The Biomechanics of the Human Foot

by

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DEDICATION

In loving memory of my grandparents, each of whom saw something special in their grandson.

'For the body itself is not made of only one part, but of many parts. If the foot were to say, "Because I am not a hand, I don't belong to the body," that would not keep it from being part of the body ..... So then, the eye cannot say to the hand, "I don't need you!" Nor can the head say to the feet, "Well, I don't need you!" On the contrary, we cannot do without the parts of the body that seem weaker; and those parts of the body that we think aren't worth very much are the ones which we treat with greater care...... If one part of the body suffers, all the other parts suffer with it; if one part is praised, all the other parts share its happiness.' 1 Corinthians 12 Vs14-26. (GNB)

ABSTRACT

This thesis reports on work undertaken to study the biomechanics of the human foot during normal daily activity, particularly walking and standing.

A literature review is presented on topics related to the subject and several of the areas demanding further investigation are highlighted.

Three lines of enquiry were pursued to consider the kinematics, kinetics, passive structural properties and muscle activity associated with the foot.

A dynamic pedobarograph with a synchronised video system was used to measure the forces and their distribution under the foot (based on seven marked areas) and six kinematic angles of the foot and lower leg. Sixty-one healthy subjects were assessed and the results are presented. Kinetic and kinematic parameters were found to be consistent and smooth for the test population. Several of the events of the gait cycle were found to be temporally different from values widely reported.

In the second investigation, four cadaveric foot specimens were tested dynamically to determine the role of the plantar structures during loading in various positions. A method of sequential dissection was used and the results support many of the theories regarding ligament function. Tests on the effect of three extrinsic muscles on the foot load distribution also support previous studies while a preliminary investigation of two pathological feet partially clarifies the biomechanical effects of a hallux valgus deformity.

Eight of the foot extrinsic and intrinsic muscles were assessed for the final investigation. Using electromyographic (EMG) recording techniques on six healthy subjects , the muscle EMG activity was quantified during walking a) barefoot, b) with a moulded heel plate, and c) with soft shoes. The results for the extrinsic muscles generally agree with previous work, while the intrinsic muscle activity is more variable. The intrinsic muscles were more active when shoes were worn and displayed unusual fatigue patterns.

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The work was financially supported by several groups and individuals. The Royal Commission for the Exhibition of 1851 provided a generous scholarship that enabled me to get on with the task at hand and not be too concerned with money matters. The Committee of Vice Chancellors and Principals of the UK provided an ORS Award to make the University fees more reasonable and my family made generous contributions in times of need. To all I express my great thanks.

My home has always been Australia and so a move to Scotland was a major step. Robert Barnett and Tim Barker were the closest link I had with Australia and their support and friendship was marvellous. The Chaplains of the University were wonderful in their counselling and friendship and to Jim Wilson I owe a very special debt for being there at a critical stage. Several families have been particularly welcoming and supportive including the Dixons, Hodges, Macmillans, Ross's and Smiths.

Few who have not had the experiences I have had will appreciate how much I have valued my friends and family that kept in contact while I've been away and taught me some valuable lessons:- Judy for her faith and understanding and for a stream of Email, Kathy for sorting my thinking out, and Mum and Dad for giving me faith in myself and others. Many kept me informed of home and made me particularly welcome on my holidays (including Craig, Fiona, Clayton, Megan and ALL the members of my family).

My deep thanks and love go to my family, Mum, Dad and Geoff who had faith in me and offered every opportunity that culminated in this thesis.

But by far the greatest thanks must be reserved for my God. This whole experience would have been impossible without Him to guide it. From arranging the scholarships to helping me find the words to write this document He has shown his hand in virtually every aspect. Many trials and joys have passed; many yet await; but with God's guidance and help I'm willing to step out in faith.

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CONTENTS

CHAPTER 1 INTRODUCTION

The study of human motion has continued down through the Ages beginning with the overall motion of the body and gradually investigating more specific aspects of function. In terms of the lower limbs, the gait studies of Borelli (1680) followed by the Weber brothers (1836) were general in their outlook but subsequent work (particularly in the last 30 years) has investigated the hip joint (Paul, 1967), the knee joint (Morrison, 1970), the ankle joint (Procter, 1980) and most recently the forefoot joints (Jacob, 1989). As Jacob points out it is surprising that the foot seems to be last on this list considering its very special "humanness".

Howells (1960) has placed the foot in a special place of note: " His (the human) foot is a really specialized organ. From the simple foot of his more generalized relatives it has been made over into a unique arched platform. It is solid and strong; it is able to apply a powerful force at the ball, and it is thus the only foot that can take a human step...... In all truth, our foot is our most human characteristic"[1]. As Howells notes it is this specialised foot that has allowed us to stand upright and devote our more versatile hands and arms to becoming the extension of the brain into the physical world.

Interestingly the foot and ankle have been shown to be involved in around 25% of sport injuries and the foot on its own in 15.5% (Garrick & Requa, 1989). Garrick and Requa's study also itemises injury against sport and shows that the injuries associated with walking and hiking will be foot related in 58% of cases. They also note that foot injuries are among the most poorly treated.

Perhaps it is because it is so specialised that the foot has not been studied more closely until recently. No other animal can be used as a basis for preliminary study of the structure and function and in fact most models come from the civil engineering field. It has of course been included in all studies of gait in the past though generally as a single unit articulated at the ankle. A simple truss or rigid body system will suffice for transferring the loads from the ground into the lower leg for most analyses though this simplification is a major one (as attempts to

[1] Howells W (1960), Mankind in the Making. pp30-31.

build robots utilizing such a structure show).

This investigation was initiated to study the foot in the normal daily activity of walking. As previous work has considered regions of the foot in detail (Procter, 1980; Jacob, 1989) this study aimed to consider the foot as a whole. To this end three parallel lines of enquiry were proposed:

- an in vivo study of the kinematics and kinetics of the foot
- an in vitro study of the passive support structures of the foot (in particular the plantar ligaments) to determine their role in load support
- a study of the activity of several of the extrinsic and intrinsic muscles of the foot, as illustrated by the EMG signal, to clarify their role in walking and assess their fatigue characteristics.

These areas of study will be presented in separate chapters following the review of the relevant biomechanical literature. In the final chapter the detailed findings of the three areas of study will be considered together as they relate to the walking cycle and final conclusions will be drawn.

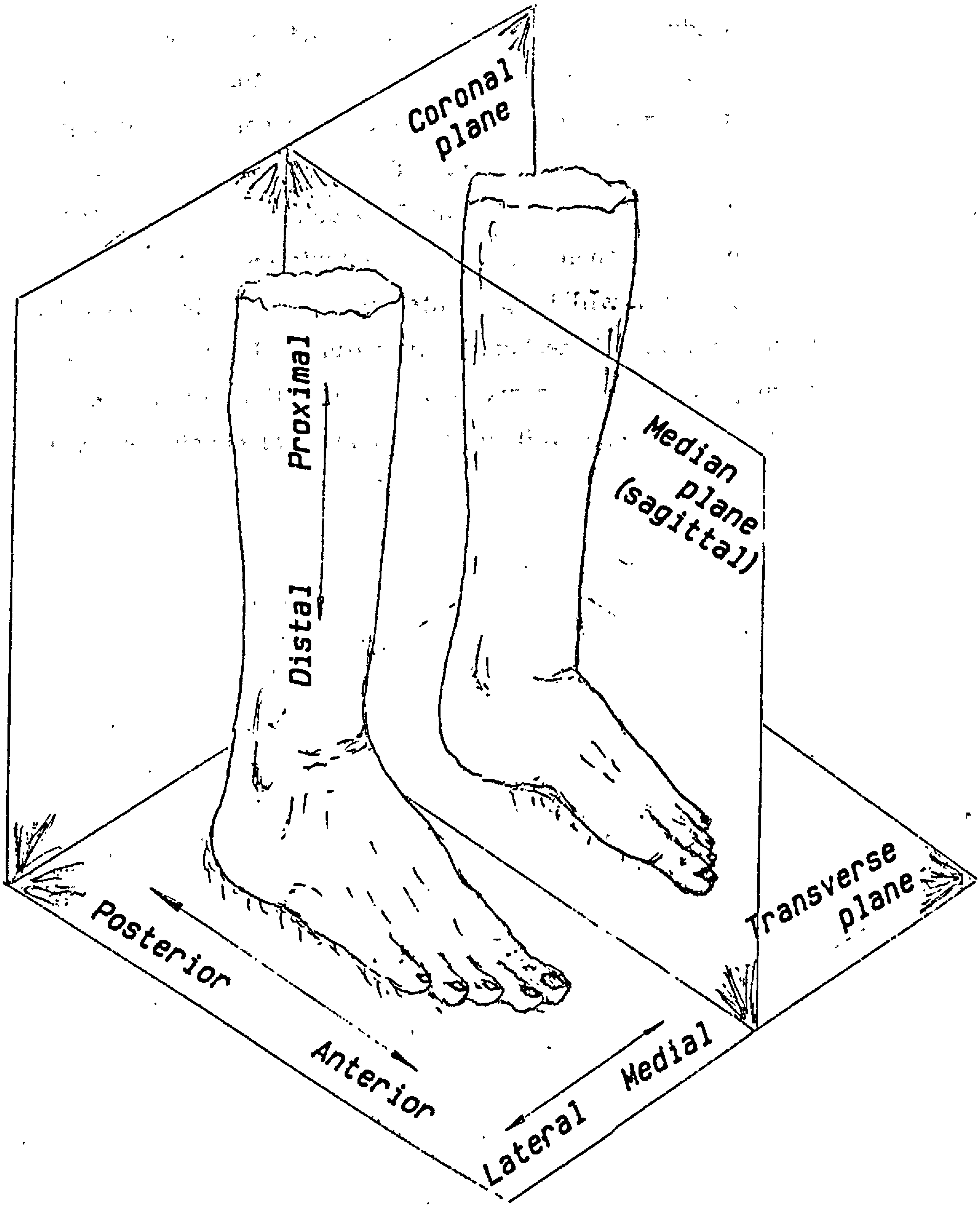


Figure 2.1 Anatomical positions and planes of reference in relation to the lower limbs.

CHAPTER 2. THE GENERAL ANATOMY OF THE FOOT

2.1 INTRODUCTION

Several texts have been prepared on this topic and offer very detailed and complete information (Sarrifian, 1983; Gray's Anatomy,1989) on the foot structure. It is not the intention to precis these texts here, but rather to provide sufficient information to assist those who are new to the area of foot biomechanics and anatomy in the understanding of the subsequent detail presented in this thesis. Those readers who are interested in clarifying detailed points or who seek a greater understanding of foot and ankle anatomy are referred to the general (Gray's Anatomy, 1989; although as Jacob (1989) has pointed out several aspects are very briefly covered) or the specific anatomy text books (Sarrafian, 1983; Romanes, 1987).

2.2 ANATOMICAL TERMS OF POSITION AND MOTION

The foot is unusual when compared with nearly every other body segment in that its major length is in the transverse (horizontal) plane. As a result several specific terms have been presented to reduce the confusion that may result from using the more general anatomical names.

The positions and planes of reference remain unchanged (fig 2.1). Median means middle and so the median plane is the vertical plane that passes through the spinal cord dividing the body into left and right regions. When referring to the foot, plantar refers to the support surface while dorsal (or back) refers to the upper or top region (fig 2.2). In terms of position within the body superficial refers to areas close to the skin while those deep are closer to the centre of the structure.

Movement in the body takes place at the joints and can occur in any plane. In general, motion in a sagittal plane is known as flexion if it moves the distal segment anteriorly or folds it and extension if the segment moves posteriorly or straightens. In order to avoid confusion at the ankle plantar flexion is used for movement toward the sole and dorsiflexion for movement toward the dorsum (fig 2.2a). Abduction is used for movement away from the medial plane while adduction is movement toward it. For the toes however this is modified so that abduction means the spreading of the toes while adduction means their closing, relative to the line of the second ray and toe (2.2b).

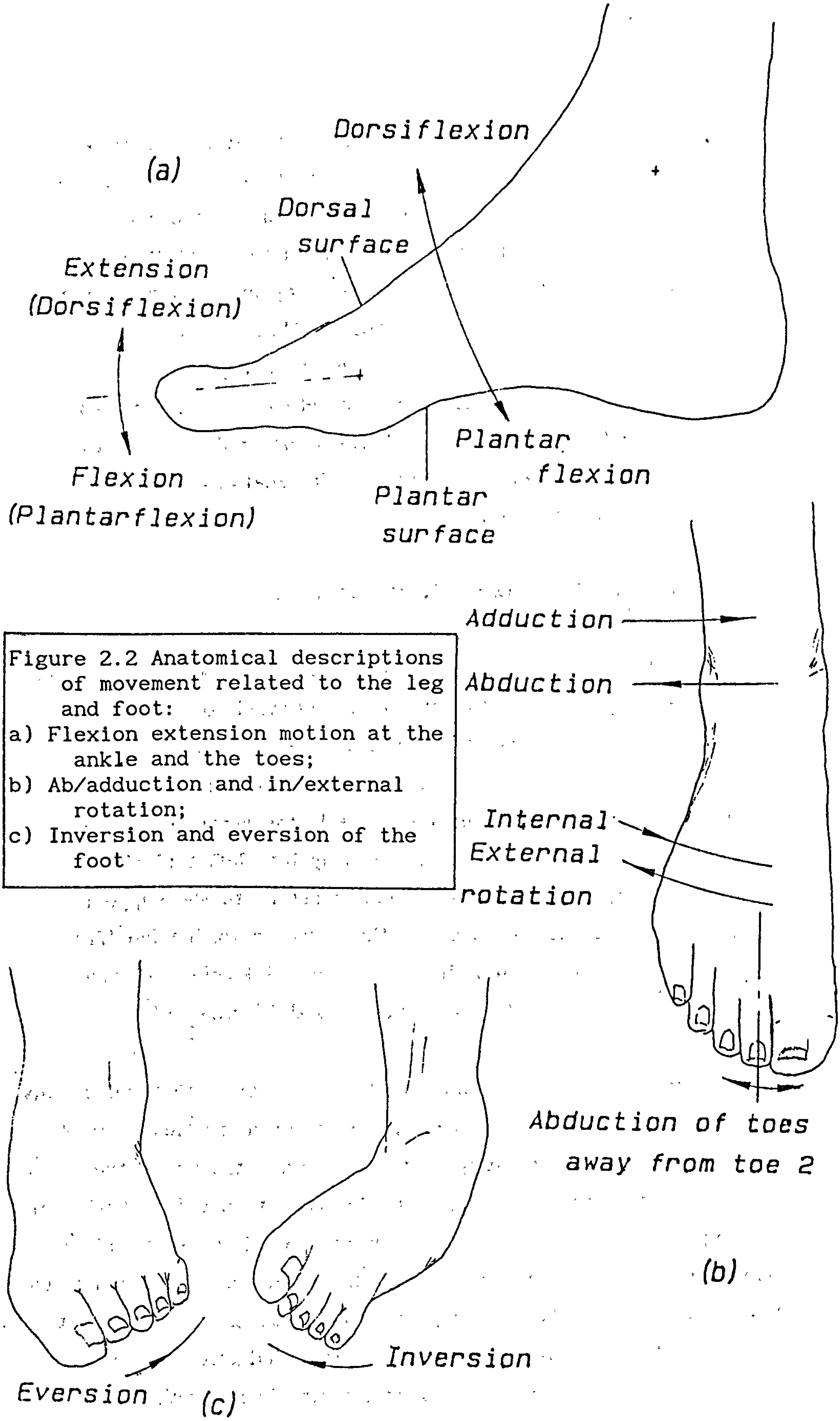


Figure 2.2 Anatomical descriptions of movement related to the leg and foot:
a) Flexion extension motion at the ankle and the toes;
b) Ab/adduction and in/external rotation;
c) Inversion and eversion of the foot

Rotation is used for a twisting action about the long axis of a limb (fig 2.2b). A particular joint axis system in the foot (which will be discussed shortly) permits the forefoot to turn in (so that the sole faces medially) and also to a lesser extent to turn out. These two motions are generally given the terms inversion and eversion respectively (fig 2.2c). The terms pronation and supination are also used with little clear definition of their true meaning. Pronation and supination refer to motion about the long axis of the foot (normally an anterior-posterior axis). A common convention (Gray's Anatomy, 1989; Jacob, 1989) is to use pronation and supination for the components of the eversion and inversion respectively that occur about an anterior-posterior axis. For the complete foot, inversion refers to the combined motions of supination, plantar flexion and adduction while eversion refers to the combined effect of pronation, dorsiflexion and abduction. For the purposes of this thesis pronation and supination will be defined as motion (particularly in the rear foot) about an anterior-posterior axis. Inversion and eversion will be considered as motion tending to twist the forefoot in or out respectively. In certain circumstances these definitions will be equivalent.

2.3 BONES OF THE FOOT

There are twenty-six bones in the foot itself excluding the sesamoid bones (fig 2.3). The articulations allow a very diverse range of motion that permit the foot to adapt to uneven and changing support surfaces.

The foot articulates with the tibia and fibula through the talus which is generally considered a conical hinge joint. The talus is linked to the calcaneus inferiorly, and in a ball and socket configuration (bound by the plantar calcaneonavicular or spring ligament) to the navicular and calcaneus bones distally. The calcaneus forms the major skeleton of the heel while the navicular articulates with the three cuneiform (wedge shaped) bones distally. The final bone that makes up the tarsal bones, the cuboid, articulates on the anterior surface of the calcaneus. These articulations mean that the talar joint itself is generally limited to plantar and dorsiflexion and the complex motions of in and eversion are the result of the ball and socket nature of the articulation between the calcaneus, talus and navicular bones (talocalcaneonavicular [Tcn] joint). The axis of rotation of this joint is now accepted as passing from the plantar, posterior aspect of the calcaneus up through the navicular bone

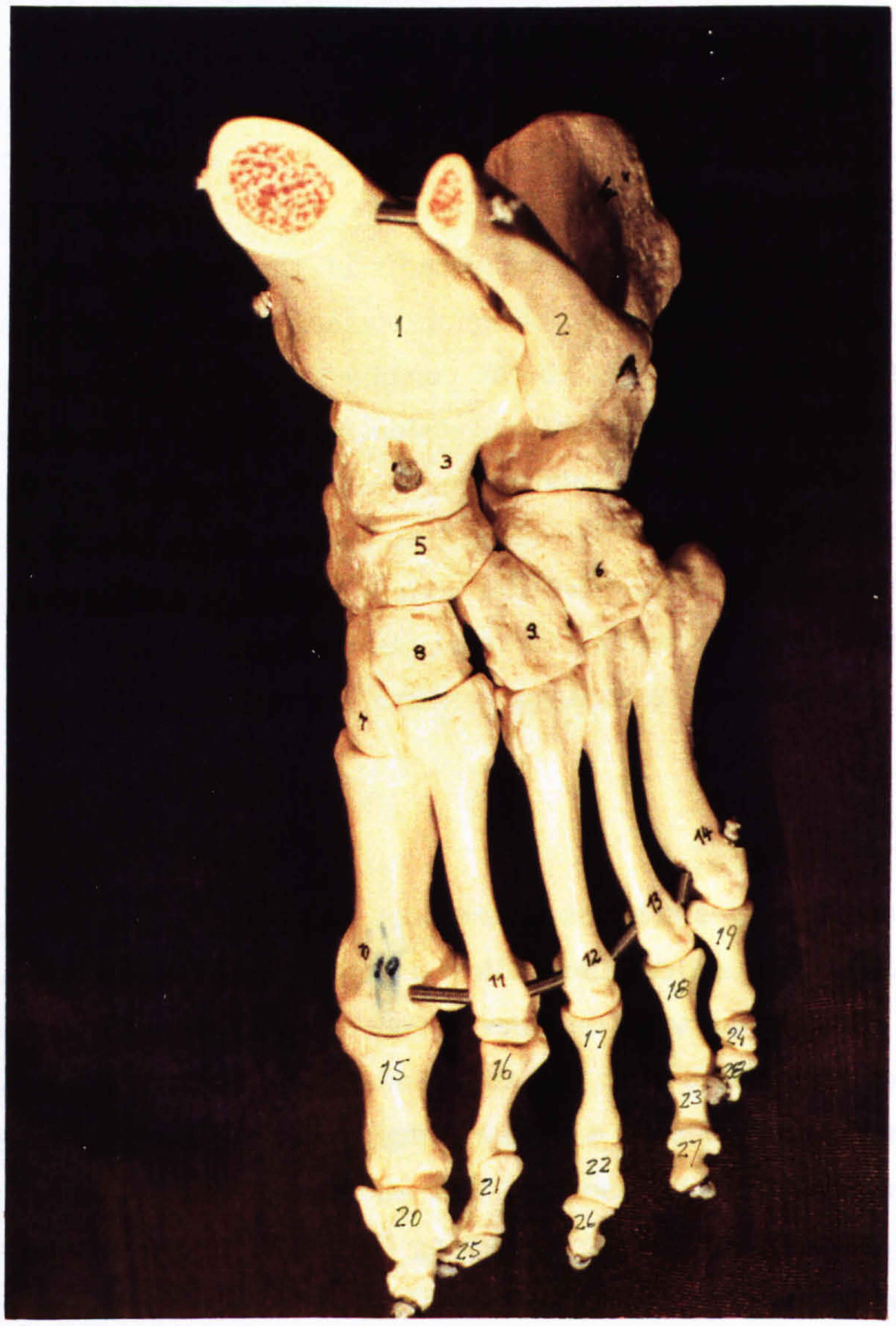


Figure 2.3 The skeleton of the foot and lower leg: (1) tibia, (2) fibula, (3) talus, (4) calcaneus, (5) navicular, (6) cuboid, (7) medial cuneiform, (8) intermediate cuneiform, (9) lateral cuneiform, (10-14) first to fifth metatarsals respectively, (15-19) first to fifth proximal phalanges, (20) first distal phalanx, (21-24) second to fifth middle phalanges, (25-28) second to fifth distal phalanges. (The two sesamoid bones under the 1st metatarsal head are not visible; see fig 3.8 for a medial view)

at about 46° to the horizontal.

The five metatarsal bones are set side by side and their proximal ends (bases) articulate with the cuneiform and cuboid bones as appropriate and with each other. The articulations are such that the second metatarsal is very firmly restricted in its range of motion as a result of the many articulations it supports. Each of the metatarsals has a distal end (head) which articulates with the proximal phalanx of the appropriate toe. There is a series of support pads beneath each metatarsal for load support. For the first metatarsal this also includes two sesamoid bones that are imbedded in the plantar ligaments of the major toe or hallux. They run in grooved articulations about the base of the first metatarsal. Each toe consists of three phalanges with the exception of the hallux which has only two. The distal phalanges of the fifth toe may be fused together.

2.4 LIGAMENTS OF THE FOOT

There are more than fifty ligaments associated with the foot. For this study only the major ligaments transmitting loads about the ankle (fig 2.4) and the plantar aspect of the foot (fig 2.5) are presented.

At the ankle the major ligaments are lateral and medial. The medial (deltoid) ligament is a particularly strong connection from the medial malleolus to the three rear foot bones and also forms the medial border of the calcaneonavicular ligament. The lateral ligament has three branches. The anterior talofibular ligament connects the lateral malleolus to the neck of the talus and is often the ligament affected in acute inversion injuries of the foot. Its posterior counterpart is stronger and passes back to insert in the posterior tubercle of the talus. The calcaneofibular branch inserts into the lateral surface of the calcaneus and restrains medial motion of both the calcaneus and talus.

The anterior and posterior tibiofibular ligaments bind the tibia and fibula together at the malleoli and maintain the tibiotalar joint. They permit the separation of the two malleoli to allow for the varying width of the trochlea of the talus but restrict motion to plantar and dorsiflexion.

The most superficial of the plantar ligaments is the plantar aponeurosis. This ligament can appear as a very thick layer of fascia and is attached proximally to the calcaneal tubercle (fig 2.8). Distally the aponeurosis widens and branches into four to five slips which insert

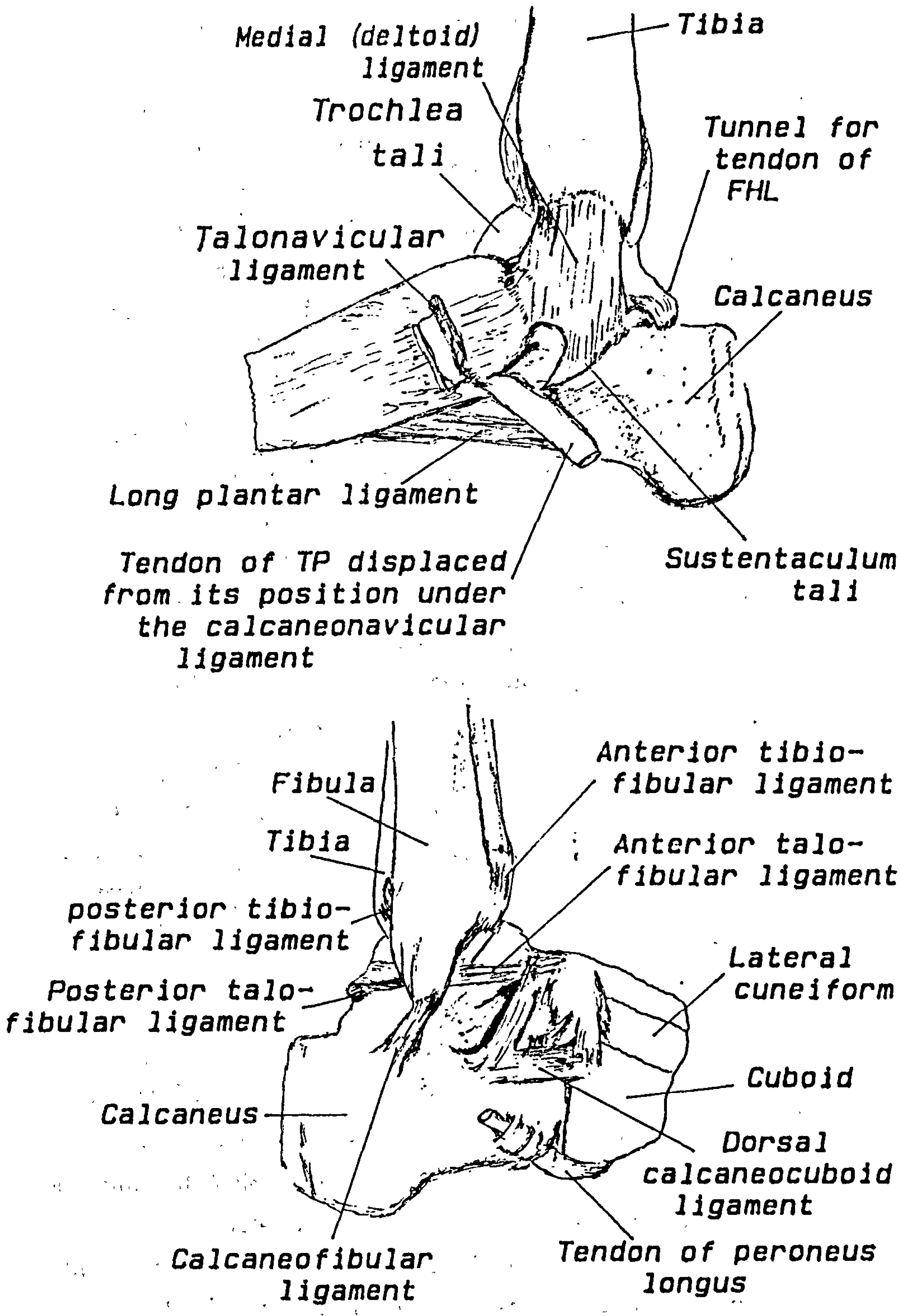


Figure 2.4 The ligaments of the ankle: medial view (top) and lateral view (bottom)

partially in the fibrous flexor sheathes and also into the plantar ligaments of the metatarsophalangeal joints of the toes. As a result each branch of the aponeurosis is firmly bound to the proximal phalanges and dorsiflexion of the toes adds to the tension in the structure. The extent of the aponeurosis over the full length of the foot places it in an ideal position to act as a tie beam at the base of the longitudinal arch.

One of the major support ligaments in the rearfoot is the plantar calcaneonavicular ('spring') ligament. This thick, fibrous sheet connects the sustentaculum tali of the calcaneus to the plantar aspect of the navicular forming the inferior border of the support capsule for the talus. It has extensions to link with the medial ligament of the ankle and receives some support from the tendon of tibialis posterior which runs inferiorly to it. It is this ligament which plays a particularly important role in the maintenance of the medial arches. Rupture or bisection of this ligament results in a complete collapse of the medial arch as the talus is evulsed medially.

Two ligaments are generally responsible for the maintenance of the lateral longitudinal arch. The long plantar and the plantar calcaneocuboid ligament both arise on the inferior surface of the calcaneus. The calcaneocuboid ligament merely spans the calcaneocuboid joint and inserts on the base of the cuboid. The long plantar ligament arises more proximally on the calcaneus and spans the calcaneocuboid joint to insert on the middle three or lateral four metatarsals. The long plantar ligament also forms the cover of the tunnel for peroneus longus.

Together these four ligaments (plantar aponeurosis, plantar calcaneonavicular, long plantar and the plantar calcaneocuboid) represent the primary ligamentous support of the longitudinal arches. All of the metatarsocuneiform and cuboid joints are firmly bound by plantar, dorsal and interosseus single joint ligaments that limit the mobility of these bones. The lateral two metatarsals tend to be slightly more mobile than the medial three. The second metatarsal, by virtue of its wedged articulation and strong ligamentous restraint is considered almost rigid. The next section will consider the muscular support of the arches.

The metatarsals are also united at their heads by the deep transverse metatarsal ligament. This ligament unites the plantar ligaments of the metatarsophalangeal joints and helps to prevent the spread of the bases of the toes. It also limits the amount of independent plantar/dorsiflexion of the individual metatarsals.

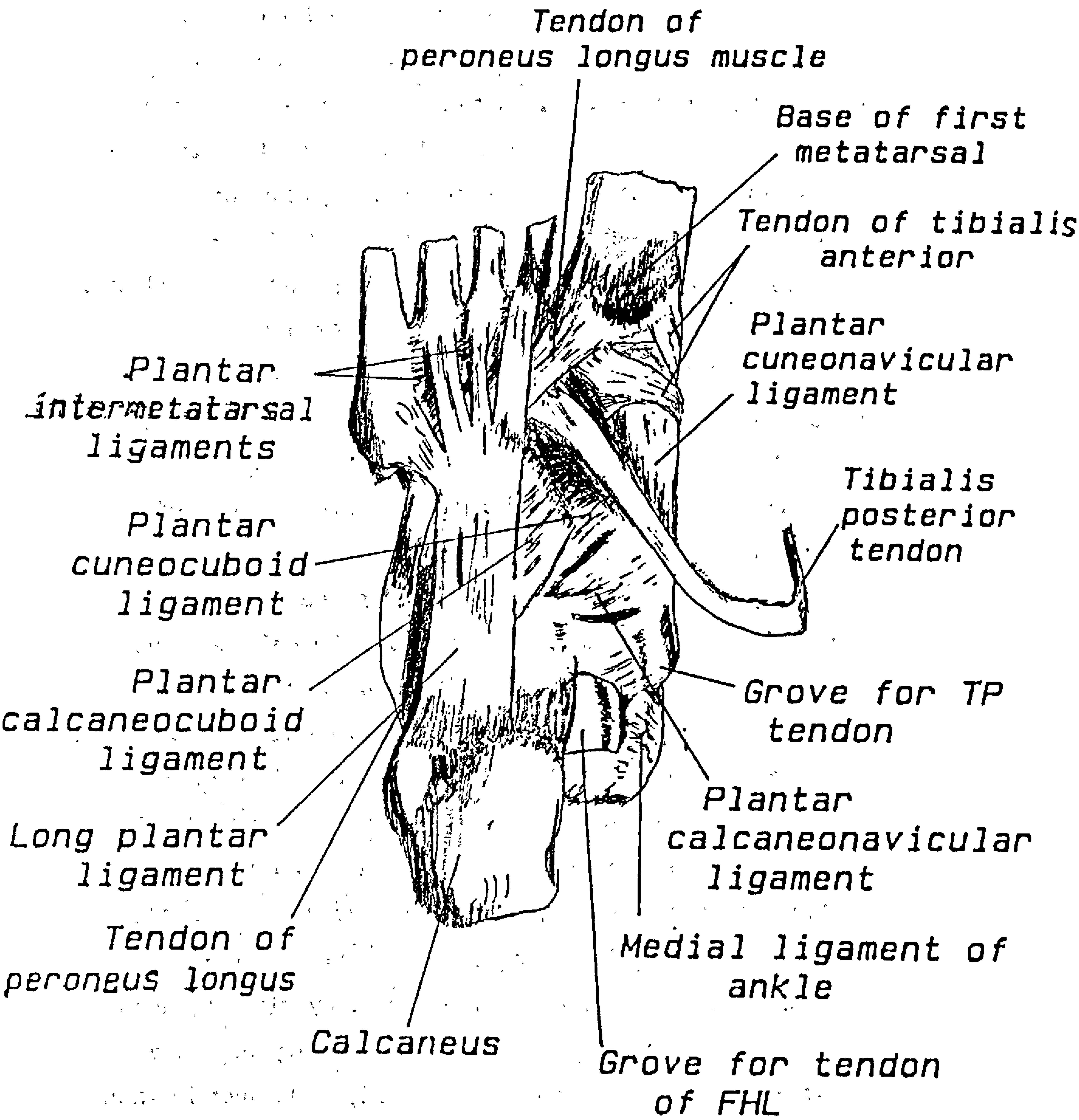


Figure 2.5 The plantar ligaments of the foot (muscles, blood vessels and nerves not shown for clarity)

The phalanges are mostly restrained by the fibrous joint capsules around the metatarsophalangeal joints. The sides are thickened to form collateral ligaments while the plantar section thickens to become the plantar ligament which is attached to several surrounding structures including the plantar aponeurosis and the deep transverse metatarsal ligament.

2.5 THE MUSCLES

There are several muscles with actions in the foot. They range in size from the two heads of gastrocnemius down to the small interossei between the metatarsals. In general a simple subdivision is made into those muscles that originate outside the foot (extrinsics) and those that originate within the foot itself (intrinsics). Figures 2.6 and 2.7 present the extrinsic group of muscles while figure 2.8 shows the plantar intrinsic muscles. Tables 2.1 and 2.2 list the position and action of the major muscles in the two groups. It is important to appreciate that several of the muscles (particularly the intrinsic muscles that insert into the toes) comprise multiple heads although these heads appear to respond together during activity.

2.6 SUMMARY

As stated at the outset, this brief overview should be considered a very simple summary of some of the relevant anatomical detail of the foot. The foot is a particularly complicated anatomical structure to which complete texts have been devoted (Sarrafian, 1983) and it is to these that the reader is directed for more detail on those areas presented here and information about blood and nervous supply etc.

It will be apparent that some of the complex articulations can be obscure initially and the reader is encouraged to study a complete foot skeleton to gain a better understanding of the anatomical layout and function.

