. In India alone, there are approximately 60 million physically handicapped persons, with between 0.7 to 1 million being added to the statistics each year. By the year 2000 AD, there would be more than 75 million disabled in India (Murkherjee,1983). The picture is relatively better in the affluent Western society, for example, there are only 27 million within the European Communities suffering from some kind of impairment, of whom the largest group are the elderly (European Communities,1981).

A1.2 THE PROBLEM OF DISABILITY FACING DEVELOPING COUNTRIES.

Although the cause of impairment may be similar throughout the world, their consequences do vary across the globe in terms of the disability experienced in their daily life and in terms of the handicap as a result of expectations and attitudes by the society. There seems to be a direct relationship between a nation's level of development, its political-economic structure, the types of disabling disease prevailing, its rehabilitation services and policies, and the nature and implication of disability experienced by its citizens (R.I.,1981; Tate & Letsinger,1983). In the industrialised society, with the exception of congenital disabilities, occupational hazards, and road accidents, disability is normally associated with aging process. Disability in the developing countries, in contrast, is caused mainly by malnutrition, uncontrolled infectious diseases, poverty, ignorance, pseudo-religious beliefs, and low quality of elementary medical facilities.

To understand the reasons behind the very serious lack of disability prevention measures and rehabilitative services in a large number of developing countries, it is necessary to study their priorities

for utilization of the very limited resources. Planning for the disabled population in the industrialised nations is very much different as compared to that of the developing countries, where the needs of the disabled are very different and they live in entirely different socio-economic environments, with a very harshly constrained economy and limited manpower. As the country's emphasis is placed mainly on the development of industry and agriculture, and on a basic infrastructure of health, transport, communication, and education facilities, very negligible, if any, resources will be left for expenditure on disability and rehabilitative services. Moreover, it also tends to give priority on the development of systems and services that will be beneficial to the greatest number of people. This have the knock-on effect of keeping disabled population and their families amongest the lowest strata of society, both socially and economically. For most it is this exclusion by society which makes the impairment into a true disability.

The definition of disability and rehabilitative services is also restricted by the resources available to deal with the problem. Prevention of disability condition and primary health care are emphasised whenever the economical situation allows the country to do so. Comprehensive rehabilitation is a luxury for the rich and privilaged few. Each year, approximately 15 million people, mostly children, are being disabled by impairments which could have been prevented (Sinha & Yadava, 1983). The extent of disability that follows the impairments is often multiple, and is more functionally handicapping than could have been dictated by the impairment alone. Many of those being disabled are within reach of rehabilitative services, and yet are beyond reach. Because of this, they are often not identified early enough for action to be taken which could alter their progress towards severe disability in the later stages of their lives.

A1.3 DISABILITIES CAUSED BY POLIOMYELITIS.

Poliomyelitis is still today one of the major health problems in the school going age group in many developing countries. The United Nations Children's Fund gave a conservative estimate of an worldwide incidence of 1.5 million, with around 75,000 cases a year (UNICEF News, Issue 105,1980). Victims of the disease are mainly children under five in the poorer developing countries of the tropics and subtropics. It

is normally endemic (i.e. regularly present in an area) in an insanitary environment. Widespread occurrance of the disease is prevented because most children are immunised by faecal contamination in their early life. Poliomyelitis becomes epidemic when public health standard rises and children are not naturally immuned. This usually occurs when the infant mortality of a country falls below 75 per 1000 live births, if widespread vaccination is not carried out at the same time as hygiene and health standards are being improved.

The residual effects of polio are due to destruction of the anterior horn cells of the spinal cord and brain stem. This lower motor neuron lesion produces flaccid paralysis with normal sensation. Muscles affected are dictated by the level on which the spinal cord is affected, but the paralysis tends to affect some muscles more than others, with lower limb paralysis occurring in 75 to 90% of all cases (see Section 1.2.5). The exact quantification of the degree of paralysis is difficult because functional loss or disability does not always accurately relate to the degree of muscle paralysis due to various compensatory actions of the non-paralysed muscles.

For ambulation, the anti-gravity muscles must be functioning or substituted for by other non-paralysed muscles. Provided the hip have a relatively unrestricted range of motion, a patient with bilaterally affected lower limbs can stand upright using 2 KAFOs with fixed knee and ankle joints. However, he will require good strength in the trunk, upper extremities, and glutei, in addition to the ability to advance the leg forward in order to ambulate functionally. Advancement of the leg can be accomplished by hip flexors of poor strength, and/or by knee flexors of fair to good strength. Of all the losses of anti-gravity muscles, absence of quadriceps can be said to be of the least handicapping. Absent quadriceps can be replaced by strong gluteus maximus action or by strong plantar flexor of the ankle.

As a consequence of imbalanced muscle power, contractures are liable to occur. These deformities are frequently seen as flexion contractures of the hip and knee, and equinus deformity of the ankle. The reason is that flexors of the hip, knee, and ankle are often less paralysed than the extensors. Furthermore, in a weakened limb the lack of muscle action on the bone causes retarded growth which results in inequality of leg length. A shortened leg, or a flexed joint contracture causes the pelvis to tilt, if not corrected, a compensatory

scoliotic deformity may develop in the spine. To add to the list of deformities, sustained weight bearing on weak joints can lead to either a genu recurvatum or a valgus knee, and a valgus ankle. All the above deformities are progressive if left uncorrected in the growing children, unfortunately, the lack of elementary rahabilitative facilities will mean that many will be deprived of even standing upright, let alone walking freely.

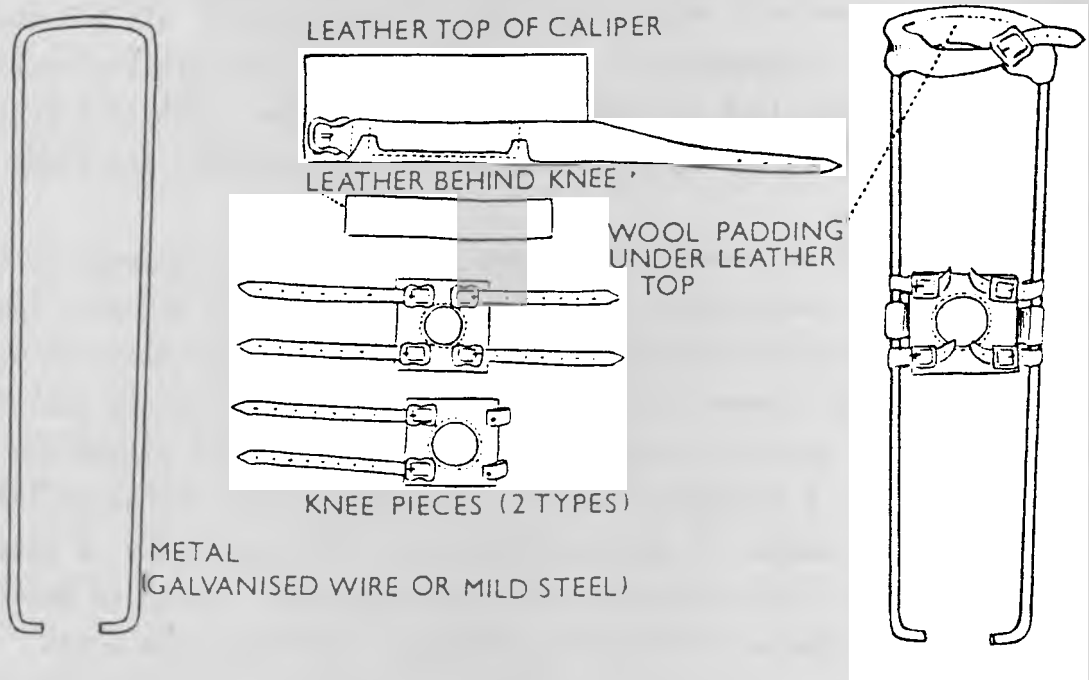
The number of severely paralysed patients with poliomyelitis in many developing countries is very worrying indeed. For example, there are more than 30,000 severely paralysed poliomyelitis patients in Uganda alone. The total number with some form of residual paralysis following a poliomyelitis attack is probably 90,000 out of a total population of 10 million, i.e. a chilling 9 out of every 1000 people (Huckstep,1975). In Nigeria, the number of untreated paralysed polio patients is estimated at between 200 to 300 thousand (Huckstep,1975), while in Morocco, there are approximately 50 thousand children suffering from polio (Florence,1983).

A1.4 THE USE OF LOW COST TECHNOLOGY IN REHABILITATION.

Within the past 30 years, there have been remarkable developments in the technologies related to rehabilitation. As a result of this technological advancement, many of the major disability conditions can be compensated to provide more mobility to the patients. However, only a small proportion of the total disabled population have benefited from the most innovative technological developments. The reason behind this being the imbalance in research emphasis, particularly between high and low technology, with a strong bias towards high technology. There are also far less research done on the underlying principles of a disability than on research in faviour of application to developments. Many research centres are reluctant to undertake work on simple,low-cost technology because it has always been regarded as more prestigious to develop complex and high technological solution. Research is urgently needed to provide solutions which are compatible with ways of making devices available and affordable to the mass of disabled population throughout the world.

A quick survey of the literature has shown that very little has been written on the modern management of severe deformities using simple

MANUFACTURE OF A SIMPLE CALIPER



COMPLETED SIMPLE ABOVE AND BELOW KNEE CALIPERS

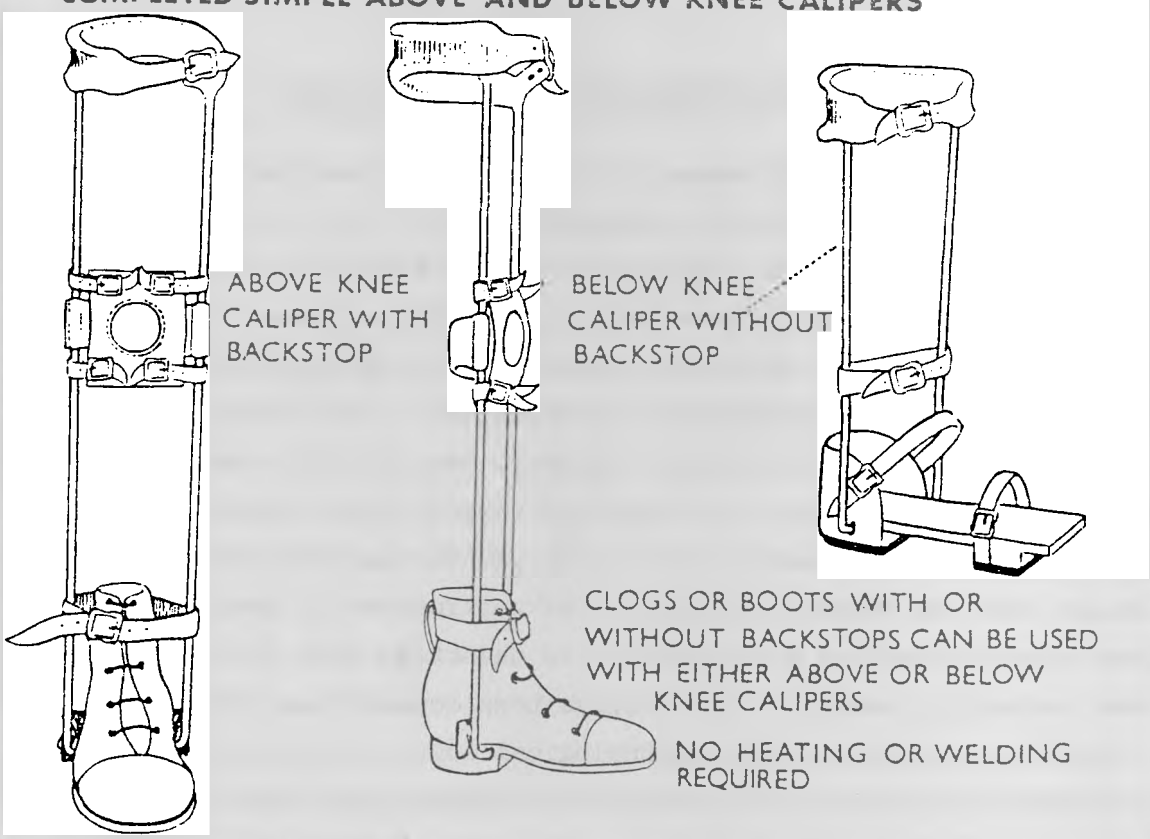


FIG. A1-1

LOW COST ORTHOSES.

(From: HUCKSTEP, 1975)

methods in developing countries, where millions of patients with poliomyelitis and other deformities still remain untreated. The kind of technical appliances that people in industralised countries associated with rehabilitation for the disabled tend to be sophisticated and expensive. Apart from the exorbitant prices of these devices, they are often quite unsuitable to the needs and inappropriate to the cultures and customs of two-thirds of the world's population.

An emphasis should be made for using locally available indigenous materials to make orthoses, crutches, wheelchairs, and other aids needed for activities of daily living by the physically handicapped. The inventiveness and creative concepts of a practical-minded designer can often lead to cheap and effective supports of various kinds. Manufacturing processes that involve 'modern' skilled craftmenship such as welding, heat curing, and other joining processes is to be avoided to keep the cost down to a minimum. Another advantage of using simple manufacturing technique is that virtually any unskilled worker can be employed in the production of the orthotic appliances. Ideally, the production team should consists of as many disabled unskilled persons as possible. This will, on one hand provide much needed employment to the disabled, but will also give them an oppotunity to help and serve their fellow disabled sufferers.

Numerous projects and appliances utilising low-cost technologies to improve the lives of the disabled have been reported. The most outstanding example can be attributed to Professor Huckstep who started the first orthopaedic workshop in Uganda with a negligible cost to produce KAFOs, AFOs, wheelchairs, and crutches. Other workshops were subsequently set up to produce several thousand orthoses every year, each costing only one-fiftieth of the price of an imported orthosis. Galvanised wires, mild steel or concrete reinforcing iron bars were used in the making of these simple orthoses (Fig A1.1)(Huckstep,1975). In India, Sengupta et al (Sengupta, Bhattacharya & Dutta,1983b) used clothes pegs to provide dynamic power for drop-foot orthoses. Simple but effective crutches can be made by splitting a length of hardwood (Huchstep, 1975) or from a simple frame of branches (UNICEF News,Issue 105,1980; Helander et al,1983). Rattan and bamboo are of abundant supply in Malaysia, and they were used for building stabilised cane in the shape of a sturdy quadruped walking stick. Children's walkers and trolleys

were also designed and built from rattan (Disabilities Study Unit,1979). In the Philippines, studies had been made on the feasibility of utilizing materials and technology available locally on the fabrication of technical aids. Prototypes of short-leg shoe inserts, hand splints, as well as protheses were designed and fabricated (Publico et al,1982).

APPENDIX 2.

THE AMERICAN ACADEMY OF ORTHOPAEDIC SURGEONS' LOWER-LIMB TECHNICAL ANALYSIS FORM.