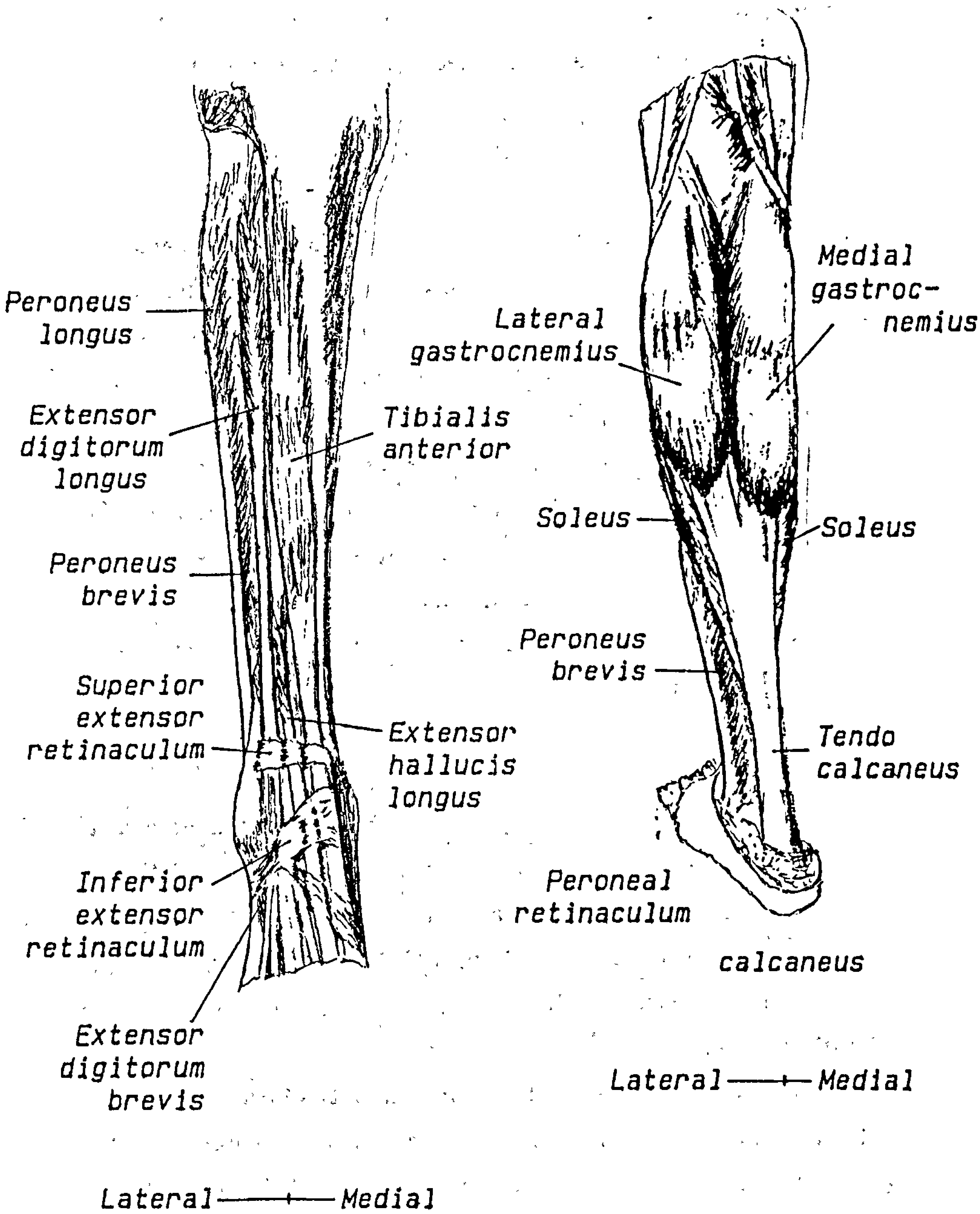


Figure 2.6 The extrinsic muscles of the foot:
(left) anterior view & (right) posterior view

TABLE 2.1 The Major Extrinsic Muscles

NAME	ORIGIN	INSERTION	ACTION
Gastrocnemius	Femur, lateral & medial condyles	Calcaneus	Knee flexion Ankle PF
Soleus	Tibia, mid 1/3 Fibula, upper 1/3	Calcaneus	Ankle PF
Peroneus -Longus	Lateral fibula, upper 2/3	Base Met 1 Med cuneiform	Ankle PF Tarsal eversion
-Brevis	Lateral fibula, lower 2/3	Met 5 base	Ankle PF Tarsal eversion
Tibialis - Anterior	Lateral tibia, upper 1/2	Base Met 1 Med cuneiform	Ankle DF Tarsal inversion
- Posterior	Posterior interosseus memb. upper 1/2 & near tibia & fibula	All tarsals except talus, Bases of Met 2-5	Ankle PF Tarsal arch support & inversion
Extensor hallucis longus	Anterior fibula mid 2/4	Hallux, base of distal phalanx	Ankle DF Tarsal inversion Hallux extension
Extensor digitorum longus	Anterior fibula upper 2/3	Extensor exp. toes 2-5	Ankle DF Toe 2-5 DF
Flexor hallucis longus	Posterior fibula mid 2/4	Hallux, base distal phal.	Ankle PF Hallux PF
Flexor digitorum longus	Posterior tibia mid 2/4	Toes, distal phalanges	Ankle PF Toe 2-5 PF

Note: PF = plantar flexion; DF = dorsiflexion.

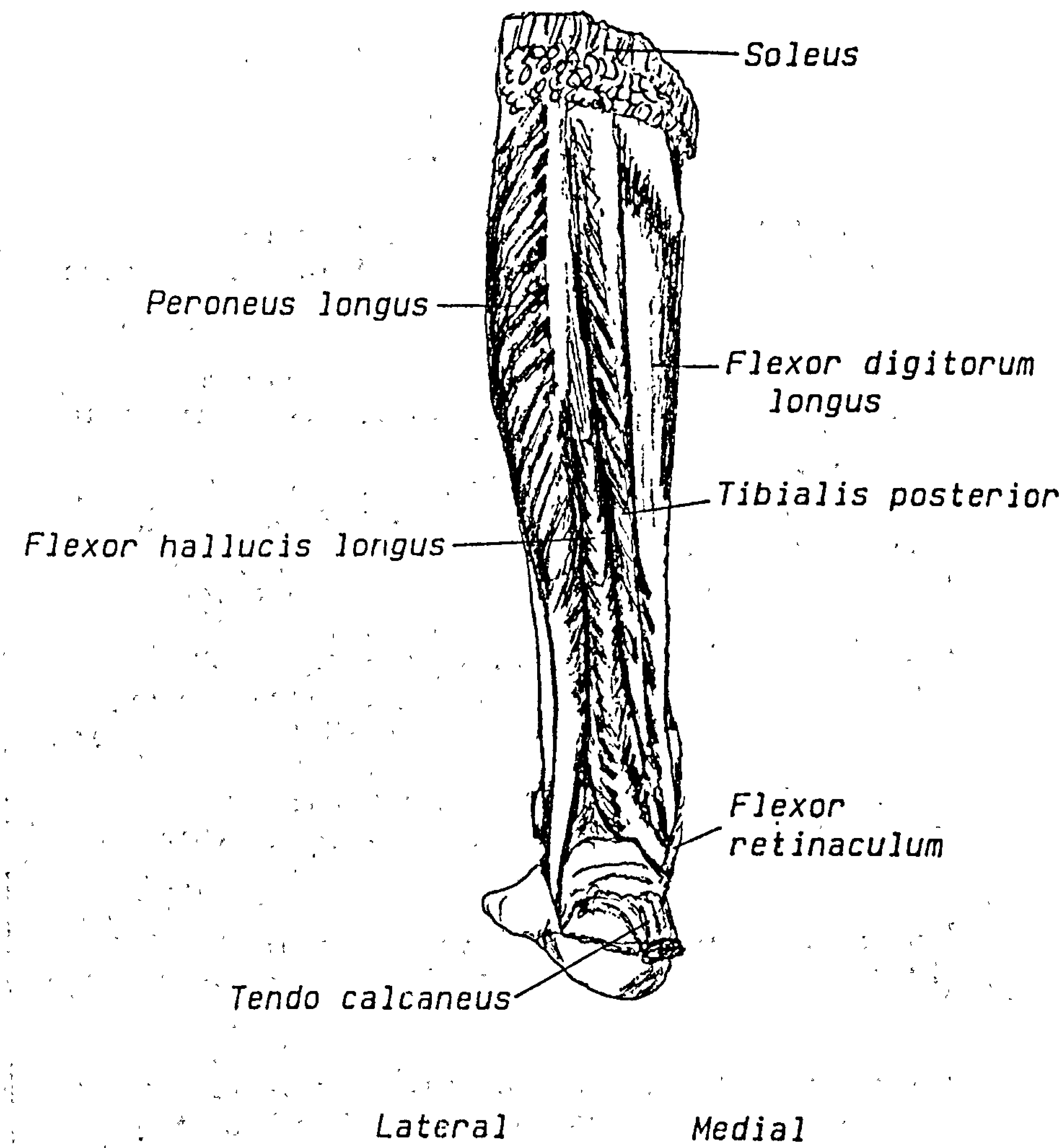


Figure 2.7 The posterior view of the deep extrinsic muscles of the foot.

Table 2.2 The Major Intrinsic Muscles

NAME	ORIGIN	INSERTION	ACTION
Extensor hallucis brevis*	Calcaneus, anter. superior surface	Hallux, prox. phalanx	MTP dorsiflexion
Extensor digitorum brevis	As above	Toes 2-4, extensor exp.	Dorsiflexion all toe joints
Flexor hallucis brevis	Medial, plantar cuboid	Hallux, prox. phalanx	Hallux, MTP plantar flexion
Flexor digitorum brevis	Tuber calcanei, medial process	Toes 2-5, middle phalanges	Tarsus support Toe, MTP & PIP plantar flexion
Flexor Digiti Minimi Brevis	Met 5, base	Toe 5, lat. base of phal.	MTP plantar flex.
Abductor hallucis	As above and flexor retinaculum	Medial base of hallux & med. sesamoid	MTP abduction or plantar flexion
Adductor hallucis	Mets 2-4, bases plantar ligaments of MTP 2-4	Lateral base of proximal phalanx	MTP adduction Accentuates ant. transverse arch
Abductor digiti minimi	Tuber calcanei, both processes	Toe 5, lat. side, proximal phalanx. Base Met 5	Toe 5 MTP abduction Met 5 abduction
Flexor accessorius	Plantar calcaneus	Tendon of FDL	Straightens pull of FDL
Lumbricals	Tendons FDL	Toes 2-5, extensor exp.	Toes 2-5, MTP plantar flexion, IP dorsiflexion
Interossei -Dorsal	Adjacent sides of two metatarsals	Prox phalanx lat. side toe 2-4, med toe 2	Toes 2-4 abduction at MP from line of toe 2
-Plantar	Met 3-5, medial side	Prox. phalanx medial side toes 3-5	Adduct toes to line of toe 2

Notes: * Extensor hallucis brevis is often considered a branch of extensor digitorum brevis.
Met = metatarsal; MTP = metatarsophalangeal joint;
IP = interphalangeal joints; PIP = Proximal interphalangeal joint.

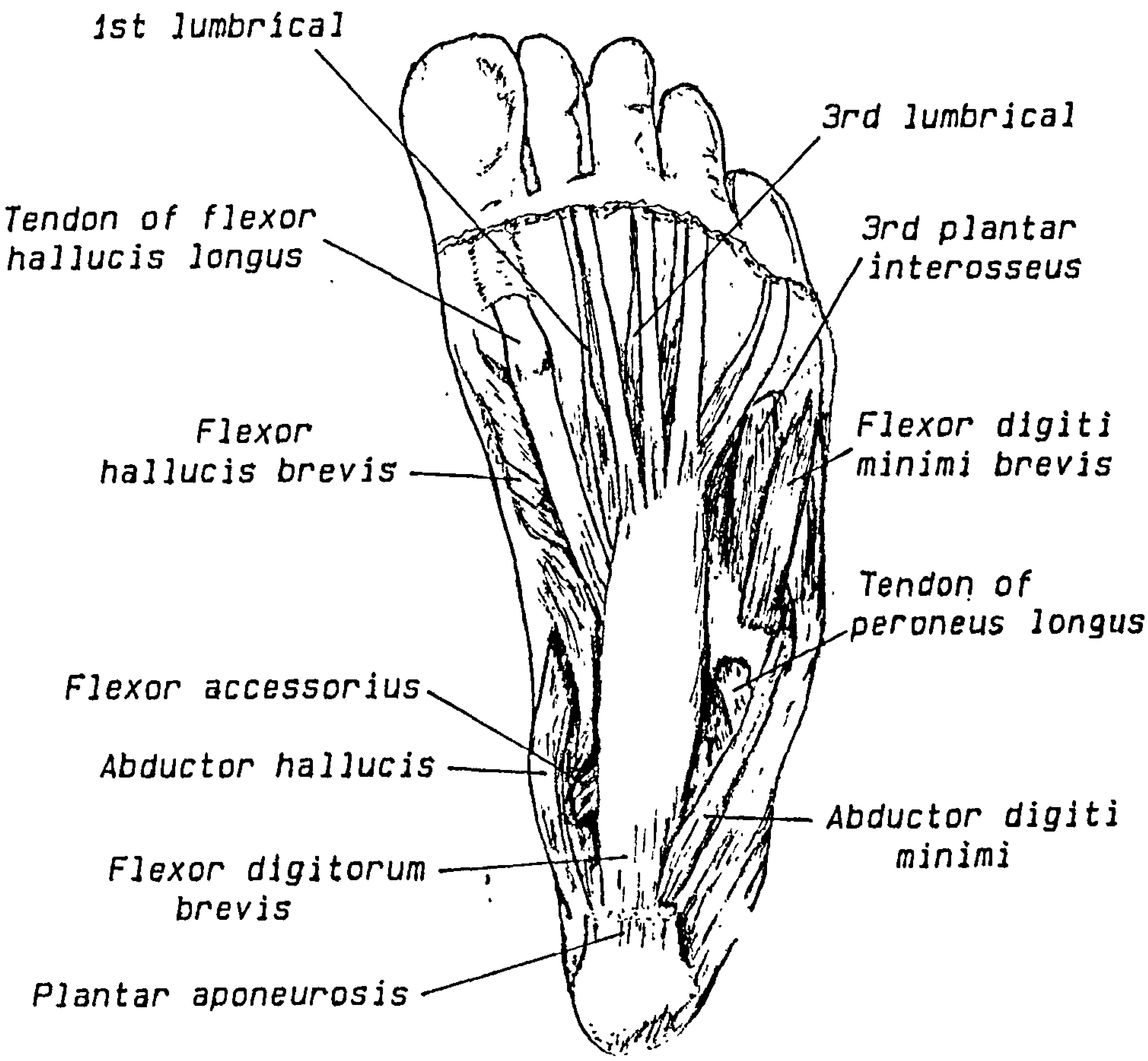


Figure 2.8 The intrinsic muscles of the plantar aspect of the foot.

CHAPTER 3. REVIEW OF THE LITERATURE RELATING TO THE FOOT AND ANKLE.

3.1. INTRODUCTION

There has long been a fascination with the human form. From the biblical descriptions (Song of Songs), through the artists (da Vinci et al.) and on to the scientific understanding of more modern times. It is interesting to note that through all these musings the observer has always been amazed and delighted by what was seen. Perhaps it is surprising then, that many scientific investigators continue to be amazed by the practical efficiency they see when such brilliant engineering has delighted since the earliest times.

3.2. EARLY STUDY IN HUMAN BIOMECHANICS

Scientific interest in the movement of humans dates to at least the Greeks some 200 years BC. Aristotle in his writings, particularly in "De Incessu Animalium" (tr 1984), discusses several principles of human movement that, although crude, show a clear understanding of basic function. He is probably the first to talk of joint articulations, anatomical planes and patterns of motion. Galen (c 200AD) was a physician to the gladiators of Roman times and clearly noted the muscles, bones and sinews that make up the human physique. It would be surprising if he had not made suggestions to the gladiators on techniques for improvement based on what he saw.

Leonardo da Vinci (1452-1519) was instrumental in opening up the study of human anatomy, physiology and movement during the renaissance period. Unfortunately da Vinci's scientific work remained hidden for over 200 years and it passed to others, notably Andreas Vesalius (1514-1564) to advance scientific understanding. Vesalius' work "De Humani Corporis Fabrica" published in 1543 heralded the beginning of modern science, particularly in the medical world. Thankfully kinesiology was not long in following this lead being born with the work of Giovanni Borelli (1608-1679), a pupil of Galileo's. His work "De Motu Animalium" of 1680 combined the sciences of mathematics, physics and anatomy in the description of human movement - a combination still used and expanded for modern study. Of particular note for this present work is Borelli's assessment of motion as the result of muscles acting about joints, clearly presented in descriptive plates (Borelli, 1680). His understanding of the role of the feet, and toes especially in maintaining balance and during the push off phase of gait has also been noted (Jacob, 1990).

From Borelli's work it is clear he was aware that force was a major aspect of both static and dynamic activity. The forces applied through the lower limbs during daily activities must have been indirectly recognised almost since our ancestors stood on two feet. Those who provided shoes, sandals and other foot supports would be in a good position to note the effects of these forces while anyone who has had someone stand on their fingers or toes knows that the force is considerable.

The Weber brothers (Wilhelm and Eduard) took up gait studies following the industrial revolution and investigated, among other things, the centre of pressure, the function of all the major leg muscles and the maintenance of the arches of the foot during the gait cycle (Weber & Weber, 1836). Both Jacob (1989) and Cooper (1976) give a good summary of the Weber's work. Several of the mathematical models the brothers proposed from their observations have since been rejected in the light of more quantitative measurements (Schwartz & Heath, 1932). The first quantitative measurements of the forces under the feet followed less than 50 years later when G. Carlet (a student of the kinesiologist, Marey) used pneumatic soles to indicate the loads during a full gait cycle (Carlet, 1872).

Studies of the physiology and function of individual active elements of the human system were continuing in parallel with this work. The behaviour of muscles became an area of much interest after Luigi Galvani (1737-1798) noted their irritability. Following this work Duchenne (1806-1875), Fick (1829-1901), Roux (1850-1924) and others formulated detailed information on the muscular system.

The advent of photography from 1826-1840 heralded the beginning of quantitative kinematic analysis. Marey (1830-1904) and Muybridge (1831-1904) introduced and refined the photographic technique for recording motion. (As an amusing aside Paul (1989) notes that this interest in recording motion began with a bet on the motion typical of horses.) This work was to lead to that of Braune and Fischer's studies of human gait in the end of the last century. The two Germans were also responsible for a detailed investigation of the anthropometric characteristics of the human body.

Clearly the major concepts of biomechanics were all in place by the beginning of this century. Subsequent work has been to refine and develop the areas of study which could broadly be classified as anatomy and the study of the behaviour of the body structures, the study of force (kinetics) and the study of motion (kinematics).

3.3. THE STUDY OF HUMAN MOTION (KINEMATICS)

The methods for assessing the mechanics of motion are either by direct measurement or measurement from an image of the zones of interest. The method of direct measurement has seen a greater use in recent years as devices such as goniometers become more accurate and simpler in function (Nicol, 1987). From the clinical aspect the ability to place a goniometer over a joint of interest and record an angle (or angles) has many advantages not least being portability (Rowe et al, 1988).

The direct measurement technique does rely on positioning the measurement device correctly to measure the normal articulation such as beside the knee or hip. In the foot these articulations are often more complex and not readily accessible from surface landmarks (most notably the subtalar joint). The foot is also more sensitive to disturbance due to instrumentation placement. In such cases measurement of the angles still revolves around the use of an image of the area of interest.

Following the work of Marey and of Muybridge who used multiple photographic units, cinematography was a simple way to use one camera to take multiple images over a period of time and so assess dynamic motion. Properly positioned, multiple cameras can quantify an object's position in 3D space.

Since these steps little has changed in the methods of kinematics. For higher speed motion the use of faster frame rates was necessary and presently cinematographic resolution is up to 10000 frames per second. Modern electronics and the development of the cathode ray tube led to the recording of images electronically or magnetically, dispensing with the reels of cine tape and their associated development and presentation difficulties. Modern video is capable of high frame rates (1000Hz) and immediate playback of a dynamic sequence. The major problem with these high frame rates is the need for a sufficient lighting level to register the image in 1ms or less. As a result recording fields for the high speed video systems often require intense lighting which can be disturbing to the subjects under investigation and prompts the question of how 'normal' the behaviour and movement is.

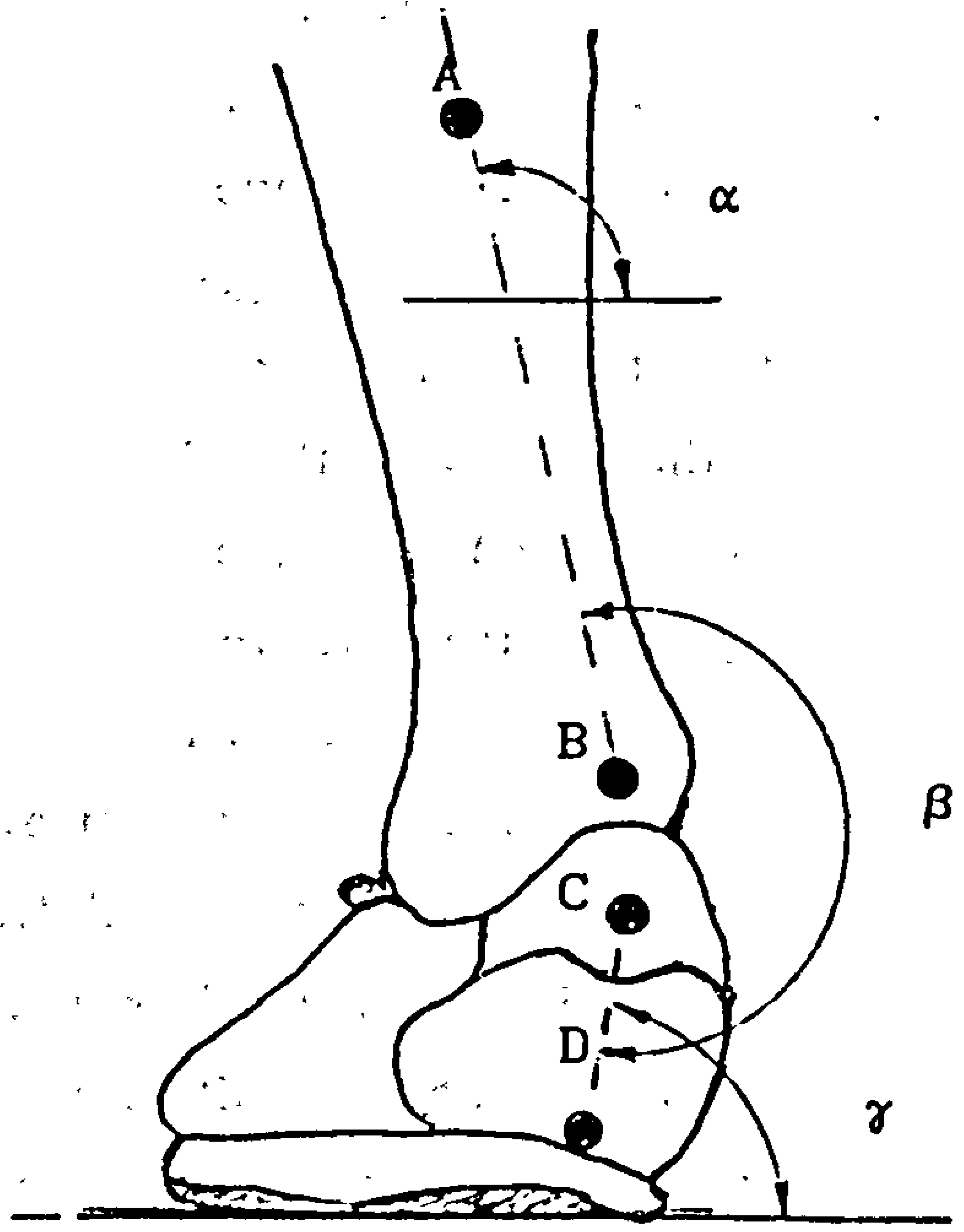


Figure 3.1 Markers and angles for kinematic analysis of the foot
(from Nigg, 1986)

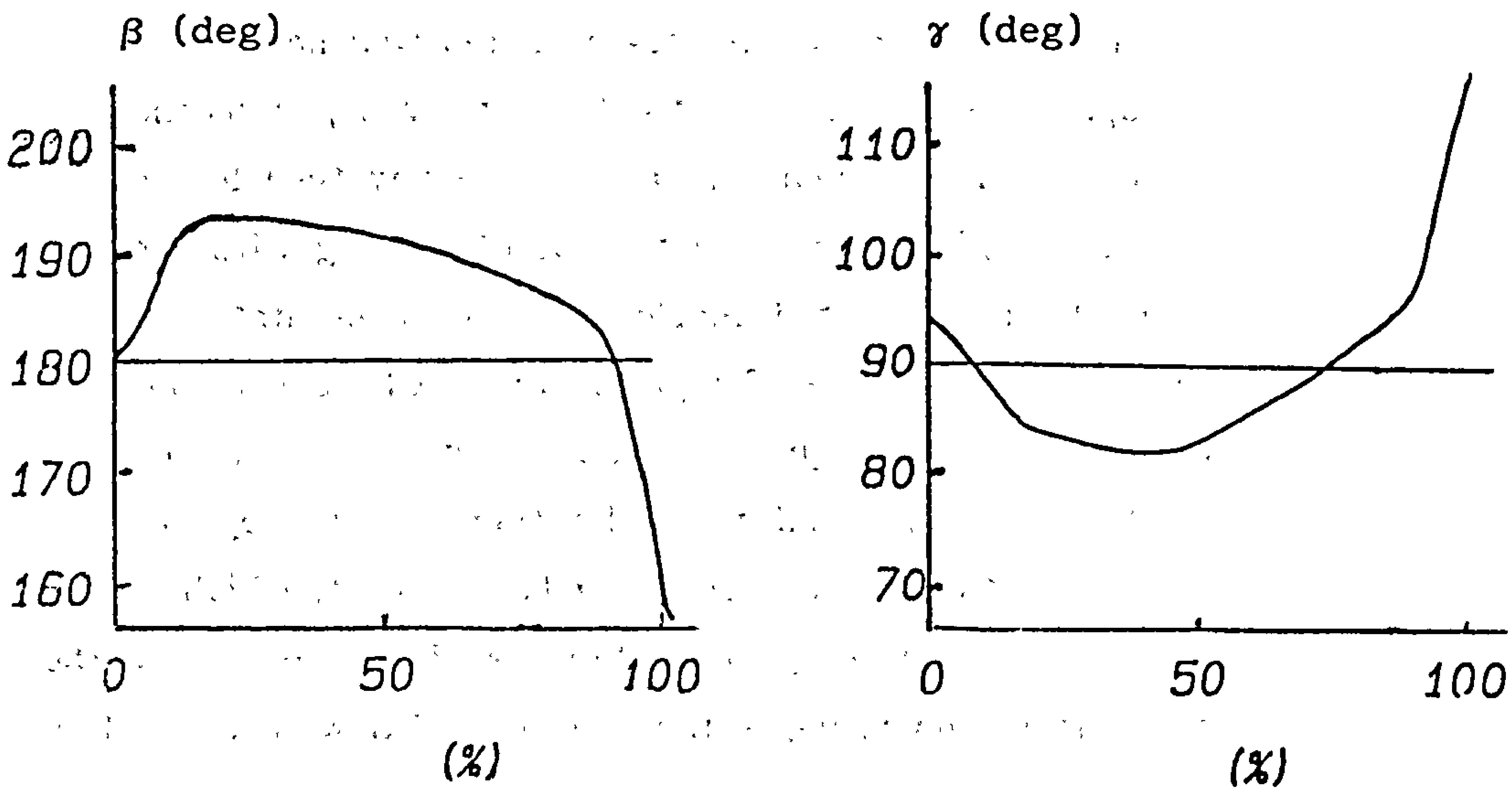


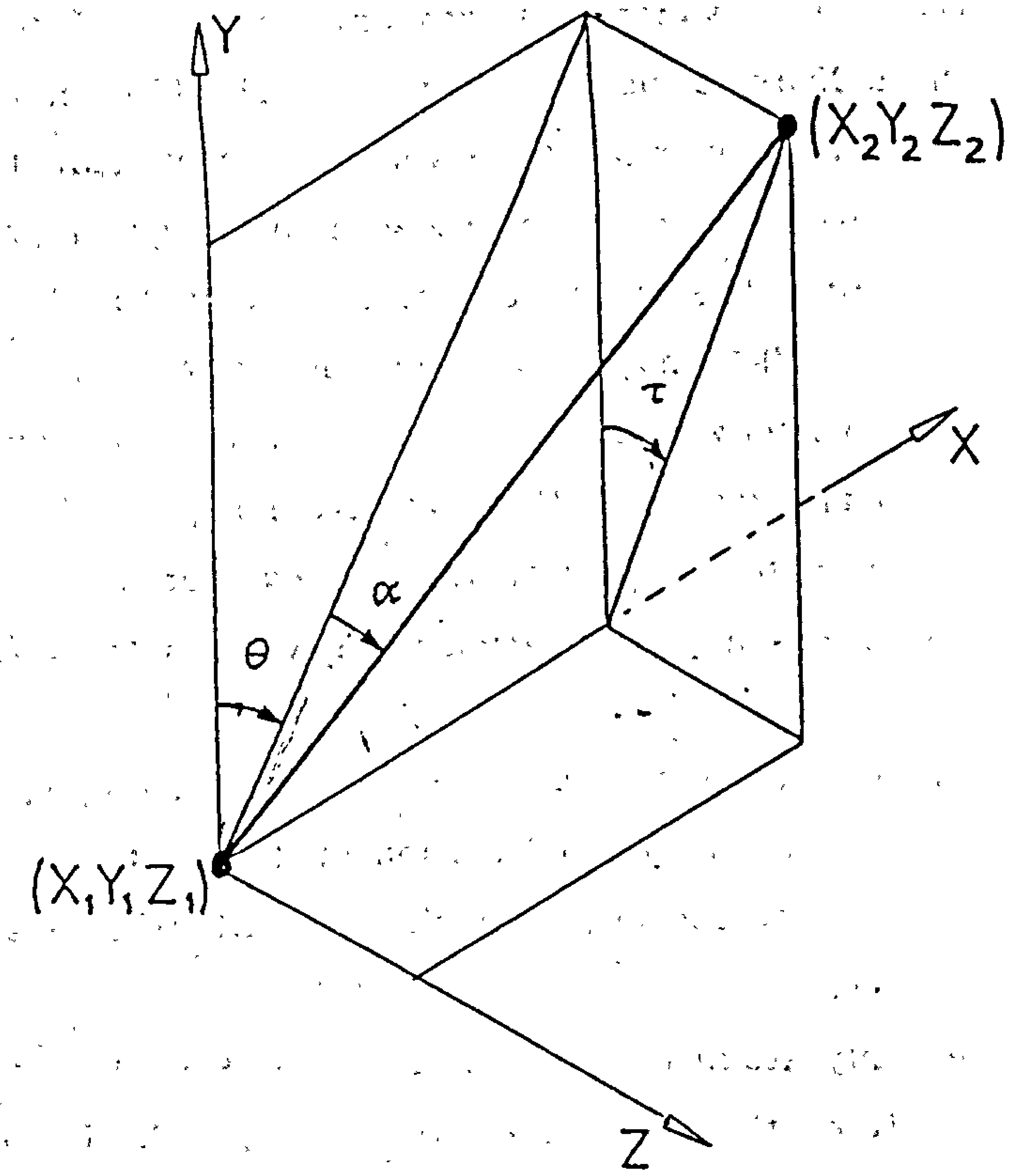
Figure 3.2 Achilles tendon (β) and rearfoot (γ) angles of a subject running at 4m/s normalised over the contact phase.
(from Nigg, 1986)

Different radiation sources can be used for imaging the human form and it is now possible to image dynamic motion of bones and soft tissue at high resolution in vivo, an advance that was only dreamed of some 15 years ago. Techniques such as Xray, computerized tomography (CT) and magnetic resonance imaging (MRI) which were developed for clinical diagnosis are now being increasingly used in kinematic research, although at present mostly for static assessment. The ability to assess structures in vivo with relative safety (as in MRI) may well supersede anatomic measurement previously done in the university and hospital dissection rooms. The digital storage and processing of these in vivo images on computer allows faster and more accurate analysis of the images than was previously possible with postprocessing digitisation used with photographic techniques.

Kinematic recording of the lower leg and foot has been relatively rare in dynamic activity. The amount of motion of the foot and ankle at high velocities has made recording and analysis difficult by comparison with more proximal joints.

One of the early studies of the lower leg was part of a particularly comprehensive study of the lower limbs produced by the University of California, Berkeley (1947). The work used markers fixed to the bones, forceplates and camera analysis that in some ways seems barbaric to modern research technique. The results are, however particularly complete, with data (and perhaps the only detailed dynamic data yet) on the axes and rotations of the ankle. Bojsen-Møller and Lamoreux (1979) show two techniques used in their assessment of foot motion. They first use cinematic recording at 200 and 400 frames per second and for further study, still pictures with light tracks formed by light emitting diodes placed on selected landmarks. The diagrams of the light tracks are particularly interesting in their indication of the speed of the motion.

Procter (1980) used double beaded markers over several landmarks which he then filmed through three cameras at a frame rate of 50Hz. His tests involved several different walking surfaces which produced different ankle motions. He subsequently digitized the film images manually and processed the data to produce a three dimensional record. In the recent years continuing research has considered the details of the foot motion. Work has been done in Canada and the USA and reported in Nigg (1986). His work details the position of a number of markers to define the foot (Fig 3.1). Nigg also includes a very detailed list of the typical angles that can be



$$\cos\theta = \frac{Y_2 - Y_1}{\ell} \quad \Rightarrow \quad \ell = \frac{Y_2 - Y_1}{\cos\theta} \quad --- \; *$$

$$\tan\tau = \frac{Z_2 - Z_1}{Y_2 - Y_1} \quad \Rightarrow \quad \text{projected angle}$$

$$\tan\alpha = \frac{Z_2 - Z_1}{\ell} \quad \text{Substitute * for } \ell \quad \Rightarrow \quad \tan\alpha = \frac{Z_2 - Z_1}{Y_2 - Y_1}\cos\theta$$

$$\alpha \quad \Rightarrow \text{true angle}$$

Figure 3.3 An illustration of the relationship between the projected angle, τ (as seen by a longitudinal camera) and the true angle, α (in a leg coordinate system).

measured from these markers. The angular patterns determined by Nigg are shown in figure 3.2. Another team (Stacoff et al, 1989) used very similar markers but used only one camera behind the subject. It is important to realise that the leg does not move about a cartesian axis perpendicular to the line of progression but about an axis that connects the malleolli set in part by the degree of toe out the subject exhibits. In this situation (particularly toward the latter part of stance phase) the true angle of the leg is not measured from behind since the leg is now out of the plane of reference (fig 3.3). The investigation by Stacoff et al makes no allowance for these 'true' angles and so the results need clarification.

A subsequent Swedish study (Areblad et al, 1990) used a three dimensional Selspot® system for assessing the motion in the rear foot during running. By comparing the sensitivity of the rearfoot angles to misplacement of a single camera Areblad et al indicate that only a true 3D arrangement will provide assured results (fig 3.4). These results also highlight the shortcomings of the two dimensional study of Stacoff et al (1989). From a comparison of the curves of the two studies it would appear that the camera used by Stacoff et al was about 15° misaligned to the back of the leg. The study by Areblad et al (1990) also provides details of two different sequences of Euler angles for describing the leg to rearfoot motion. Their results show little difference in the angles calculated using the two methods though as they note more extensive motion in the rearfoot would result in greater differences. The authors do not conclude which of the rearfoot angles that they have calculated are related to injury though they record a maximum rearfoot eversion angle (β) of -7° with a range during stance of 17°.

Jacob (1989) used a single camera mounted in line with the walkway and then a mirror at the side of the zone of interest to provide a lateral view visible by the camera. He used a frame rate of 150 frames per second and made a special marker triad mounted on a saddle support over the midfoot. In order to minimise the disturbance of this marker triad, Jacob custom made each support in bone cement, although it is of concern that in strapping the support 'firmly onto the foot' some abnormality in the gait pattern would result. Jacob makes no comment on this subject though in a subsequent personal communication he indicated that the subjects did find the arrangement reasonably comfortable. In this study Jacob concludes that the helical axis of the forefoot passes though the first and third metatarsal heads and has a 3° inclination down laterally. He notes only

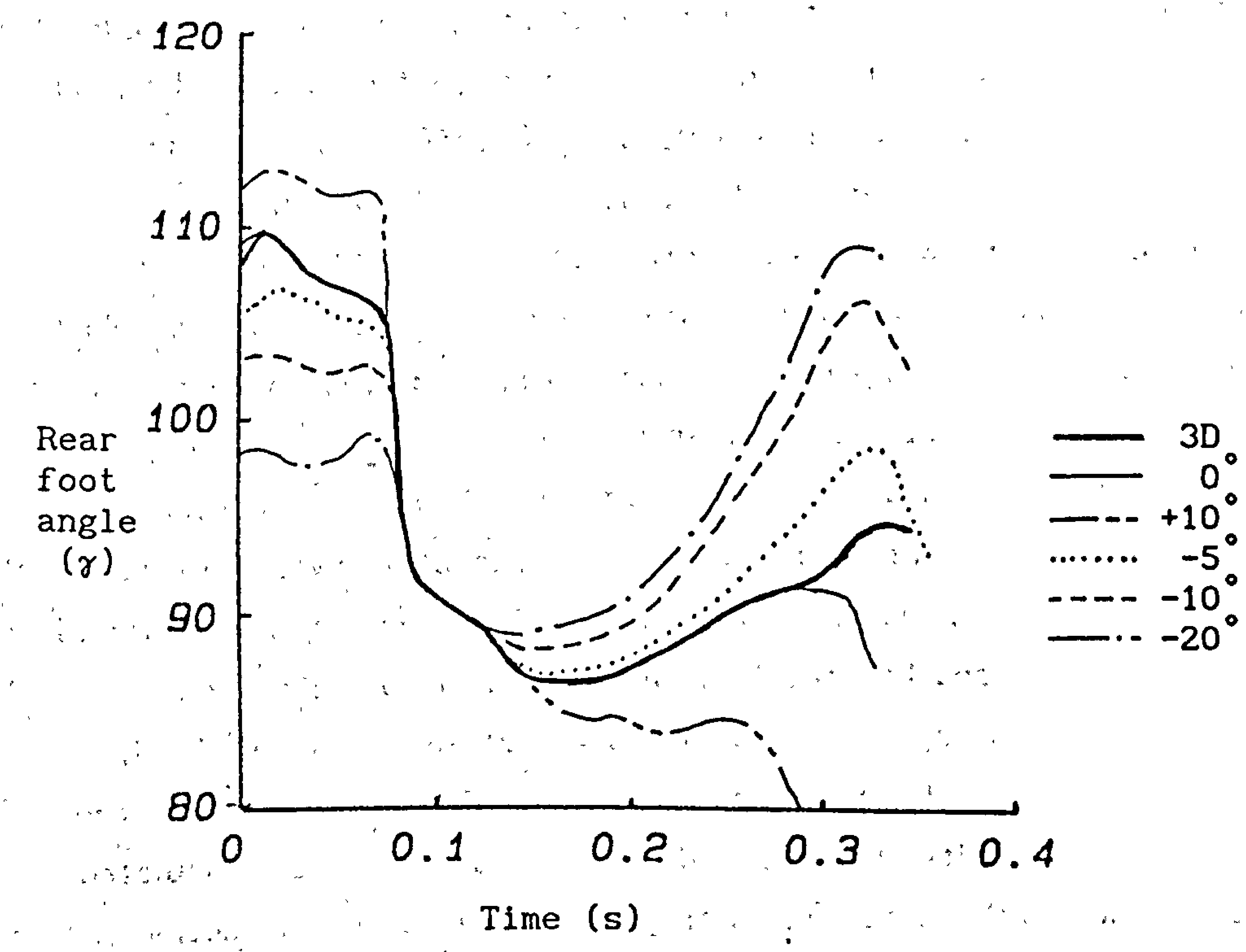


Figure 3.4 The rearfoot angle, γ (as in figure 3.2) as a function of time (running at 3.8 m/s) illustrating the effect of 2D camera misalignment with foot axis. (from Areblad et al, 1990)

3.5mm translation along the axis during the push off phase and notes the 'hinge' nature of the movement. Surprisingly there is only three lines of discussion on the result that toe-in and toe-out do not significantly alter the position of the helical axis to the direction of progression. From the results it appears that the axis rotates about a point between the first and second metatarsal in order to maintain the angle to the line of progression of around 80°. Jacob appears to express surprise that his tests do not indicate a second helical axis through the second to fifth metatarsal heads which has been suggested by Bojsen-Møller (1978, and with Lamoreux 1979). The original articles by Bojsen-Møller and Lamoreux make note that the 'oblique' axis is relevant only for high power applications (eg uphill walking or carrying heavy weight) while the primary axis is involved for normal gait. Jacob's work supports Bojsen-Møller and Lamoreux's 1979 experiments and no conflict is apparent.

3.4. STUDIES OF THE FORCES AND PRESSURES UNDER THE FOOT (KINETICS)

3.4.1. Studies Presented Before 1930

The attempts to quantify the level of support force and its distribution stretch back into the last part of the 19th century. The earliest recording of the vertical force trace during the gait cycle is attributed to Carlet (1872) who used a pneumatic bladder under the foot as the transducer for a kymograph. The resulting traces are typical of the vertical ground reaction forces recorded today though they lack the high frequency components. Carlet's mentor, Marey showed a keen interest in these studies of gait and developed a hand held system for a more mobile measurement device.

Soon after Carlet, Beely (1882) used bags filled with plaster of Paris as a qualitative indication of the load distribution of a subject's step. The method is based on the assumption that a higher force in one area will result in a greater depression. Unfortunately the distortion caused by the settling of the plaster has a significant effect on the load characteristics of the normal foot.

In a very clever experiment Seitz (1901) utilised blood pressure to indicate the degree of pressure below the plantar foot. By standing his subjects on a glass plate he was able to observe the change in skin colour that resulted from pressure induced ischaemia.

The area of foot pressure measurement has been well reviewed by Lord (1981) and Lord et al (1986). The subsequent sections present an overview of some recent techniques and should be considered as complementary to the work of Lord et al.

3.4.2. Printing Methods

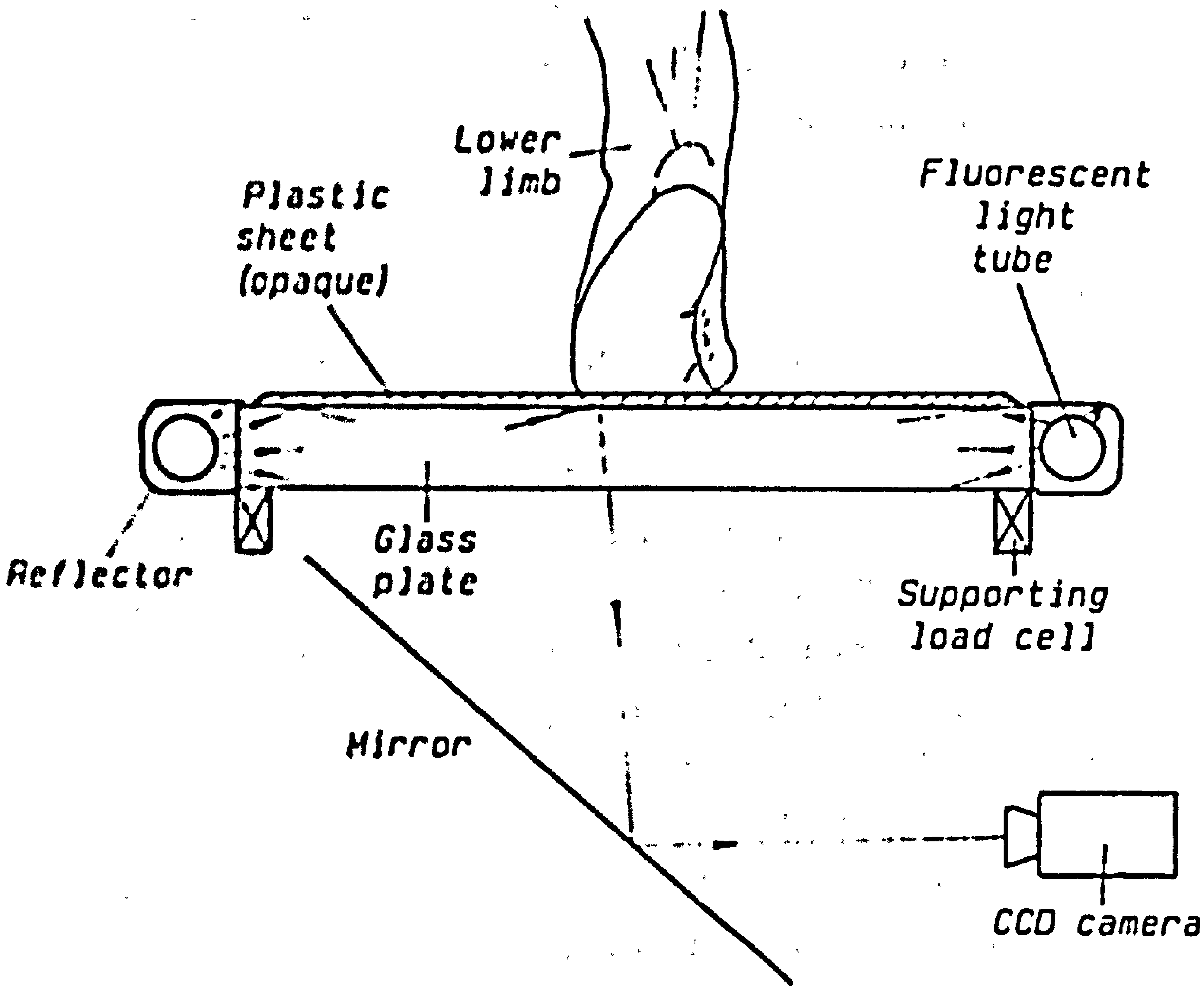
One of the earliest recordings of the pressure distribution following the plaster bags of Beely (1882) was presented by Frostell (1926) who laid an array of fine iron wires onto an inked pad and covered it with a sheet of paper. When pressure was applied the net sank into the pad exposing the paper to the ink. In 1927, Abramson utilised lead plate as the recording surface which he placed over distributed steel shot. The depth of penetration provided an indication of the applied weight on the top surface. Elftman (1934) indicated that Abramson had reported that in standing only 1.8% of the load was taken by the toes and 37% by the forefoot. Abramson's paper (1927) included one of the most cited reviews of the early history of foot force measurement.

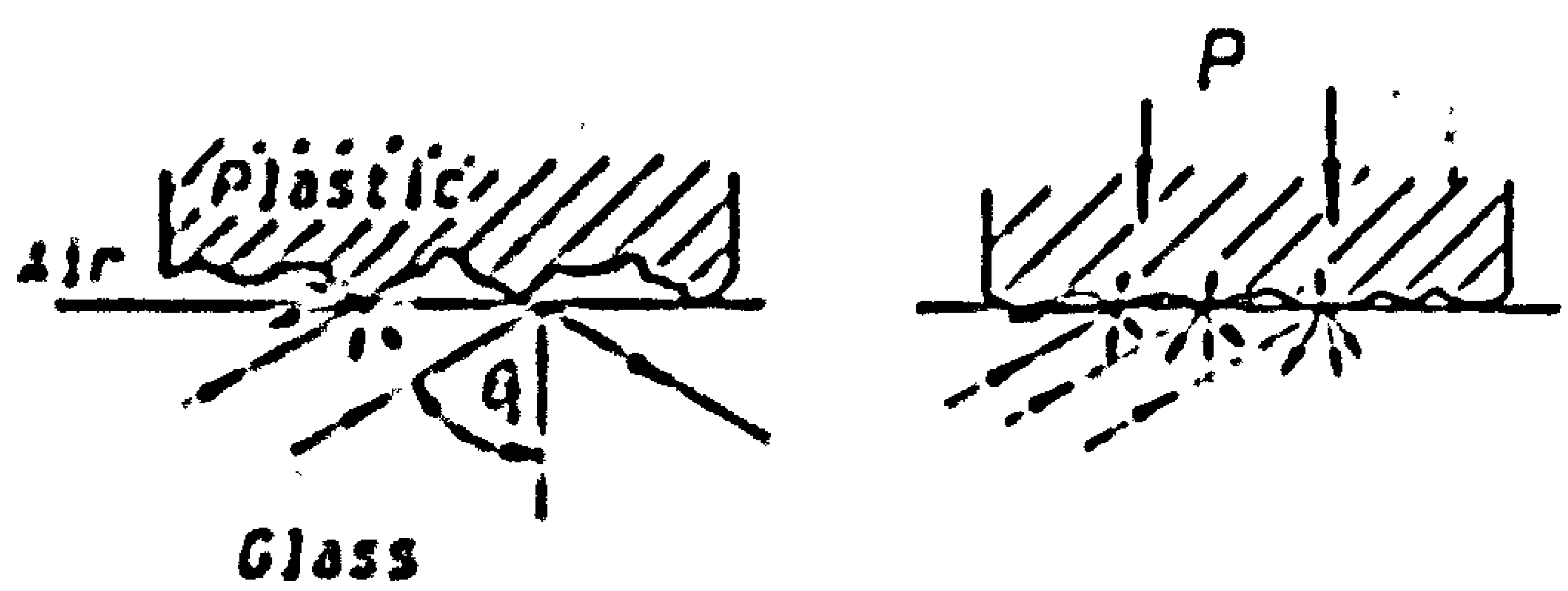
Morton reported a different system in 1930. The kinetograph, used the deformation of inked rubber ridges to provide an indication of the forces supported. Higher pressures on the mat resulted in greater deformation of the rubber ridges and a corresponding increase in the printed ridge width.

The principle of the inked mat has been refined by several researchers most notably Harris and Beath who developed their 'footprinting pad' in 1947 and used it extensively to study soldiers' feet. Others have used this mat in their research and even modified it for their own purposes (Lord, 1981). The limitation of these mats is their inability to provide an indication of the timing of the individual forces, resulting in a record of the *peak* support forces under the foot.

3.4.3. Optical Techniques

Elftman (1934) took the ideas of Morton's printing mat and adapted the ridges into support pyramids. He further enhanced the images produced by using a white fluid between the pyramids and photographing the images through a clear support deck with a cinema camera at 72 frames per second. The simplicity of the system, called the 'Barograph', belied its utility and it was clearly a significant advance in force/pressure measurement under the foot. Elftman noted the changes in pressure distribution when the arch was 'low' and also as a result of different lengths of the





$\mu_a < \mu_g < \mu_p$

$\theta_i < \theta_{crit}$ for refraction

$\theta_i > \theta_{crit}$ for total internal reflection

$\theta_{crit} = 42^\circ$ for crown glass

Figure 3.5 The basic principles of operation of the the dynamic pedobarograph: (top) overall lightbox arrangement and (below) the physics behind the glass/plastic interface metatarsal bones. He also suggested that the centre of pressure locus followed the direction of progression.

A second type of barograph appeared in the 1950s which used a different optical principle to the deformable pyramids of Elftman. Chodera (1957) filed a patent for the Pedobarograph while Hertzberg had shown a similar principle in his 1955 paper on the C-Ray Scope. Although the work was advanced steadily in the following decades it was a group at Sheffield University who completed the development. The most complete details of the system were published by this group (Betts, 1980; Betts et al, 1978; 1980a,b; Duckworth et al, 1982a). The Dynamic Pedobarograph, as it has become known, uses the principles of total internal reflection. A thick glass plate is mounted on load cells and illuminated from the edges. A cover of opaque plastic card is sealed to the top surface and the plate is viewed from below by a standard CCD camera (Fig 3.5). When contact pressure is applied to the top surface the reflective properties of the glass/plastic interface are disturbed and light is reflected out (fig 3.5), resulting in a corresponding lighter grey shading. The amount of light emitted is related to the pressure applied and appropriate computer algorithms convert the grey scales into pressure levels. The system checks the calibration using the supporting load cells to maintain a high accuracy.

Since its development several pedobarograph units have been used to investigate foot function mostly in the clinical setting. The original Sheffield team investigated a group of healthy subjects (Betts et al, 1980b; Duckworth et al, 1982a) and those with various pathologies (Betts et al, 1980b,1988; Duckworth et al, 1982b). Minns (1982) used a pedobarograph (apparently developed by Chodera and Lord (1978) and modified to operate like the Sheffield system) to assess the clinical possibilities of two materials as pressure indicators. Dreeben et al. (1989) used both radiographic and pedobarographic analysis to assess the changes following a metatarsal osteotomy procedure. In 1990, Hughes et al used the pedobarograph to assess the importance of the toes during walking. As part of a very complete clinical assessment, the pressure details produced from the pedobarograph indicated that the toes were in ground contact for over 75% of stance phase and were also subjected to pressures equivalent to the metatarsal heads. Lord et al published a full review of the clinical use of pressure analysis equipment in 1986.

The more common engineering technique of photoelasticity, used in stress analysis, has been implemented in a different system reported by Arcan and Brull (1976). The equipment is more complicated than the pedobarographic principle, using polarising, reflective, and support layers with a dimpled sheet or shoe sole covering to provide discrete indications of load. The equipment has not been as extensively used as the pedobarograph.

3.4.4. Load Cell Based Systems

Although force plates have been extensively used in kinetics, they provide no information about the pressure distribution under the foot except for the locus of the centre of pressure. The principle of measuring force as used in force plates has been modified by miniaturisation to make up load cell arrays. The extent of this miniaturisation ranges from a novel, multi bar, harp-like instrument (Basler,1927) through to some of the modern 1000 cell arrays (Rodgers & Cavanagh, 1988). The majority are based on piezoelectric transducers for their high reliability and natural frequency. The review presented by Lord (1981) contains several other examples of these transducers.

Two commercial systems based on force sensors built into an array have also been developed and tested. The German EMED system, based on the capacitive mat reported by Nicol & Henning (1978), has proved quite satisfactory in the clinical setting while the Irish Musgrave Footprint system, based on the work of James et al (1982) is still in the development stage. Both systems benefit from portability though are forced to compromise their long term accuracy and require recalibration at regular intervals. All forceplate like systems have a significant drawback. Only one footstep can be recorded in any one test, although the EMED system can make use of a thin film insole transducer to provide detail of other strides.

In 1987, Hughes et al compared the relative merits of the Harris-Beath mat, the "Dynapod" (Dhanendran, 1978) and the Dynamic Pedobarograph from a clinician's viewpoint. They concluded that the Harris-Beath mat was the simplest and cheapest solution for general use, the Dynapod the most accurate (but also the most expensive) and the Dynamic Pedobarograph the device with the best resolution. Since this test, further development has enhanced the accuracy of the pedobarograph and it is possible that it now has accuracy comparable to or better than the Dynapod.

3.4.5. Insoles and Footfitted Transducers

To measure multiple strides and to record the forces between the foot and any foot covering (shoe, sandal etc) a thin transducer is required. Several have been developed using strain gauged components (Tappin et al, 1980; Frost & Cass, 1981; Soames et al, 1982; Jacob, 1989), capacitive elements (Holden & Muncey, 1953; Bauman & Brand, 1963; Myazaki & Ishida, 1984; Walker, 1987), piezoelectric cells (Henning et al, 1982; Jacob, 1989; Rodgers and Cavanagh, 1989) and resistive rubbers and plastics (Durie & Shearman, 1979). The transducers are either incorporated into an insole or used as individual sensors under the areas of interest. Tappin et al (1980) deserve special mention since only they, to the author's knowledge, have produced a transducer to indicate shear forces under the foot. Their second paper (Pollard et al, 1983) provides evidence that the shear forces are more variable than vertical forces in different footwear. The authors deserve strong criticism for their poor use of units (kg/cm^2 for shear force and kg for vertical force) which severely hampers interpretation of their results. It appears that shear *pressure* levels of between 0.1 and 1.3 kg/cm^2 were found under the areas of the forefoot. These levels rose by up to 120% when barefoot and fell by up to 80% in a below knee plaster cast. Further studies are required to verify these findings. The authors proposed further work to determine the implication of shear in ulceration.

Unfortunately these insole and individual sensors do have their problems. For complete insoles difficulties occur with reliability and sizing. The range of individual foot sizes is wide and therefore a wide range of transducers is required to accommodate this variation. The movement of the foot both with and without foot covering is very demanding on any insole and they often fail within a relatively limited time. In the case of the small individual sensors, problems occur in attempting to place the transducer under the correct site, maintaining that set position during tests and the accuracy and frequency response of the transducers themselves (the performance of the Langer Electrodynagram has been found to leave much to be desired (F. Ross, personal communication,1990)).

3.4.6. Summary

The above review of force and pressure measurement has included some of the various advantages and disadvantages of the different systems. It is important to realise that the methods measure different parameters. In general, measurement techniques for the force or pressure distribution under the foot fall into three categories:

a) Pressure measurement : uses some method to record the distributed pressure such as fluid filled chambers, light property changes etc. A typical example is the pedobarograph (Duckworth et al, 1982).

b) Force measurement : which includes every load cell based system. Typical examples are the Dynapod (Dhanendran,1978) and the EMED system (Nicol & Henning, 1978). Any subsequent display of pressure is based on the division of the measured force by the discrete areas of the load cells supporting the forces.

c) Foot area contact : measured most commonly using foot switches that have a predetermined trigger level of applied force. The output has only two discrete steps: above or below the threshold. The records are intended mostly for temporal analysis of contact times.

It is important to realise the implications of these measurement methods particularly as they relate to *indicated* pressure. If a uniform pressure of, say 500kPa (5kg/cm^2) is applied over an area of 25mm^2 which coincides with one load cell in a device such as the Dynapod it will be indicated correctly. If however the 25mm^2 is centred midway between two load cells it will be calculated as only half the pressure acting over twice the area. Clearly the force under the area remains unchanged but such a device is quite likely to underestimate *peak* pressures. In order to reduce these problems larger numbers of smaller load cells are used so each cell measures the force under a smaller area, permitting a more accurate representation of the pressure distribution.

3.5. INVESTIGATIONS OF THE FOOT ARTICULATIONS AND LIGAMENTS

3.5.1 Introduction

Generally this has been well reported in several works that have preceded this and the reader is encouraged to investigate the detail if particular clarification is required (Sarrafian, 1983; Gray's Anatomy, 1989).
Although the foot may appear a simple structure on first glance, closer study reveals a very complex mechanism. Several authors in recent times have spent years of research studying particular aspects in detail to provide definitive descriptions. In this regard the aspects of joints of the foot have come in for the closest scrutiny.

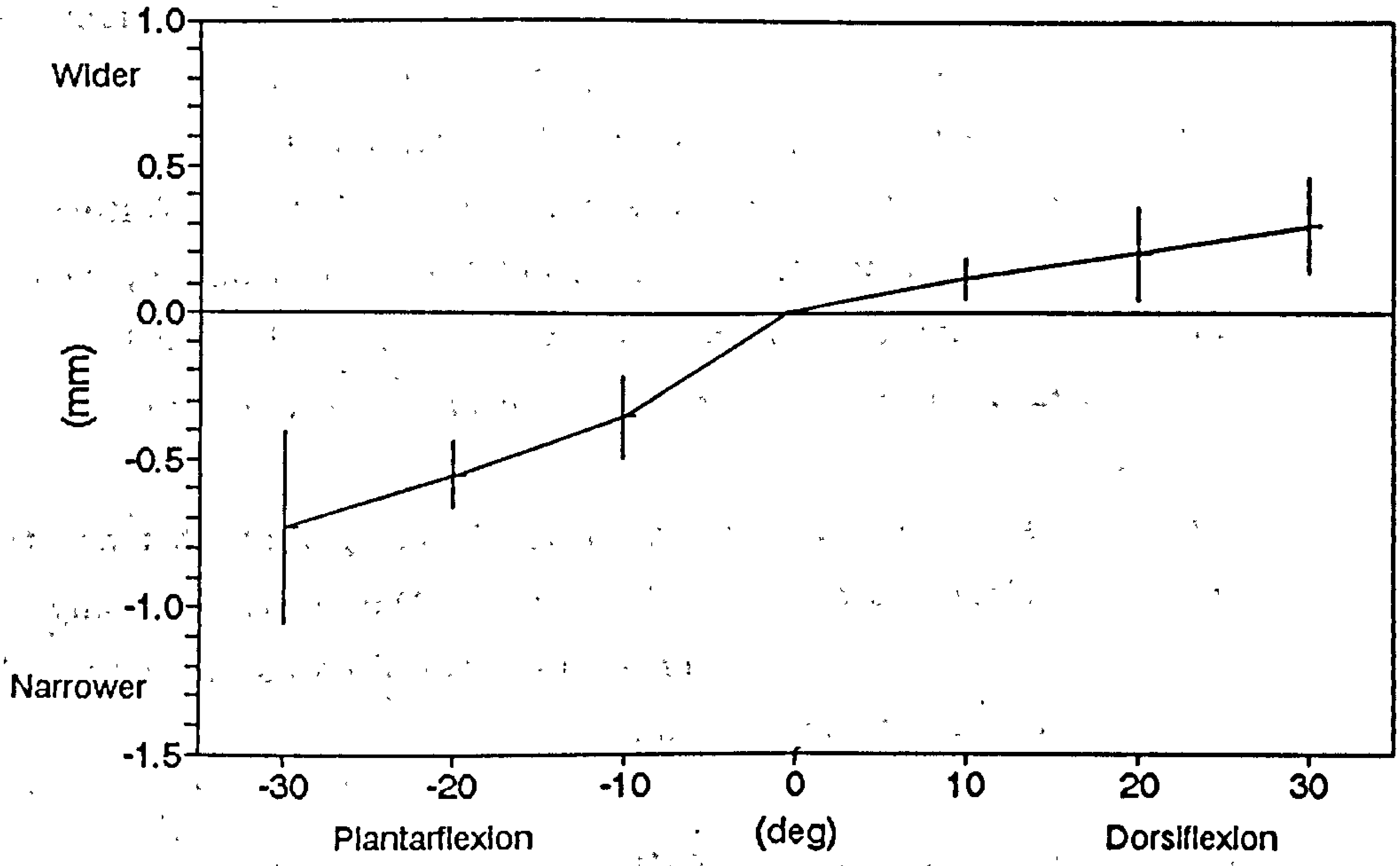


Figure 3.6 The change in the distance between the fibula and tibia (at the malleoli) due to plantar/dorsiflexion of the ankle (from Lundberg, 1989).

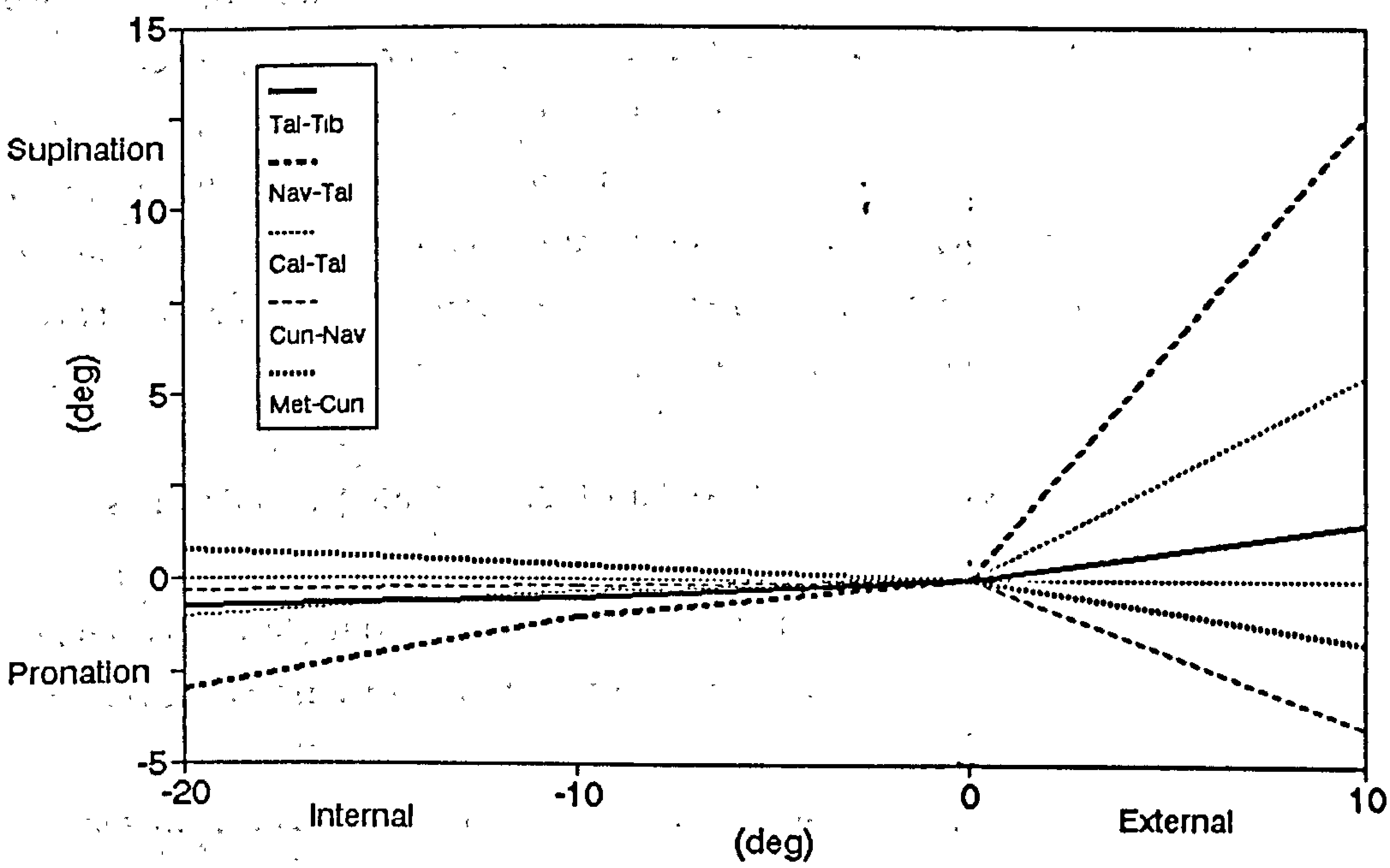


Figure 3.7 The effects of rotation of the lower leg tending to pro/supinate the joints of the rear foot (from Lundberg, 1989).

3.5.2. The Hindfoot Articulations

The talar-navicular and talar calcaneonavicular joints were studied in their in vitro state by Procter (1980) who used mechanical profile gauges and dividers to describe the articulations for subsequent use in his ankle models. Langelaan (1983) extended this work using preserved cadaveric specimens while Benick (1985) considered the articulations in both frozen and freshly amputated specimens and was among the first to conclude that freezing did not adversely affect the soft tissue structures. Both of these studies used tibial rotation to induce motion in the rear foot and define the helical axes of motion of several of the joints. The next logical stage from these in vitro studies was to study the articulations in vivo.

Lundberg (1989), using the techniques of Roentgen stereo photogrammetry on healthy volunteers, produced very detailed results on the articulation axes of the joints and showed the complexity of the joint function. Through this work Lundberg has quantified in vivo behaviour that has generally been accepted qualitatively. Figure 3.6 shows the effects of dorsiflexion and plantarflexion of the tibio-talar joint on the width between the malleoli. The narrowing of the malleoli (over 0.7mm) when the foot plantarflexes would reduce the tension in the constraining ligaments resulting in decreased ankle stability. Figure 3.7 highlights the amount of interaction between the rotation of the leg and in particular the talo-navicular and talo-calcaneal joints, quantifying the behaviour of the subtalar joint. As Lundberg points out it is interesting to note that external rotation produces considerably greater foot rotation (supination, so raising the arches) than does internal rotation. The opposite interaction (ie pro/supination on leg rotation) was similar, with inversion of the foot producing the greater rotation of the leg. This same study showed that *all* the joints of the midfoot were affected by pro/supination of the forefoot relative to the talus.

3.5.3 The Fore and Midfoot Articulations

The forefoot joints have been closely scrutinized by Jacob (1989) who used Moiré fringes to map the joint surfaces and orthogonal photography to quantify the joint axes. He mapped the joints of the first and second rays and offered some detail on the joint axis of the first MTP joint. In this work Jacob notes that only Stokes et al (1979) had previously attempted a quantified description of the MTP joints and their model is significantly simplified. No work appears to have provided more detail than that of Jacob on the joints forward of the cuneiform and cuboid bones. The three lateral rays are yet to be investigated and further detail of the helical axes of all the joints would be a significant addition to the field of knowledge.

The midfoot articulations, involving the cuneiform and cuboid bones, have been assumed to be almost rigid with no movement. Jacob (1989) mapped the topography of the cuneiform-metatarsal joint surfaces (Lisfranc's joint) and also performed a test on one cadaveric specimen to determine the helical axes of the first and second joints. He only provides the details of the first joint (explaining that the results of the second 'are probably strongly influenced by the amount of bending in the slender bone') where he finds an helical axis that is tilted in the frontal plane and passing medial to the joint. The amount of motion for an increase of force of 40N is only 1.5°. For this work the assumption will remain that the joints are semi-rigid based on cadaveric observation.

3.5.4 Foot and Ankle Ligaments

With the large number of bones in the foot and the repetitive nature of the applied loading regime's the role of the ligaments must not be overlooked. The maintenance of the arches of the foot was recognised as due to tension in the plantar ligaments by the Weber brothers (1836) and others (Meyer 1873; Fick, 1911) though they had their opponents who credited the muscles as being primarily responsible (Abramson, 1927; Keith, 1929; Lake, 1937). In 1941, RL Jones undertook a particularly detailed investigation into the role of the muscles in arch support. The major aspects of this work will be presented later (3.6.2) but for the present discussion, in his conclusion Jones clearly states that there was limited or no muscle activity when standing statically and that the extrinsic muscles were not capable of restoring collapsed arches. He also proposed that low arches would only be associated with pain if there were concomitant higher forces on the medial side of the foot.