(Adapted from McCollough, 1975)

TECHNICAL ANALYSIS FORM LOWER LIMB

Name _____ No. _____ Age _____ Sex _____

Date of Onset _____ Cause _____

Occupation _____ Present Lower-Limb Equipment _____

Diagnosis _____

_____

_____

Ambulatory ☐ Non-Ambulatory ☐

MAJOR IMPAIRMENTS:

A. Skeletal
1. Bone and Joints: Normal ☐ Abnormal _____
2. Ligaments: Normal ☐ Abnormal ☐ Knee: AC ☐ PC ☐ MC ☐ LC ☐
Ankle: MC ☐ LC ☐
3. Extremity Shortening: None ☐ Left ☐ Right ☐
Amount of Discrepancy: A.S.S.-Heel _____ A.S.S.-MTP _____ MTP-Heel _____

B. Sensation: Normal ☐ Abnormal ☐
1. Anaesthesia ☐ Hypaesthesia ☐ Location: _____
Protective Sensation: Retained ☐ Lost ☐
2. Pain ☐ Location: _____

C. Skin: Normal ☐ Abnormal: _____

D. Vascular: Normal ☐ Abnormal ☐ Right ☐ Left ☐

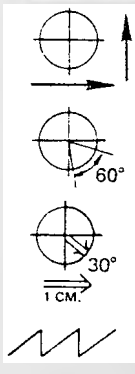
E. Balance: Normal ☐ Impaired ☐ Support: _____

F. Gait Deviations: _____

_____

G. Other Impairments: _____

LEGEND



= Direction of Translatory Motion

= Abnormal Degree of Rotary Motion

= Fixed Position

= Fracture

Volitional Force (V)
N = Normal
G = Good
F = Fair
P = Poor
T = Trace
Z = Zero

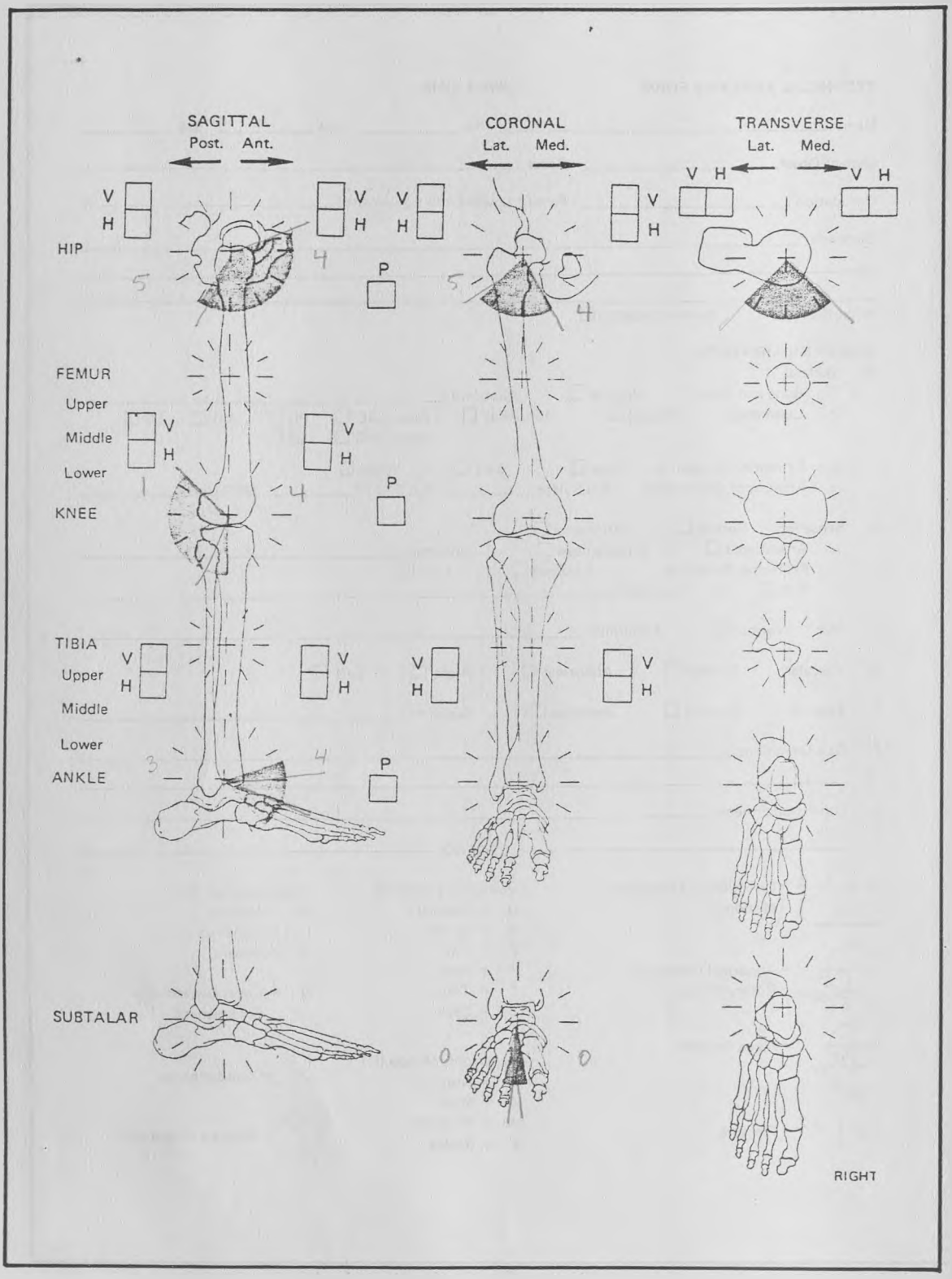
Hypertonic Muscle (H)
N = Normal
M = Mild
Mo = Moderate
S = Severe

Proprioception (P)
N = Normal
I = Impaired
A = Absent

D = Local Distension or Enlargement

= Pseudarthrosis

= Absence of Segment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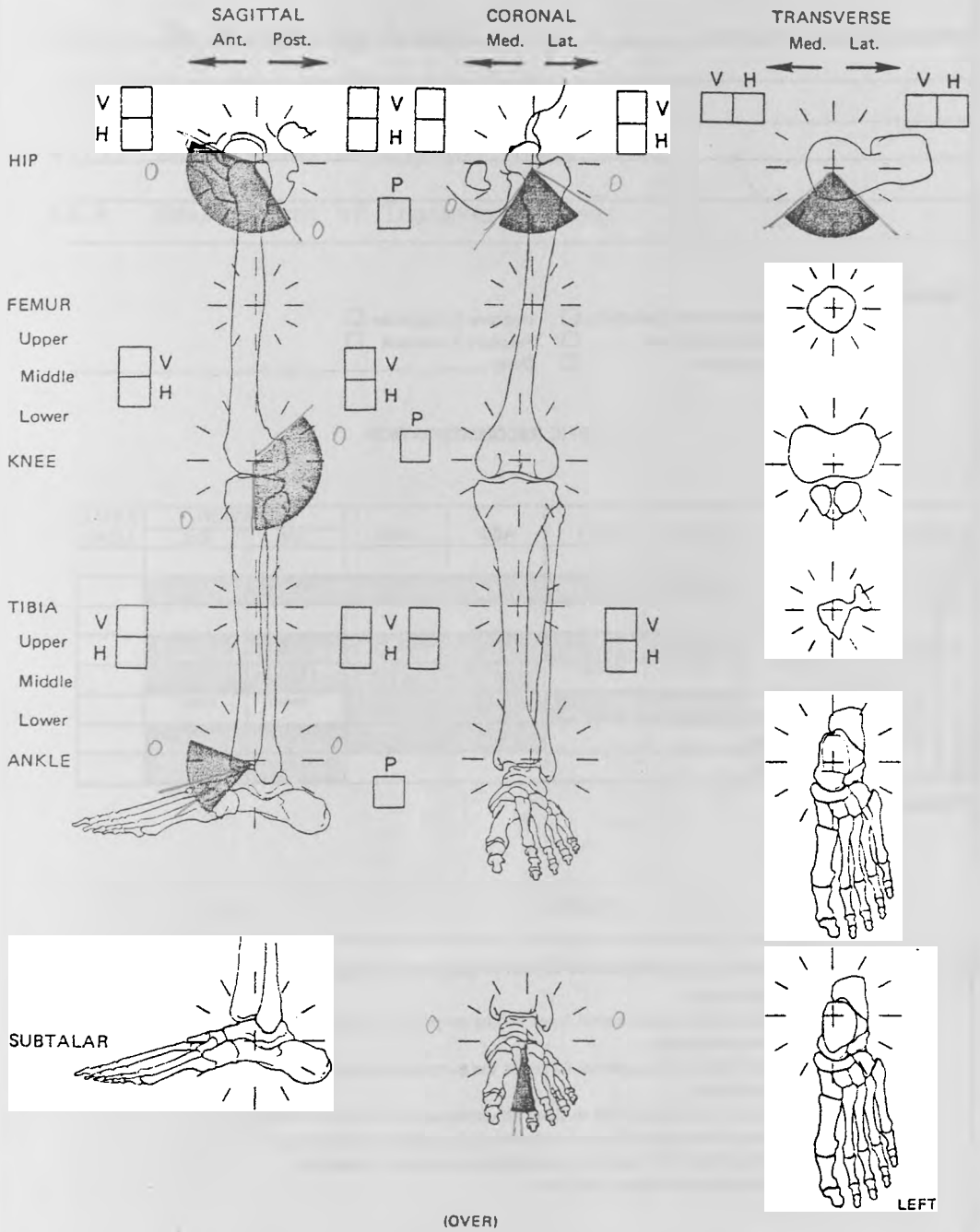
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Middle
Lower
ANKLE
SUBTALAR
LEFT

(OVER)

Summary of Functional Disability _____

_____

_____

_____

_____

_____

Treatment Objectives:

Prevent/Correct Deformity ☐ Improve Ambulation ☐
Reduce Axial Load ☐ Fracture Treatment ☐
Protect Joint ☐ Other _____

ORTHOTIC RECOMMENDATION

LOWER LIMB		FLEX	EXT	ABD	ADD	ROTATION		AXIAL LOAD
						Int.	Ext.	
HKAO	Hip							
KAO	*Thigh*							
	Knee							
AFO	*Leg*							
	Ankle	(Dorsi)	(Plantar)					
FO	Foot: Subtalar					(Inver.)	(Ever.)	
	Foot: Midtarsal							
	Foot: Met.-phal.							

REMARKS:

_____ _____
Signature Date

KEY: Use the following symbols to indicate desired control of designated function:

F = FREE — *Free* motion.
A = ASSIST — Application of an external force for the purpose of increasing the range, velocity, or force of a motion.
R = RESIST — Application of an external force for the purpose of decreasing the velocity or force of a motion.
S = STOP — Inclusion of a static unit to deter an undesired motion in one direction.
v = Variable — A unit that can be adjusted without making a structural change.
H = HOLD — Elimination of all motion in prescribed plane (verify position).
L = LOCK — Device includes an optional lock.

APPENDIX 3:

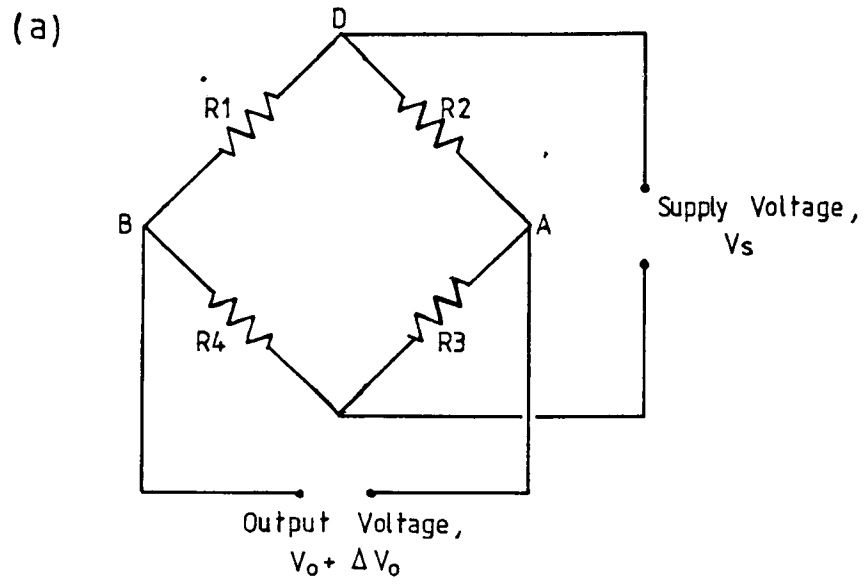
THE TRANSDUCER MEASURING CIRCUITS.

A3.1 The Wheatstone Bridge.

A3.2 Multiple Gauges in Each Bridge Arm.

A3.3 Measurement of Applied Loads.

A3.4 Measurement of Transverse Shear.



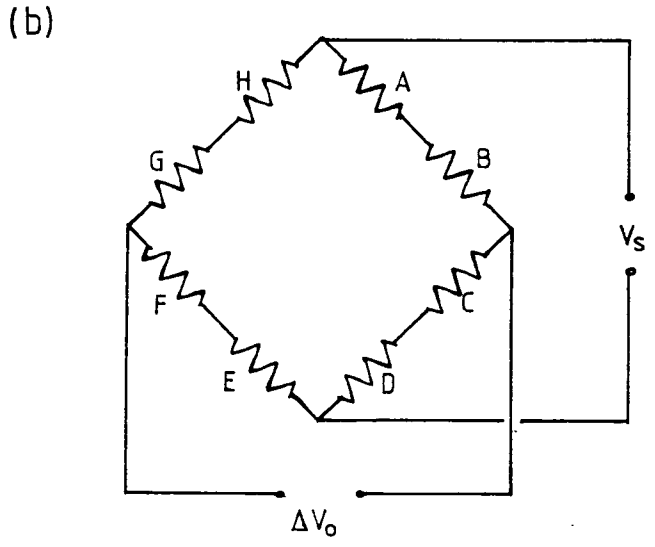


FIG A3·1	The Wheatstone Bridge.

A3.1 THE WHEATSTONE BRIDGE.

Consider a Wheatstone bridge (Fig A3.1a) consisting of 4 resistances, any number of which may be strain gauges. For an initially balanced bridge condition, V_o is zero, and V_o is a function of the strains in the gauges which made up the bridge, i.e.

$$dV_o = \frac{\partial V_o}{\partial R_1} dR_1 + \frac{\partial V_o}{\partial R_2} dR_2 + \frac{\partial V_o}{\partial R_3} dR_3 + \frac{\partial V_o}{\partial R_4} dR_4$$

If the resistance changes are small compared to the total value of the resistance, the equation becomes:

$$\Delta V_o = \frac{\partial V_o}{\partial R_1} \Delta R_1 + \frac{\partial V_o}{\partial R_2} \Delta R_2 + \frac{\partial V_o}{\partial R_3} \Delta R_3 + \frac{\partial V_o}{\partial R_4} \Delta R_4$$

Solving the above equation and considering an initially balanced condition where $V_o=0$, and $R_1R_3 = R_2R_4$, we have:

$$\Delta V_o = V_s \left[\frac{R_1R_3}{(R_1+R_2)(R_3+R_4)} \left(\frac{\Delta R_1}{R_1} + \frac{\Delta R_3}{R_3} - \frac{R_2R_4}{R_1R_3} \left\{ \frac{\Delta R_2}{\Delta R_2} + \frac{\Delta R_4}{\Delta R_4} \right\} \right) \left(1 - n \right) \right]$$

where n is a nonlinearity factor for the finite quantities. However, n is usually so small as to be negligible, then for practical purposes,

$$\Delta V_o = V_s \frac{R_1R_3}{(R_1+R_2)(R_3+R_4)} \left[\frac{\Delta R_1}{R_1} - \frac{\Delta R_2}{R_2} + \frac{\Delta R_3}{R_3} - \frac{\Delta R_4}{R_4} \right] \quad \cdots\cdots\cdot(A3.1)$$

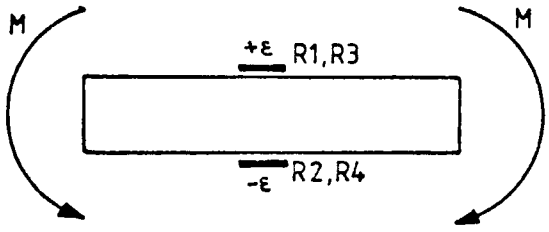
Since the unit change in resistance of a strain gauge is proportional to strain, therefore,

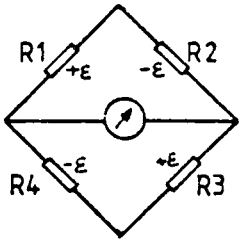
$$\varepsilon = [\Delta R/R]/K$$

where K is the gauge factor. For a nominal case where $R_1=R_2=R_3=R_4=R$, equation A3.1 becomes,

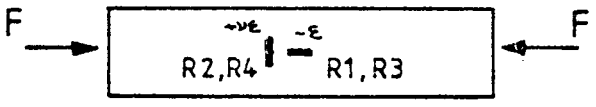
$$\Delta V_o = V_s \cdot \frac{K}{4} \left[\varepsilon_1 - \varepsilon_2 + \varepsilon_3 - \varepsilon_4 \right] \quad \cdots\cdots\cdots\cdots\cdots\cdot(A3.2)$$

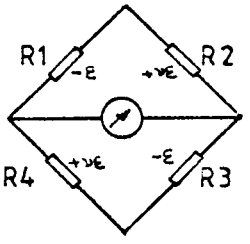
This equation is true for all full bridge circuits, whatever their purpose may be (Holister,1967; Massachusetts Institute of Technology, 1957; Perry & Lissner,1962).