A new discovery on ligament support was shown in the work of Hicks (1954) where he indicated the tensioning effect that occurs in the plantar aponeurosis when the first and second toes are dorsiflexed (fig 3.8). The resulting pretensioning effect has become known as the Hicks 'windlass' mechanism. Basmajian and Stecko (1963) clarified the muscular role in arch support with their EMG studies of both the extrinsic and intrinsic muscles with the subject seated and controlled passive loading applied over the knee. The conclusions were that none of the muscles was required for arch support until the applied load rose above 200 pounds (890N).

A direct in vitro tensile test of the plantar fascia was undertaken by Wright & Reynolds (1964). They used a static testing regime with loading (using static weights) up to 125-150lb (555-665N) for one cycle. The subsequent loading tested the specimen to failure. Wright & Reynolds found the modulus of elasticity to range from 0.05×10^6-0.12×10^6lb/in^2 (345-827MPa).

As part of ongoing research into the mechanics of movement of the legs (Alexander, 1984), Ker et al (1987) produced a very detailed report on a study of the plantar structures (eg plantar aponeurosis, long plantar ligament etc) of the foot during repetitive loading. By successive removal of the structures they were able to show the flattening of the structure under load and a reduction in its energy storing capability. Ker and his coworkers suggested that for a 70kg man running at 4.5m/s, 17J of the 100J required at each stance phase could be stored as strain energy in the plantar structures with 35J stored in tendo calcaneus.

Jacob (1989) also tested the plantar aponeurosis using a relatively slow extension (1.3mm/min) and reported an elastic modulus of 1236 MPa.

The ligaments that support the ankle have been studied in some detail, generally as they are involved in injury. The work of Procter (1980) provides a large quantity of detail on the origin, line and insertion of several of the ankle ligaments. From the sporting and dance perspective the ligaments that control forced ankle inversion are commonly involved in injury, particularly in acute sprains (Garrick & Requa, 1989; Duddy et al, 1989; Hardaker, 1989; Sohl & Bowling, 1990). Garrick & Requa (1989) note over half of all ankle sporting injuries were sprains and Hardaker (1989) notes ankle sprains (usually involving the anterior talo-fibular ligament) are the most common acute injury among classical dancers. As all these papers point out, incorrect diagnosis or treatment of these injuries can leave the patient with lifelong ankle instability and may seriously curtail their career. Renstrom et al (1988) performed an in vitro study of the lateral ankle ligaments and found that only the calcaneo-fibular ligament was directly strained by inversion/eversion motion. Most authors note that failure of the anterior talo-fibular ligament is most likely in forced inversion with the foot still markedly plantarflexed, a configuration difficult to achieve in vitro. Further work is clearly required and

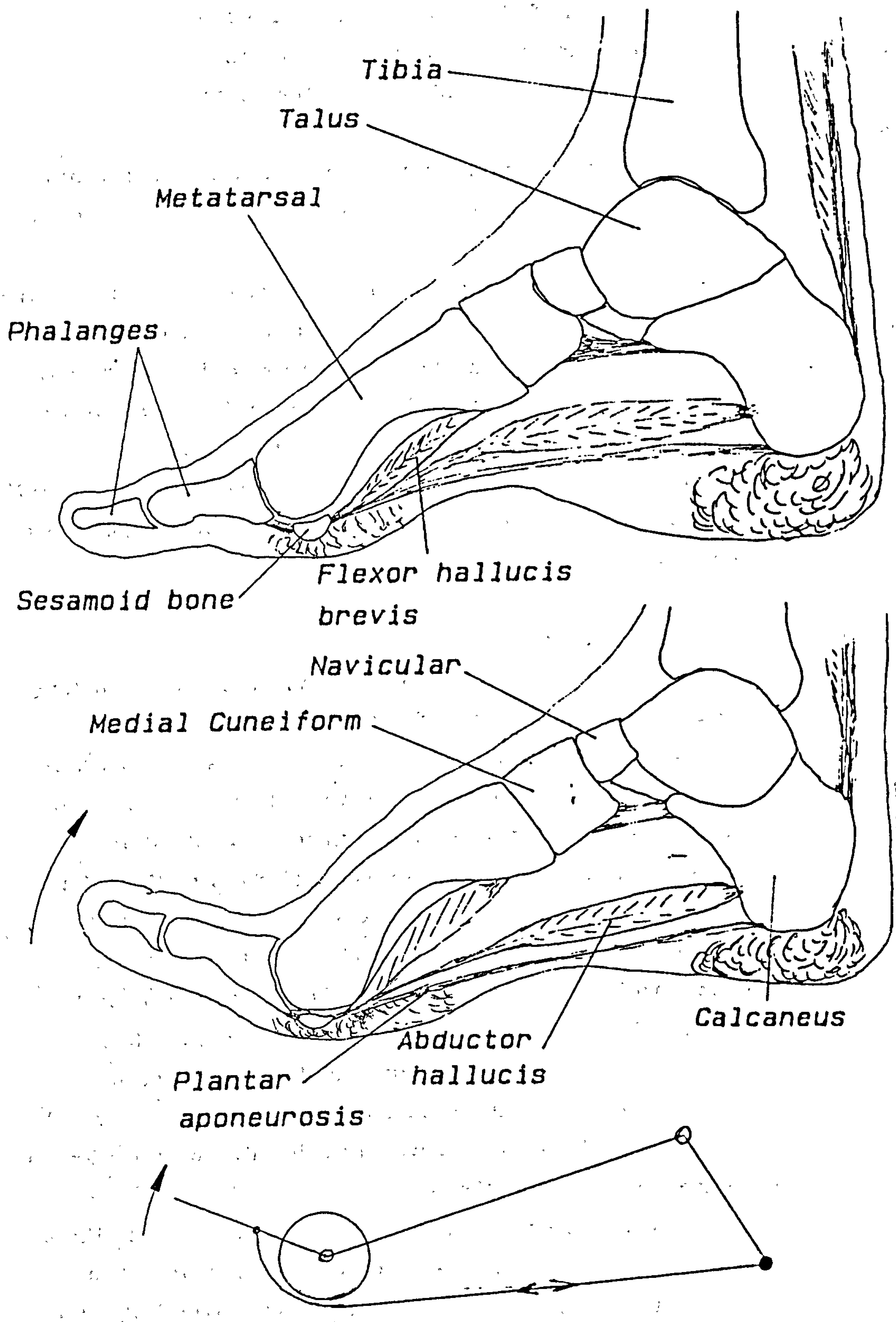


Figure 3.8 Sagittal section through the medial foot to illustrate the tensioning effect of dorsiflexion of the toes (the Hicks 'windlass' effect)

particularly in the area of computer modelling.

3.6 STUDIES OF THE MUSCLES OF THE FOOT AND ANKLE

3.6.1. Introduction to the Muscle Physiology and EMG Analysis

The foot is controlled by a number of muscles originating both from within its structure (the intrinsics) and from the leg (the extrinsics). The number of muscles involved allows a wide range of movement but also complicates the study of the phasic muscular activity. Striated muscle, as the type distinct from smooth muscle, is the major power source for locomotor activity. Each muscle consists of a collection of motor units. A motor unit is composed of a number of muscle fibres (10-1000 in leg muscles) which are innervated by a single α type motorneuron. The motorneuron axon branches to terminate on each muscle fibre. When a stimulus arrives along an α motorneuron all the fibres of a motor unit contract together and then relax (within 1ms to 0.2s). This contraction and relaxation is generally referred to as a 'twitch'. Three types of motor unit have been classified. Type A are fast twitch, high tension units, while type C are slow twitch, fatigue resistant units. Type B motor units are intermediate units with some characteristics of both the other unit types. In a skeletal muscle the proportions of these three types determine the muscle characteristics.

When a muscle is to be fired, stimuli are sent down the motorneurons to the appropriate motor units. The units will not all be stimulated at once but cycle to share the load and allow each unit to 'twitch' on a duty cycle allowing a smooth sustained contraction. The conduction of the stimulating signals through the muscles is registered externally as a changing level of charge and it is this changing charge that is recorded as the electromyogram (EMG). In order to produce stronger contractions from a muscle it is necessary to stimulate motor units more rapidly and in greater numbers. The EMG signal as a result of this greater activity increases and with appropriate electrodes, voltages of 0.5mV or more can be recorded. It is important to note that the stimulating electrical impulses will subside in general much faster than the muscle contractions they excite. Depolarizing potentials from one twitch will have subsided within 5 to 12 ms while for slow fibre motor units, contraction is sustained for 100ms or more. Several texts are available on muscle physiology which contain more detail on the behaviour of muscles (eg Basmajian, 1985).

Table 3.1 Tendon mean cross sections

Muscle	Mean cross section (mm^2) (Procter, 1980)	(Jacob, 1989)
- Extrinsic		
Gastrocnemius } Soleus	58.5	
Peroneus longus	18.9	
Tibialis anterior	20.6	22.4
- Intrinsic		
Flexor hallucis brevis		22.0
Abductor hallucis		18.1
-oblique head		18.2
-transverse head		15.0
Flex. digit. brev. II		3.4
III		3.6
IV		2.5
Ext. digit. brev. II		2.5
III		2.0
IV		1.6

Other techniques often used in parallel with EMG are based on muscle bulk and tendon cross section. These two areas are intended to serve as a base to calculate the force production ability of a muscle and to determine the proportion of forces that a group of muscles can carry. When calculations are based on the muscle itself difficulties arise from three major variables: the fibres themselves (type,number,orientation,area of cross section and standard length), the contraction type (isometric, eccentric and concentric) and the degree of muscle fatigue. Despite ongoing research since the Weber brothers proposed a correlation between muscle size and force production, the estimates of force from a given muscle still vary by up to 800% (Brand et al, 1986). Both Procter (1980) and Jacob (1989) present a good account of the work in this area. An alternative to measurement of the muscle is the use of the tendon cross sectional area. Since the tendon transfers muscle force into the bones it would be reasonable to presume (as proposed by Haughton, 1867) that the cross section is related to the force it must carry. Procter (1980) used this method as the basis for his models and the results are presented in Table 3.1. Jacob (1989) also used this technique in his study of the foot muscles and tendons of five feet, the results of which are also included in Table 3.1.

3.6.2. Muscle Response During Static and Dynamic Activity

The action of the muscles in the leg and foot was noted even by the Weber brothers in 1836. Jones (1941) used bathroom scales and several mechanical tensioners to quantify the effects that several of the extrinsic muscles would have on the forces (although Jones constantly referred to force as pressure) under the forefoot. Table 3.2 contains the details of the effectiveness of each of the extrinsic muscles in loading the forefoot based on four cadaveric tests. Jones concludes that only gastrocnemius and soleus have any major contribution to supporting forefoot forces. As a result of palpating the muscles and tendons in several subjects performing various manoeuvres, Jones retested the muscle ability to vary the load distribution on the forefoot between the first metatarsal and the remaining rays (Table 3.3). This work showed that several of the extrinsic muscles were well placed to alter the forefoot load distribution. In a further aspect of his study Jones considered the effects of tilting the leg bones above the cadaveric feet under load so simulating the various foot positions. It is difficult to understand how Jones interprets the results

Table 3.2 Resulting force at the forefoot from 445N (100lb) tension applied to the selected tendons (from Jones, 1941)

Muscle	Force at ball of foot (N(lb))
TP	18 (4.0)
FDL	39 (8.7)
FHL	44 (10.0)
PL	40 (8.9)
G & S	213-222(48-50)

Table 3.3 Load support at the metatarsal heads with 445N (100lb) tension in selected tendons. (from Jones, 1941)

Muscle	Met 1 (%)	Met 2-5 (%)
None	50	50
PL	74	26
FDL	45	55
TP	44	56
FDL & TP	44	56
FHL	50	50

Notes: TP = tibialis posterior; FDL = flexor digitorum longus; FHL = flexor hallucis longus; PL = peroneus longus; G = gastrocnemius; S = soleus.

Table 3.4 The distribution of force across the forefoot in the various foot positions (shown as ratios) (from Jones, 1941)

	Met 1	Mets 2-4
Cadaveric -		
Tibia vertical	1.0	1.0
5° Inverted	1.0	0.75
5° Everted	1.0	1.34
10° Inverted	1.0	0.66
10° Everted	1.0	1.6
10° Dorsiflex	1.0	0.86
10° Plantarflex	1.0	1.07
40-220lb in tendo achilles	1.0	1.0
In Vivo -		
Normal stance	1.0	2.5
120lb on flexed knee	1.0	1.67

(Table 3.4) as indicating that the forces under the foot tend to redistribute to "limit overturning effects". On the contrary, the increase of force on the side of the acute angle (eg medial side for inversion) would tend to increase the "overturning effect".

Basmajian & Bentzon (1954) considered six muscles (tibialis anterior, peroneus longus, lateral head of gastrocnemius, abductor hallucis brevis, flexor digitorum brevis and abductor digitii minimi) in their study of the static activity of the feet of 17 men and 16 women. Even from this early study the authors confidently indicated no important role for the muscles in normal static arch support. The study also considered the effect of wearing high-heeled shoes for the female subjects. In the static position the shoes resulted in an increase in the activity of both the lateral gastrocnemius and peroneus longus. Basmajian and Stecko (1963) continued this work by loading the lower leg through the proximal condyles while the subject was seated. Their conclusions that no muscular activity is required in a healthy subject for static loads less than 200 pounds (950N) was a major advance in establishing the load carrying capability of the foot and confirmed the qualitative tests of Jones (1941).

The first detailed EMG study of the muscles of the legs was undertaken by the University of California (1953). Their work presented the EMG patterns for 28 muscles for six different activities. Six subjects wearing 'low heeled shoes' were assessed using fine wire implanted electrodes. The EMG signals were RMS rectified and filtered (time constants and cut off frequencies not specified) to produce a series of rectified envelopes. These were normalised, literally using the 'rubber sheet' method, to a consistent time base. The results for several of these muscles during normal walking are presented in figure 3.9. The investigators made no comment on the implications of their data.

Mann and Inman (1954) used bipolar fine wire implanted electrodes to measure activity in the intrinsic muscles of the foot as well as tibialis anterior and gastrocnemius (the implanted head was not specified). The results of Mann and Inman's study were presented as pictures of raw EMG with no amplitude scale and so are generally used as indicators of temporal activity during a number of locomotor activities. Mann and Inman also considered the muscle activity when standing quietly and on the toes. In the former case the intrinsic muscles showed minimal activity while in the latter case all the muscles were reported active.

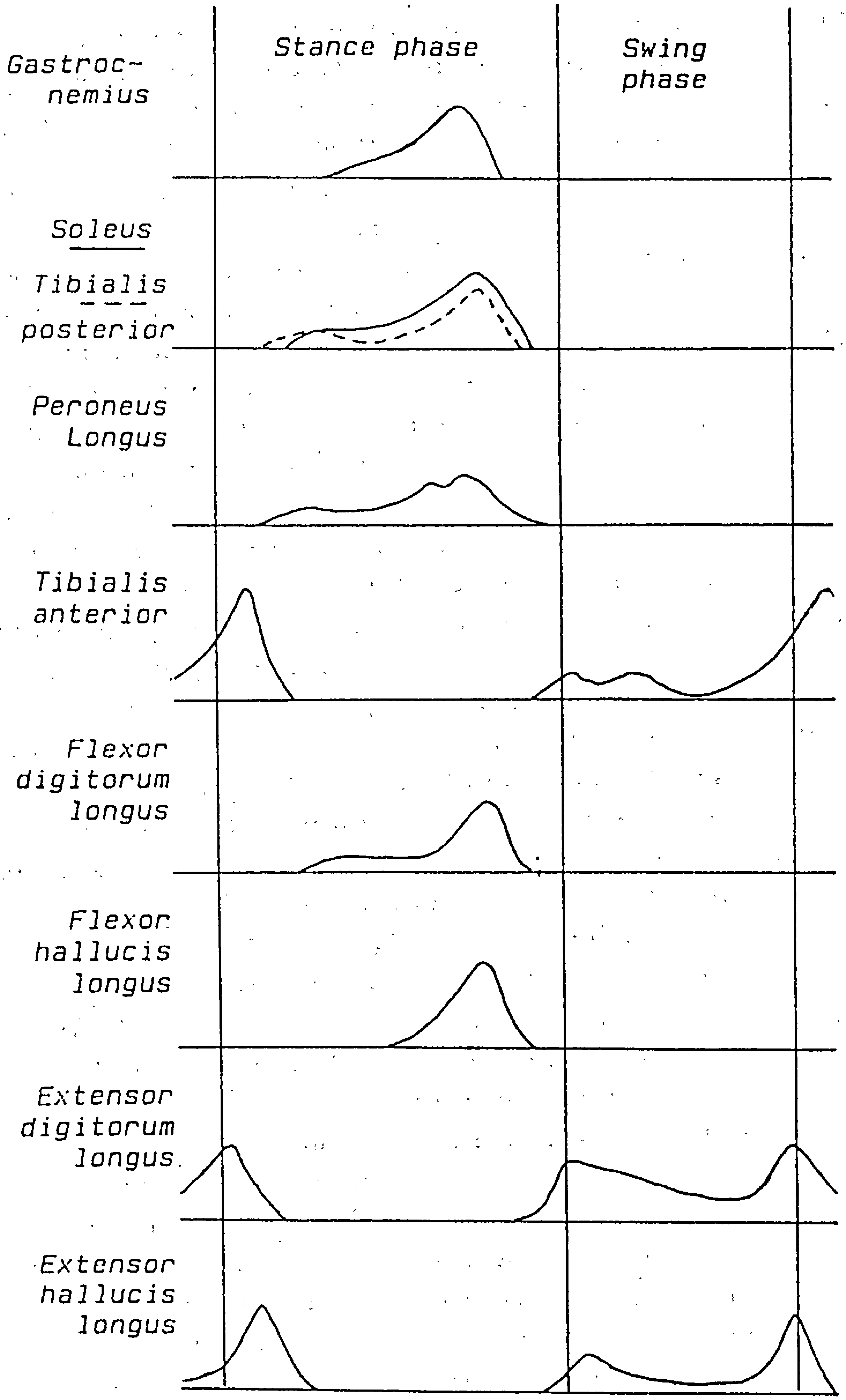


Figure 3.9 The RMS EMG signal of several muscles of the lower leg during normal gait (from University of California, 1953)

The work of Sheffield et al (1956) extended Mann and Inman's work. They investigated 12 muscles of the lower leg and foot in a group of male subjects using both surface and needle electrodes. There is no clear statement in the paper as to the effect of the needle electrodes on the subject's gait style despite the fact that they were also wearing shoes. Despite efforts by the authors to reduce interference to their signals, an inspection of samples of the raw EMG signal presented in the paper show quite noticeable shifts in the EMG baseline; an indication of motion artefacts. Although the authors do not present the details of their calibration procedure they list out their results in terms of maximum and minimum levels. The activity levels are illustrated in figure 3.10 for the gait cycle reported to be 60 steps/minute (although time scales on the graphs suggest times of 2.2s for one gait cycle). The temporal aspects of these graphs are similar to those of Mann and Inman. A further part of the tests involved standing their subjects on one or both feet. With only one foot balance the authors noted large levels of general activity while balance on both forefeet activated all the flexors and abductor hallucis brevis while no activity was seen in the extensors.

In 1966, Battye & Joseph studied eight muscles, including tibialis anterior and soleus, in 14 subjects using a telemetry system. The system was contained in a backpack worn by the subject and allowed the authors to record consistent temporal patterns during the gait cycle. Gray and Basmajian (1968) were still concerned with the support of flattened arches when they studied two groups of 10 subjects. The muscular activity was classified into four levels: nil, slight, moderate and marked. They found moderate or marked levels of activity only necessary for arch support when the subject had flat feet. Their work also contained a detailed analysis of the muscular activity and the corresponding kinematics. Interestingly they note no change associated with exaggerated toe out and toe in. The muscle activity associated with standing was further investigated recently when Rao et al (1987) considered the EMG activity of six lower leg muscles (including four foot extrinsics) in twenty male subjects. Despite a series of rather confusing diagrams their data seem to indicate that the muscle activity is quite stable when standing on the forefoot. There is also some indication that the muscles play a role in the support of the normal arches during the transition from sitting to standing although the biomechanical reasoning for this is poorly presented.

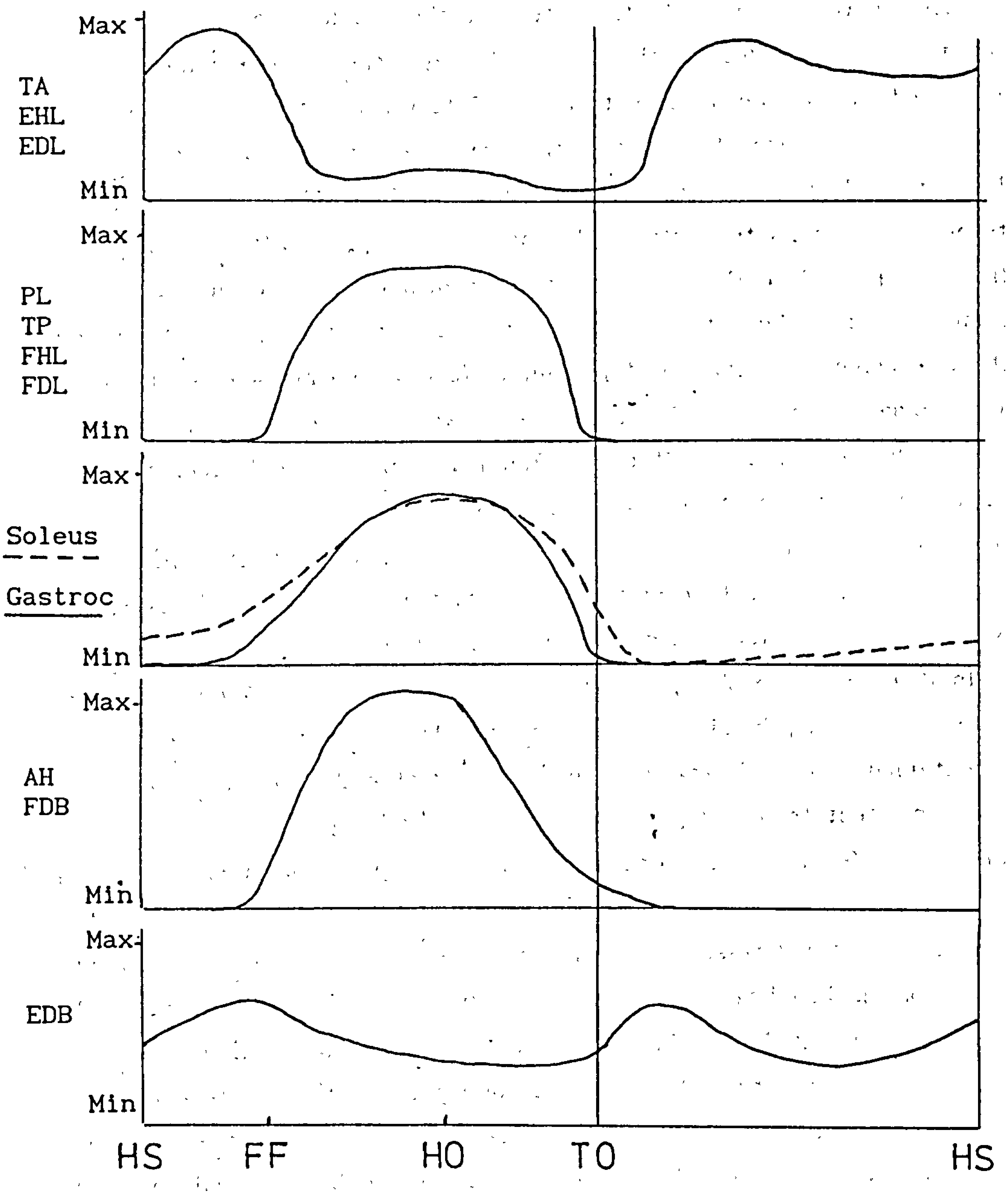


Figure 3.10 The rectified EMG signal of several of the lower leg muscles during walking (60 strides/min).
(from Sheffield et al, 1956).

Notes: TA = tibialis anterior; TP = tibialis posterior;
PL = peroneus longus; Gastroc. = Gastrocnemius;
EHL = extensor hallucis longus; EDL = extensor digitorum longus;
FHL = flexor hallucis longus; FDL = flexor digitorum longus;
EDB = extensor digitorum brevis; FDB = flexor digitorum brevis;
AH = abductor hallucis

Most of these studies have considered only the timing of muscular activity. The amplitude of the EMG signal has only recently come in for scrutiny. The reason for recording an EMG signal is, after all to provide some indication of muscle activity. This activity is nearly always presumed to be related to 'force producing' activity although the EMG record is only of the neurological activity that induces muscular contraction. This presumption is one that is often ignored as investigators use muscle force interchangeably with muscle and neurological activity, a practice soundly condemned by Paul (1971). In 1981, Perry & Bekey published a review on this subject alone, citing over 100 papers reporting on studies in the area. At the outset Perry & Bekey note the review of Bouisset (1973) and treat their work as supplementary, covering work prepared after about 1970.

As with many biological systems, relationships between EMG signal and the resulting force and activity are not conveniently expressed with simple mathematical relationships. As Perry & Bekey (1981) note, only isometric contractions seem to obey a near linear rule and even then the constant of proportionality is dependent on the articulating joint position. With investigation into the muscle/EMG relationship such an important area of biomechanics it is astounding that experimental groups are not more rigorous in their experimental protocol and reporting. Perry & Bekey bemoan the lack of a universally accepted method for collecting and analysing EMG data and with a sense of desperation call for more attention to acquisition and processing detail in future work particularly in the areas of electrodes, sampling rates and signal processing (filtering and amplification). Ideally standards are needed (such as normalizing all EMG records as proposed by Perry & Bekey) and it appears that such standards are slowly being implemented. Two of the most complete studies in this area are those by Hof & van den Berg (1981) and Hatze (1977). In neither case is the modelling simple though it does seem to stand up to experimental results. Wells & Evans (1987) considered the recruitment patterns of one and two joint muscles of the lower leg and noted higher activity levels in the muscles than had been expected. They suggested that this may be due to out of plane motion that their transducers were not recording. Their work also supported the premiss that developed isometric force is proportional to the amplitude of the recorded EMG.

A number of investigations have been undertaken to assess the ability of EMG signals to indicate pathological gait patterns. Guth et al (1979) and Gueth et al (1984) used surface electrodes to assess fourteen muscles including tibialis anterior, peroneus longus and the medial head of gastrocnemius. In their experiments they found significant differences between the EMG signals of normal subjects (Guth et al, 1979) and those with cerebral palsy (Gueth et al, 1984). The work of Arsenault et al (1987) presents a general review of EMG analysis and concludes that although the gait cycle appears automatic in nature there is considerable 'voluntary' control. They do note however that although EMG temporal relationships are reasonably consistent the overall patterns of activity show limited variability for a particular subject but have greater intersubject variability. Patla (1986) notes the changes induced in the EMG activity of six leg muscles due to different inclinations of a treadmill. While there is a noted change in activity, Patla suggests that the limb is controlled as a unit and that the distal muscles are particularly tightly controlled.

For routine clinical assessment some account must be made of the variability of the EMG signal for individual subjects. An initial study to investigate this topic was undertaken by Arsenault et al in 1986. Their proposal was to assess the number of steps that would be required in order to provide a reliable measure of an individual's EMG pattern and concluded that three strides provided an adequate representation of the patterns. Kadaba et al (1989) extended the work of Arsenault et al to look at the reliability of EMG recordings for one subject recorded over several different occasions. They assessed 40 subjects three times in one day and then over three separate days and considered the variations in several biomechanical measurements including the EMG records. The results indicate that parameters associated with gait (cadence, velocity, stride length and to an extent EMG activity) were generally quite repeatable with only moderate increases in variation as a result of testing on several separate occasions. In particular the coefficients of variation (CV, defined in Eqn 3.1 below) for the EMG records of tibialis anterior and gastrocnemius were 39% and 54% respectively for the records made over a single day, and 39% and 52% respectively for records made over separate days.

Wootten et al (1990a&b) studied 35 subjects with surface electrodes over 10 leg muscles. They reported a detailed calculation method to determine the mathematical descriptions of the EMG signal for each muscle

Table 3.5 Coefficients of variation (Eqn 3.2) for data normalised to: 1) 50% maximum voluntary contraction (MVC)
2) Mean EMG/ unit isometric moment of force
3) Peak of within subject ensemble average
4) Mean of within subject ensemble average
(from Yang & Winter, 1984)

Muscle	Unnorm.	1	2	3	4
Rectus Femoris	81	119	167	54	56
Tibialis Ant.	49	52	51	37	32
Soleus	52	82	61	35	34

over the full gait cycle. It is not clear what normalisation method was used but by applying principal component analysis the authors are able to categorise the patterns into three representative traces for each muscle. This, to the author's knowledge, is the only study to attempt this categorisation with any appropriate method and the results are encouraging.

The variation between subjects has been of concern for some time. In an attempt to reduce this variability Yang & Winter (1984) considered four methods of normalisation. The authors tested 5 muscles (including tibialis anterior and soleus) in 11 subjects. The four normalisation methods used were

a) calibrating to activity equivalent to 50% maximum voluntary contraction (%MVC)

b) calibrating to the mean EMG / unit isometric moment of force (μV/Nm)

c) calibrating to the peak of intra subject ensemble average

d) calibrating to the mean of intra subject ensemble average.

In order to assess the relative merits of the different methods Yang and Winter introduce the term Coefficient of Variation previously used by Winter (1983). The coefficient is calculated as :

$$CV_{(\text{per samplestep})} = \frac{\sigma^{\prime}}{\bar{X}} \qquad \text{- Eqn 3.1}$$

$$CV_{(\text{complete trace})} = \frac{\sqrt{\sum_i \sigma^2/\text{num}}}{\sum_i X_i/\text{num}} \quad \text{where } i = 1 \text{ to num (num = number of steps)} \qquad \text{- Eqn 3.2}$$

(Equation 3.2 is presented with the summation sign outside the square root in Yang & Winter (1984) by error in the author's opinion)

The tests assessed the effect of using the four methods of calibration by calculating their overall CV using Eqn 3.2. The results for tibialis anterior and soleus are shown in Table 3.5. As Yang and Winter note, although the last two methods reduce the variability (but not to zero) they contain no indication of the relative level of muscular activity. The average may be equivalent to 10% MVC or 90% MVC and the results would show the same curves. There have only been a (relatively) small number of studies where the level of muscular activity of the muscles of the lower leg is normalised to a defined level. Gray and Basmajian's work (1968) has already been noted with the use of four quantised activity levels. This provides an indication of the extent of muscular activity but is still

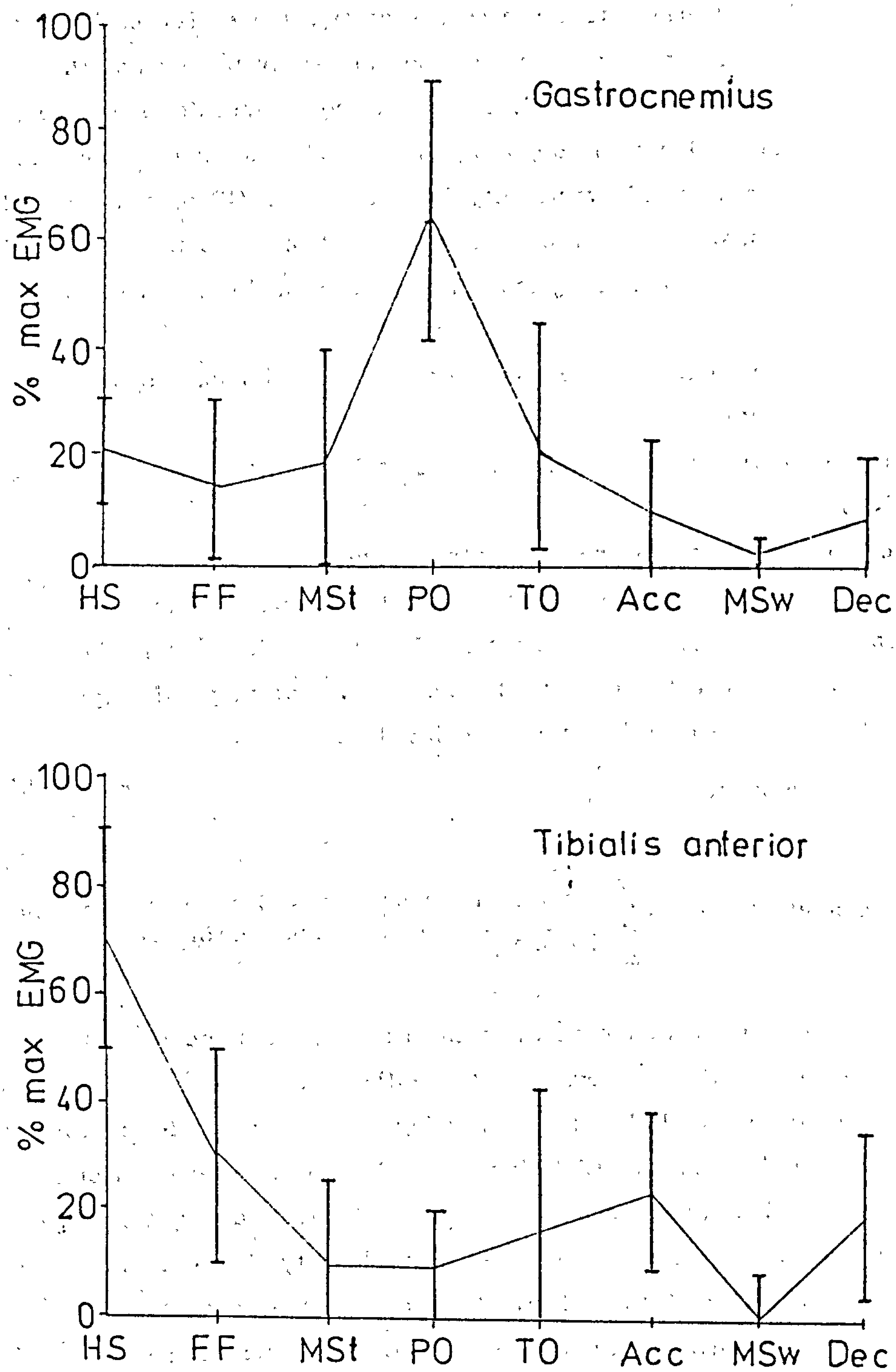


Figure 3.11 The EMG signals of two of the lower leg muscles normalised to Maximum Isometric Voluntary Contraction (MVC). HS, FF & TO have their usual meaning. MSt = Mid Stance; PO = Push Off; Acc = acceleration phase; MSw = Mid Swing; Dec = deceleration phase. (from Dubo et al, 1976)

rather crude. Dubo et al (1976) considered the RMS integrated EMG signal as an indication of the activity of tibialis anterior, the lateral head of gastrocnemius, the medial hamstrings group and vastus lateralis. The levels of EMG activity are all normalised to the peak EMG signal recorded during a maximal isometric contraction (MVC). With 20 subjects the results (Fig 3.11) are a good indication of the muscle activity when wearing soft shoes. The work was further enhanced by Ericson et al in their 1986 study which included the previous four muscles as well as the medial head of gastrocnemius, soleus and five other upper leg muscles. The results for the four lower leg muscles from this study again normalised to MVC are shown in figure 3.12.