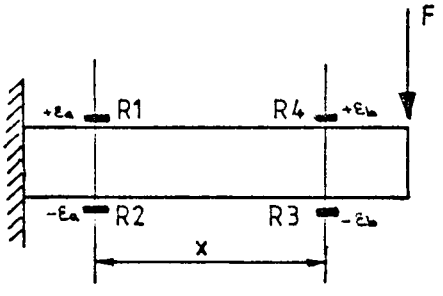


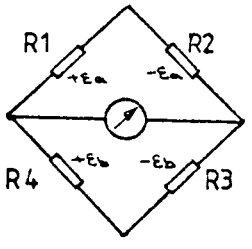
(a) Bending load.





(b) Axial load.





$$F = \frac{Ma - Mb}{x}$$

(c) Transverse shear.

FIG A3·2	The Measuring Circuits.

A3.2 MULTIPLE GAUGES IN EACH BRIDGE ARM.

Consider the circuit shown in Fig A3.1b, where there are two strain gauges in series in each arm of the bridge, and assuming that the gauges in each arm are subjected to the same strain in magnitude and sign, equation A3.1 becomes,

$$\Delta V_o = V_s \left[\frac{2R_1 \cdot 2R_3}{(2R_1+2R_2)(2R_3+2R_4)} \right] \left[\frac{2\Delta R_1}{2R_1} - \frac{2\Delta R_2}{2R_2} + \frac{2\Delta R_3}{2R_3} - \frac{2\Delta R_4}{2R_4} \right]$$

i.e. $$V_o = V_s \left[\frac{R_1 \cdot R_3}{(R_1+R_2)(R_3+R_4)} \right] \left[\frac{\Delta R_1}{R_1} - \frac{\Delta R_2}{R_2} + \frac{\Delta R_3}{R_3} - \frac{\Delta R_4}{R_4} \right]$$

Therefore, the use of multiple gauges in each arm of the bridge does not increase the output signal. However, the current passing through each gauge have halved, implying that the supply voltage can be doubled without exceeding the maximum safe current rating of the gauge. This increase of supply voltage will double the output of the bridge.

A3.3 MEASUREMENT OF APPLIED LOADS.

For a beam subjected to bending moment as shown in Fig A3.2a, the surface strain on the upper surface is equal and opposite to those on the lower surface of the beam, i.e.

$$[\varepsilon_t] = [\varepsilon_c] = [\varepsilon_b]$$

where the symbol [x] signifies the magnitude of x, and the subscripts t, c, and b represents the tensile, compressive, and bending strains respectively. Therefore,

$$\varepsilon_1 = \varepsilon_3 = +\varepsilon_t = +\varepsilon_b$$

and $$\varepsilon_2 = \varepsilon_4 = -\varepsilon_c = -\varepsilon_b$$

Substituting into equation A3.2,

$$\Delta V_o = V_s \cdot \frac{K}{4} [\varepsilon_b - (-\varepsilon_b) + \varepsilon_b - (-\varepsilon_b)]$$

i.e. $$\Delta V_o = V_s \cdot K \cdot \varepsilon_b$$

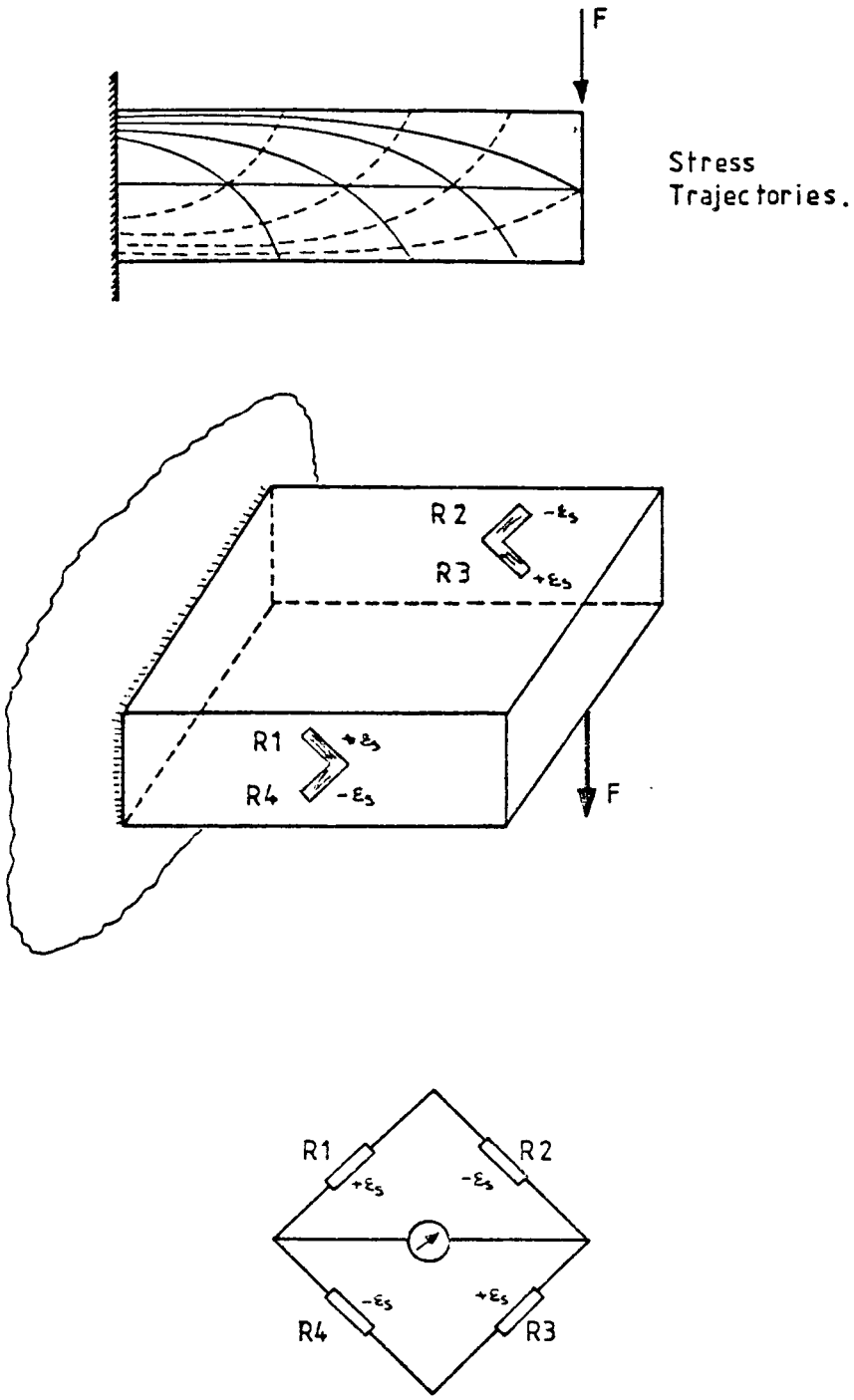


FIG A3·3	Measurement of Transverse Shear by Principal Strains.

It shows that the output of the bridge is 4 times as large as the corresponding output of a single gauge. If a strain level of 1500 $\mu\varepsilon$ were to be recorded at every gauge location in a transducer with full bridge circuits, and assuming a gauge factor K of 2.0, the nominal output would be:

$$\Delta V_o = 1 * 2.0 * 1500\ \mu\varepsilon$$

i.e. $\Delta V_o = 3$ mV per bridge voltage.

Similarly, for a beam subjected to an axial load (Fig A3.2b), the output can be shown to be:

$$\Delta V_o = V_s \cdot \tfrac{1}{2} K \cdot (1+\nu) \varepsilon_a$$

where ν is the Poisson's ratio, and ε_a the axial strain.
The value of ΔV_o for torsional moment is,

$$\Delta V_o = V_s \cdot K \cdot \varepsilon_t$$

where ε_t is the principal strain proportional to the applied torque.

A3.4 MEASUREMENT OF TRANSVERSE SHEAR.

Measurement of transverse shear can be achieved by either of the following method, i.e.

1. derived from the difference of bending strains at 2 levels, (Fig A3.2c);
2. monitoring of the principal strains (Fig A3.3).

The bridge output obtained as from Equation A3.2 are:

for method (1): $\Delta V_o = V_s \cdot \tfrac{1}{2} K \left[\varepsilon_a - \varepsilon_b \right]$

for method (2): $V_o = V_s \cdot K \cdot \varepsilon_s$

where ε_a and ε_b are the two bending strains, and ε_s the principal strain.

A comparison of the signal output level between a transducer that uses method (1) above and one that uses method (2) is given below. For one that utilises a rectangular cross-sectional area in the transducer element, the ratio of the signal output with method (1) to method (2) is:

$$\Delta V_{ab} = [2 \cdot x]/[h \cdot (1+\nu)]$$

where x is the distance apart between the two bending strain monitoring positions, h is the depth of the beam, and ν the Poisson's ratio. For a circular cross-sectional area with diameter d, the ratio becomes,

$$V_{ab} = [3 \cdot x]/[d \cdot (1+\nu)]$$

APPENDIX 4.

ASSESSMENT OF MARK I KAFO TRANSDUCERS.

A4.1 Theoretical Analysis.

A4.2 Experimental Assessment.

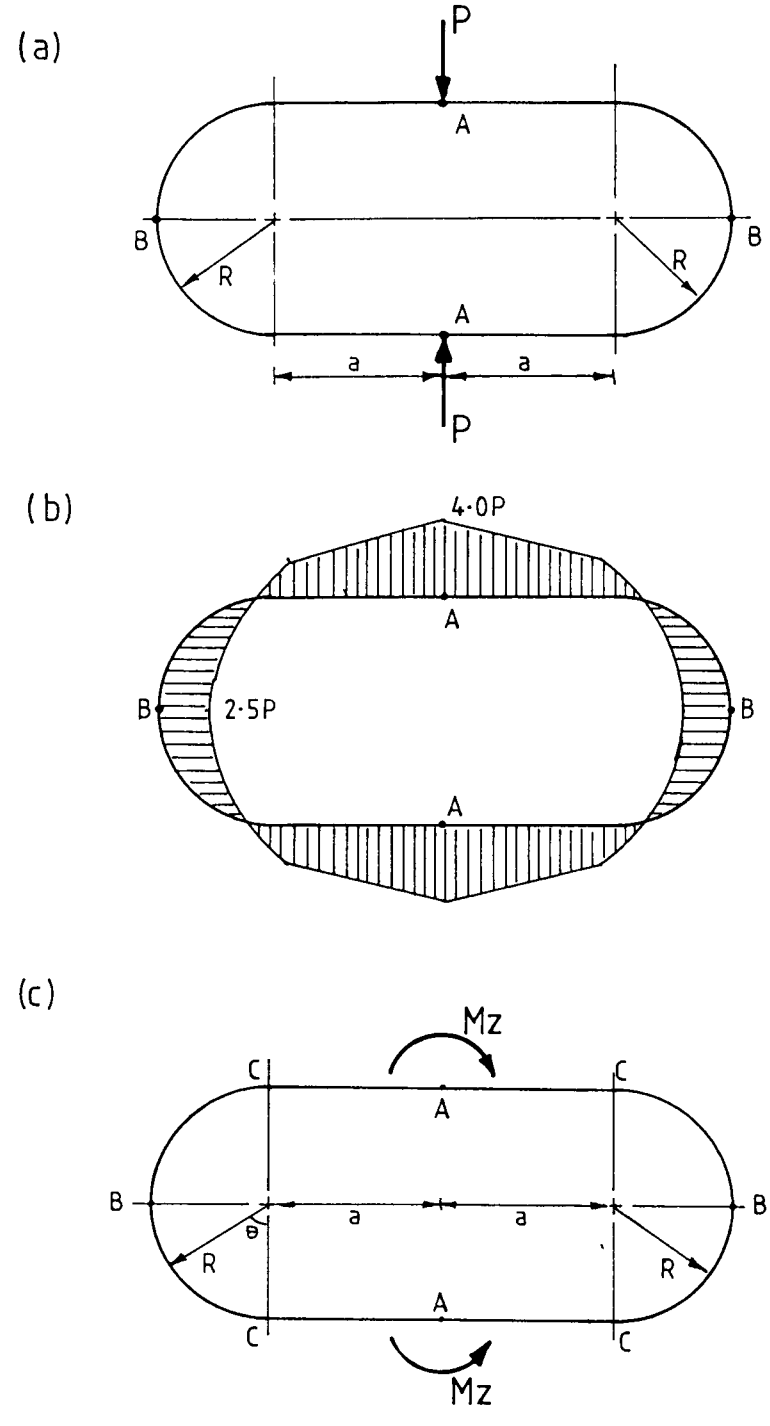


FIG A4·1	Theoritical Analysis.

A4.1 THEORETICAL ANALYSIS.

An extended ring of the form shown in Fig A4.1a was considered for the axial loading condirion. Using the unit-load method of analysis, which is a derivative of the Principle of Virtual Forces, on a symmetrical quarter section of the ring, the induced bending moment on the ring can be shown to be:

at point A,

$$M_A = \tfrac{1}{2}P\left[\frac{a^2 + \pi \cdot a \cdot R + 2 \cdot R^2}{\pi \cdot R + 2 \cdot a}\right]$$

at point B,

$$M_B = -\tfrac{1}{2}P\left[\frac{R^2(\pi - 2) + 2 \cdot a \cdot R + a^2}{\pi \cdot R + 2 \cdot a}\right]$$

The dimensions of the ring in the transducer are 8 mm and 5 mm respectively for R and a, giving a bending moment distribution as shown in Fig A4.1b, where the loading is in newton-millimetres. However, the magnitude of M_A will be considerably smaller in the actual case, since the load is not a point force, but applied over an area of 8.7 mm in diameter.

The properties of the transducer material were established with two specimens tested in a Hounsfield Tensometer (supplied by Monsanto Ltd, Swindon). The average values of the yield stress and ultimate tensile stress were found to be 450 MPa and 700 MPa respectively.

Using 450 MPa as the maximum allowable stress, the permitted axial load at point A is found to be 3000N from the straight beam theorem. However, it is required to use the curve beam theorem at point B, and from Seely (1952), the stresses at B is given by,

$$\sigma_B = \frac{P}{2 \cdot b \cdot h} + \frac{M_B}{b \cdot h \cdot R}\left[1 + \frac{1}{Z} \cdot \frac{y}{R+y}\right]$$

and

$$Z = -1 + (R/h)\left[\log_e\left(\frac{R+c}{R-c}\right)\right]$$

where h is the thickness of the beam, c is the half-thickness ($c=\frac{1}{2}h$), b is the width of the beam, and y, the distance from the centroid to the point where stress is to be determined (negative towards the concave surface). This gives a maximum axial load of approximately 4500N and 5000N for the innermost and outermost fibres at section B respectively. The circular cross-sectional beam adjoining the ring can

withstand loads of up to 27 kN. Hence, the transducer will be most unlikely to fail by virtue of the patient's body weight.

Similarly, the bending moment in the XY-plane induced by an externally applied Z-directed moment Mz can be derived from symmetry and energy method of solution, giving,

$$M = Mz\left[\frac{1}{2}-\left[\frac{\frac{1}{4}a^2+\frac{1}{4}(\pi\cdot a\cdot R)+\frac{1}{2}(R^2)}{\frac{1}{3}(a^3)+\frac{1}{2}(\pi\cdot a^2\cdot R)+2.a.R^2+\frac{1}{4}(\pi.R^3)}\right][a+(R\cdot\sin\theta)]\right] \quad \cdots\cdot(A4.2)$$

When applying to the transducer, the moments are (Fig A4.1c),

$$M_A = 0.500Mz$$
$$M_B = -0.148Mz$$
$$M_C = 0.251Mz$$

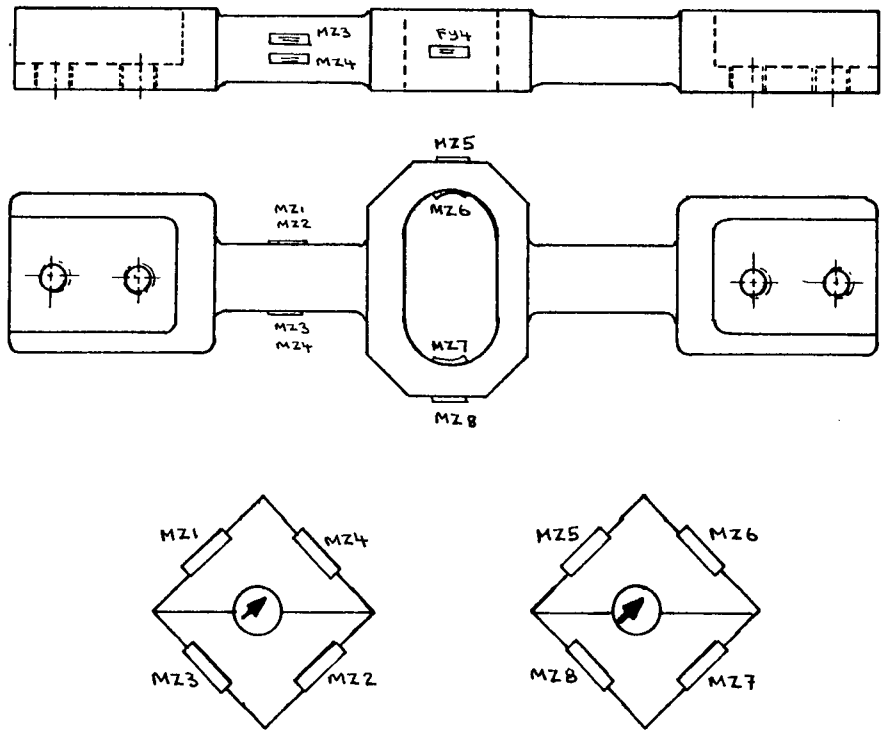
However, Loewen and Cook (1956) reported that the stress level of an octagonal ring are 30-40% lower than those in a circular ring. Furthermore, the somewhat evenly distributed loading pattern at A and the slightly larger minimum cross-sectional area of the flattened surface nearest to C (11% larger than the other surfaces) all contributed to further reducing the stress levels at points A and C. Thus for analytical purposes, the section passing through B can be considered to be the most critical.

The maximum permissible bending moment Mz at the beam sections adjoining the ring was found to be 29 Nm, which give rise to an M_B of 4.3 Nm. This is further reduced to 3 Nm when the reported minimum reinforcement effects (30% stronger) produced by the octagonal extended surfaces are taken into account. From equation A4.1, the stress levels at B can be shown to be 136 MPa and 96 MPa at the innermost and outermost fibres of the ring respectively. When compared with stresses at the adjoining beam section, the stresses on the beam are approximately 3.3 times and 4.7 times those on the innermost and outermost fibres of section B at the ring.

Therefore, the critical sections on the transducer are the two beam sections adjoining the ring, and would probably fail under a bending condition of 29 Nm.

A4.2 EXPERIMENTAL ASSESSMENT.

For the purpose of verifying the potential loading capacity of the transducer experimentally, a transducer was machined from the same



GAUGE TYPES:

(MICRO - MEASUREMENT GAUGES).

MZ1 - MZ4

EA-06-125MK-120.

MZ5 - MZ8

EA-06-062AA-120.

FIG A4·2	Assessment of Transducer.

material and with the dimensions as shown in Fig 4.9. It was then strain-gauged at 2 levels, one at the beam section and the other at section B of the ring (FigA4.2). The gauges were wired into 2 full bridges to measure the moment Mz at the two levels.

Uniform bending moment over the whole length of the transducer was applied through a 4-point loading system, and both the loading and unloading characteristics were recorded. To simulate the worst case, a 'reverse loading cycle' procedure was adopted, where the magnitude of the applied load increased fractionally with an alternatively directed load cycle. Within the elastic limit, the load-strain curves showed very good linearity. The ring section was found to be 3 times stronger than the beam section, which is in close agreement with the theoretical analysis.

The transducer showed its initial sign of yielding at around 30 Nm, resulting in a large residual strain. Subsequent loading in the opposite direction produced a reduction in the limit of proportionality and an increased nonlinearity of the output signal.

APPENDIX 5.

THEORETICAL ANALYSIS OF MARK II KAFO TRANSDUCER.

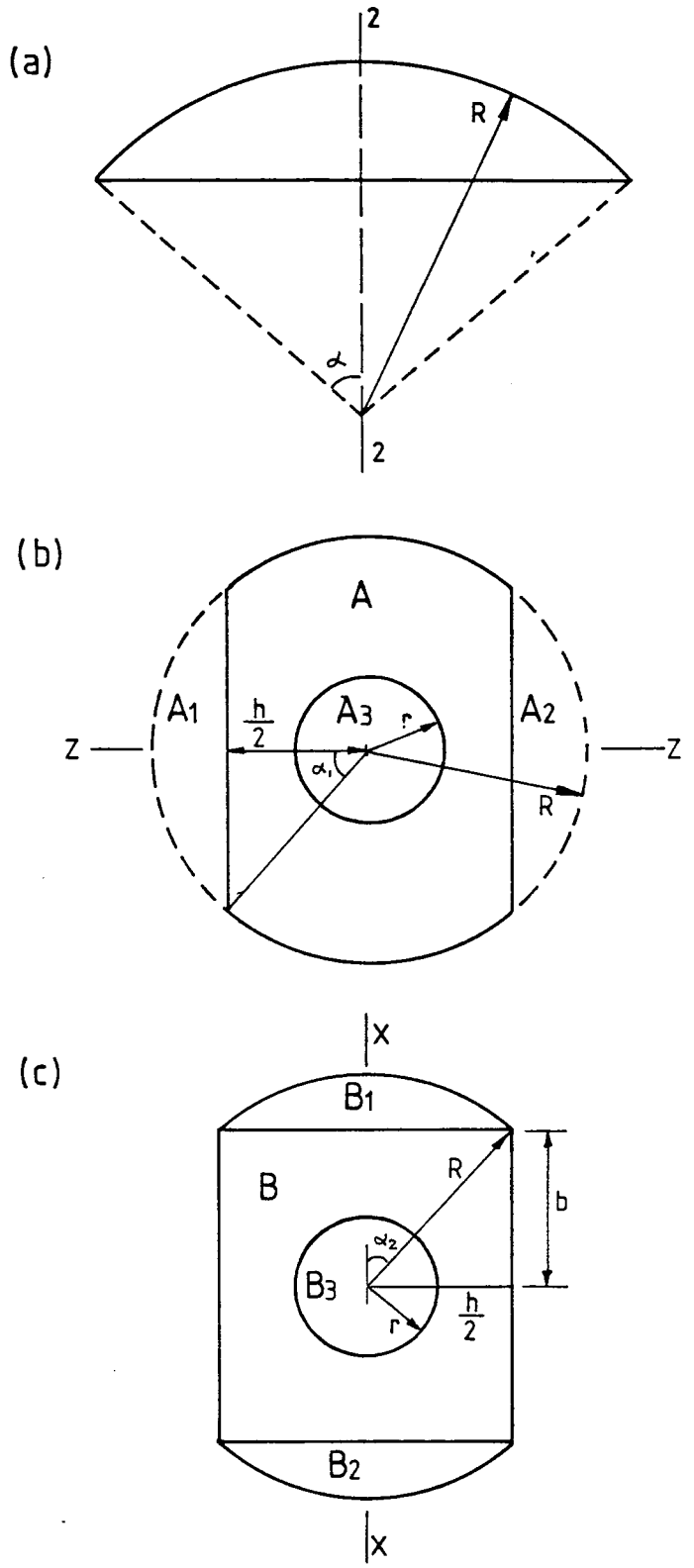


FIG A5·1	Dimensions for the Calculation of Sectional Properties.

THEORETICAL ANALYSIS OF MARK II KAFO TRANSDUCERS.

With reference to Fig A5.1a, the properties of a circular segment are given by Roark (1965) to be:

$$I_{22} = R^4\left[\frac{1}{8}(2\alpha-\sin 2\alpha)-\frac{1}{12}\frac{(2\alpha-\sin 2\alpha)\sin^3\alpha\cos\alpha}{\alpha-\sin\alpha\cos\alpha}\right]$$

where α is in radian.

For a cross-sectional area consisting of a hollow cylinder with 2 flattened external surfaces, the second moment of area about the Z-axis is (Fig A5.1b),

$$I_{zz(A)} = I_{zz(A+A1+A2+A3)}-I_{zz(A1)}-I_{zz(A2)}-I_{zz(A3)}$$

i.e. $$I_{zz} = R^4\left[\frac{\pi}{4}-\tfrac{1}{4}(2\alpha_1-\sin 2\alpha_1)+\frac{1}{6}\frac{(2\alpha_1-\sin 2\alpha_1)\sin^3\alpha_1\cos\alpha_1}{\alpha_1-\sin\alpha_1\cos\alpha_1}\right]-\frac{\pi r^4}{4}$$

.....(A5.1)

Also, the second moment of area about the X-axis is (Fig A5.1c),

$$I_{xx(B)} = I_{xx(B+B3)}+I_{xx(B1)}+I_{xx(B2)}-I_{xx(B3)}$$

i.e. $$I_{xx} = \left[\frac{1}{12}h^3(4R^2-h^2)^{\frac{1}{2}}\right]-\left[\frac{\pi r^4}{4}\right]$$

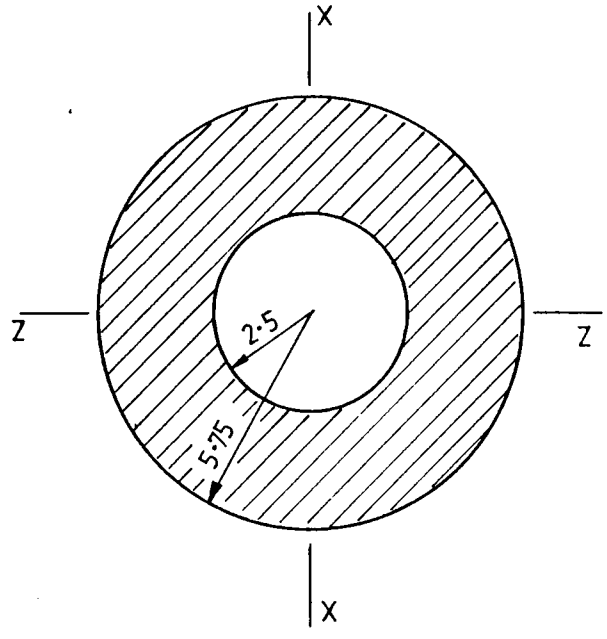
$$+ R^4\left[\tfrac{1}{4}(2\alpha_2-\sin 2\alpha_2)-\frac{1}{6}\frac{(2\alpha_2-\sin 2\alpha_2)\sin^3\alpha_2\cos\alpha_2}{\alpha_2-\sin\alpha_2\cos\alpha_2}\right] \cdots(A5.2)$$

For an ordinary hollow cylindrical cross-section, the second moments of area are:

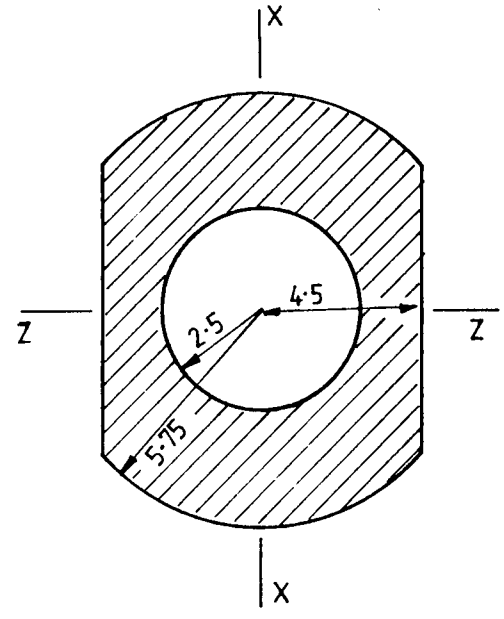
$$I_{xx}=I_{zz}=\tfrac{1}{4}\pi(R^4-r^4)$$

R and r being the external and internal radius respectively.

The effects of torsion My on the flattened section cannot be easily calculated because the theory of torsion for shafts of non-circular cross-section is very complex, and the assumptions that are valid for circular shafts no longer apply. Roark (1965) again provides a set of values, according to the size of cross-sectional dimensions, for the estimation of the torsional strength of a circular shaft with opposite sides flattened. No estimates were given for the case of a tube section with flattened external surface. For the purpose of this analysis, an approximately 10% increase in shear stress was assumed for the latter case. Carter and Cliphint (1952) reported that the maximum stress in such a cross-section would most probably be found at



UPPER SECTION



LOWER SECTION

Dimensions in mm.

FIG A5·2	Cross-sections of the Transducer Elastic Element.

the centre of the flat side.

The dimension chosen for the cylindrical section of the transducer elastic element were 11.5 mm and 5 mm for the external and internal diameters respectively. For the flattened section, the 2 plane surfaces are parallel to the X-axis, and a distance of 9 mm apart (Fig A5.2). The stresses imposed onto the transducer by an externally applied load can therefore be calculated as:

for bending stress,

$$\sigma = [M/Z]$$

and maximum torsional stress,

$$\tau = [T/J_1]$$

where M and T denotes the applied moment and torque respectively. Z is the sectional modulus, and J_1 is a modulus obtained from the division of polar moment of inertia J by the radius of the shaft. Thus, for the cylindrical section,

$$Z_{xx}=Z_{zz}= 144 \text{ mm}^3$$
$$J_1 = 288 \text{ mm}^3$$

for the flattened section,

$$Z_{xx} = 114 \text{ mm}^3$$
$$Z_{zz} = 137 \text{ mm}^3$$
$$J_1 = 169 \text{ mm}^3$$

The magnitude of the stresses at the transducer can be tabulated below:

Channel.	Maximum Service Load.	Stress at Flattened Section.	Stress at Cylindrical Section.
Mz	35 Nm	255 MPa	243 MPa
Mx	25 Nm	220 Mpa	174 MPa
My	15 Nm	89 MPa	52 MPa

If both the maximum Mx and Mz moments were to apply simultaneously, the maximum stress experienced by the transducer is 375 MPa, at the edge of the convex surfaces.

The nominal limiting loads of the transducers are as follows, and are for singularly applied loads:

Channel	Nominal Limiting Load.
Mz	103 Nm
Mx	86 Nm
My	128 Nm

As a result of the requirement for higher strength, larger cross-sectional areas were designed. This consequently reduced drastically the sensitivity of the transverse shear channels Fx and Fz.

For the chosen geometrical configuration, the sectional moduli of the transducer and that of a KAFO upright (16 mm x 5 mm) are fairly close. In the xx direction, the ratio of the sectional moduli of the upright to that of the flattened section and cylindrical section of the transducer are 0.59 and 0.47 respectively. In the zz direction, the ratios are 1.56 and 1.47 .