In 1987, Winter & Yack studied 16 leg muscles including gastrocnemius (medial and lateral heads), soleus, peroneus longus, extensor digitorum longus, and tibialis anterior. Their results were based on the use of the normalisation to average activity (Eqn 3.1) and are presented in figure 3.13. The studies of Dubo et al (1976), Ericson et al (1986) and Winter & Yack (1987) will be considered in more detail later for comparison with the EMG study performed as part of this project. To the author's knowledge no study has been done into the normalised amplitude of the EMG signal of the intrinsic muscles of the foot (except for the two considered by Gray & Basmajian (1968)).

3.6.3. Muscular Fatigue

The EMG signal has also been applied in recent years to the study of the fatigue properties of the muscles. As we have already noted the muscle force-EMG relationship is dependent on the degree of muscle fatigue and so the fatigue properties of these muscles is an important area of investigation. The research has mainly been conducted in the Scandinavian countries where Lindström et al (1970, 1977), Kadefors et al (1976) and others have been studying how the EMG signal changed as muscles fatigued. In 1984, De Luca published a review of the topic with a very complete description of the changes and their possible causes. The mathematical descriptions of the EMG signal make rather heavy reading but two important points emerge from all the studies: the power density spectrum of the EMG signal shifts toward the lower frequencies and subsequently the amplitude of the EMG tends to increase as the muscle fatigues. The reasons for these changes are covered by De Luca but generally the accepted ones are a change in the conduction velocity of muscle fibres, a change in the regularity of

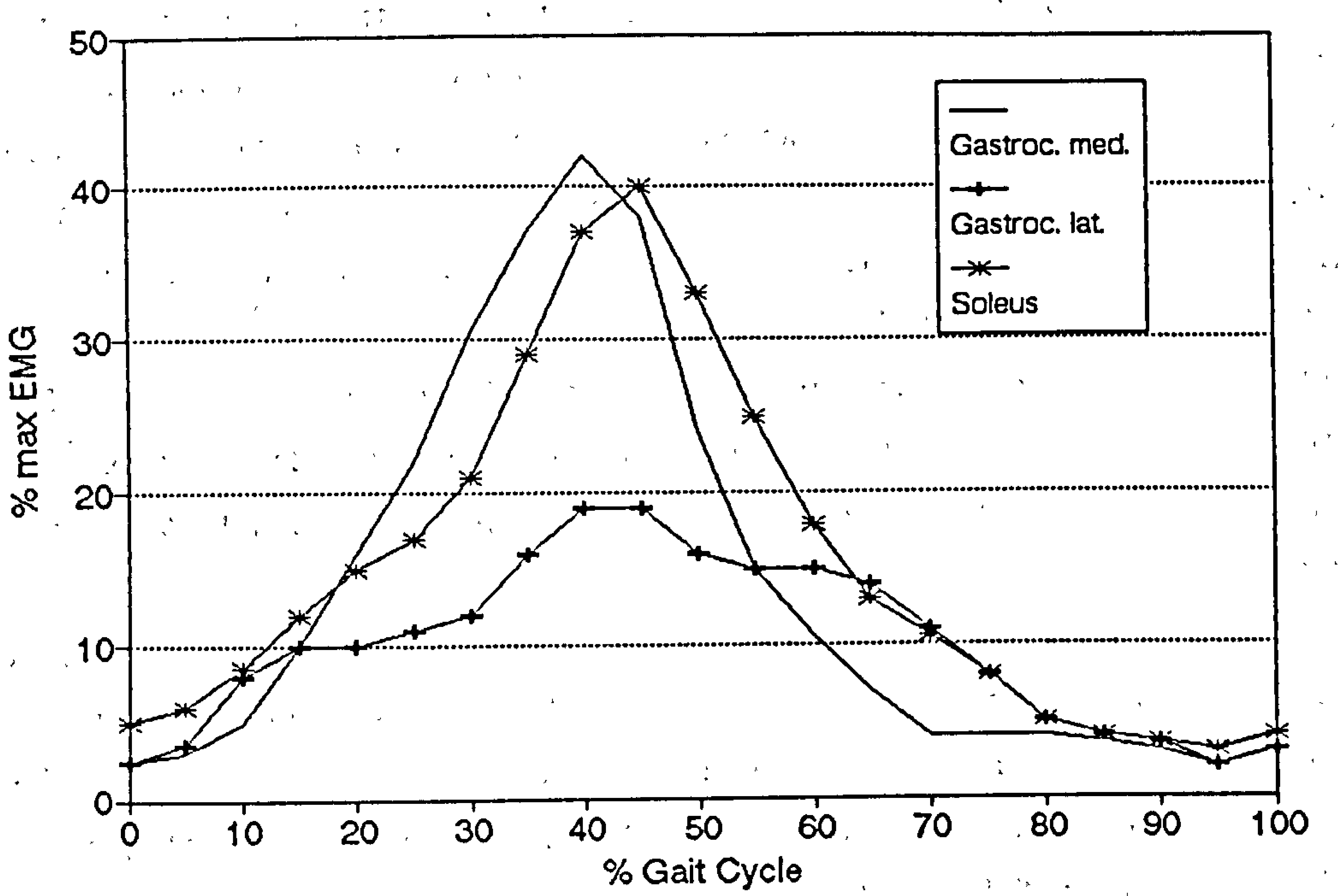


Figure 3.12a: EMG activity of the triceps surae group (Ericson et al, 1986).

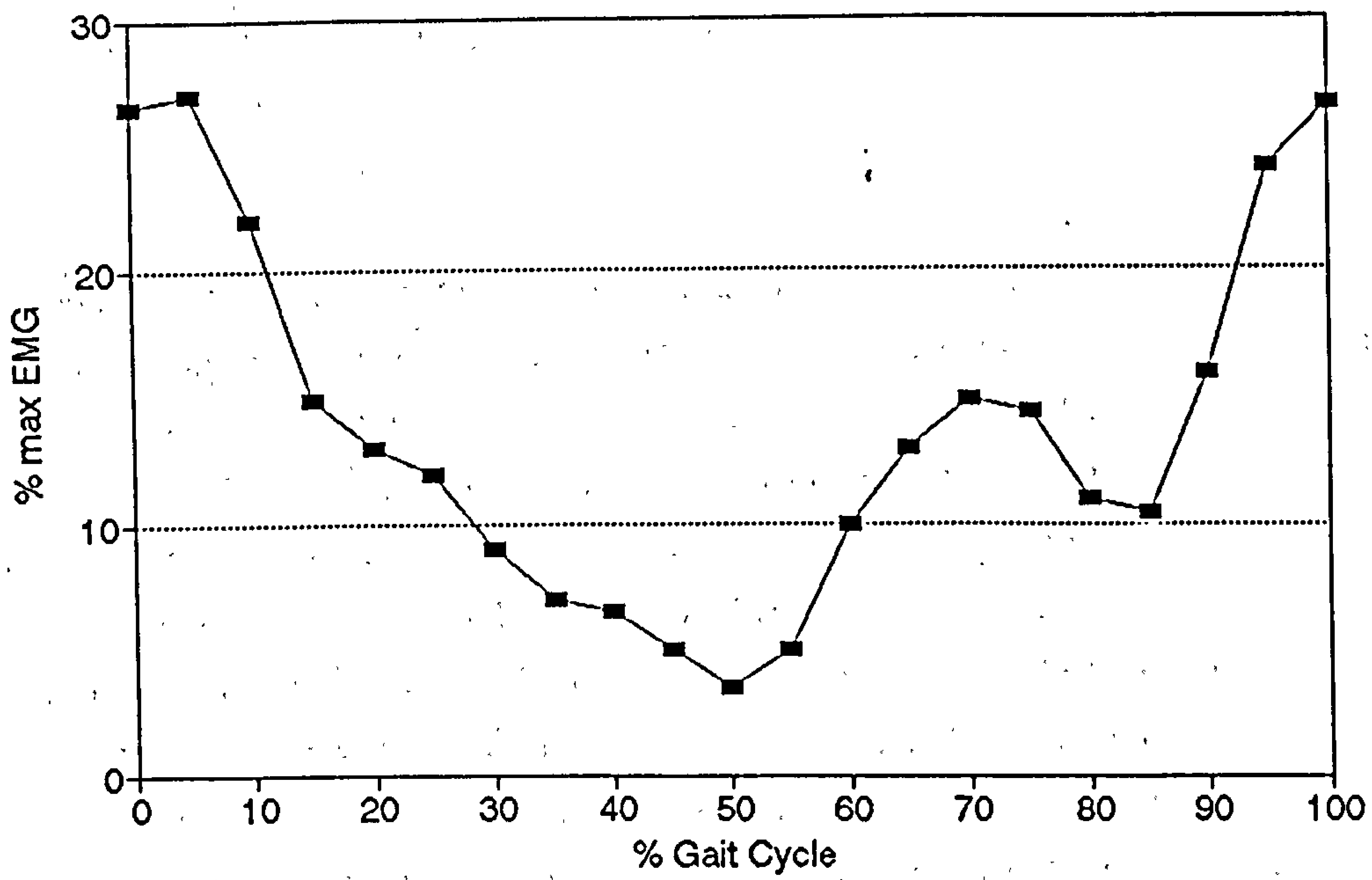


Figure 3.12b: EMG activity of tibialis anterior (Ericson et al, 1986).

the motor unit discharge and to a lesser extent a change in the synchronization of the motor units. These three points are in some ways related as De Luca points out. De Luca cites two well accepted methods of detecting the change in frequency: the mean (average) frequency and the median frequency (where the power density spectrum has equal power above and below the frequency). The mode frequency (the point at which the spectrum peaks) is rejected since the EMG signal is stochastic and does not have a sharply defined peak. Variance of the spectrum would strongly affect the peak detected. In the studies of the frequency changes that occur with fatigue it is important to make allowance for the filtering effects of the electrodes and the processing equipment. Lindström & Magnusson (1977) provided clear details of the effects of electrode placement on the frequency spectrum (Fig 3.14). Figure 3.14a indicates the attenuation that depends on the distance between surface electrodes while figure 3.14b illustrates the importance of locating surface electrodes as close to the muscle as possible to minimise attenuation. As a secondary effect the figure 3.14b also shows that the EMG signals from the larger muscles will have less attenuation at distant electrodes than smaller muscles giving rise to serious 'cross-talk' effects. Muscle cross-talk is an important consideration both for frequency analysis and activity patterns as De Luca & Merletti (1988) clearly show. As part of their paper these two authors indicate methods to reduce cross-talk effects using double differentiation techniques.

To the author's knowledge no studies have been undertaken on the fatigue characteristics of the muscles of the lower leg and foot. Most studies of fatigue in the past have been related to ergonomic studies (Kadefors et al, 1976), though De Luca (1984) cites work in physiotherapy, diagnosis of neuromuscular disorders, assessment of diaphragm fatigue, and the physiological changes associated with fatigue. The simplicity and convenience of the technique of assessment of EMG frequency change makes it ideal for muscular study as an alternative to other invasive techniques (muscle biopsy, blood tests for lactic acid levels, etc.)

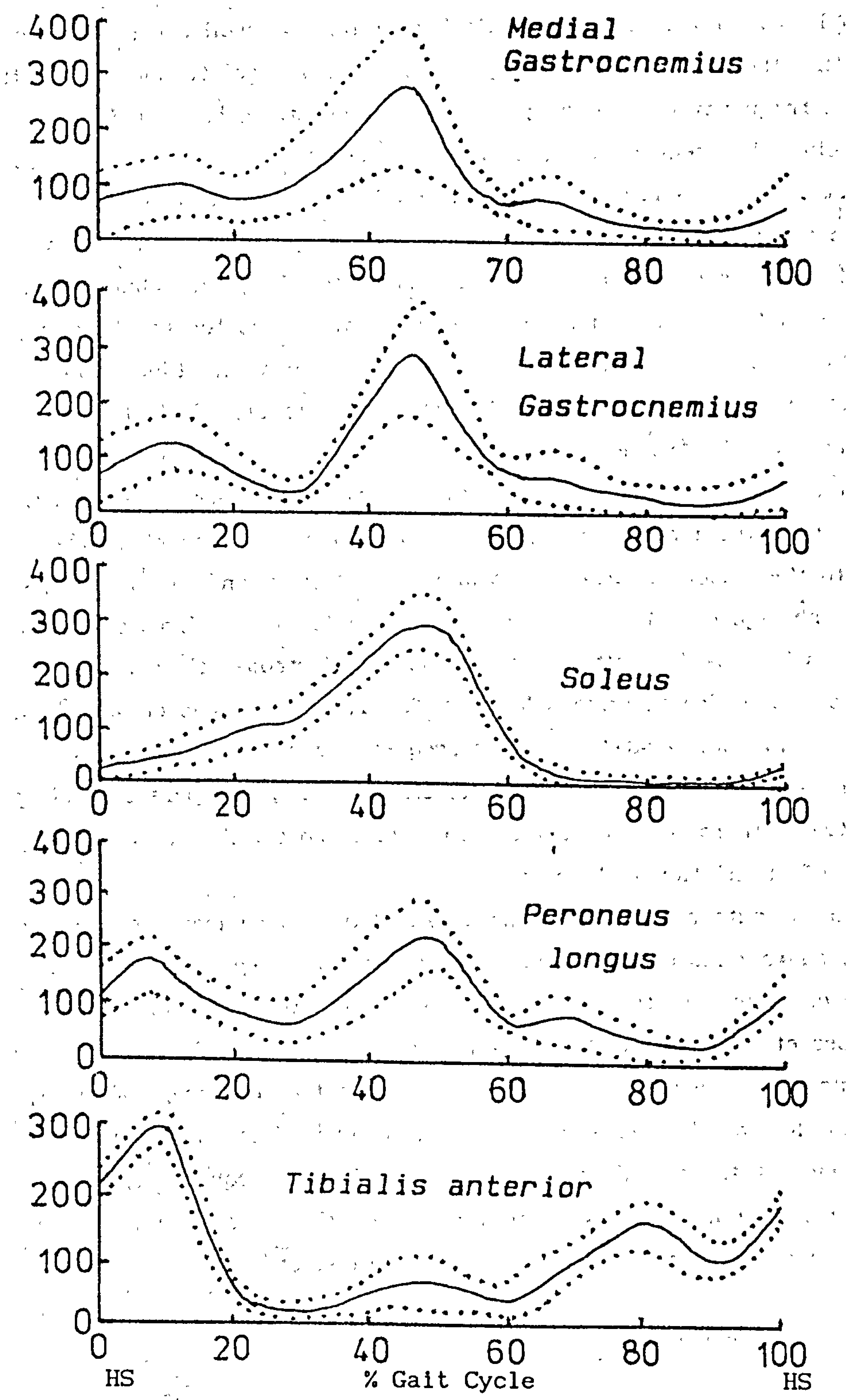


Figure 3.13 The EMG signals of five of the lower leg muscles normalised to the mean EMG signal (method (d) in text). Expressed as % of the mean signal. (n=18) (from Winter & Yack, 1987)

3.7 BIOMECHANICAL REVIEWS AND MODELLING

3.7.1 Reports on the Biomechanics of the Foot

Perhaps as a result of the complexities of the foot, reviews of the biomechanics of its function are quite numerous. They cater for professionals from all disciplines in varying levels of complexity. At the rather basic level many textbooks on biomechanics (Cooper & Glassow, 1976) include a section on foot biomechanics though it is generally as part of the biomechanics of the lower leg with little detail on the foot itself. A few articles try to offer some simple insight into foot biomechanics and often end up producing more confusion than clarification (Foulston, 1987; Katoh et al, 1983). There are examples of excellent introductions and reviews presented for therapists (Riegger, 1988; Oatis, 1988), sports professionals (Mann et al, 1981; Adelaar, 1986; Nigg, 1986; Czerniecki, 1988; Nuber, 1988; Ting et al, 1988) and surgeons and clinicians (Morris, 1977; Mann, 1982; Perry, 1983; Donatelli, 1990). The categories presented above are not discrete and most of these articles would be appropriate for any health professional.

Some of the above articles provide in-depth detail on the biomechanics of the foot. R Mann, who has been closely involved with foot research for several years, targeted a surgical audience with his paper in 1982. He considered the articulations in detail particularly of the hindfoot and then proceeded to discuss the implications of biomechanics on surgical procedures. M Rodgers appears to have had a much wider audience in mind for her 1988 article. This is perhaps one of the most comprehensive reviews of the field of foot function including discussion of kinematics and kinetics (including EMG work, pressure studies etc). The article then considers the changes that occur during running. Other works dealing with the biomechanics of the foot are the books of Nigg (1986), providing a wealth of detail in relation to running and the running shoe for a wide audience, Gould (1988) and Donatelli (1990), focused clearly at the medical audience, and those dealing with foot deformity and pathomechanics (Klenerman, 1982). Some books are now being published that form comprehensive references on the foot (Helal & Wilson, 1988) but in most cases are outdated by the time of publication due to continuing research.

As an alternative to these comprehensive works some authors have investigated one part of the foot in detail. Not surprisingly the areas considered are those most closely associated with pathology and injury. The hindfoot has been presented by Perry (1983) and many others who

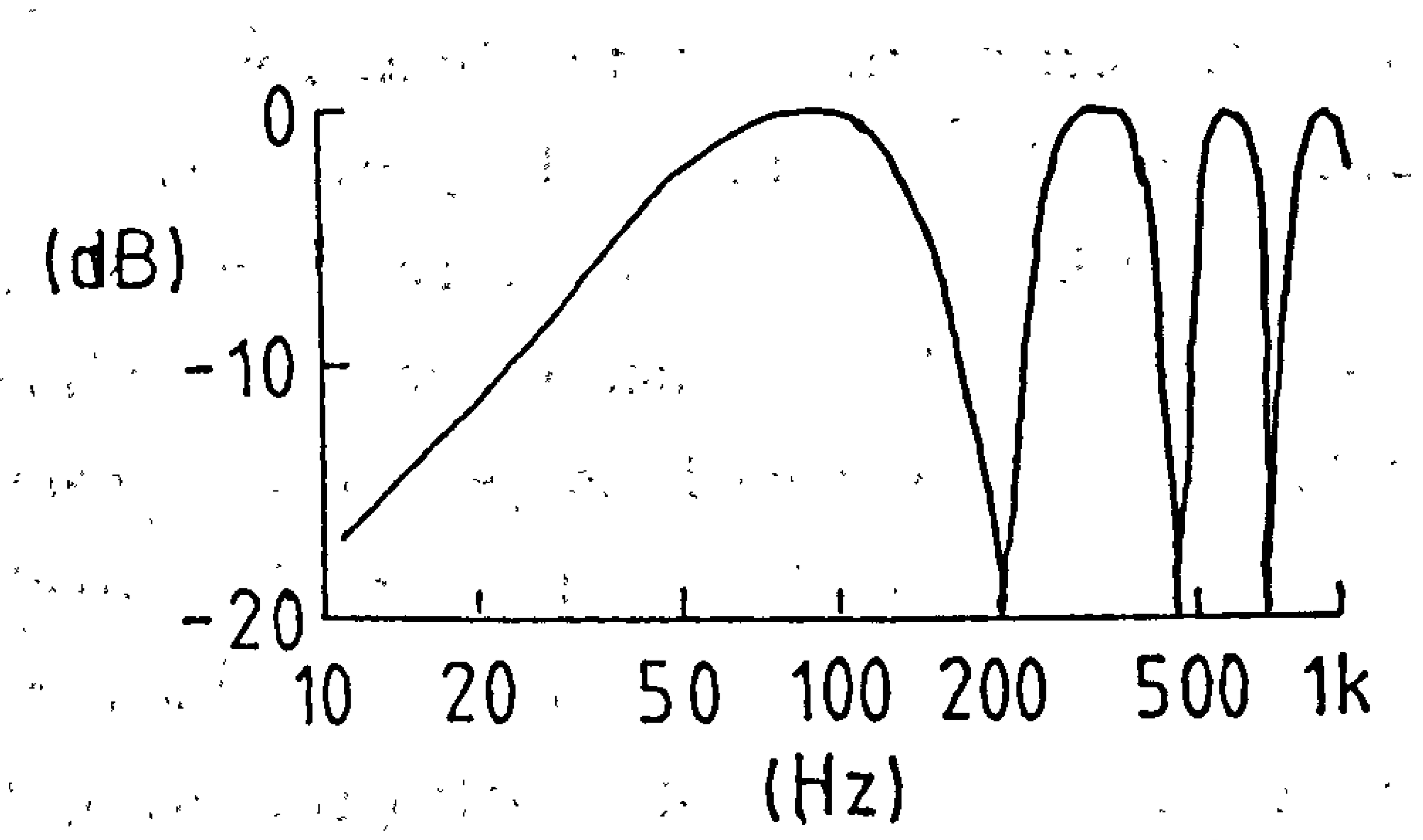


Figure 3.14a Plot of the surface electrode filter function $|F_{bipol}(j\omega)|^2$. Parameters: electrode separation (2d) = 20mm and conduction velocity (v) = 4 m/s. (from Lindström & Magnusson, 1977)

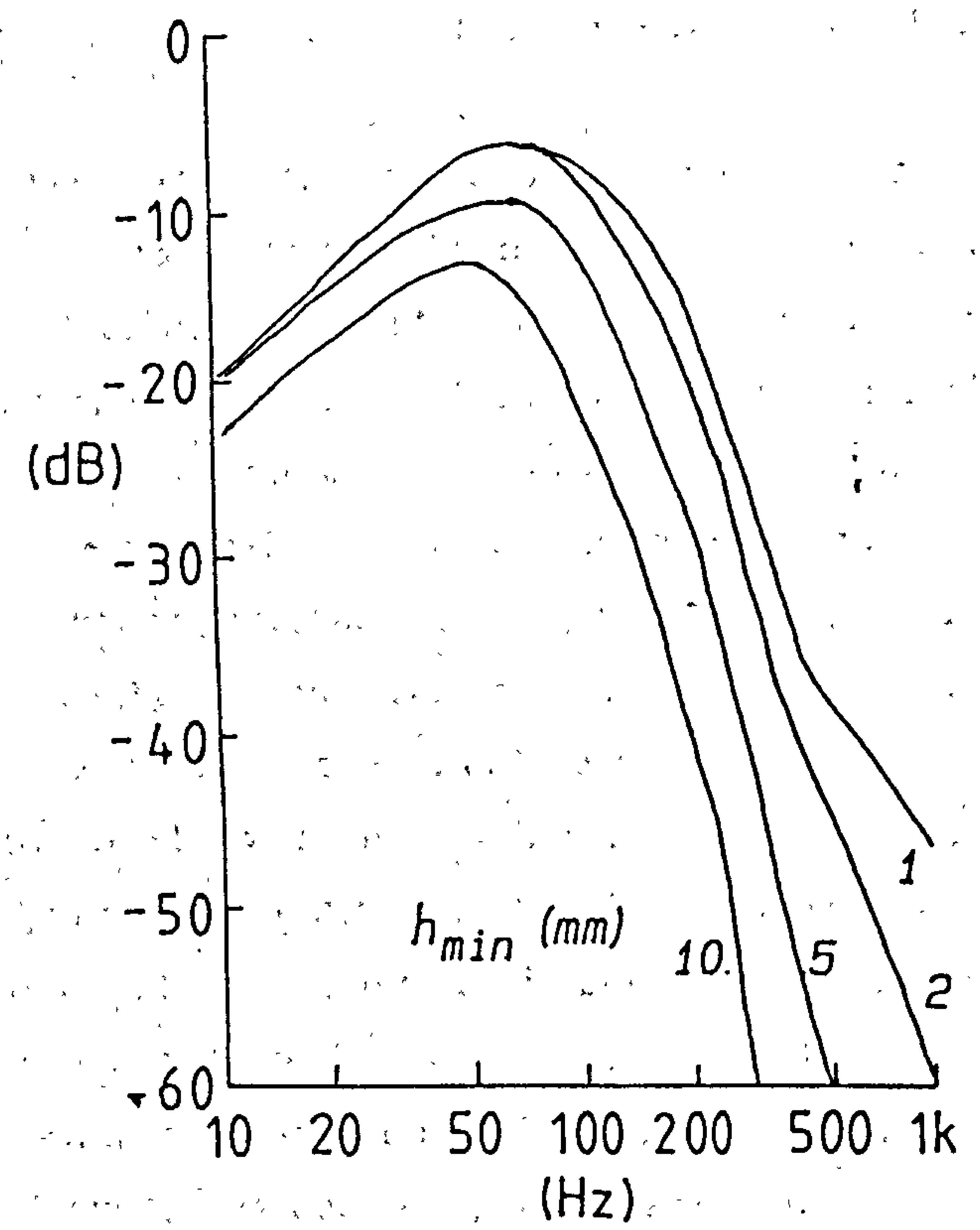


Figure 3.14b Attenuation of the EMG signal for different values of the distance to the nearest active muscle fibre (h_{min}).
Parameters: 2d = 30mm; h_{max} = 15mm; fibre radius = 20μm; number of fibres per unit = 200.
(from Lindström & Magnusson, 1977)

investigated sporting (Ting et al ,1988; Evans, 1990) and dancing related injuries (Hamilton, 1988; Hardaker, 1989). At the front of the foot the metatarsophalangeal joint and the articulations associated with the first two toes are dealt with by Bojsen-Møller & Lamoreux (1979) in relation to walking, Shereff et al (1986) and Mann (1989) dealing with normal and pathomechanics and Mann et al (1988) considering the effects of toe amputation. It is interesting to note from Mann et al (1982), that the loss of the hallux produces a decrease in medial stability of the foot and an inability of the first metatarsal to actively support load. The ankle complex is itself an articulation of extensive research with works ranging from the models of Procter and Paul (1982), through the anatomical descriptions of Brodsky & Khalil (1986) to the clinical experience of Carmines et al (1988).

3.7.2 Modelling of the Foot

The foot has been modelled in several different ways (Jacob, 1989). Early work proposed a vaulted truss structure though this was altered to consider the possibility that the metatarsal bones were supporting some of the foot load in bending. The debate as to which of these propositions was correct seemed to parallel that associated with the role of the muscles in the plantar support of the foot. Having proposed his model of the plantar aponeurosis, Hicks (1954) was clear that the most likely structure was the truss with the aponeurosis as the lower tension element. To determine if Hicks had been correct, research continued along both the experimental and mathematical path.

In 1982, Simkin considered the foot mathematically based on a simplified structure without the muscular action. He assessed seven cadaveric feet to determine nodal points in three dimensions and also tested each joint of the foot to determine the flexibility matrix for use in the modelling calculations. Using an incremental loading pattern he was able to determine the forces and displacements of the components in steps up to 500N. It is rather surprising that having noted the quiescence of the muscles Simkin includes a substantial tension on tendo calcaneus in an attempt to ensure the centre of pressure corresponded to static in vivo records. With this loading pattern the support was found to be 49% at the heel and 15,12.5,12.5,9 and 2.5% on the metatarsal heads from medial to lateral side respectively. Significant changes were noted in the forefoot load distribution that were dependent on the position of the applied load

Table 3.6 The tensions in the plantar aponeurosis and long plantar ligaments expressed as a percentage of the corresponding vertical force under the metatarsal heads. (Based on Simkin, 1982).

Metatarsal head	1	2	3	4	5
Plantar aponeurosis	146%	128%	128%	124%	212%
Long plantar ligament	-	68%	43%	54%	-

on the talus. Simkin presents several tables detailing changes in the structure when loaded to 360N (about 50% BW). The ligamentary loading is shown to be linear with the majority of the moments from forefoot load transferred rearwards by the plantar aponeurosis (260N tension at 50% BW) and the long plantar ligament (68N tension at 50% BW) while the vertical force (shear) is mostly supported by the midfoot joints. Table 3.6 contains the details of the tensions found in the plantar aponeurosis and long plantar ligament in relation to the support forces at the metatarsal heads. These two ligaments showed reasonably linear loading patterns with up to 500N force applied to the foot. As a notable simplification Simkin considers the bones to be rigid and no calculations are made of their stress-strain behaviour. Although his structure can be considered a very detailed static model it possessed no facility to determine the question of the muscular role.

A similar model was suggested by Salathé et al (1986) that again considered the foot as an indeterminate structure that could be progressively deformed. The model included the plantar aponeurosis and the joints but ignored muscular effects. A subsequent preliminary paper (Salathé et al, 1990) indicated further investigations using this model to assess the dynamic behaviour of the foot in absorbing shock.

Jacob (1989) was obliged to consider the muscular role in his thesis on the forefoot. His work includes allowances for the muscles and is based on a cadaveric study of the bending stresses in the metatarsal bones of an amputated specimen. In a novel arrangement to load the foot from below through the heel and forefoot pads, Jacob is able to determine the load support characteristics. Unfortunately in an effort to clarify the view of the foot all the skin and the musculature is removed. Jacob makes note that all care is taken to avoid damage to the ligaments but no allowance is made for the change in the lines of action of the ligaments as a result of his dissection. Unfortunately there is no definitive test with the plantar aponeurosis. Although this is given as a major reason for conducting the tests and all care is taken to restore the aponeurosis after fitting the three strain gauges to the metatarsal bones, Jacob only tests the foot with the toes in the flat position. Even in this position Jacob's tests show that a proportion of the load is supported by the structure. It is clear that Hicks proposed his windlass effect as operational in the latter phase of the gait cycle when the toes are dorsiflexed. Jacob does mention that under load the toes allowed no free dorsiflexion at all. He also notes that loss of the aponeurosis does increase the bending strains of the first and second metatarsal by 24% and 12% respectively supporting an indication that apparently the *first* metatarsal or its proximal joint (rather than the second as Jacob suggests) is stiffer. By sectioning the intermetatarsal ligaments, the resulting load distribution of 1.3:2:1:4 is reasonable though probably somewhat artificial as Simkin (1982) had shown the ligament's significant role in load support across the forefoot.

As a result of his tests Jacob concludes that the metatarsals carry the majority of their loads in bending (and notes that fracture would occur at about 3000N total load) and subsequently dismisses the aponeurosis from his future calculations. The measurements made are the most comprehensive to date on the loading distribution in the forefoot. Unfortunately Jacob's test equipment limited his ability to test the feet in plantarflexed positions. Jacob's own kinetic tests show the forefoot is loaded during the later part of the gait cycle and further detail of the foot behaviour during this stage would complete this already detailed, initial study.

Ottevanger et al (1989) presented the details of an experimental setup to assess the loading of the foot in vitro and the motion of the individual bones. A specialised testing frame was developed and used to load the foot through the tibia and fibula while supports under and beside the plantar aspects of the foot indicated the load distribution. Some important simplifications were developed into the device:

- the heel was restrained in a cup support
- the foot was only supported beneath the metatarsal heads, the base of the fifth metatarsal head and the heel
- all metatarsal friction forces were supported at the fifth metatarsal head
- inversion/eversion motion was produced by internally/externally rotating the tibia and fibula bones.

The specimens used are said to be preserved in toto citing a reference to Benink (1985), who used both formalin preserved and fresh specimens. It is difficult to envisage the authors performing their tests on preserved as opposed to fresh specimens due to the significant changes the preserving agents have on the soft tissue (crosslinking of the collagen). Only preliminary results are presented indicating the resulting twisting moment developed in the tibia and fibula bones as a result of foot in/eversion and opposition by several of the tensioned lower leg muscles.

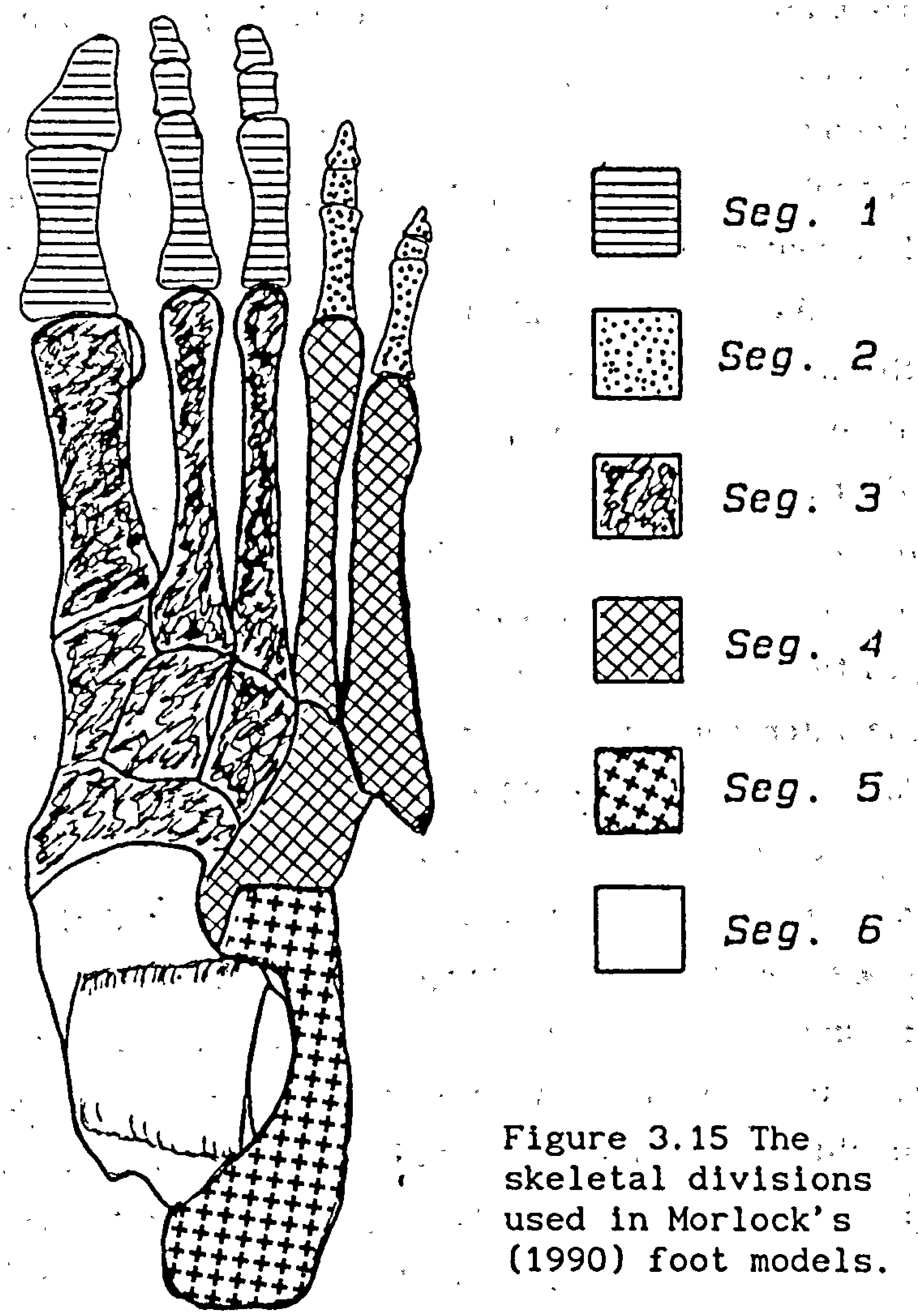


Figure 3.15 The skeletal divisions used in Morlock's (1990) foot models.