APPENDIX 6.

PROTOCOL FOR KAFO NORMAL LEVEL WALKING TESTS.

EQUIPMENT:

1. KAFO with transducers;
2. Amplifiers and multiplexer;
3. Cine cameras;
4. Force platforms and associated equipment;
5. PDP 11, MFRUN sampling programme, disk DL1: ;
6. Storage oscilloscope;
7. Digital Volt Meter(DVM);
8. LEDs with their stands;
9. Interconnecting cables;
10. Kodachrome ASA40 cine films (2 off);
11. Calibration Grid Board;
12. Markers;
13. KAFO test recording sheets;
14. Cine test numbering system;
15. Tapes: masking, double-sided, black non-reflective;
16. Light meter;
17. Additional floor lights;
18. Large vernier caliper;
19. Scissors, allen keys, miniature screw drivers, etc.

PRETEST PREPARATION:

1. Lighting:
   1) Check height of cine light gantry;
   2) Ensure all lights functioning properly;
   3) Position pair of floor lights;
   4) Check light intensity with light meter;
   5) Adjust gantry light configuration, if necessary;
   6) Check through camera view-finders.

2. Walkway:
   1) Check force plates are in correct positions;
   2) Remove any obstacles from the full length of the walkway;
   3) At a point near to the beginning of walkway, about 5 paces from the force plate, place lengths of masking tape in the shape of a 'T' across the walkway to mark starting point for patient;
   4) Place another length of masking tape across the walkway at about 5 paces away from the FP towards the front camera to mark stopping point for patient.

3. KAFO:
   1) Check metal parts of the instrumented KAFO have been sprayed with matt black paint;
   2) Ensure all transducers are fully secured;
   3) Cover all other exposed metal surfaces with black non-reflective tapes.

4. Amplifiers and Multiplexer:

1) Ensure KAFO/amplifiers/LEDs/multiplexer/storage-oscilloscope/computer-console have been properly connected;
2) Switch on amplifiers and multiplexer;
3) Set synchronising switch on multiplexer to INTERNAL;
4) Set bridge voltages with DVM;
5) Zero balance bridges with oscilloscope and DVM;
6) Set amplifier gain to 500 for all channels.

5. Force Platform:

1) Select the FP to be used in the test, with reference to patient's affected limb;
2) Ensure FP has been properly connected to multiplexer (back of FP console to channel 25-30 of multiplexer);
3) Switch on charge and buffer amplifiers;
4) Switch remote control switch to OPERATE;
5) Check 'Trands. Sens.' settings are correct;
6) Set charge and buffer amplifiers to:

   C1: 20 mech units/volt;
   C2: 100 mech units/volt;
   Buffer: 1, 1, 2, 1.25, 2, 1.25

6. Cine Cameras:

1) Check cameras have not been loaded by other users;
2) Switch on power at console, with RUN switch at OFF;
3) At camera units, switch to FORWARD and REMOTE;
4) Load film, run film to ZERO position using REMOTE/LOCAL switch;
5) Leave REMOTE/LOCAL switch in REMOTE position;
6) Check focusing on cameras;
7) Position synchronising LEDs;
8) Close view-finder flaps;
9) Set f-stops to:

Front:	f1.5
Left :	f1.6
Right:	f1.3

7. PDP 11/34:
   1) Schmidt trigger: set potentiometer to 580, 'switch up', cable from Input 1 to Biomech V7;
   2) Log into PDP 11;
   3) Mount disk onto Drive DL1: ;
   4) Check for free-spaces on DL1:, if full, transfer data to magnetic tapes, or use DL2: as a temporary measure.

8. Ground Markers:
   1) Position ground static reference markers at:
      (i) beside and forward of FP2,
      (ii) beside and just forward of the rear of FP2.
      Ensure that both the markers are more forward in the X-axis than the GRS origin.

PREPARING PATIENT FOR WALKING TESTS:

1. KAFO:
   1) Patient put-on the instrumented KAFO;
   2) Link KAFO with amplifiers, secure transducer connectors with 8BA bolts and nuts;
   3) Run cable from medial transducers via the back and below the brim of ischial seat to meet with cables from lateral transducers;
   4) Use belt to run cables to other side of patient.

2. Markers:
   Position and secure the following markers with double-sided adhesive tapes:
   1) Orthotic knee markers : KL, PL, KM
   2) Orthotic ankle markers : AL, AM
   3) Shank markers : BS, BI, BM
   4) Hip markers : H1, H2
   5) Tail marker stick : HTR-HTS-HTT

3. Amplifier Settings:
   1) Ask patient to walk freely around laboratory (with an assistant holding the cables behind patient);
   2) Set oscilloscope to STORE mode;
   3) Check and adjust amplifier settings to ensure the output voltages do not exceed $\frac{2}{3}$ of a volt (maximum allowable voltage: $\pm$1.0 V).

4. KAFO-Patient Static Calibration:

This ahould be carried out in as short a time duration as possible because of the full 'heating effects' of the cine lightings.

1) Stand patient on FP facing front camera;
2) Position anatomical joint centre markers on patient,
   (i) knee centre : KC1, KC2
   (ii) ankle centre : AC1, AC2
3) Ensure all other markers are well attached to patient or orthosis;
4) Run cameras for $\frac{1}{2}$ second;
5) Remove anatomical markers;
6) Measure hip and tail marker distances.

5. Patient/Force-Plate:

1) Existence/Location of FP should be concealed from the patient;
2) Position patient at starting point of walkway;
3) Ask patient to walk towards the front camera;
4) Readjust starting point if patient miss the FP, or did not step centrally on it.

6. Final Checks:

1) Activate MFRUN on DL1:
   (MFRUN can be activated by simply typing 'MFR');
2) Press DISCHARGE button on FP remote control;
3) Start patient walking;
4) Press RETURN to start sampling on MFRUN, and flip sampling switch to ON;
5) Check MFRUN sampling;
6) Check storage oscilloscope display;
7) Run through channels on MFRUN display to check amplifier and FP outputs;
8) Readjust amplifier or FP settings, if necessary.

WALKING TESTS:

1) Set cine test number for film reference;
2) Activate MFRUN on DL1:
3) Record PDP filename and film test number;
4) Press FP DISCHARGE button;

5) Start patient walking;

6) Switch on cine cameras at mid-stance prior to FP strike;

7) On toe-off prior to FP strike, press RETURN and sampling switch ON;

8) On heel-strike after FP, switch off cameras;

9) After MFRUN has stopped sampling, flip sampling switch to OFF;

10) Check MFRUN sampling and oscilloscope display;

11) If patient dropped or displaced any marker, replace marker and note test number;

12) For next test run, repeat procedure 1 - 10 above.

Note: a) Check length of cine film after 6 to 7 runs;

b) If hip markers are displaced, reposition them and measure hip-tail marker distances;

c) If shank or KAFO markers displaced, reposition marker and recalibrate Static 'KAFO-Patient' with anatomical joint centre markers .

AFTER WALKING TESTS:

1. Patient:

1) Measure hip-tail marker distances;

2) Remove hip, tail, and shank markers;

3) Remove instrumented orthosis, with the markers still attached;

4) Return patient's own orthosis.

2. Static KAFO Calibration:

1) Attach transducer triangle markers;

2) Ensure all markers are securely positioned on KAFO;

3) Position KAFO on FP;

4) Check through camera viewfinders to ascertain all markers are clearly seen by both cameras;

5) Run both cameras for $\frac{1}{2}$ second.

3. Grid-Board Calibration:

1) Position calibration grid board immediately above GRS origin, facing SIDE camera;

2) Run side camera for $\frac{1}{2}$ second;

3) Repeat the above procedure with FRONT camera.

4. Equipment: 1) Remove films from cameras;
2) Record the amplifier settings;
3) Delete unwanted files from PDP 11;
4) Dismount DL1: from PDP 11;
5) Measure and record all distance vectors of KAFO.

MARKERS USED IN THE TESTS:

1. KAFO-Orthosis Calibration:

1) static markers	: SM1, SM2
2) orthotic knee markers	: KL, PL, KM
3) orthotic ankle markers	: AL, AM
4) shank markers	: BS, BI, BM
5) knee centre markers	: KC1, KC2
6) ankle centre markers	: AC1, AC2

2. Walking Tests:

REMOVE:	KC1, KC2 AC1, AC2
KEEP:	SM1, SM2 KL, PL, KM AL, AM BS, BI, BM
ADD:	H1, H2, HTR-HTS (hip-tail markers)

3. KAFO Calibration:

1) static markers	: SM1, SM2
2) orthotic knee markers	: KL, PL, KM
3) orthotic ankle markers	: AL, AM
4) transducer triangle markers.	

4. Grid Board Calibration:

1) static markers	: SM1, SM2
2) grid board markers	: BX, BZ

++

APPENDIX 7.

SAMPLES OF TEST AND SURVEY FORM.

A7.1 NORMAL LEVEL WALKING TEST RECORDING SHEETS.

KAFO TEST DATA.

Date:_____.

Subject Code:_____.

Cine Run	PDP 11 File Name	Test Run OK?	Comments	.ICL Created?	VAX Name AB**C*DEF	Data OK? @RUNREAD	Analysis Completed?	Graphs Plotted?

ORTHOSIS SECTION:

1) Transducer option : (ITRX)
- ☐ 1 = AET KAFO Transducers.
- ☐ 2 = New KAFO Transducers.

2) Position option : (IOPT)

	TLD	TMD	TLP	TMP
☐ 1 =	1	2	3	4
☐ 2 =	1	4	3	2
☐ 3 =	3	2	1	4
☐ 4 =	3	4	1	2

3) Side option : (ISIDE)
- ☐ -1 = Left Limb
- ☐ +1 = Right Limb.

4) Inter-strap axial force option : (SFOPT)
- ☐ 1 = Prox strap weight-bearing; dist strap non-weight-bearing.
- ☐ 2 = Dist strap weight-bearing; prox strap non weight-bearing.
- ☐ 3 = Dist strap weight-bearing; no prox strap.
- ☐ 0 = Apron 70% weight-bearing; dist strap 30% weight-bearing.

5) Orthosis knee and atraps positions option : (SOPT)
- ☐ 1 = Knee between straps.
- ☐ 2 = Knee below straps.
- ☐ 3 = Knee above straps.
- ☐ 4 = Calf band only.
- ☐ 0 = Apron and calf band only.

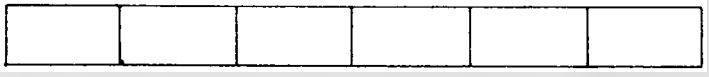
6) T-strap option : (IST)
- ☐ 1 = Strap on medial upright.
- ☐ 2 = Strap on lateral upright.
- ☐ 0 = No T-strap.

FORCE PLATFORM SECTION:

1) Top charge amp setting: C1 = _____ mech unit/volt.

2) Bottom charge amp setting: C2 = _____ mech unit/volt.

3) Buffer amp settings, (BMF) :



4) Force platform used in test:
- ☐ FP1
- ☐ FP2

AMPLIFIER SECTION:

1) Bridge voltages, (VOLTS) :

Fx	Fy	Fz	Mx	My	Mz	
						(volts)

2) Amp settings:

	Fx	Fy	Fz	Mx	My	Mz
GAIN1 =						
GAIN2 =						
GAIN3 =						
GAIN4 =						

CAMERA SECTION:

1) Camera distances (from GRS origin) : Front (X) =_____m.
Side (Z) =_____m.

2) Camera heights: Front (HCAMF) =_____m.
Side (HCAMS) =_____m.

3) Aperture settings : Front: f_____.
Side : f_____.

4) Type of film used:_____.

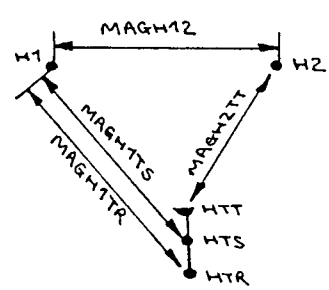
5) Length of film used: Front :_____m.
Side :_____m.

DISTANCE VECTORS:

1) Hip-tail markers:

MAGH12 :_____m.
MAGH1TR :_____m.
MAGH1TS :_____m.
MAGH2TT :_____m.

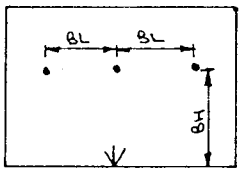
*Marker H1 is nearer to the side camera.



2) Calibration Grid Board:

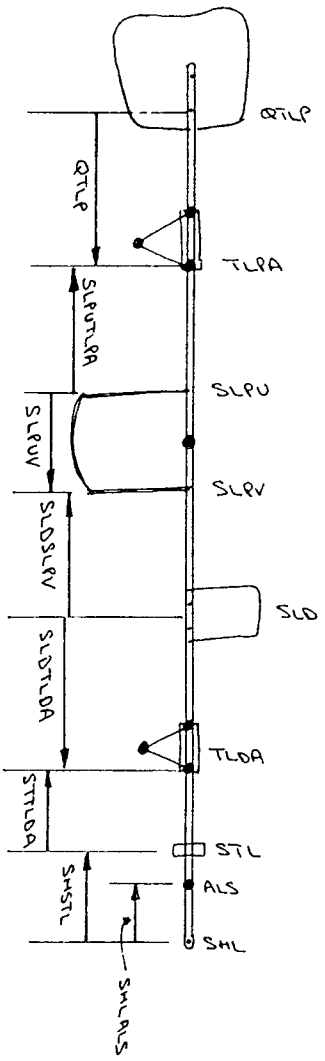
BH :_____m.

BL :_____m.



3) Orthosis:

	x	y	z (m)
QTLP :			
QTMP :			
SLPTLPA :			
SMPTMPA :			
SLDSLP :			
SMPSMD :			
SLPUTLPA :			
SMPUTMPA :			
SLPUV :			
SMPUV :			
SLDSLPV :			
SMDSMPV :			
SLDTLDA :			
SMDTMDA :			
SHSTL :			
SHSTM :			
STTLDA :			
STTMDA :			
SHLALS :			
SHMAMS :			

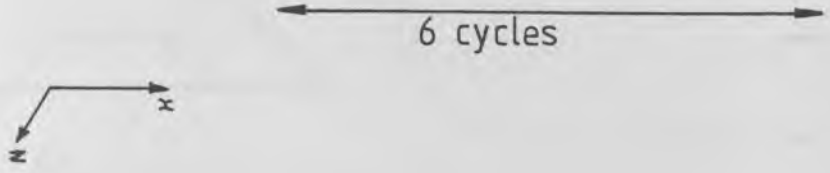


A7.2 LIMITED RANGE FREE WALKING TEST RECORDING SHEETS:

Name : _____.

Date : _____.

LEVEL WALKING IN A STRAIGHT LINE.



Direction of walk	Sampling name	Sampling ok ?	Approx. no. of strides	Comments	VAX file name

Name : _____.
Date : _____.

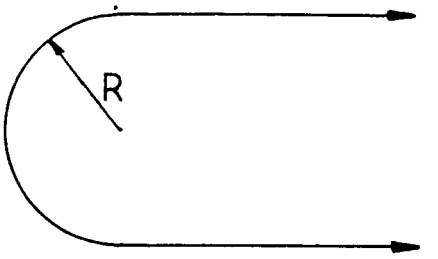
LEVEL WALKING (DOUBLE-BEND)

Direction of walk	Sampling name	Sampling ok ?	Approx. KAFO HS	Comments	VAX file name

Name : _____

Date : _____

LEVEL WALKING (SEMI-CIRCULAR)



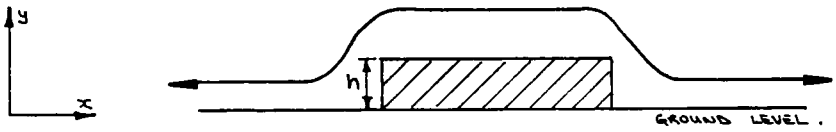
R= _____ .

Direction of walk	Sampling name	Sampling ok ?	Approx. KAFO HS	Comments	VAX file name

Name : _____

Date : _____

UP / DOWN LOW PLATFORM



h = _____.

Direction of walk	Sampling name	Sampling ok ?	Approx. KAFO HS	Comments	VAX file name

Name : _____

Date : _____

UP / DOWN RAMP.

ϕ = _____.

Direction of walk	Sampling name	Sampling ok ?	Approx. no. of strides	Comments	VAX file name

Name : _____

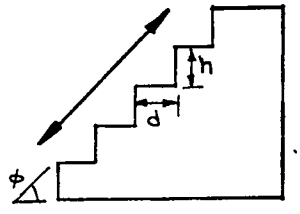
Date : _____

UP / DOWN STAIRS.

pitch, ϕ = _____.

rise, h = _____.

going, d = _____.



Direction of walk	Sampling name	Sampling ok ?	Approx. no. of strides	Comments	VAX file name

Date :_____.

Subject Code :_____.

RECORDED DATA:

1) Transducer option: (ITRX)

☐ 1 = AET KAFO Transducers.
☐ 2 = New KAFO Transducers.

2) Position option : (IOPT)

	TLD	TMD	TLP	TMP
☐ 1 =	1	2	3	4
☐ 2 =	1	4	3	2
☐ 3 =	3	2	1	4
☐ 4 =	3	4	1	2

3) Side option : (ISIDE)

☐ -1 = left Limb.
☐ +1 = Right Limb.

4) Analysis option : (IDIRCOS)

☐ 0 = Direction cosine matrices from the NLW tests.
☐ 1 = Kinematic data from cine films.
☐ 2 = Results expressed in transducer reference systems.

5)Bridge voltages (VOLTS) :

Fx	Fy	Fz	Mx	My	Mz	
						volts

6) Amp settings:

	Fx	Fy	Fz	Mx	My	Mz
GAIN1 =						
GAIN2 =						
GAIN3 =						
GAIN4 =						

A7.3 KAFO SURVEY FORM.

Confidential.

KAFO SURVEY.

Ref. No._____.

(Please tick '✓' as appropriate).

GENERAL INFORMATION.

1. Height :_____m.
   Weight :_____kg.
   Sex : Male / Female.
   Age :
   - ☐ 0 - 15 years.
   - ☐ 16 - 30 years.
   - ☐ 31 - 45 years.
   - ☐ 46 - 60 years.
   - ☐ 61 - 75 years.
   - ☐ 76+ years.

2. KAFO on : Left / Right / Bilateral.
   Other walking aids :_____.

3. Diagnosis/disability relevant to need for KAFO: _____
   _____.
   Date of onset of disability :_____.
   Other disability affecting ability to walk :_____
   _____.
   Any other disability affecting ability to put on/off KAFO :_____
   _____.

ORTHOSIS:

1. Can you put on the caliper yourself?
   - ☐ Yes, Always.
   - ☐ Yes, Sometimes.
   - ☐ No.

   If NO, what reason ? _____
   _____.

If YES, how difficult do you find it to put on the caliper?

- ☐ Very difficult.
- ☐ Difficult.
- ☐ Managing.
- ☐ Easy.
- ☐ Very easy.

If DIFFICULT, what are the reasons ? _____

_____.

2. How comfortable/uncomfortable do you find the caliper?

	Walking	Standing	Sitting.
Comfortable:	☐	☐	☐
Uncomfortable:	☐	☐	☐
Very uncomfortable:	☐	☐	☐

3. Where are the points of discomfort ?

- ☐ Thigh band/top.
- ☐ Knee joint.
- ☐ Knee straps.
- ☐ Calf band.
- ☐ Others.

4. Have you ever tried to alter the caliper in any way to make it more comfortable/safe ?

- ☐ Yes
- ☐ No.

If YES, where and how ? _____

_____.

5. Do you find the caliper too heavy ?

- ☐ Yes
- ☐ No.

6. Do you have any "frightening" experience with the caliper, e.g. caliper broken, etc?

- ☐ Yes.
- ☐ No.

If YES, a) how many times ? _____.

b) Describe the cause/incidents: _____

_____.

c) were you hurt in those incident ?

- ☐ Yes.
- ☐ No.

if YES, describe :_____

_____.

d) Other comment on those experiences :_____

_____.

7. How frequently does your caliper need repair/servicing ?_____

_____.

8. Does the caliper damage your clothings ? _____

_____.

9. How much confidence do you have on your caliper?

☐ Very confident.
☐ Confident.
☐ Not confident.
☐ Very worried.

10. What other problems do you have to face with the caliper ? _____

_____.

11. Any other comments on your caliper ? _____

_____.

12. What would you like to see in a new caliper ?
(including weight, shape, size, new design, new ideas, etc.)

_____

_____.

ACTIVITIES:

1. Are you in employment?
☐ Yes, full time.
☐ Yes, part time (_____hrs/week)
☐ No.

If YES, a) what is your occupation ?_____.
b) How do you get to work?
☐ Walk.
☐ Bus.
☐ Underground.
☐ Train.
☐ Driving.
☐ Passenger in car.
☐ Others.
c) is the caliper a hinderance to your work?
☐ Yes.
☐ No.

If NO, a) are you....
☐ retired ?
☐ housewife ?
☐ others ?

b) how often do you get out of the house (incl. shopping, visiting friends) ?

- ☐ More than twice a day.
- ☐ Once a day.
- ☐ Several times a week.
- ☐ Once a week.
- ☐ Seldom.

2. How many hours a day do you stand, with the caliper on (incl. in employment) ?

- ☐ Not at all/unable to stand.
- ☐ 0 - 1 hour.
- ☐ 1 - 3 hours.
- ☐ 3 - 6 hours.
- ☐ 6 - 8 hours.
- ☐ 8 + hours.

3. How far do you walk each day ?

- ☐ Very little walking.
- ☐ Just in the house.
- ☐ To the nearest shop.
- ☐ ½ to 1 mile.
- ☐ 1 to 2 miles.
- ☐ More than 2 miles.

4. Can you walk/stand without your caliper?

- ☐ Yes.
- ☐ No.

5. How much of the day do you wear your caliper?

- ☐ At all times.
- ☐ Most of the time.
- ☐ Occasionally.
- ☐ Not at all.

6. Can you perform the following activities without any walking aids other than your normal ones?

a) up/down stairs (or steps)
- ☐ easily.
- ☐ O.K.
- ☐ just managing.
- ☐ not at all.

b) up/down slopes :
- ☐ easily.
- ☐ O.K.
- ☐ just managing.
- ☐ not at all.

c) sitting down/getting up :
- ☐ easily.
- ☐ O.K.
- ☐ just managing.
- ☐ not at all.

d) walking on gravel/uneven pavement :
- ☐ easily.
- ☐ O.K.
- ☐ just managing.
- ☐ not at all.

++++++++++++++++++++++++++++++++

APPENDIX 8.

THE DATA PROCESSING SYSTEM.

A8.1 File Store Designation.

A8.2 Flow Charts.

A8.1 FILE STORE DESIGNATION.

This file store designation system was devised for identification of data or results for a particular patient, or a group of patients with similar characteristics, or the results of patients who have undertaken a particular activity. Due to the differences in the operating systems of the ICL and VAX computers, the file store designation system was subsequently modified during the change-over. Both the systems will be briefly described below.

The ICL 1904S George III system allowed a maximum of 12 characters for the naming of a file, and it was fully utilised with a file name typified by :

AB*CD**EFGH

where: A denotes the sex (M/F);
B denotes the affected side (L/R);
CD denotes the activity undertaken;
EFGH denotes the nature of the file content; and
** are numerical numbers, the first set of ** indicates the patient number, while the second set identifies the test.

For example, the name "ML15NWO9DATL" contains a file for a Male and Left limb affected patient number 15. The activity undertaken was Normal level Walking, test number 09, and contains the DATa of the transducer and force-plate Loadings for the test.

In the VAX system, the file name is restricted to not more than 9 characters. The designation system that had been used in the ICL system was therefore shortened and amended. It can be represented by:

AB**C*DEF.XXX

where: AB identifies the patient;
** denotes the test number;
C contains the information of the patient's sex, affected side, and the transducers used;
* denotes the activity taken;
DEF identifies the nature of file content; and
XXX is the file type designation required by the VAX operating system.

For example, "DC20A8DDT.DAT" denotes the DATa file containing the transducer loading data (DDT) for patient 'DC', test run number 20, with the patient who is male and left side affected using the Mark I

KAFO transducers (A) taking the low platform manoeuvring activity (8).

A8.2 FLOW CHARTS:

The simplified flow charts of the following programmes are presented in this section:

"READNLW"
"COORSNLW"
"COORDNLW"
"TRXFPNLW"
"BRACENLW"
"TRXLR"
"KSTRAP"

In the flow charts, the symbol '%' is used to represent the patient test code (i.e. AB**C*) section of the file name.

CHART A8.1 | FLOW CHART OF "READNLW".

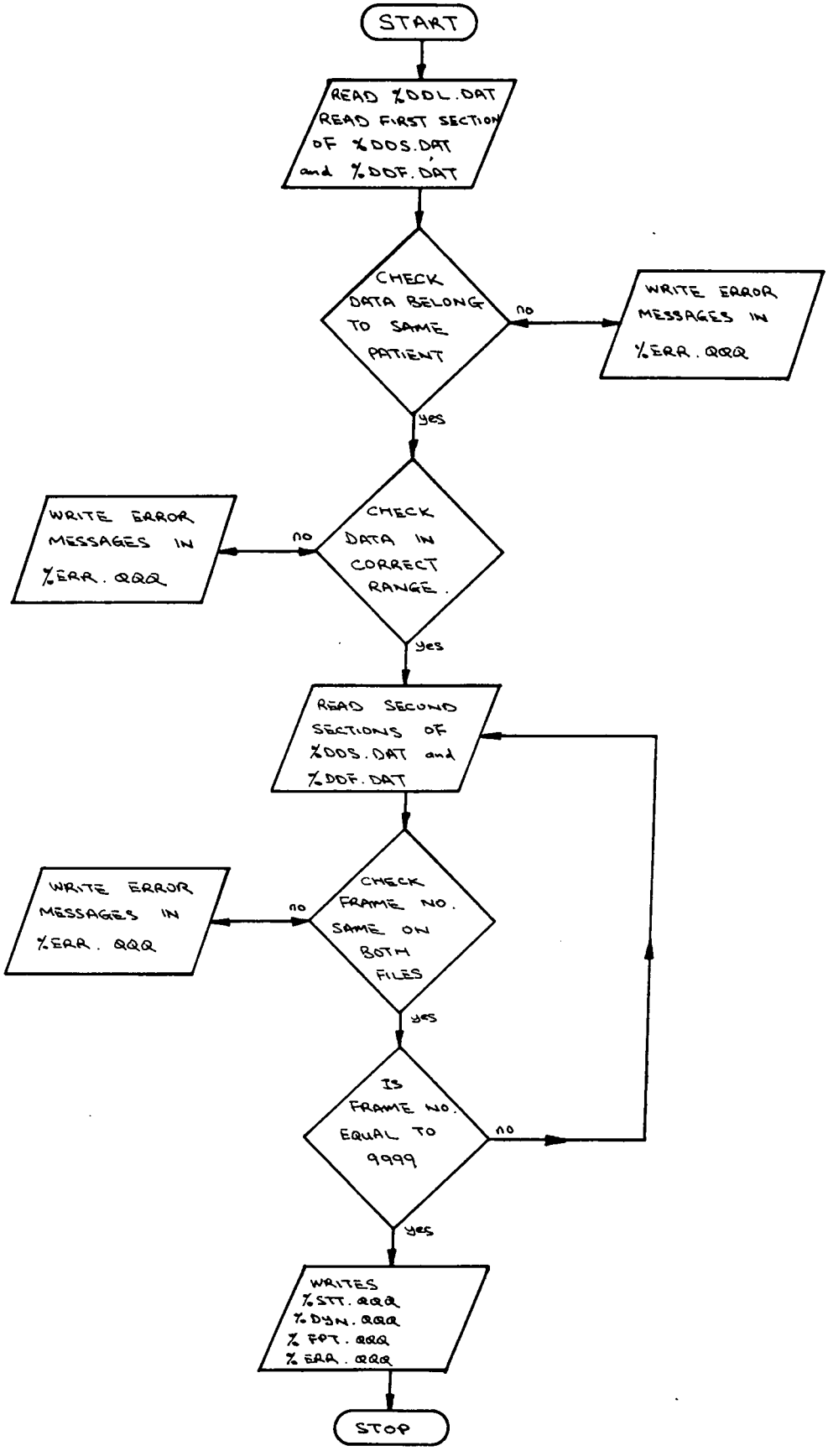


CHART A8.2	FLOW CHART OF "COORSNLW".

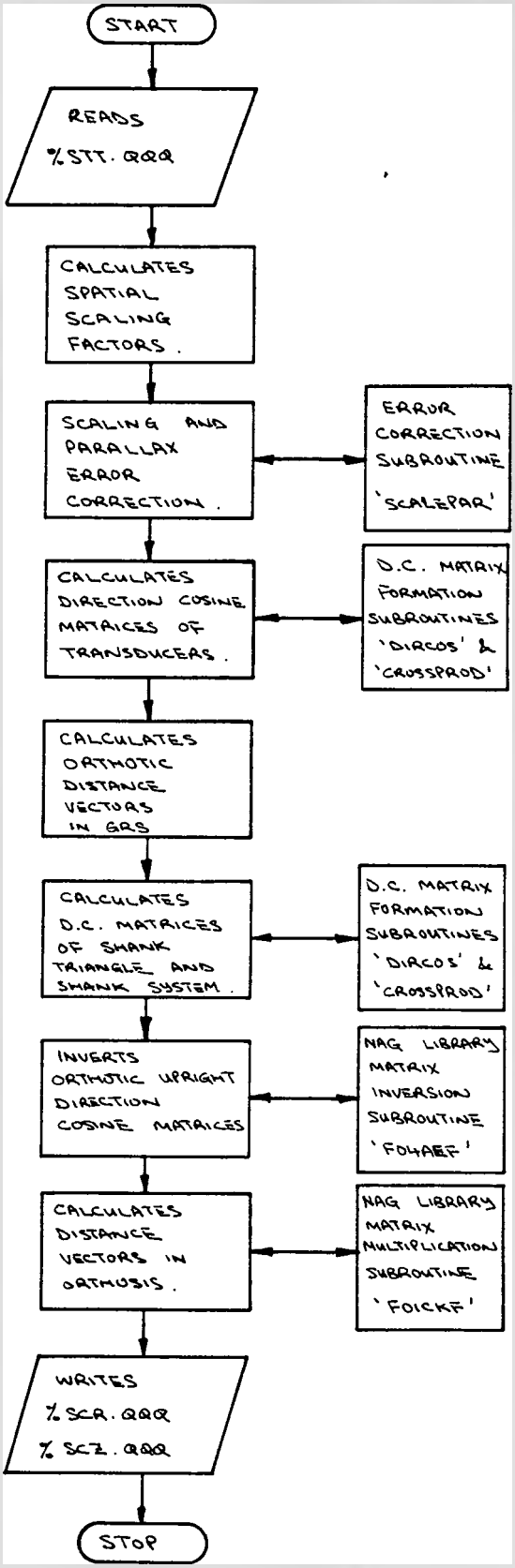
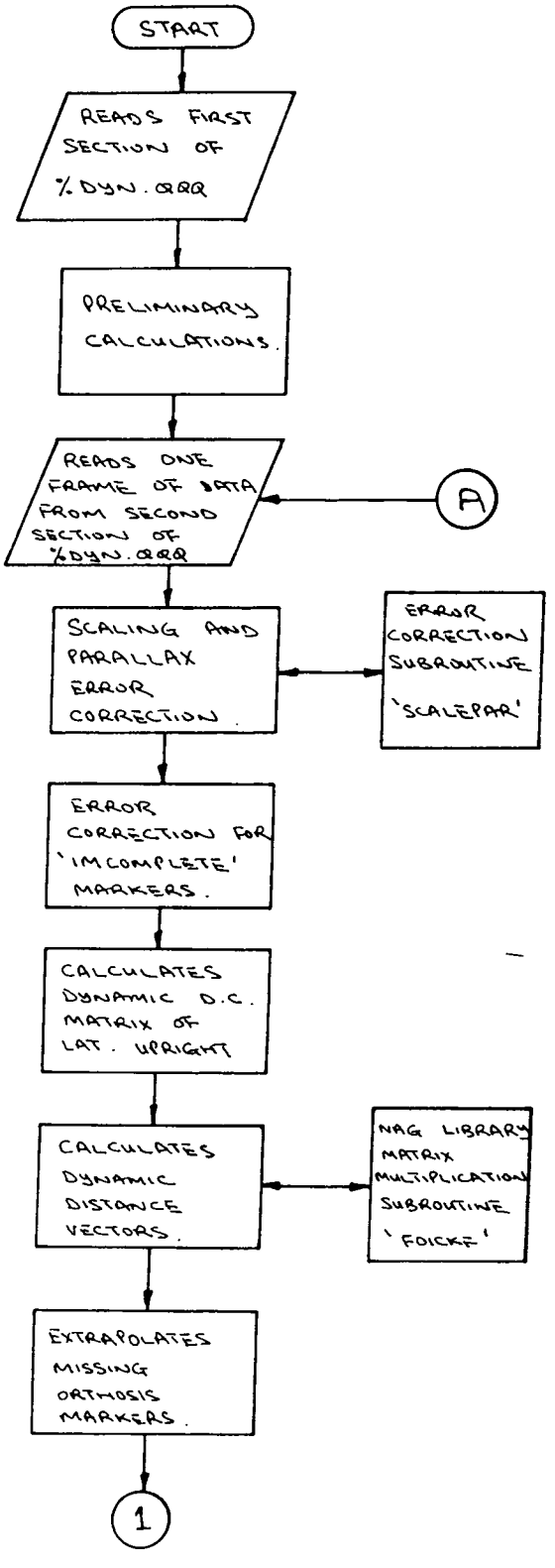