Table 3.7 The joint numbers and their corresponding degrees of freedom as used in the four anatomical models (classified by their degrees of freedom; DOF) assessed by Morlock (1990)

No	Joint desciption	Connecting segments	Degrees of freedom 18 DOF	12 DOF	6 DOF	1 DOF
1	Metatarso-phalangeal	1->3	3	1	1	0
2	Metatarso-phalangeal	2->4	3	1	1	0
3	Midtarsal- Navicular & talus	3->6	3	3	1	0
4	Midtarsal- Cuboid & calcaneus	4->5	3	3	1	0
5	Subtalar	5->6	3	3	1	0
6	Tibio-talar	6->leg	3	1	1	1

Before the end of 1989 another study, produced under the supervision of Nigg, was published (Morlock, 1989 & 1990) on another model (or series of models) of the foot. The work was to assess the likely forces and motions about the ankle joint during sporting manoeuvres. Morlock considered the foot as a one, two and six segment model (Fig 3.15) with selected ligaments and muscles active. In order to test his model Morlock looked at the behaviour of the foot in a 'side step' motion. It is not particularly surprising that in this lateral bracing motion the two segment model with all its simplifications was found to be the closest to experimental data. It is the author's opinion that the greater degrees of freedom would be necessary when the components of the foot act more independently during normal walking and running, particularly when this occurs on uneven surfaces.

3.8 SUMMARY

This review of the previous work associated with the foot provides an overview of the present state of biomechanical knowledge of normal foot function. There are other aspects of study associated with the foot, such as medical conditions and surgical procedures, that have been excluded from this review although they constitute an extensive proportion of the literature associated with the foot. Several journals are now available that deal specifically with foot research and the reader is referred to these as well as the general surgical journals for detail of those areas not discussed in this review.

It should be clear that in many aspects of foot research, a lack of quantitative data results in a wide range of opinion regarding the various theories used to describe its behaviour. The role of the ligaments and muscles in support of the foot arches is a typical example. Until the EMG studies of Basmajian and Stecko (1963), opinions varied widely from the suggestion that no muscular action was required for static arch support to highly active intrinsic muscles.

There are also the difficulties encountered when investigating a structure that is very complex, quite small but also subjected to high body forces. Under these circumstances the wide variation seen in several of the reported results is not particularly surprising. There remains much still to discover about the function of the normal foot.

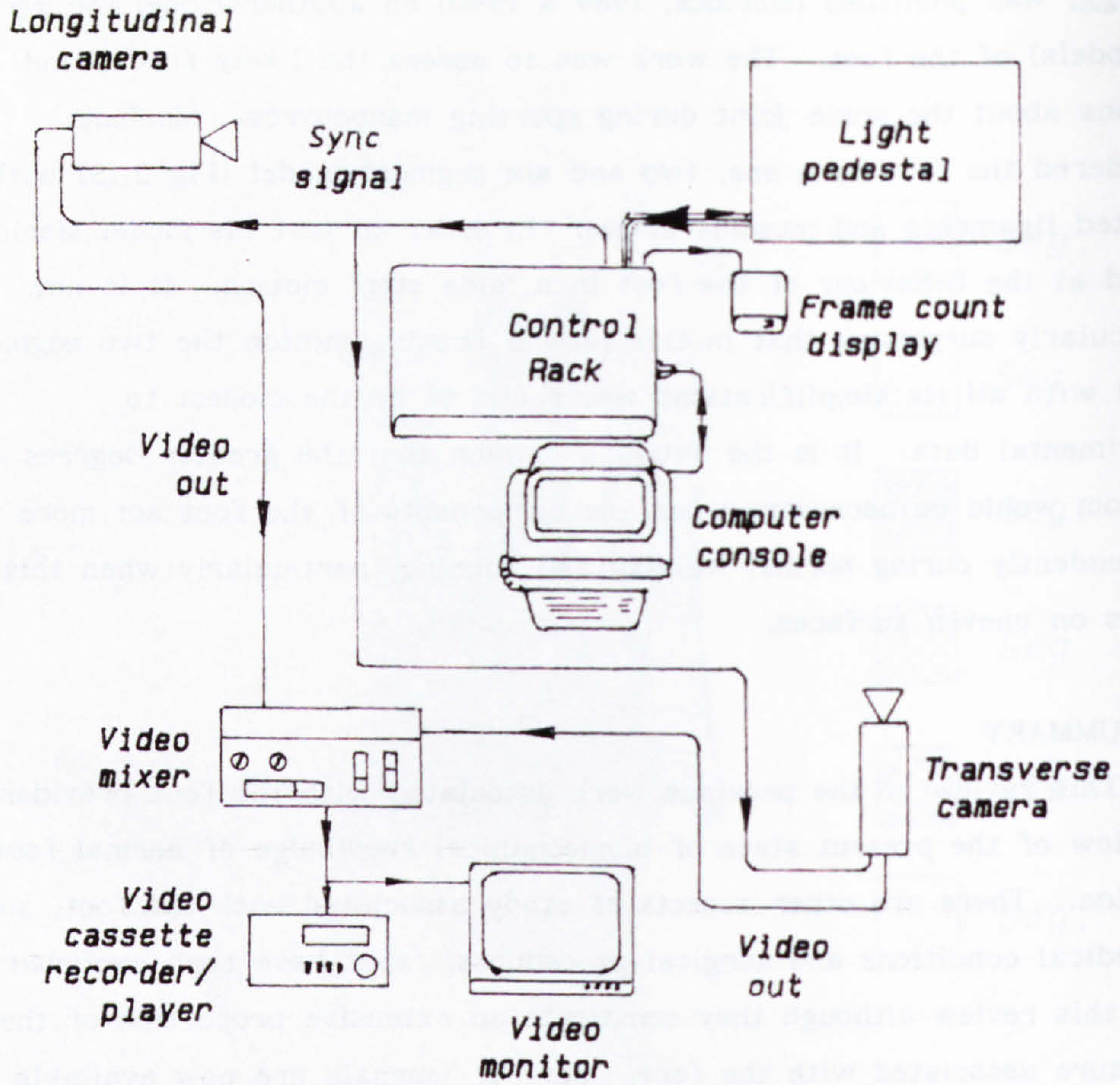
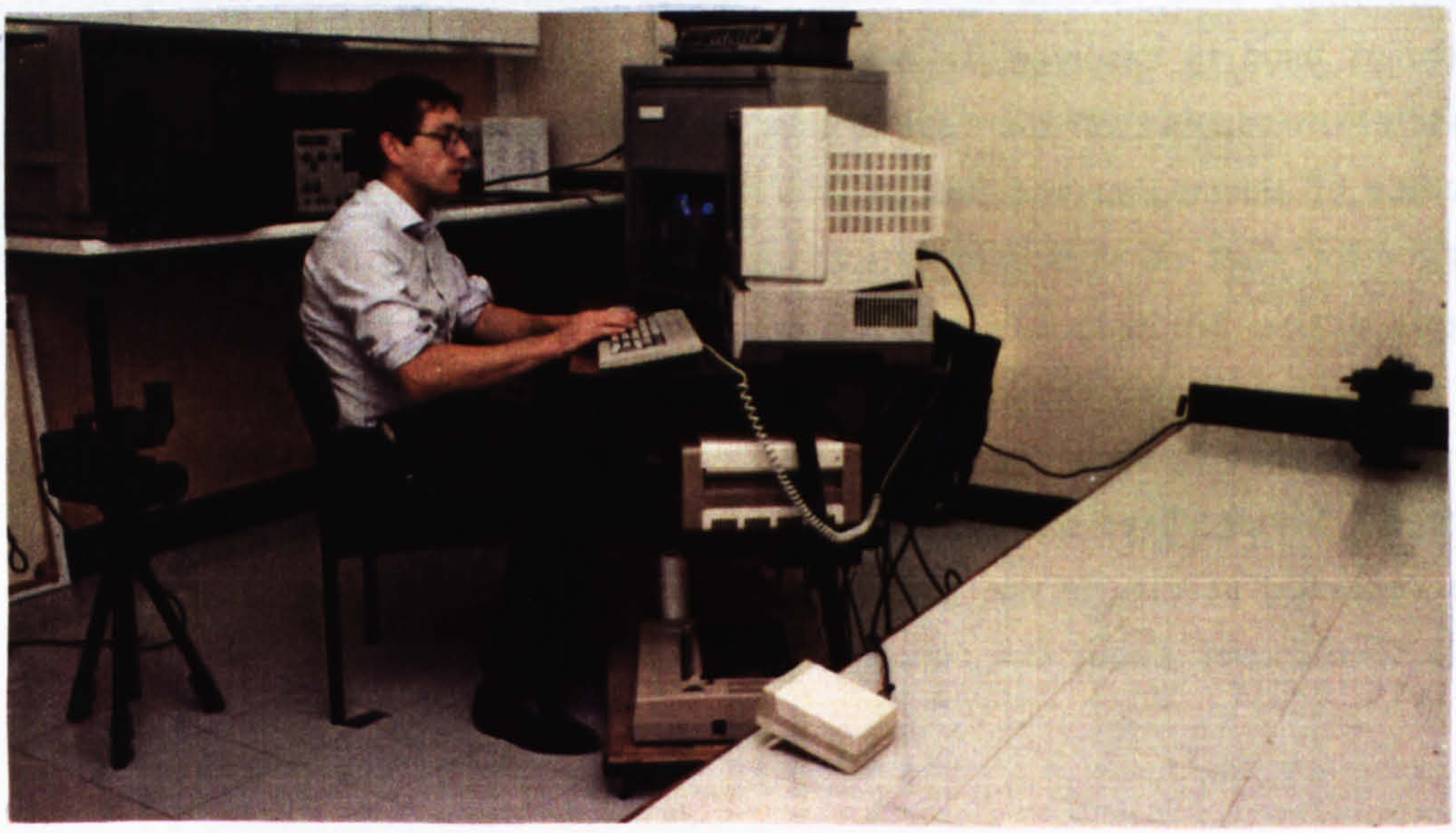


Figure 4.1 Schematic diagram of the pedobarograph/video connections (top) and illustrated (below) looking from the light box. Items shown are (l-r): lateral camera, video monitor, video cassette recorder/player (with video mixer on top behind technician), computer console (with control rack below) and longitudinal camera. Frame counter is in the foreground



CHAPTER 4 STUDIES OF THE NORMAL FOOT IN VIVO

4.1 INTRODUCTION

This chapter presents details of the study of the discrete forces under the foot and the angular changes of the articulations that occur simultaneously during a steady walking pattern. Special adaptations were made with a dynamic pedobarograph and video analysis to provide details at 0.04s intervals during the stance phase of gait of sixty one subjects.

4.2 METHOD

The Sheffield Dynamic Pedobarograph has already been introduced in the literature review. This device offers a high degree of precision and has featured prominently in past research on the pressures under the foot during dynamic activity. A complete version of the pedobarograph was available for use in the Gartnavel General Hospital in Glasgow. The system was operational and supported by the staff from Sheffield who had been responsible for its development.

Although several investigators (including the Sheffield staff) had already used the pedobarograph to study a wide group of healthy subjects there was no previous work done to link the dynamic force records with the kinematic motion of the foot. A video recording system was available for use that had previously been used to take qualitative images from the front and side positions of the foot while the subject walked over the pedobarograph. In order to make further use of this system it was necessary to synchronise it to the pedobarograph and utilize markers on the feet for anatomical identification.

4.2.1 Synchronization of the Systems.

The video system used in this test consisted of :-

2 CCD colour cameras (Panasonic WVP-F10E) fitted with appropriate lenses (WV-KT100 & 200) for a clear view and placed in line with and to the side of the pedobarographic plate.

1 Video mixer (Panasonic WJ-MX10)

1 Video monitor (Sony PVM-2010QM)

1 Professional U-Matic video recorder/player (Sony VO-5800PS).

This system was cabled as shown in fig 4.1 to allow orthogonal images to be recorded simultaneously as a subject passed over the pedobarograph

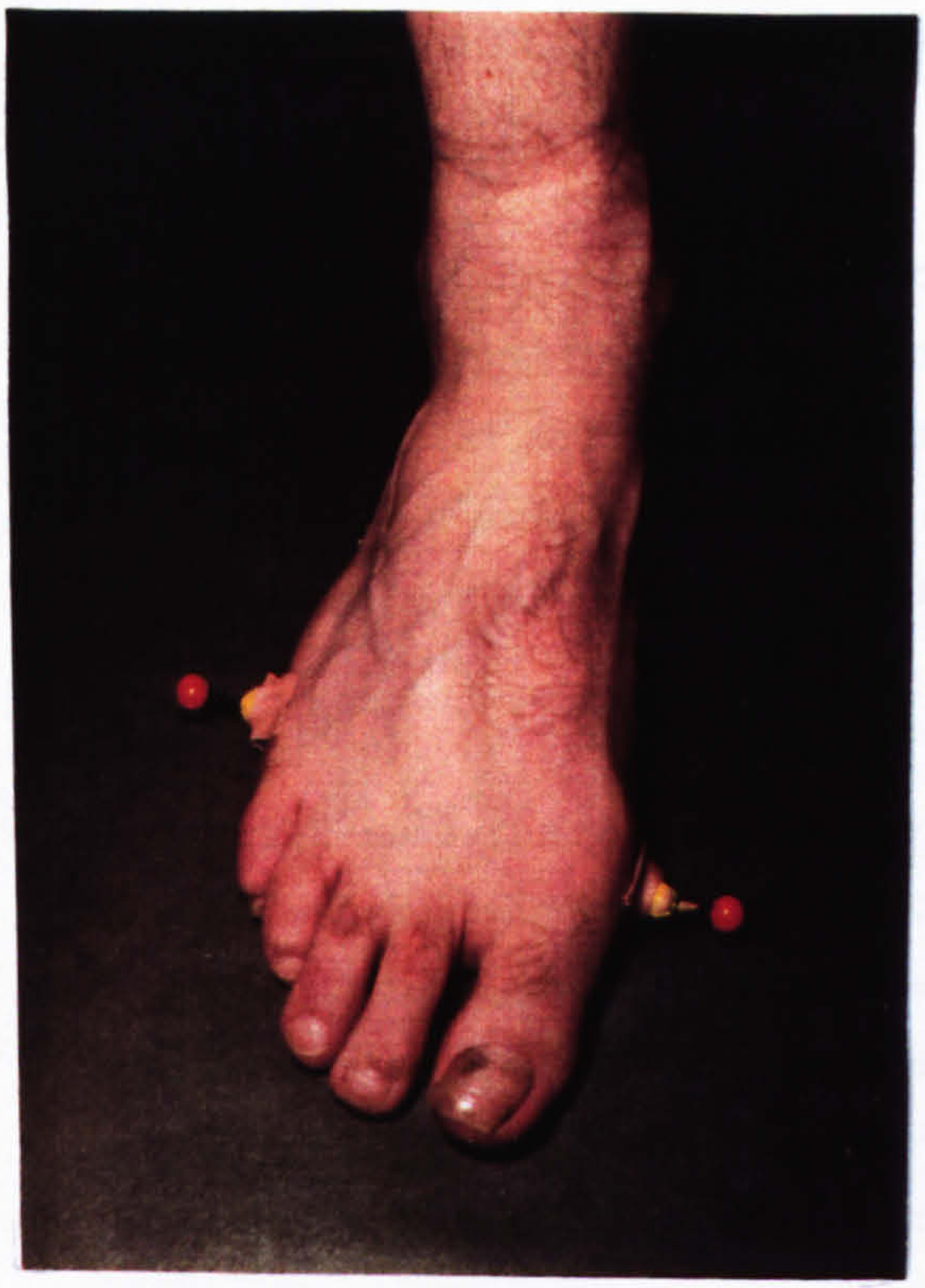
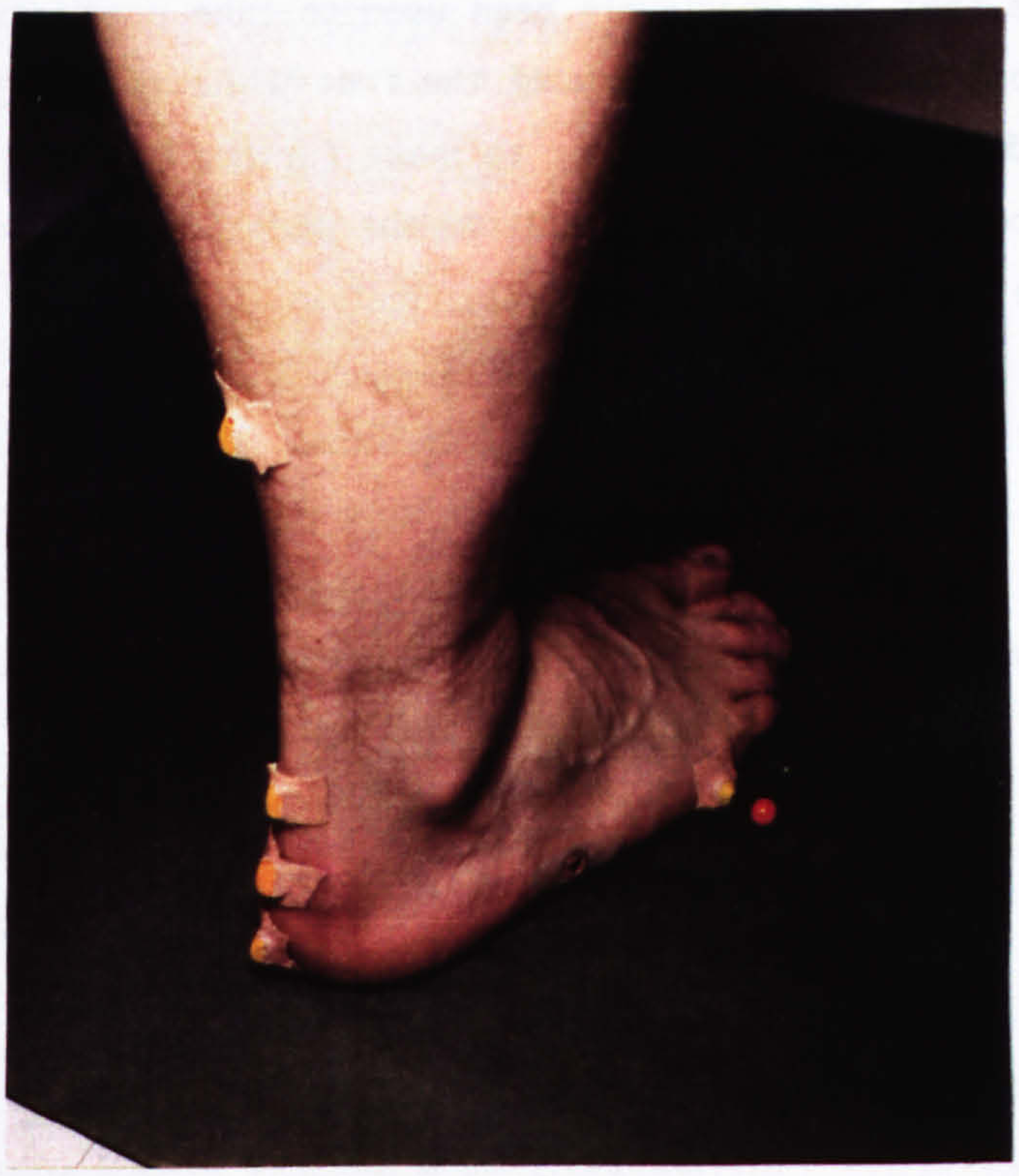


Figure 4.2 Placement of the markers on the leg of a subject: anterior view (above left) and lateral rear view (below right)



light box in the middle of a 6m walkway.

These cameras were adapted (Panasonic WV-AD36E Gen-loc adapter) to accept an appropriate synchronization pulse to control the camera shutters together at 50Hz. The pedobarograph control system required a very stable video signal from the pedestal camera it used to image the pressure plate. To achieve this, circuitry in the computer control system provided the synchronization pulses for the pedestal camera. With some minor adjustment the composite video signal from the pedestal camera (controlled by the pedobarograph system) was used as the synchronization signal for the two CCD video cameras.

When the appropriate connections were made the computer controlled the pedestal camera which in turn controlled the two kinematic imaging cameras. The pedobarograph operated at 25Hz while the two CCD cameras operated at exactly twice this rate (50 Hz).

A final requirement was to provide an indication to the video cameras of the beginning of a pedobarograph record in real time. A digital counter was connected to a control point in the control computer that provided a signal pulse for every frame recorded after the first. This simple incrementing counter was placed in the field of view of one of the cameras and indicated the exact progress of the pedobarograph record.

4.2.2 Markers

As the foot is not a solid structure but is capable of various deflections it is important to define several points of the bony skeleton. The marker positions used in these studies were similar to those that have been used before (Nigg, 1986). Positions used were :-

1. Midline upper extent of tendo calcaneus
2. Midline lower extent of tendo calcaneus
3. Midline point of insertion of tendo calcaneus
4. Midline lower posterior aspect of calcaneus
5. Midline lateral aspect of the leg
6. The malleolus nearest the transverse camera
7. Base of the 5th metatarsal
8. Medial side of the head of the 1st metatarsal
9. Lateral side of the head of the 5th metatarsal
10. Medial side of the first inter-phalangeal joint.

These points were marked with fluorescent triangular markers and in the case of the metatarsal heads a special two beaded marker. Figure 4.2

Table 4.1 The details of the subjects of the in vivo foot tests.

Subj. number	Age	Sex	Mass (kg)	Relevant history[1]	Left foot Away[2] P[3]	V	Left foot Toward P	V	Right foot Away P	V	Right foot Toward P	V
1	21	F	61	Disloc. L hip	x	–[4]	x	x	x	–	x	x
2	20	F	62		–	x	x	x	x	–	x	x
3	37	F	76	Sprain, >18yr	x	–	x	x	x	x	x	x
4	10	M	40	Shin splints	x	x	x	x	x	x	x	x
5	30	F	51		–	x	x	x	x	x	x	x
6	15	F	49		x	–	–	x	–	–	x	x
7	15	F	54		x	–	x	x	x	x	x	x
8	37	M	86		–	x	–	x	–	x	–	x
9	16	F	44		–	x	x	x	x	–	–	x
10	15	F	47	Recur. R sprain	x	x	x	x	x	x	x	x
11	15	F	49		x	x	–	x	x	x	x	x
12	61	F	56	R THR, 5 mths	–	x	x	x	x	x	x	x
13	27	M	85	R knee pain	x	x	–	x	–	x	x	x
14	23	F	51		x	x	x	x	x	x	x	x
15	18	F	59		x	x	x	x	x	x	x	x
16	56	M	89		–	x	x	x	x	x	–	x
17	29	M	70		x	x	x	x	x	x	x	x
18	49	M	70	Plantar corns	–	x	–	x	–	x	x	x
19	22	F	55		x	x	x	x	x	x	x	x
20	21	F	58		–	x	x	x	x	x	x	x
21	27	F	60		–	x	x	x	x	x	x	x
22	62	F	57		x	x	x	x	x	x	x	x
23	37	M	92		x	x	–	x	–	x	x	x
24	44	F	64	Leg oedema	x	x	x	x	x	x	x	x
25	44	M	73		–	x	x	x	x	x	x	x
26	56	F	69		x	x	x	x	x	x	–	x
27	21	F	58		x	x	x	x	x	x	x	x
28	58	F	51		x	x	x	x	x	x	x	x
29	52	M	80		x	x	x	x	x	x	x	–
30	20	F	60		x	–	x	–	x	x	x	x
31	35	F	60		x	x	x	x	x	x	x	x
32	44	M	75		x	x	x	x	x	x	x	x
33	23	F	60		x	x	x	x	x	x	x	x
34	45	F	99		–	x	x	x	–	x	–	x
35	41	M	76		x	x	x	x	x	x	x	x
36	28	F	61		x	x	x	x	x	x	x	x
37	32	F	60		x	–	x	x	x	x	–	x
38	44	M	80		x	x	x	x	x	x	–	–
39	42	F	53		x	x	–	x	–	x	x	x
40	35	F	62		–	–	x	x	x	x	x	x
41	53	F	68		x	x	x	x	–	x	x	x
42	24	F	62		x	x	–	x	x	x	x	x
43	32	M	81		x	x	x	x	x	x	x	x
44	34	M	70		x	–	x	x	x	–	x	x
45	23	F	60		–	x	x	x	x	x	x	x
46	16	M	65		x	x	x	x	x	x	x	x
47	42	M	78		x	x	x	x	–	x	x	–
48	36	F	57		x	x	x	x	x	x	x	x
49	59	M	73	R Hammertoe	x	x	x	x	x	x	–	x
50	25	M	67		x	x	x	x	x	x	x	x
51	31	M	70		x	x	x	x	x	x	x	x
52	19	F	60		x	x	x	x	x	x	x	x
53	50	F	65		x	x	x	x	x	x	x	x
54	20	F	63		x	x	x	x	x	x	x	x
55	25	F	55		x	x	x	x	x	x	x	x
56	29	F	67		x	x	x	x	x	x	x	x
57	24	M	85		x	x	x	x	x	x	x	x
58	24	F	66	R Ant cruc. torn	x	x	x	x	x	x	x	x
59	29	M	73		x	–	x	x	x	–	x	x
60	26	F	61		x	x	x	x	x	x	x	x
61	27	M	70		x	x	x	x	x	x	x	x
				Totals	48	52	54	60	52	55	53	58

Notes: [1] History of orthopaedic or neurological disturbances to the lower legs.
[2] Away and Toward are related to the longitudinal camera
[3] P = Pedobarographic force record; V = Video analysis
[4] x = the records were used in the analyses

illustrates the positions of the markers on a subject.

4.2.3 Subjects

Sixty one healthy subjects who showed no orthopaedic or neurological disorder participated in the study. They came from a general population group associated with one of the main orthopaedic hospitals of the city and colleagues from one of the university's postgraduate departments. The subjects' details are tabulated in Table 4.1. All subjects gave their consent for the tests after a general explanation of the procedure and the aims.

4.2.4 Procedure

Each subject was asked to remove their shoes and socks, where worn, and to lie prone on an examination couch. A general biomechanical assessment was made of each foot and then the markers were applied using double sided tape to the appropriate landmarks. If the subject was wearing trousers or a skirt that concealed the lower leg they were asked to roll up the trousers or hitch up the skirt to provide a clear view of the lower leg markers.

The pedobarograph pedestal was located in the centre of a 5m walkway. Subjects were asked to stand at a series of markers at one end of this walkway and then, beginning with a particular foot, walk at a comfortable pace down the walkway. This procedure was repeated, with necessary changes to the starting point, until the subject put the foot of interest directly onto the pedobarographic plate. Every effort was made to avoid the subject targeting the plate.

Each subject was recorded four times, twice for each foot as the subject walked toward and away from the longitudinal walkway camera. Only one subject was so inconsistent in their walking patterns that it was not possible to get a complete set of records in under ten minutes and so the subject's record was dismissed.

A log was kept of all the subjects' details including the biomechanical assessment, pedobarographic filenames and the tape counter details for the motion video recorder.

4.2.5 Analysis

Following the recording of all the data, two paths are followed for the analysis process. The details recorded from the pedobarograph and

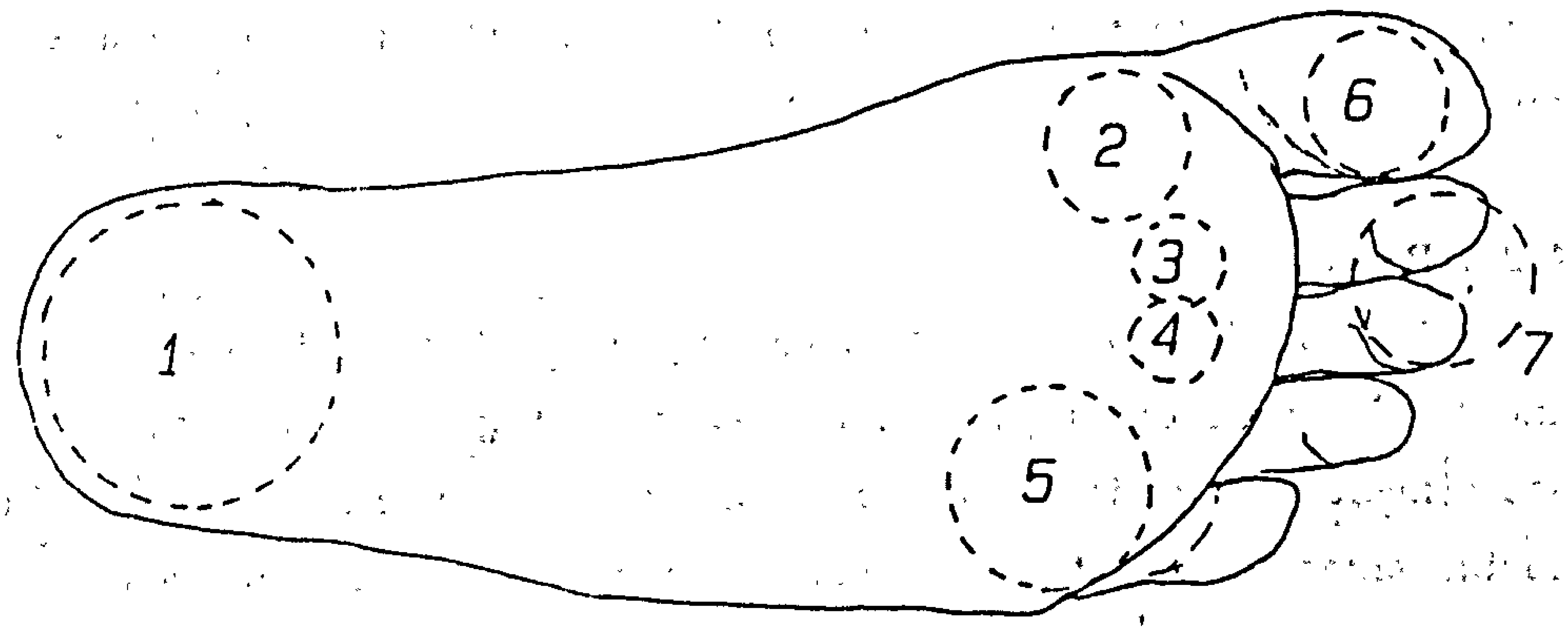


Figure 4.3 The seven areas beneath the foot marked for the indication of force records.

those from the video cameras were initially analysed separately before the final stages where they were both studied using the same program.

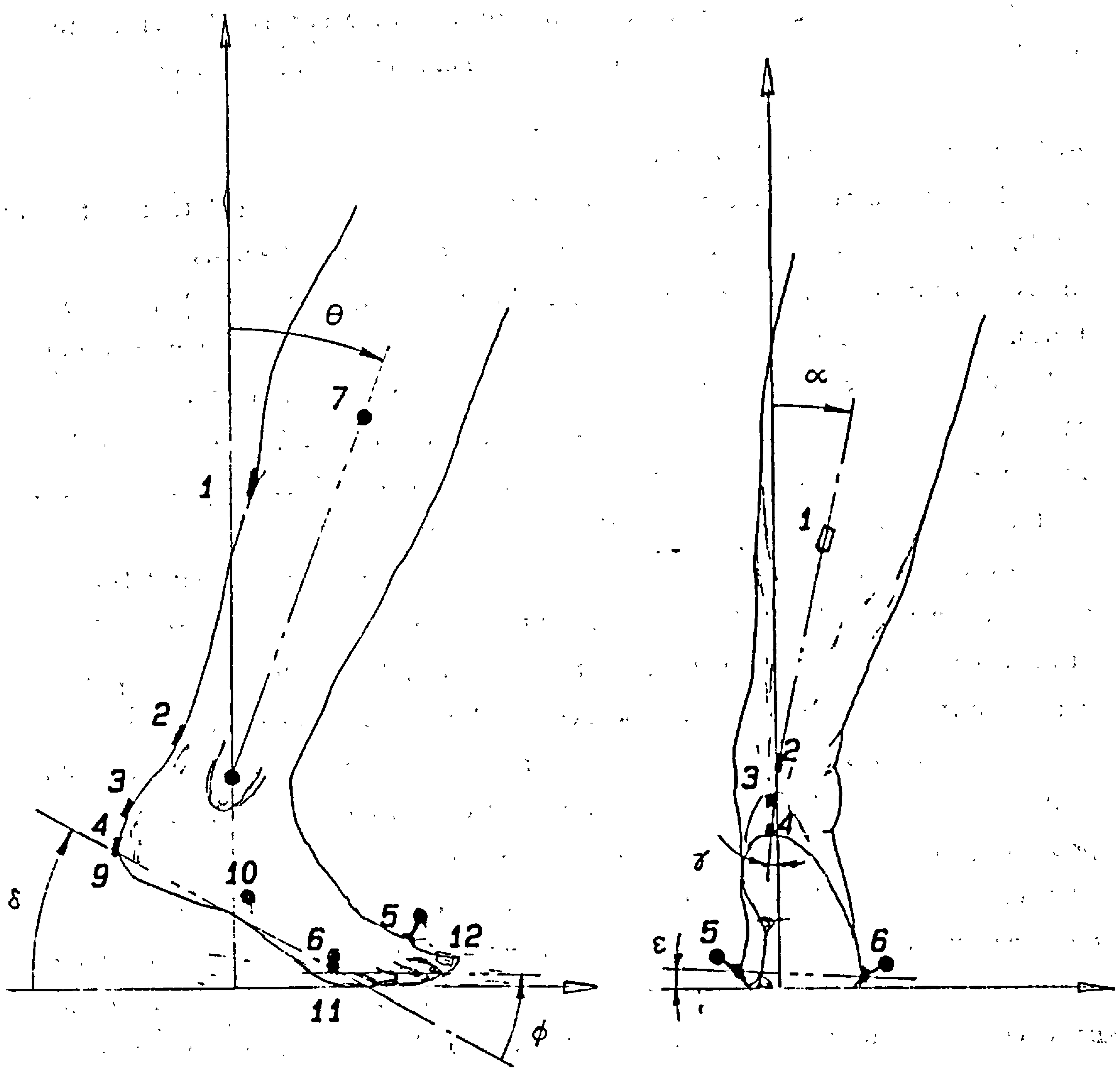
4.2.5.1 Pedobarograph data analysis

Data from the pedobarograph computer was stored in a data file format so the subject's original file could be replayed in full. Various analysis routines are available in the pedobarograph software. For these tests, areas of interest were ringed on the screen and the computer would calculate the force, maximum or average pressure in the circles. A standard order of seven circles were used for each foot (fig 4.3):- heel, 1st metatarsal (MT) head, 2 MT head, 3 MT head, 4 & 5 MT heads, the head of the hallux and toes 2 & 3 together.

Output from the routines was in the form of structured text files and a Pascal program (FEETALL; Appendix 4) was written for a desktop AT standard personal computer to open these files and provide a graphical representation of the forces in the areas relative to a normalised time base of 18 frames. The routines were capable of normalising the forces to a subject's body weight or to the total vertical force measured by the pedobarograph plate. By graphically illustrating all the records on the screen, it was possible to visually assess the variation and remove damaged records (ie those where the computer had prematurely truncated a record before toe off etc). At the end of this process the program would produce a file containing the median, upper and lower quartiles and the minimum and maximum values for each area over each frame based on all the accepted records (Table 4.1).