CHART A8.3 | FLOW CHART OF "COORDNLW".



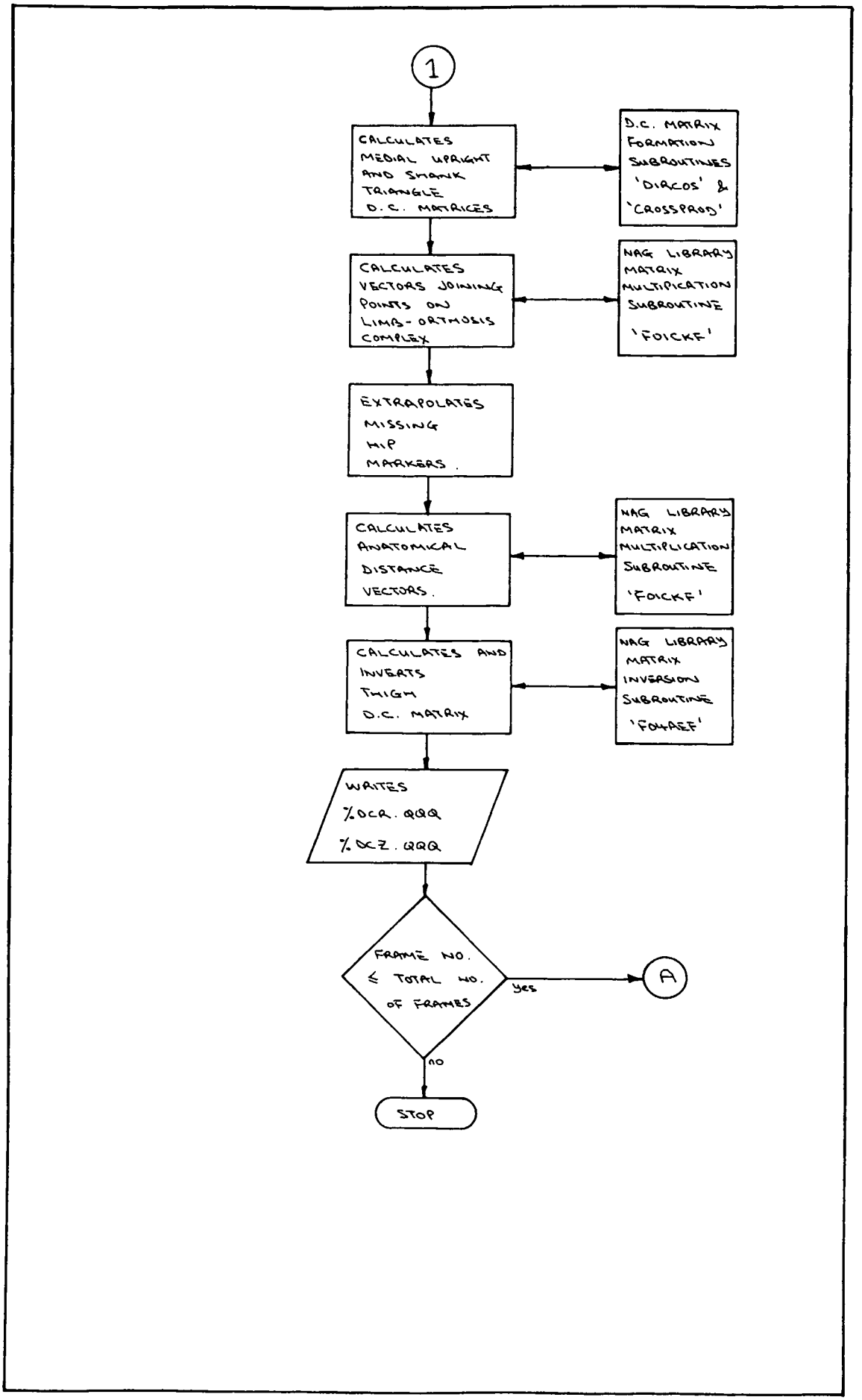

1
CALCULATES MEDIAL UPRIGHT AND SHANK TRIANGLE D.C. MATRICES
D.C. MATRIX FORMATION SUBROUTINES 'DIRCOS' & 'CROSSPROD'
CALCULATES VECTORS JOINING POINTS ON LIMB-ORTHOSIS COMPLEX
NAG LIBRARY MATRIX MULTIPICATION SUBROUTINE 'F01CKF'
EXTRAPOLATES MISSING HIP MARKERS.
CALCULATES ANATOMICAL DISTANCE VECTORS.
NAG LIBRARY MATRIX MULTIPLICATION SUBROUTINE 'F01CKF'
CALCULATES AND INVERTS THIGH D.C. MATRIX
NAG LIBRARY MATRIX INVERSION SUBROUTINE 'F01AEF'
WRITES %DCR.QQQ %DCZ.QQQ
FRAME NO. ≤ TOTAL NO. OF FRAMES
yes
A
no
STOP

CHART A8.4	FLOW CHART OF "TRXFPNLW".

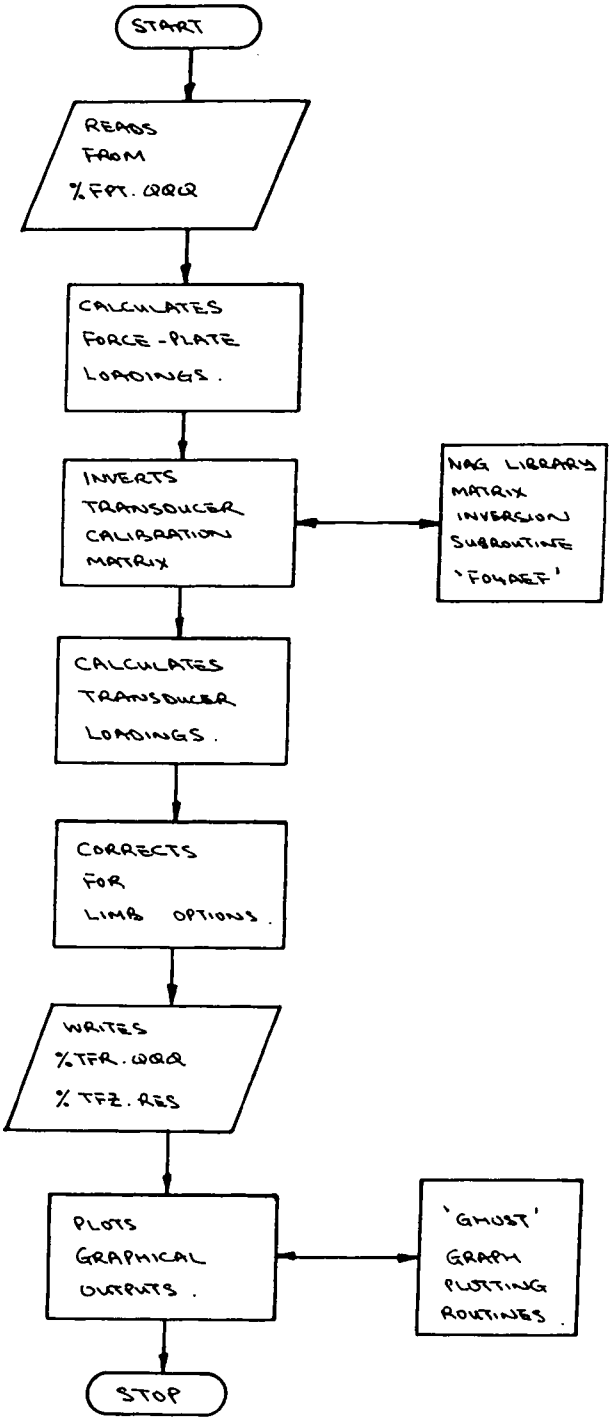
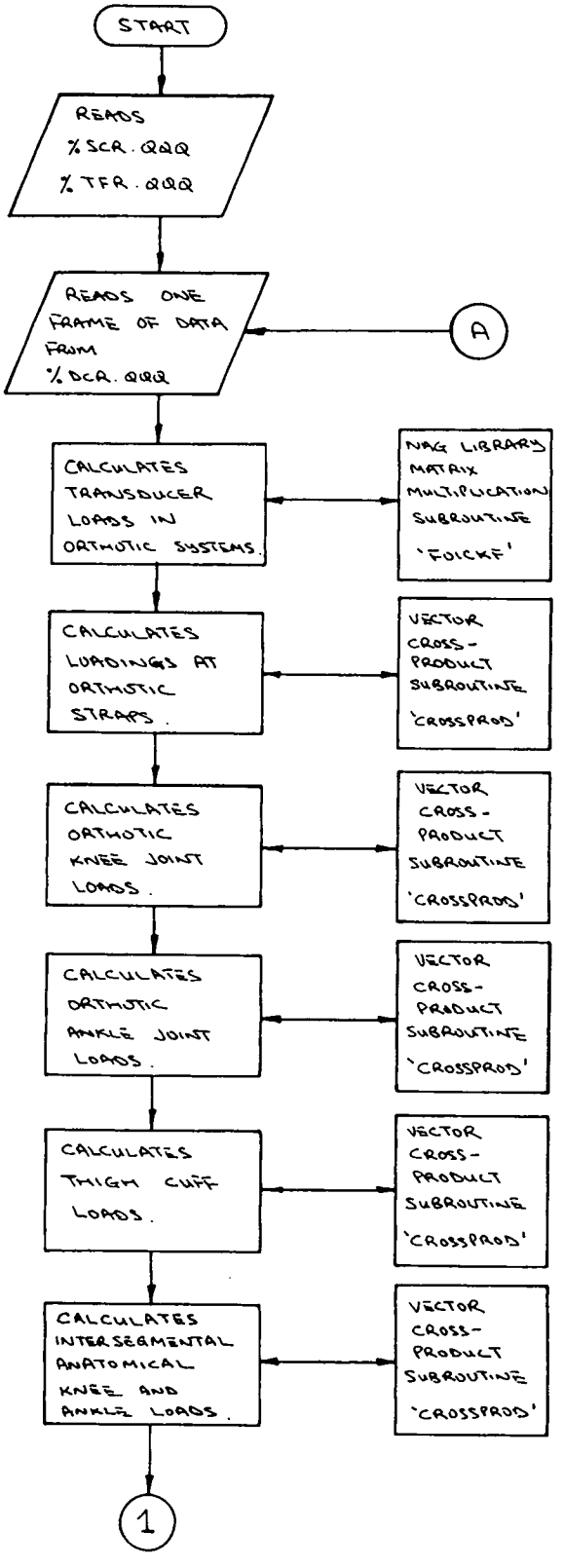


CHART A8.5	FLOW CHART OF "BRACENLW".