4.2.5.2 Video data analysis

The video digitizing system comprised the U Matic video cassette player, a colour monitor and an Archimedes computer. The player was used to replay the recorded tapes one frame at a time and hold each frame for digitizing. The computer was fitted with a video digitizing card (Watford Electronics, Archimedes Video Digitiser) which allowed a video frame to be displayed on the graphics screen. The Archimedes computer uses simple Basic language commands to interact with the video images. These commands were included in the digitisation program (MDIGIT3; Appendix 4) which enabled the positions of the markers on the foot to be recorded using the computer mouse and an on-screen cursor.



$$\tan\alpha = \frac{Z_1 - Z_2}{Y_1 - Y_2}\cos\theta \qquad \tan\theta = \frac{X_7 - X_8}{Y_7 - Y_8}$$

$$\tan\gamma = \frac{Z_3 - Z_4}{Y_3 - Y_4}\cos\delta \qquad \tan\delta = -\left(\frac{Y_{11} - Y_9}{X_{11} - X_9}\right)$$

$$\tan\varepsilon = -\left(\frac{Y_5 - Y_6}{Z_5 - Z_6}\right) \qquad \tan(\phi - \delta) = \frac{Y_{12} - Y_{11}}{X_{12} - X_{11}}$$

Figure 4.4 The calculated angles (in terms of a leg coordinate system; see fig 3.3) produced from the digitized video images.

In order to provide an indication of the true cartesian positions of the markers a special marker board was placed over the pedobarograph plate in the field of view of the two cameras. These images of the fixed points were recorded each day after the cameras had been installed and before subjects were assessed. These points were digitised and used to calibrate marker records to real positions. Calibrated marker points were stored on magnetic disk against marker number and pedobarograph frame.

A second program (STATPROCB; Appendix 4) was produced to read user selected files produced from MDIGIT3 (Table 4.1), process the data and calculate various distances and angles between markers. The important points marked, and angles calculated are illustrated in fig 4.4. It should be noted that all the angles calculated are *projected* angles (see fig 3.3 for the graphical illustration of this). In order to calculate true angles (about limb coordinated systems) it was necessary to correct for the inclination of a segment relative to the plane in which the angle was measured (sufficiently accurately by multiplying the tan of the projected angle by the cosine of the inclination). For all the angles recorded by the transverse camera the inclination of the segment to the sagittal plane of reference was seldom more than 5-10°. In this case $\cos 10^\circ = 0.985$ and so the calculated angle is within 98.5% of the true angle and the extra calculation involved to correct for the projection was considered unwarranted. Both α and γ are measured in a plane that will move from typically -20° to 40° to the coronal plane. In this case the correction for the projected angle was considered essential particularly in early and late stance when the inclination of the limb was maximal. α and γ were corrected using the values of θ and δ respectively, to indicate the limb inclination. These values were normalised to 18 frames and the median, upper and lower quartiles and minimum and maximum values were calculated from the data for each angle for each frame and stored.

4.2.5.3 Final Analysis

The output from the initial analyses were both read into a final program (BGRAPHHP; Appendix 4). This program provided graphical output of the statistical details previously calculated in the form of 'Box-Whisker' plots (fig 4.5). It also contained the routine for plotting the details on an HP pen plotter.

The output from BGRAPHHP indicated that some regression techniques might have provided further information on the data. The program FILSTRIP

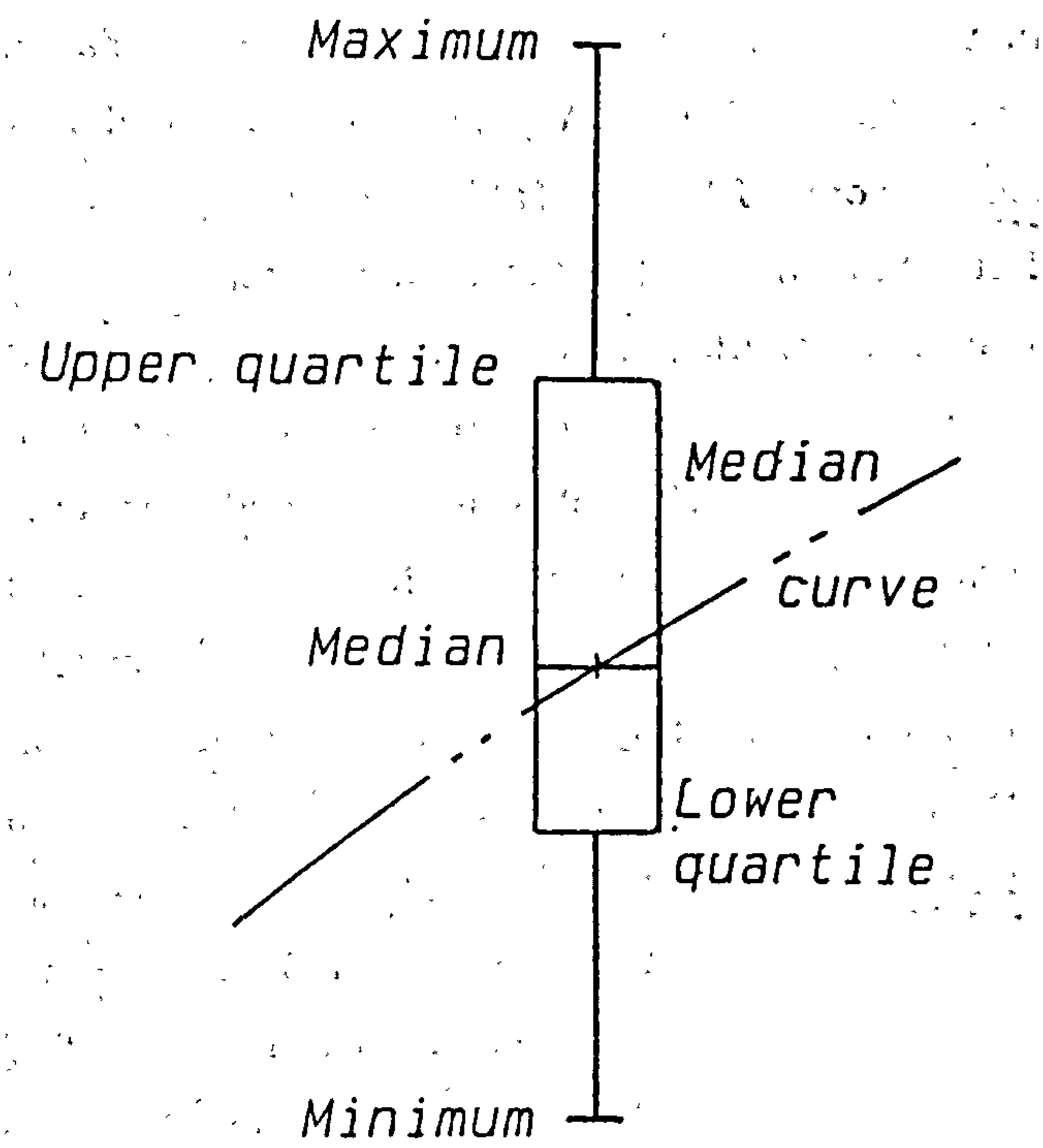


Figure 4.5 An explanation of the "box-whisker" method of displaying data as used in the diagrams of this chapter.

(Appendix 4) was used to collect the appropriate details from both the Pedobarograph records and the angle files from the STATPROCB program and file them to ensure that both the foot and the frame number corresponded. These files were assessed using the MINITAB statistical package. Firstly stepwise regression was used to establish dependent variables and subsequently multivariate regression provided more detailed analysis of the interdependency that is presented in the next section.

4.3 RESULTS

The graphical results of the tests are presented in figures 4.6-4.8. These diagrams show the typical or median levels of force or angle and the variation about that typical value. The following review of these curves discusses the changes in the median values for simplicity.

4.3.1 The Pedobarograph

The results from the pedobarograph are a new representation of previously reported information. Earlier studies that have used the pedobarograph have indicated the results in terms of maximum or average pressure in an area of interest (Betts et al, 1980b; Duckworth et al, 1982a) while figure 4.6 shows the total *force* in the areas. As a result average pressure is related to the force by the equation:

Force = Pressure x Area of application - Eqn 4.1.

In general, pressure has been used to assess the effect of these loads on the skin (Duckworth et al, 1982a) while in the present work the forces transmitted into the end of the metatarsal bones were considered more appropriate.

The force transmitted by the heel rose rapidly to over 55% body weight (BW) within the first three frames (0.1s). Unfortunately although the plate support load cells would be able to measure the vertical force at over 50Hz the pedobarograph had a recording frequency of only 25Hz. It would be unlikely that the peak level of the heel strike force would be recorded due to its higher frequency though the maximum 'whiskers' indicate that some subjects produced forces that were measured up to 95%BW. It is important to realise the frequency limitation of the pedobarograph since very high heel forces can be a good indication of a subject who has targeted the plate. The force level fell steadily after the peak and was zero between frame 12 and 13 (0.4s).

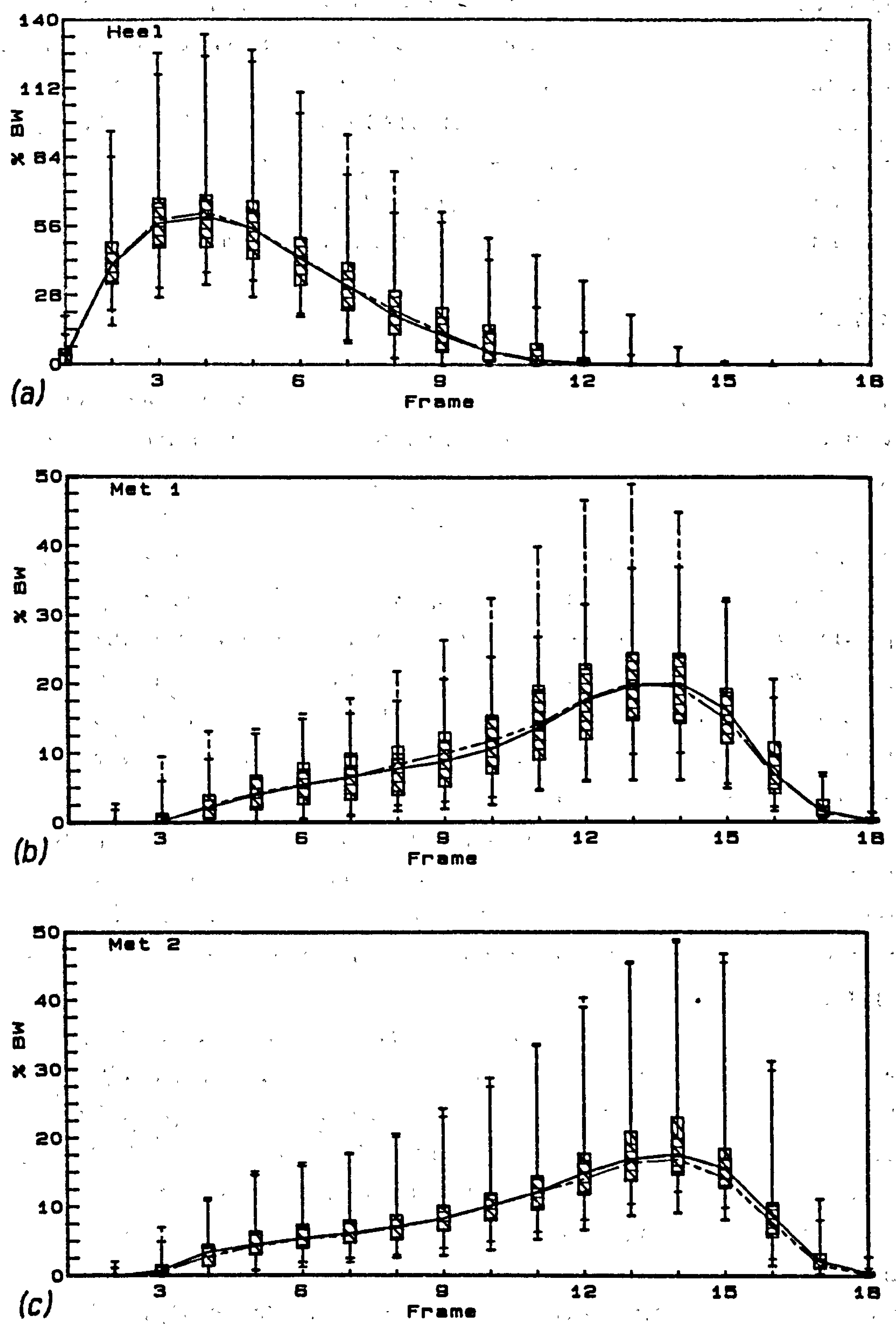


Figure 4.6: Forces under the areas of the left (-- ——— --) and right (———) feet for all 61 subjects.

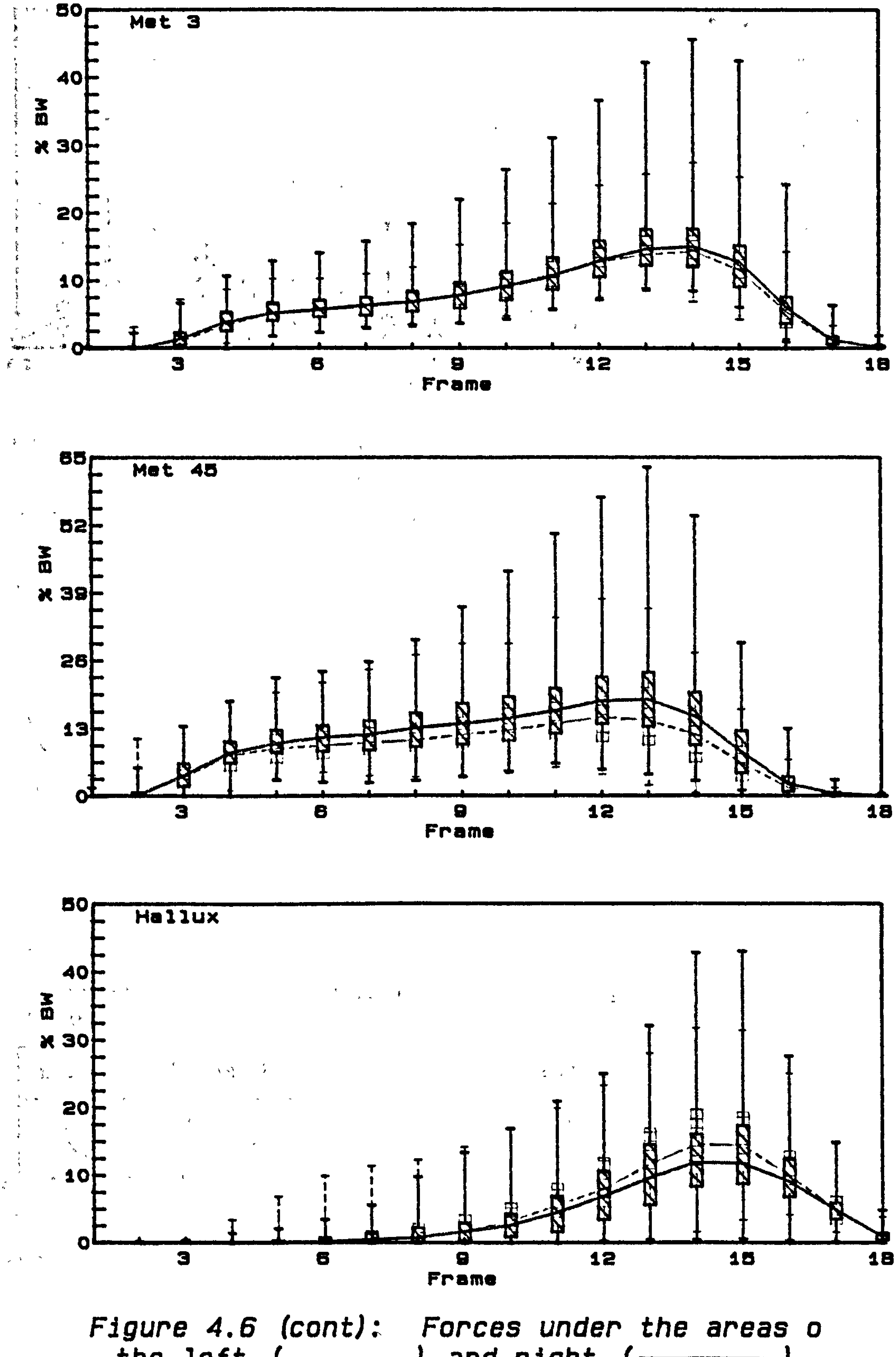


Figure 4.6 (cont): Forces under the areas o the left (--———--) and right (————) feet for all 61 subjects.

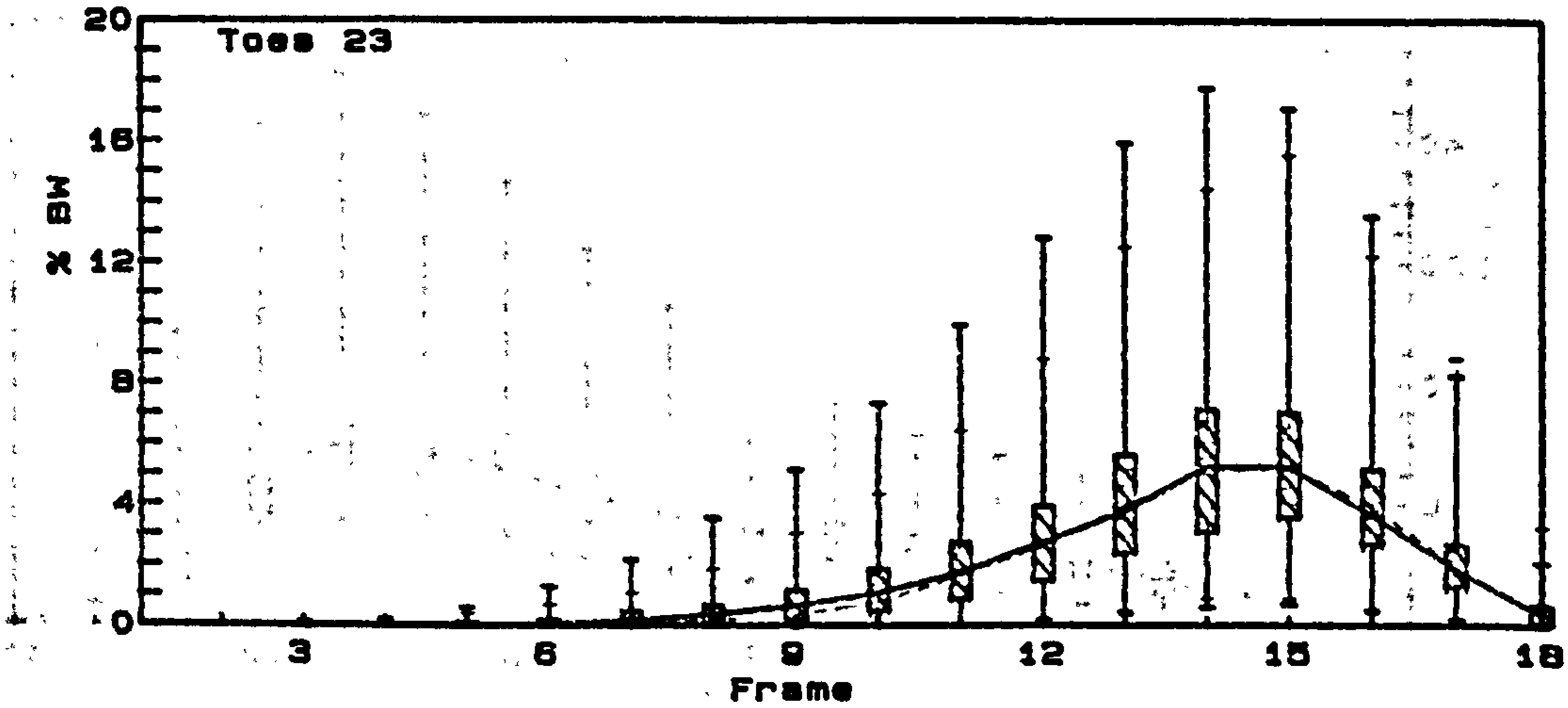


Figure 4.6g The vertical force under toes two and three.
Left foot (-- ——— --); right foot (———).
(N = 61)

Table 4.2: Percentage of total vertical force supported by each foot area during periods of mid and late stance. (% applied force)

Frame	Heel	Met 1	Met 2	Met 3	Met 4&5	Hallux
7 (Mid stance)	40	7	7	7	10	2
16 (Late stance)	0	15	18	12	16	26

The force under Metatarsal 1 began to rise as the heel force peaked (at or after frame 3) and exhibited a reasonably linear increase until a peak of 15-20%BW at frame 13 to 14 (0.45s). A rapid decrease in force then occurred to bring the level to less than 2%BW at frame 17.

Metatarsal 2 followed a similar loading pattern to metatarsal 1. The peak level was more consistently around 15%BW and the loading regimes indicated almost a 50% reduction of the variation of those for Met 1.

The middle metatarsal followed a very similar pattern to that of the second. The initial plateau around frame 5 is accentuated and the peak is only 12-14%BW but the variation is of the same level.

On the lateral side of the foot there is a much greater degree of variation in the loads carried by the fourth and fifth metatarsals. Some subjects reached their peak loads before frame 9 and subsequently decreased to zero. The majority had a pattern not unlike the third metatarsal load. There appeared to be no correlated variable that could be used to predict which subjects would load their lateral rays earlier than others. Without corroborating evidence it was felt to be statistically invalid to arbitrarily divide the subjects into groups.

The overall curve shown for the lateral two metatarsal rays shows considerable variation as a result of these varying loading patterns. The plateau was quite marked and the region from frame 5 to 13 had only a 3-4%BW increase in load to a peak of 11-14%BW. The decrease in load after the peak was still rapid falling to zero within four frames.

Forces under the big toe or hallux are rarely investigated. Figure 4.6(f) shows that the toe did not bear load before frame 5 to 6, then increased to 9-11%BW between frames 14 and 15 (0.48s) and decreased quickly to zero in the last frame (push off).

Toes 2 and 3 follow a loading pattern very similar to the hallux reaching a peak of 6%BW again between frames 14 and 15.

A different perspective on the data can be obtained by normalising the forces to total vertical force applied to the foot as measured by the pedobarograph load cells. The resulting diagrams (fig 4.7) are based on scales of "% Total Shank Load". The vertical load percentages carried by each support area at midstance (frame 7) and late stance (frame 16) are noted in Table 4.2.

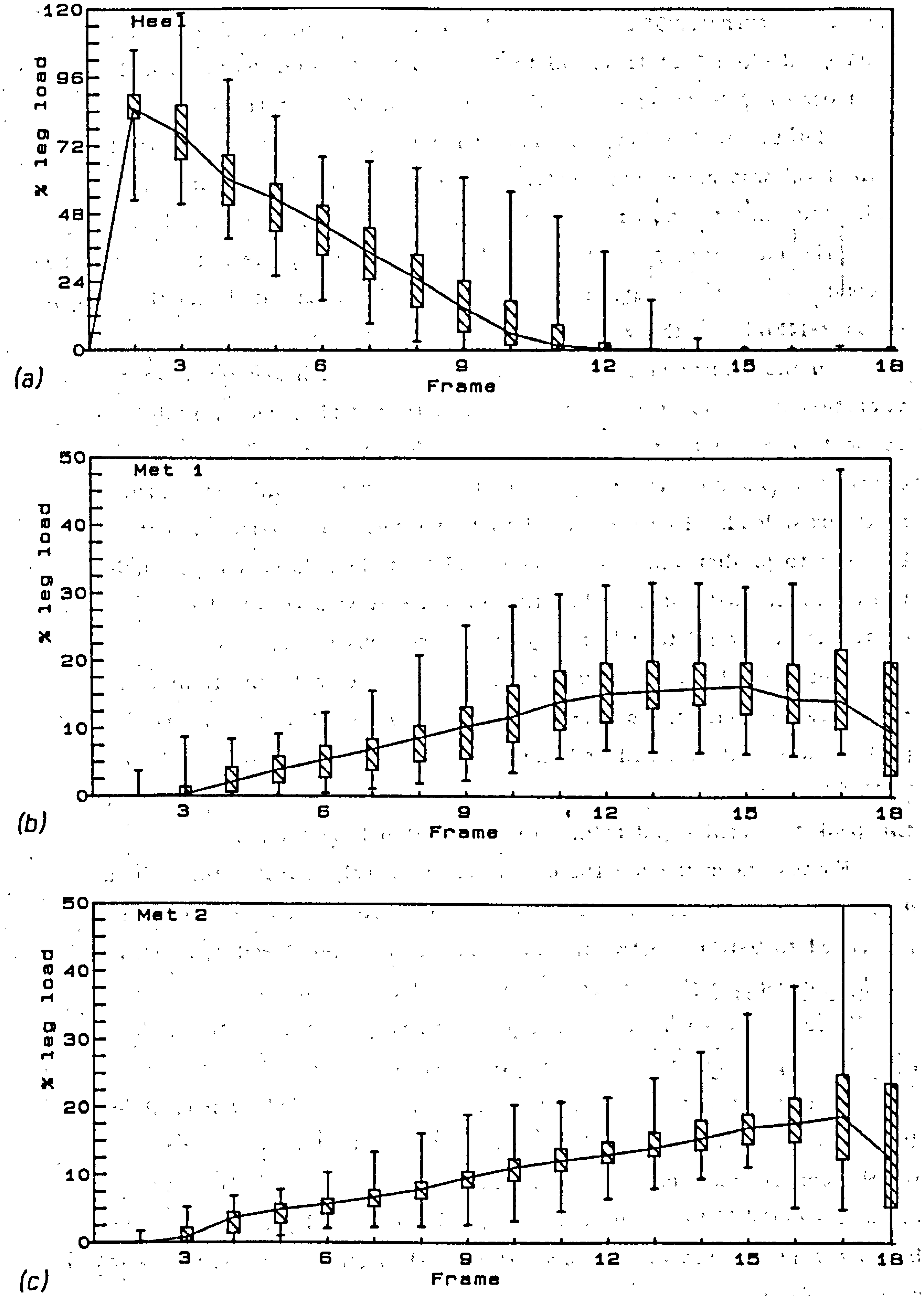


Figure 4.7 Force in the areas beneath the right foot normalised to the total vertical force carried by the leg (N = 60).

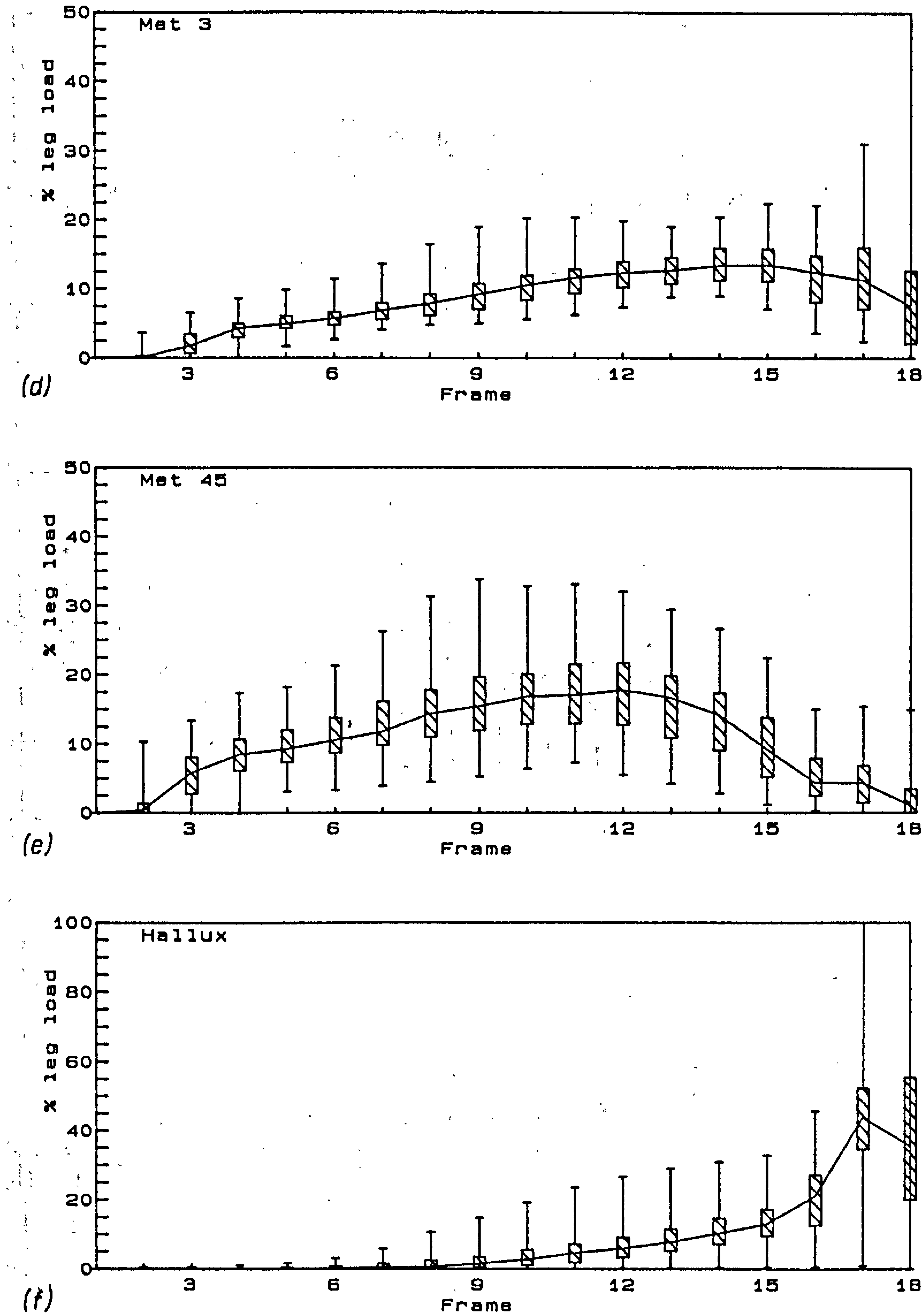


Figure 4.7 (cont) Force in the areas beneath the right foot normalised to the total vertical force carried by the leg (N = 60).

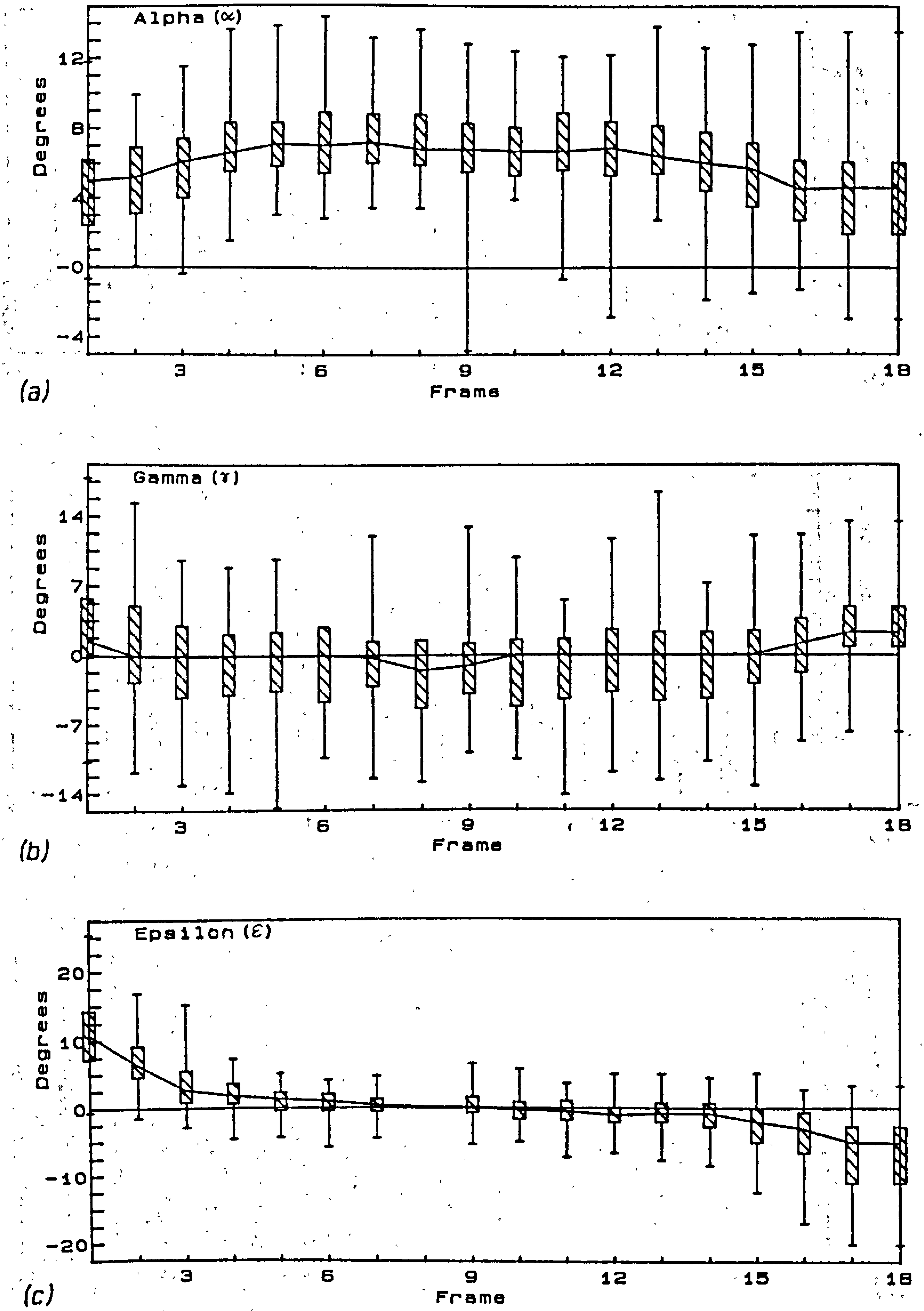


Figure 4.8: Angles of the left leg and ankle taken from the most reliable views. (N=55) Epsilon (ε) set to zero in frame 8.

4.3.2 Measured Foot Angles

The six angles calculated are each displayed against the frame number in figure 4.8.