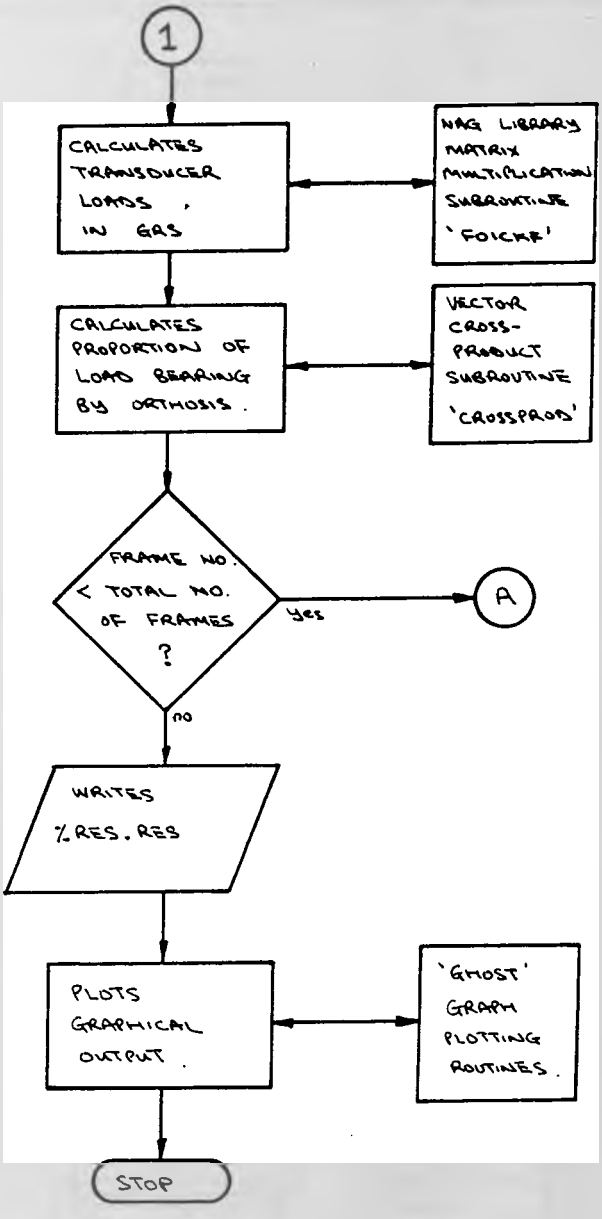
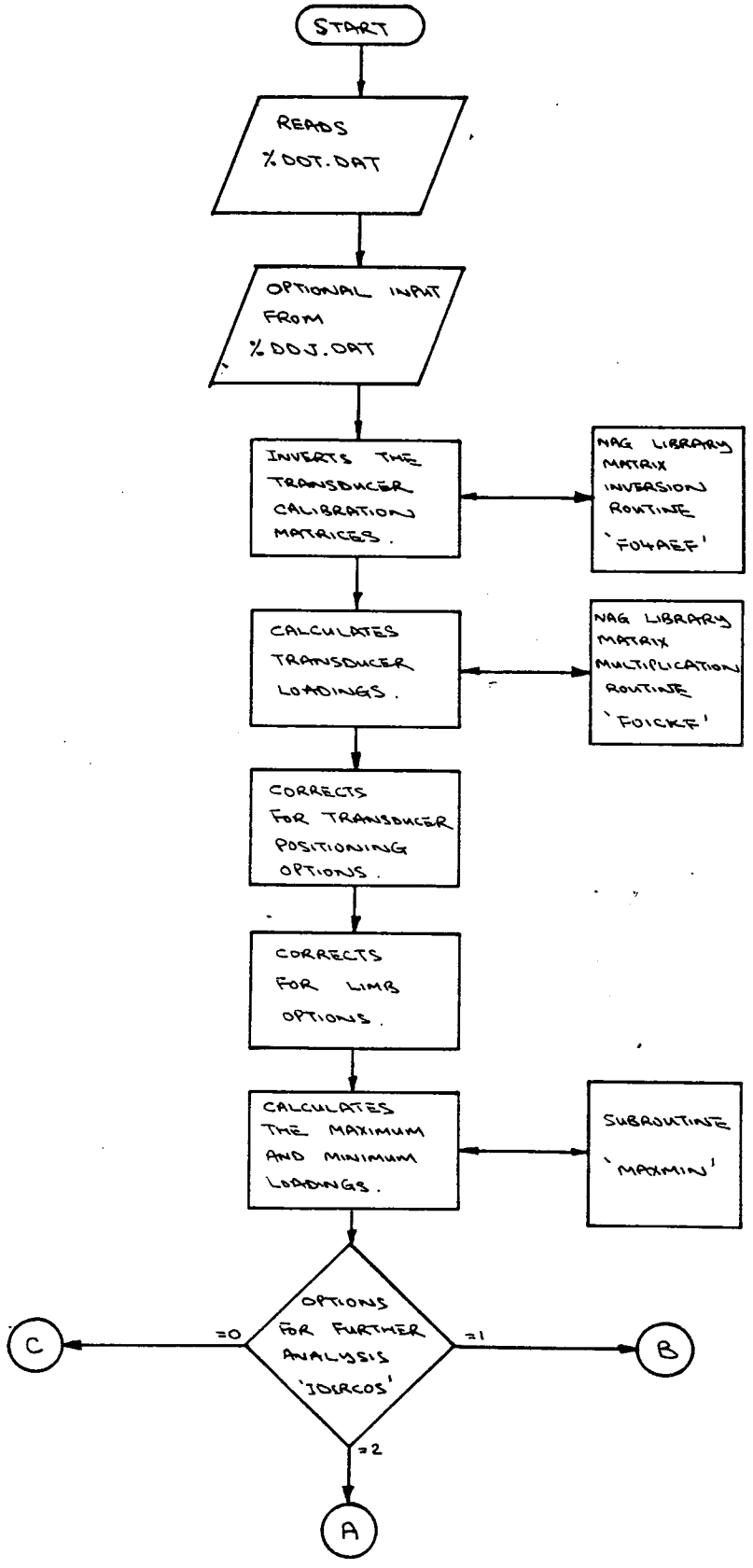

1
CALCULATES TRANSDUCER LOADS, IN GRS
NAG LIBRARY MATRIX MULTIPLICATION SUBROUTINE 'FOICHF'
CALCULATES PROPORTION OF LOAD BEARING BY ORTHOSIS.
VECTOR CROSS-PRODUCT SUBROUTINE 'CROSSPROD'
FRAME NO. < TOTAL NO. OF FRAMES ?
yes
A
no
WRITES %.RES.RES
PLOTS GRAPHICAL OUTPUT.
'GHOST' GRAPH PLOTTING ROUTINES.
STOP

CHART A8.6	FLOW CHART OF "TRXLR".



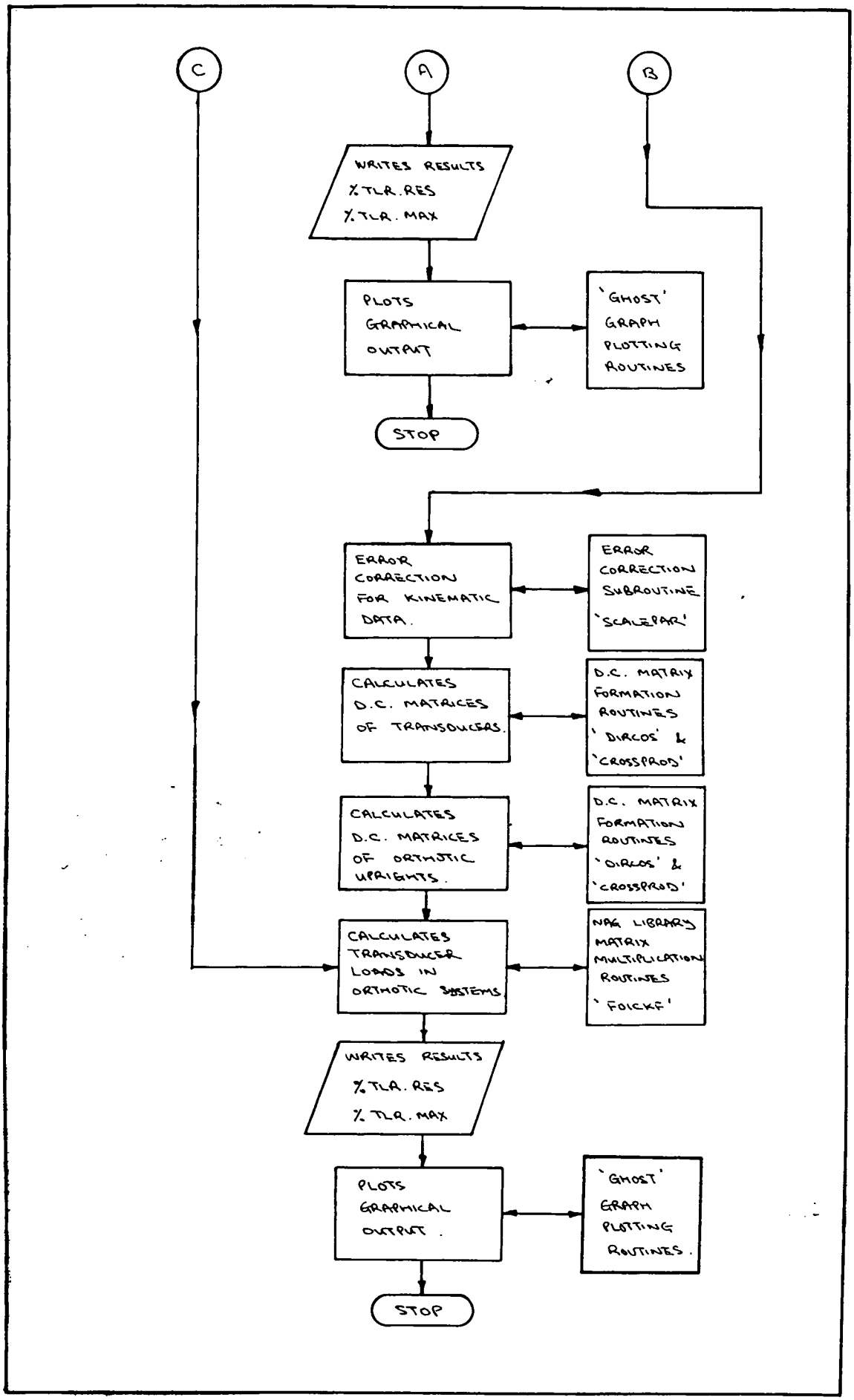
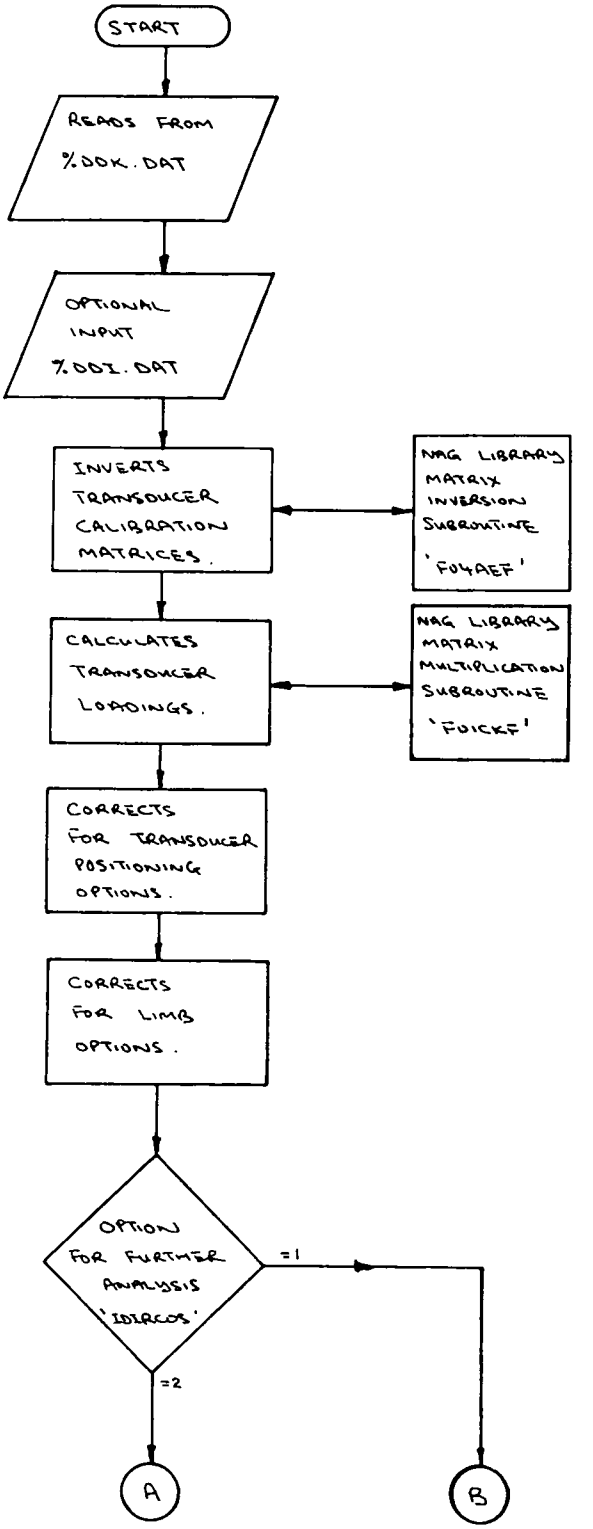
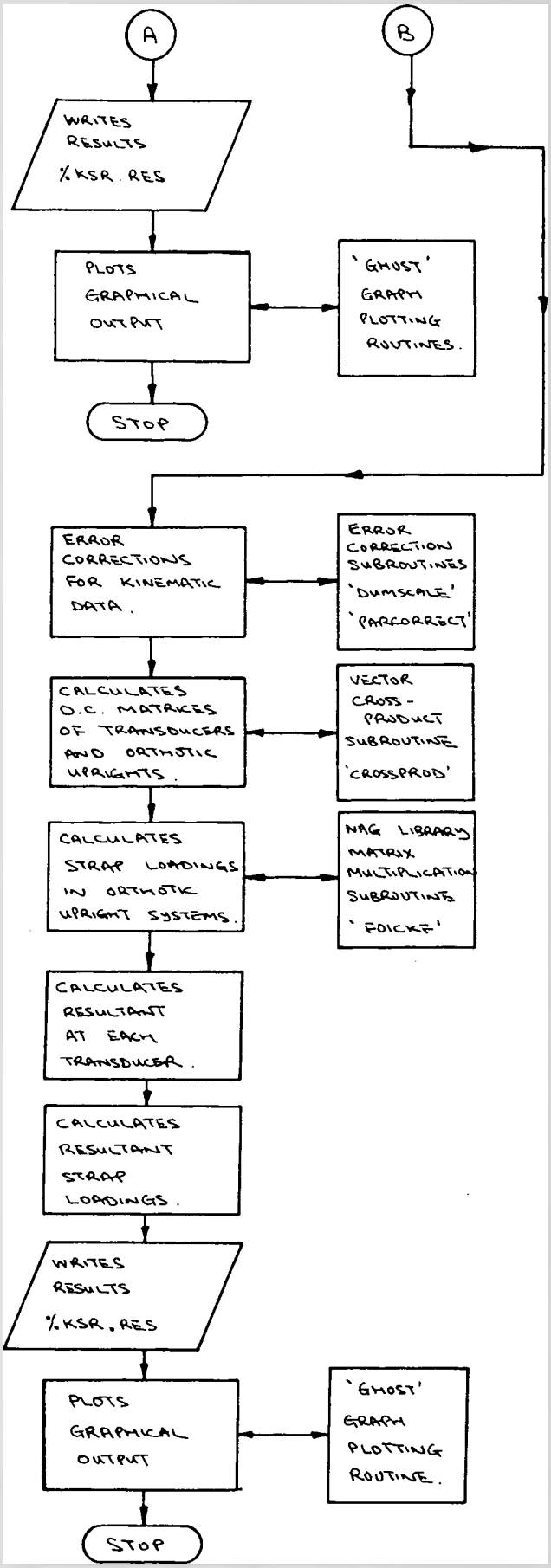

C
A
B
WRITES RESULTS
%TLR.RES
%TLR.MAX
PLOTS
GRAPHICAL
OUTPUT
'GHOST'
GRAPH
PLOTTING
ROUTINES
STOP
ERROR
CORRECTION
FOR KINEMATIC
DATA.
ERROR
CORRECTION
SUBROUTINE
'SCALEPAR'
CALCULATES
D.C. MATRICES
OF TRANSDUCERS.
D.C. MATRIX
FORMATION
ROUTINES
'DIRCOS' &
'CROSSPROD'
CALCULATES
D.C. MATRICES
OF ORTHOTIC
UPRIGHTS.
D.C. MATRIX
FORMATION
ROUTINES
'DIRCOS' &
'CROSSPROD'
CALCULATES
TRANSDUCER
LOADS IN
ORTHOTIC SYSTEMS
NAG LIBRARY
MATRIX
MULTIPLICATION
ROUTINES
'F01CKF'
WRITES RESULTS
%TLR.RES
%TLR.MAX
PLOTS
GRAPHICAL
OUTPUT.
'GHOST'
GRAPH
PLOTTING
ROUTINES.
STOP

CHART A8.7	FLOW CHART OF "KSTRAP".




A
B
WRITES RESULTS %KSR.RES
PLOTS GRAPHICAL OUTPUT
'GHOST' GRAPH PLOTTING ROUTINES.
STOP
ERROR CORRECTIONS FOR KINEMATIC DATA.
ERROR CORRECTION SUBROUTINES 'DUMSCALE' 'PARCORRECT'
CALCULATES D.C. MATRICES OF TRANSDUCERS AND ORTHOTIC UPRIGHTS.
VECTOR CROSS-PRODUCT SUBROUTINE 'CROSSPROD'
CALCULATES STRAP LOADINGS IN ORTHOTIC UPRIGHT SYSTEMS.
NAG LIBRARY MATRIX MULTIPLICATION SUBROUTINE 'F01CKF'
CALCULATES RESULTANT AT EACH TRANSDUCER.
CALCULATES RESULTANT STRAP LOADINGS.
WRITES RESULTS %.KSR.RES
PLOTS GRAPHICAL OUTPUT
'GHOST' GRAPH PLOTTING ROUTINE.
STOP

APPENDIX 9.

LISTINGS OF COMPUTER PROGRAMMES.

(in microfiche)

A9.1 "READNLW"

A9.2 "COORSNLW"

A9.3 "COORDNLW"

A9.4 "TRXFPNLW"

A9.5 "BRACENLW"

A9.6 "TRXLR"

A9.7 "KSTRAP"