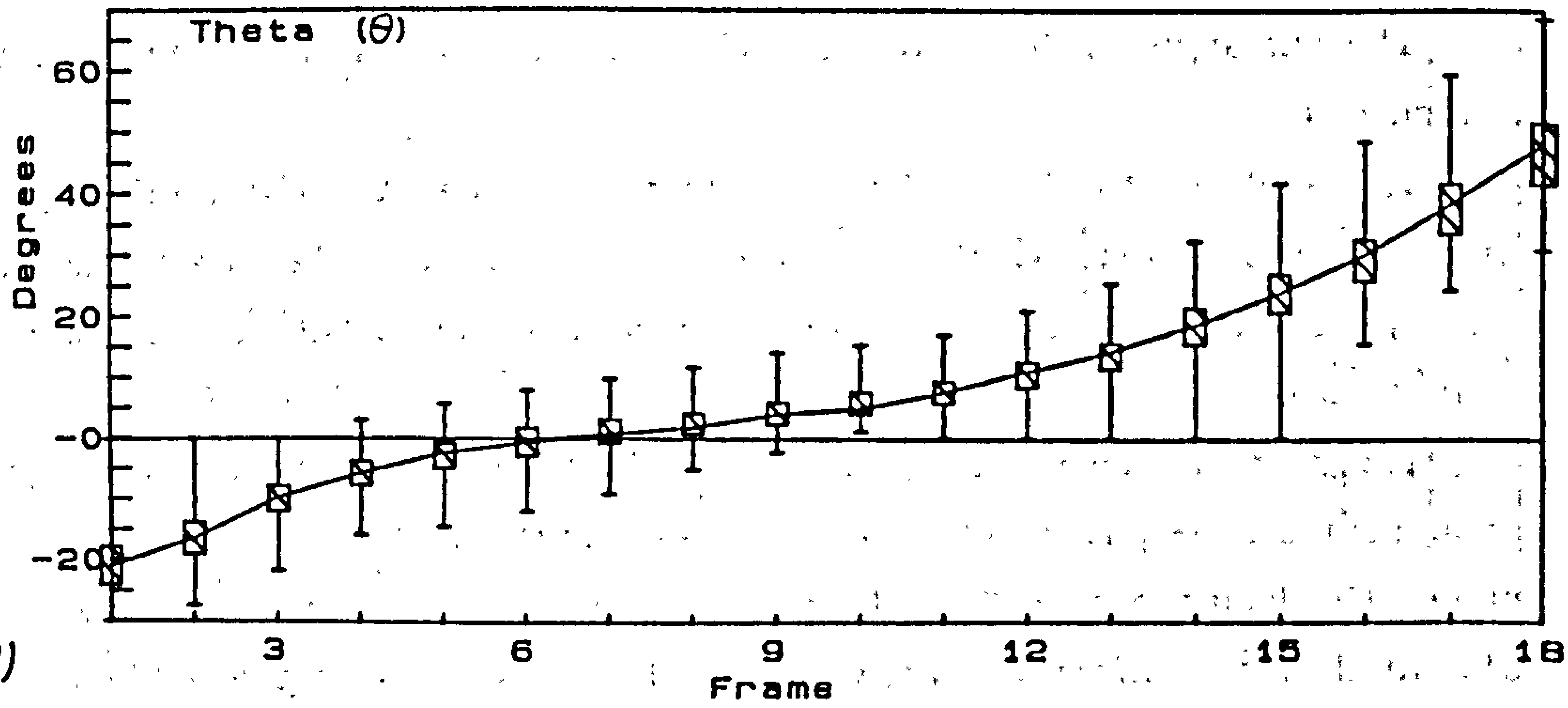
The angle of the back of the leg to the vertical, α, (fig 4.8a) remained greater than 4° throughout the stance phase. The angle increased slightly and plateaued at 6-7° between frames 8 and 13 before decreasing again to 5-6°.

The rear calcaneal angle, γ, (fig 4.8b) remains close to vertical throughout stance phase only rising to 3-4° in the last three to four frames (ie during push off). The pro/supination angle of the ankle is calculated as the difference between α and γ. As a result the ankle is less than 4° pronated at heel strike but this increases to 6-8°during midstance before falling to less than 3° at push off. Caution must be exercised in interpreting the last one or two frames of data for α and γ. The angle calculation algorithms corrected for the angle of the leg in the sagittal plane (θ) but the markers used to calculate these angles were moving into a horizontal line on the top of the heel in late stance and in this position were difficult to digitise accurately.

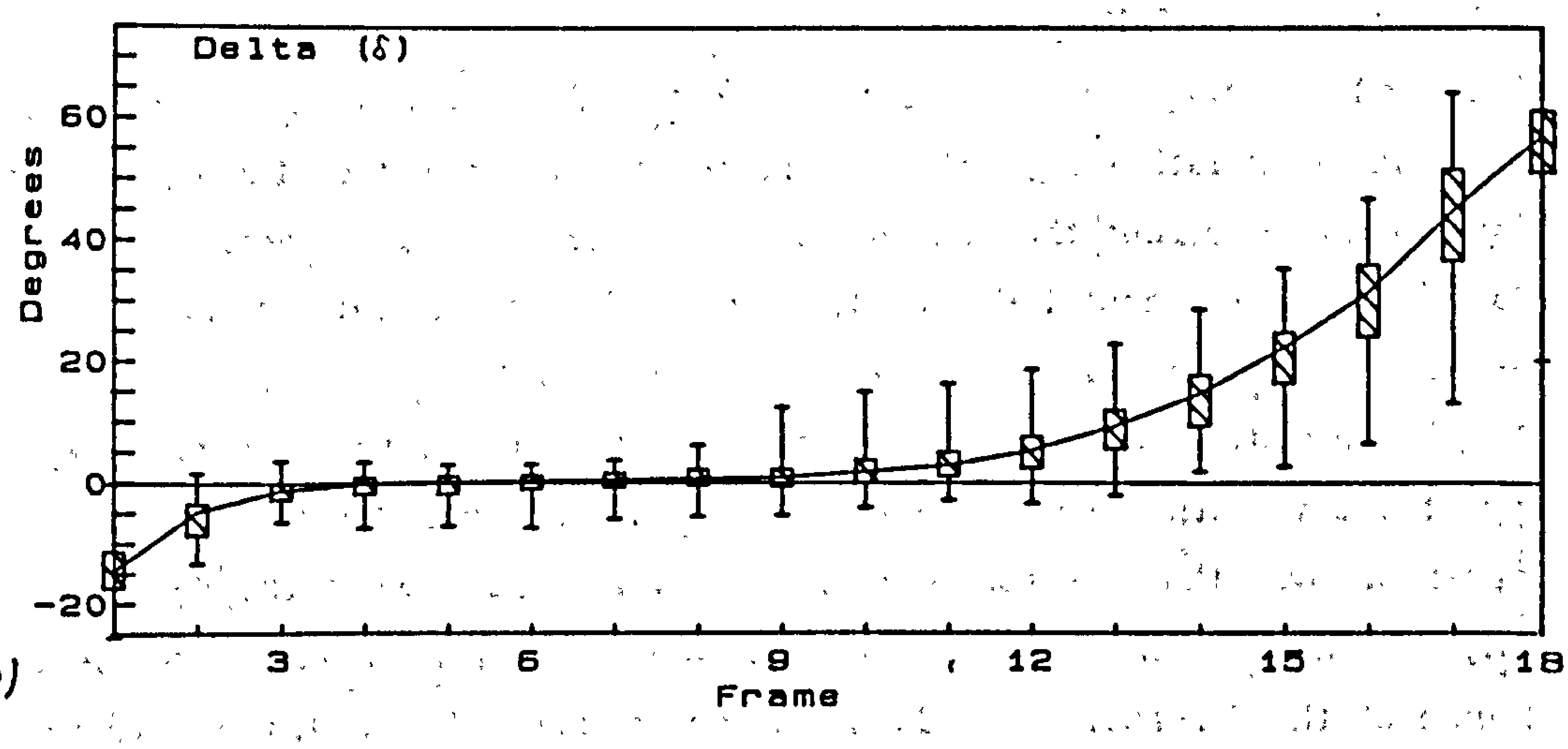
The forefoot angle, ε, was normalised to zero at frame 8 to remove any of the variation due to different sized metatarsal heads. The hypothesis was that the forefoot angle would be zero when the forefoot was fully on the ground. This technique allowed a clearer indication of the change of the forefoot angle during stance phase. Figure 4.8c shows that the forefoot was almost 10° inverted at heel strike and this rapidly decreased to less than 2° at frame 4. Apart from some variation over the next nine to ten frames the angle did not appreciably change until it increased to 4° eversion during the push off phase.

One of the most consistent curves was the graph of the angle of the tibia and fibula in the sagittal plane, θ, against time (fig 4.8d). Heel strike was accompanied by a rearward angle of 20° which decreased to an inflexion point just after the leg reached the vertical position (frame 7). The angle then increased in the forward direction reaching around 45° at toe off.

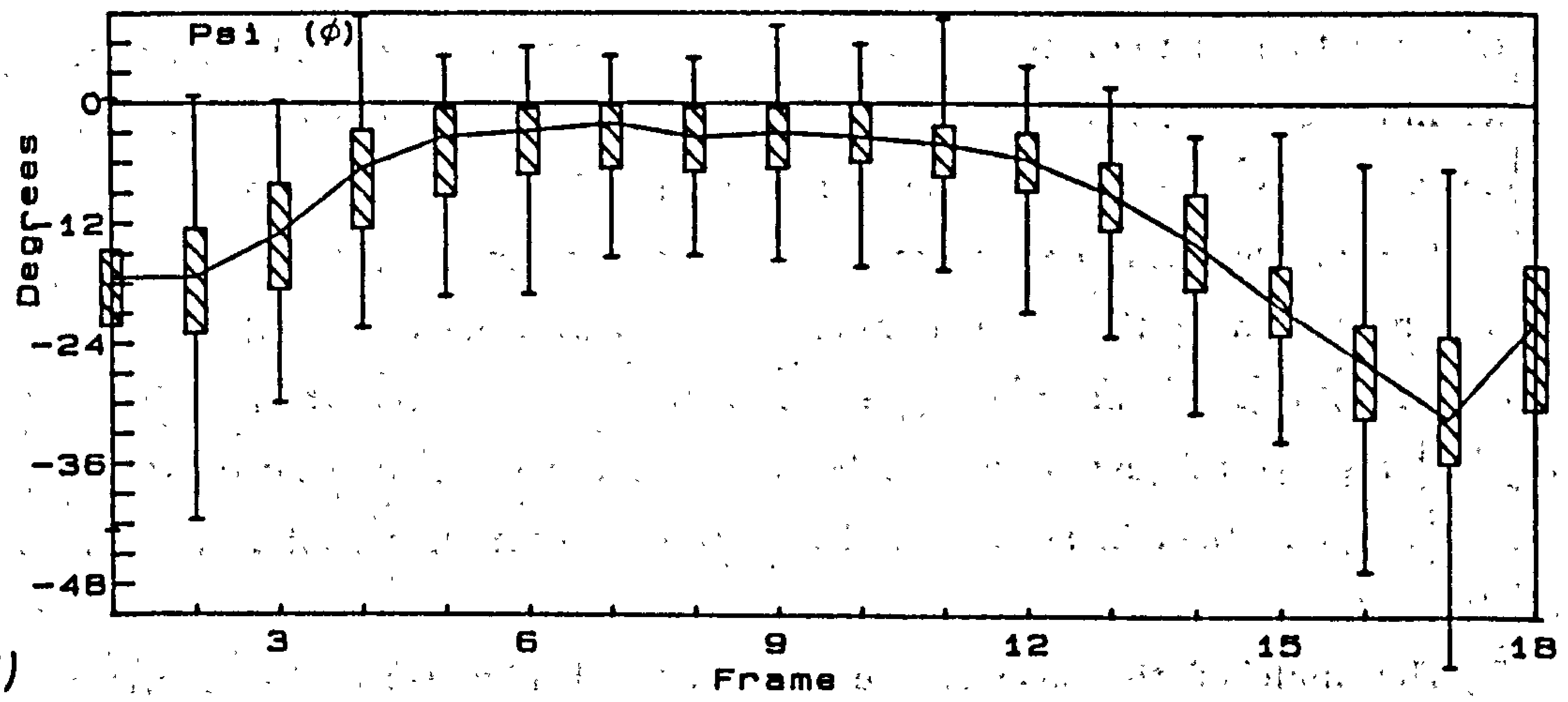
The angle of the base of the foot to the horizontal in the sagittal plane, δ, (fig 4.8e) is not unlike θ although it has a much longer region of inflection (corresponding with the foot flat phase). At heel strike the angle is around 15° which decreases to zero before frame 4. The angle remains unchanged until frame 9 when there is an increase which



(d)



(e)



(f)

Figure 4.8 (cont): Angles of the leg and ankle taken from the most reliable views. (N=60) becomes rapid in later stance reaching around 55° at toe off. Again an important anatomical angle can be calculated this time by subtracting θ from δ to give the angle of plantarflexion at the ankle. This is shown in figure 4.8g.

Toe flexion and extension (or plantar and dorsiflexion for clarity) is indicated by ϕ (fig 4.8f). Interestingly the toes are dorsiflexed up to 16° at heel strike. This angle decreases to around zero degrees from frames 5 to 12 after which the toes begin to dorsiflex to 40° at frame 17. There is a sharp decrease in dorsiflexion in the last frame to around 20-25° at toe off.

4.3.3 Multiple Regression Analysis of the Forces and Angles

Regression calculations were performed using all the variables in order to determine any underlying relationship between measured parameters. Variables are represented by either their area of action in the case of force, or by the Greek symbol defined in figure 4.4 for the angles. "Frame" is the frame number and as such represents time in the equations. The forces were either normalised to a percentage of body weight or total vertical shank load and only the regression equation with the best fit is presented here. The full details (including example scatter plots) and the statistical summaries are included in Appendix 1.

The heel was found to correlate best when variables were normalised to total vertical shank load:-

Heel = 94.0 - 4.5 Frame - 1.4 Met45 - 0.86 Met1 R^2=91% -Eqn 4.2

All other forces were represented as a percentage of body weight. The first metatarsal showed similar R^2 values for both normalisation methods:-

Met1 = 0.56 + 0.82 Met2 + 0.66 Hallux - 0.96 Toes23 R^2=74% -Eqn 4.3

Both the second and third metatarsals appeared to be dependent on the loads supported by their neighbours:-

Met2 = 0.78 Met3 + 0.20 Met1 + 0.15 Hallux R^2=87% -Eqn 4.4

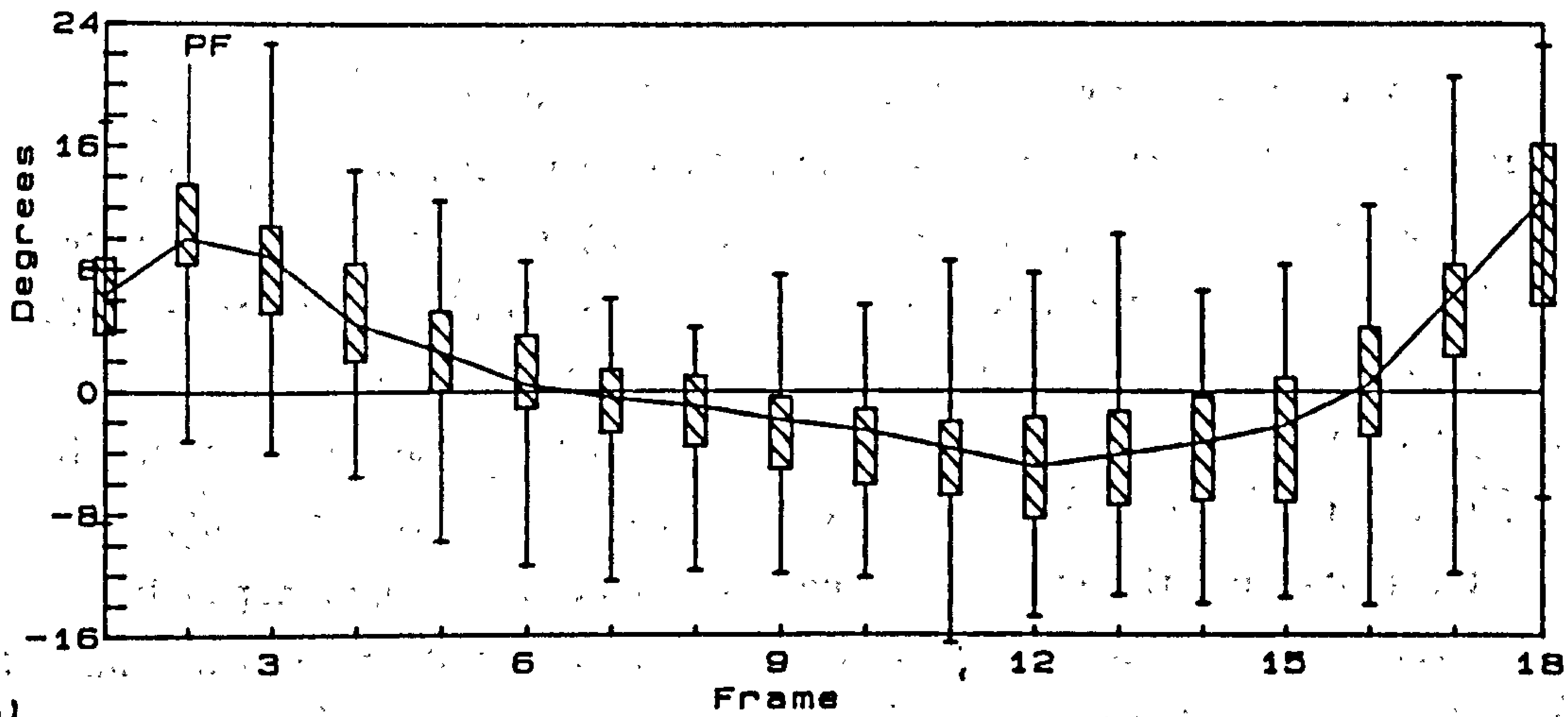
Met3 = 0.57 Met2 + 0.24 Met45 + 0.15 Toes 23 R^2=88% -Eqn 4.5

Metatarsals 4 and 5 showed the poorest fit to the regression line and normalisation to the total shank load only increased R^2 to 68:-

Met45 = 2.6 + 1.5 Met3 - 0.43 Hallux + 0.24 Met 2 R^2=62% -Eqn 4.6

The toes all showed a reasonable fit to the calculated regression lines:-

Hallux = 0.65 +1.6 Toes23 +0.35 Met1 -0.17 Met45 R^2=80% -Eqn 4.7



(g)

Figure 4.8(cont): The ankle plantarflexion angle taken from the LT records (N = 60).

$Toes23 = -0.09 + 0.33\ Hallux + 0.14\ Met3 - 0.088\ Met1 \qquad R^2=71\% \quad$ -Eqn 4.8

Neither α nor γ showed any clear relation to any of the other variables (best $R^2<15$) and so both have been omitted from this analysis. The forefoot angle, ε, was moderately correlated with three variables:-

$\varepsilon = 11.5 - 1.0\ Frame - 0.076\ Heel - 0.106\ \phi \qquad R^2=50\% \quad$ -Eqn 4.9

All the sagittal angles appeared to be interrelated:-

$\theta = -26.1 + 3.75\ Frame - 0.57\ Met3 + 0.08\ Heel \qquad R^2=90\% \quad$ -Eqn 4.10

$\delta = -15.7 + 3.50\ Frame - 1.00\ Met3 - 0.414\ Met1 \qquad R^2=83\% \quad$ -Eqn 4.11

$\phi = -5.3 - 0.80\ \delta - 0.48\ \varepsilon + 0.36\ \theta \qquad R^2=61\% \quad$ -Eqn 4.12

The ankle flexion angle ($\delta-\theta$) was related to similar variables to those found for θ and δ themselves:-

ankle PF = $22.0 - 3.54\ Frame + 1.13\ Met1 + 0.49\ Met3 \qquad R^2=90\% \quad$ -Eqn 4.13

4.4 DISCUSSION

4.4.1 Introduction

This study of normal foot function in vivo was undertaken to provide a clearer overall view of both the load distribution under the foot and the corresponding articular motion during the stance phase of normal steady gait. Several of the findings presented in this study require further discussion individually, although it is important to consider the complete picture subsequently in the final subsection.

A number of the results will be used in the next chapter to compare with results obtained from the tests on the amputated specimens. The assessment of the muscle activity in chapter 6 will also draw on some of the kinematic and, to a lesser extent, force patterns presented here.

4.4.2 Force Distribution

In general the graphs of the forces against time were encouragingly consistent with only a limited variation. Drawing on a random subsection of the general population might have easily produced curves with quite large variation bars. This limited variation strengthens theories of a 'standard' gait pattern already seen in more general observations.

The forces under the heel were slightly lower than might have been anticipated though the low frequency (25Hz) of the pedobarograph system will tend to attenuate these rapidly changing forces. Heel rise corresponds with the normally accepted time of 40% of the gait cycle. The normalisation of the heel force to the force transmitted by the tibia and fibula indicated a near linear loading pattern. The regression equation based on these normalised forces indicated a strong inverse correlation with the frame number (ie time), and the loading on the medial and lateral rays of the foot. The linear load decrease with time may indicate a simple load transfer onto the medial rays which show an increasing linear behaviour when normalised in the same fashion.

At 10% of the gait cycle, the beginning of loading on the metatarsals was ahead of the usually accepted time of 20% of the gait cycle (Rodgers, 1988; Mann, 1988). The forces are low and may have been previously overlooked with systems using a higher threshold. The first three metatarsal heads all had similar loading patterns that peak at 17,15 and 13%BW respectively. The regression equations for the forces normalised to body weight indicated that the loading patterns on these three rays were strongly influenced by the pattern on the neighbouring metatarsal(s). The first metatarsal was also correlated to the forces on the hallux and inversely correlated to toes 2 and 3.

The lateral two metatarsal heads showed a much greater variation than the medial rays. Some subjects reached their peak force on these two rays before frame 9 while the majority had a loading pattern that was similar to the other three rays. No other variable could be found that explained this variation and an arbitrary division was dismissed. There is certainly strong qualitative evidence that there are people who are 'lateral-medial' walkers (ie load laterally early but push off on the medial side) and others who walk 'straight through' (ie the locus of the centre of pressure is a relatively straight line near the second metatarsal bone) but further experimental work would need to be done specifically to prove or disprove these observations. The median peak force was recorded in late stance at around 13%BW. The inverse correlation with the hallux force pattern in the regression analysis indicated that the hallux had some ability to offload the lateral ray or was involved with it in some stabilisation of the forefoot.

The toes are a further area that has received scant attention in the past (with the exception of the work by Jacob, 1989). The results show that the hallux and the second and third toes began load bearing around 20% of the gait cycle and peaked at 48-50% of the cycle with force levels of 10% and 6%BW respectively. It is interesting to see that although the hallux and toes two and three supported less than 9%BW in frame 17 this load represented 55-80% of the total load transmitted by the tibia and fibula during that frame. The regression analysis indicated that the

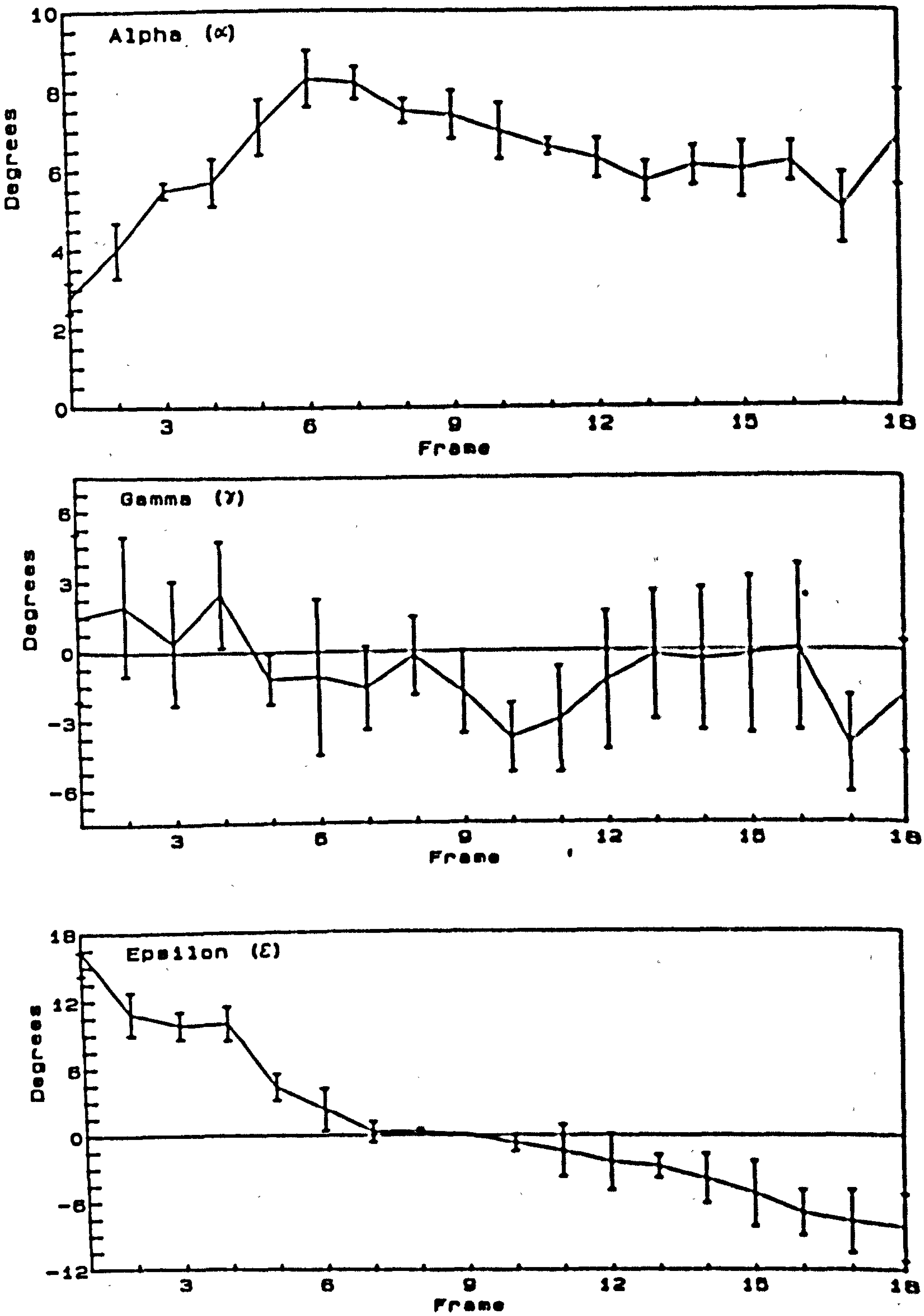


Figure 4.9: Mean and std deviations of the foot angles after digitizing the same record four times.

loading patterns of the toes are closely related and for the hallux there is also a close relation with the first metatarsal and an inverse association with the lateral rays of the foot as has already been discussed.

4.4.3 The Kinematics

Several of the angles calculated from the recorded leg markers show a large degree of variation in comparison with the force records. The difficulties associated with accurate placement of the markers, marker movement over the bony areas and digitisation of the images are the most likely causes of this variation although previous work (Kadaba et al, 1989) has shown a degree of variability in the gait cycle itself. In order to quantify the errors of digitisation one subject's file was digitised 4 times and the corresponding angle plots calculated (fig 4.9). The same subject had three records made of the same set of markers to help assess the errors associated with marker motion and gait cycle variability (fig 4.10). The results presented in these two graphs indicate the variation associated with the digitisation process. By far the poorest consistency was the curve associated with the rear foot (γ, fig 4.9b). This was due almost entirely to the difficulties in digitising the two close markers on the calcaneus. These were about 25mm apart and on the digitisation screen one pixel either side of the marker would alter the angle by up to 5°. By way of contrast with this large variation the results for all the other calculated angles show standard deviations from the mean no greater than 2.5° and generally deviations much less than this, particularly for the sagittal angles (θ,δ & ϕ). The effects of marker motion and the variability in the subject's gait pattern can be seen in figure 4.9. There is an appreciable increase in variation for all the angles with standard deviations about the means increasing to 4° maximum (although the curve for γ has higher variation, up to 8°).

In general the results of this error analysis indicate that reliable results can be achieved with careful digitisation of a single record and the effects of marker motion and the variability of a subject's gait style constitute a random error of only a few degrees for most of the calculated angles. The largest variation occurs where markers are placed close together (eg rear calcaneus). Only in the case of γ does the deviation obscure the angular changes taking place.

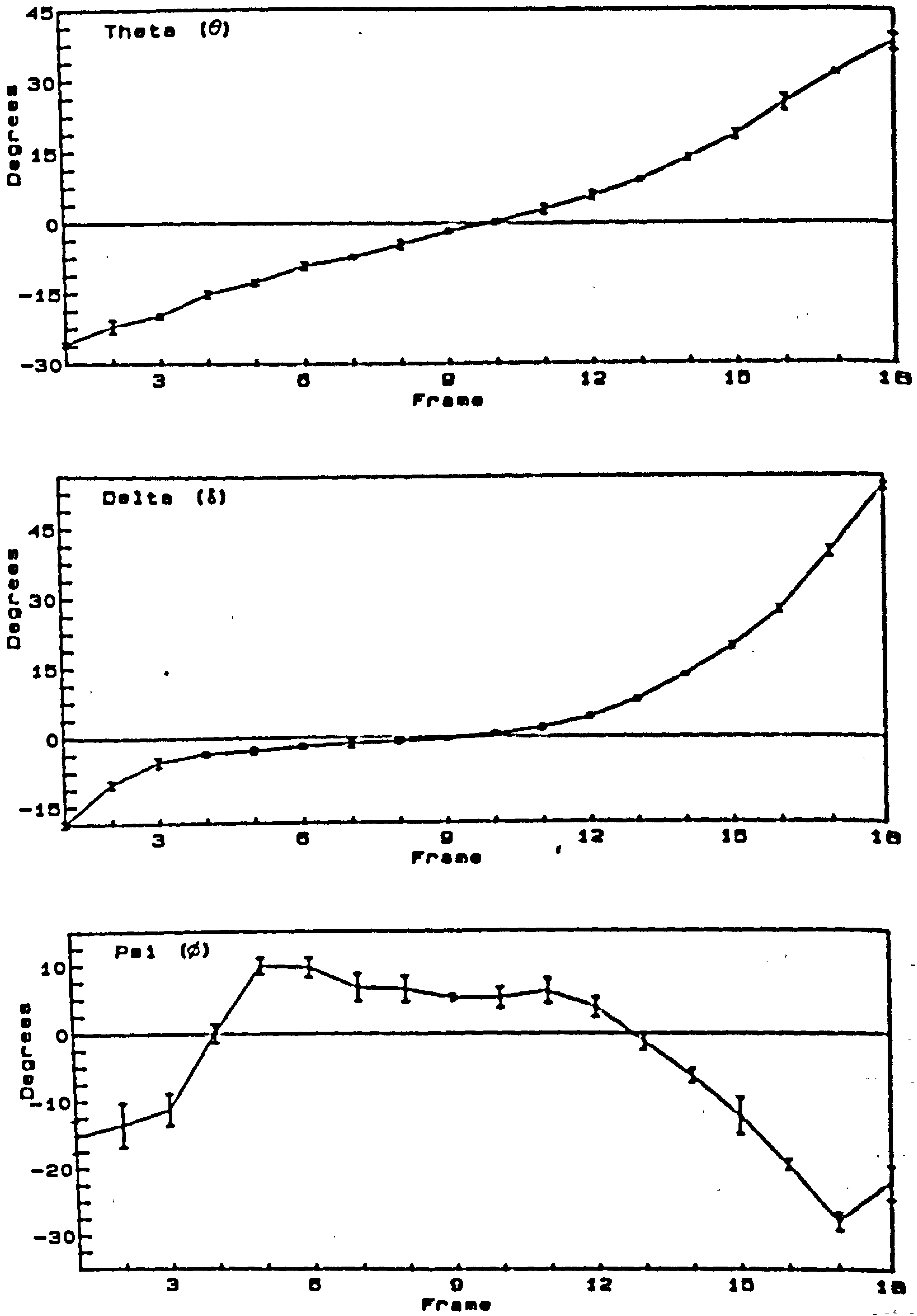


Figure 4.9: Mean and std deviations of the foot angles after digitizing the same record four times.

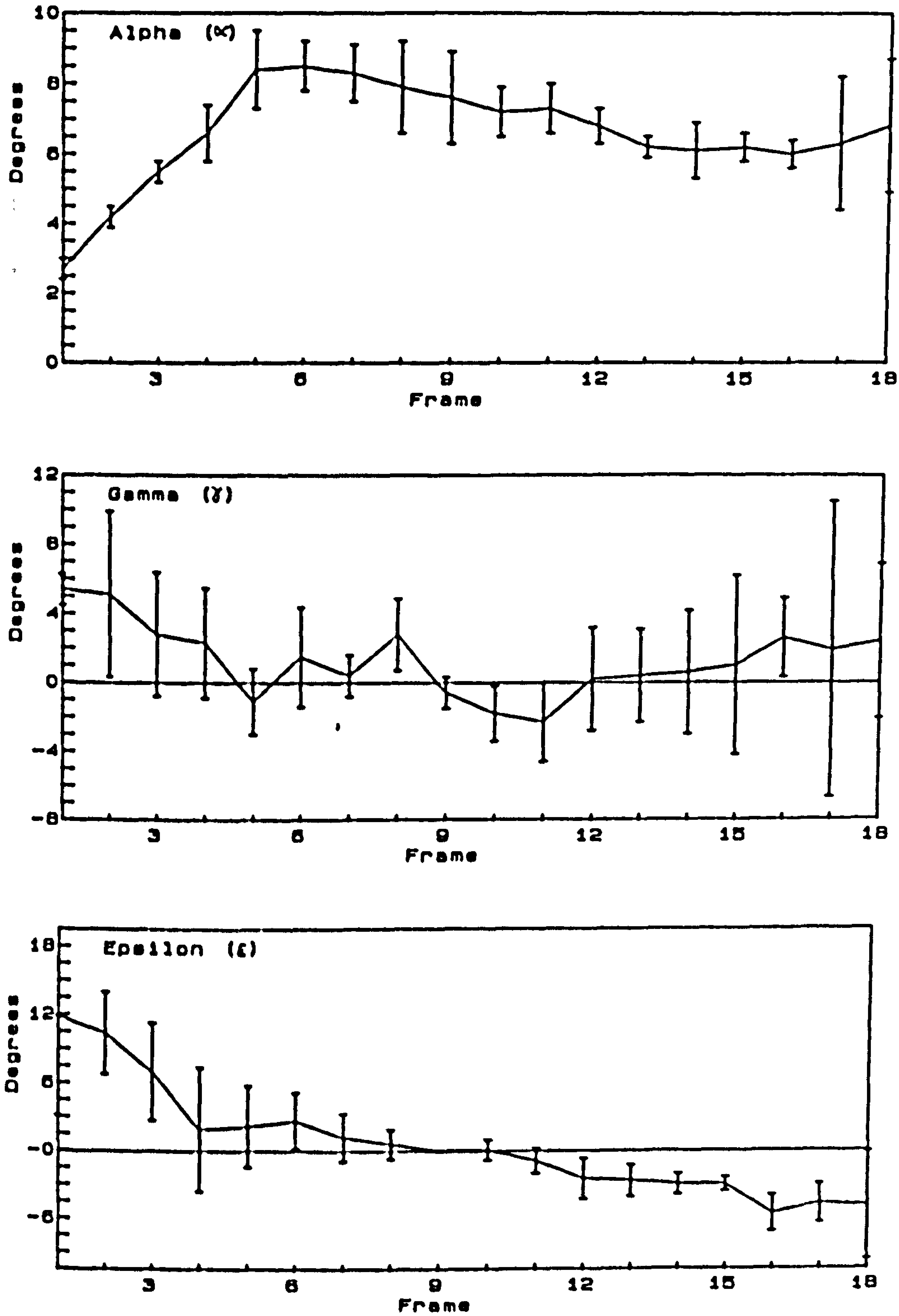


Figure 4.10: Mean and std deviations of the foot angles for the same subject tested over three different trials.

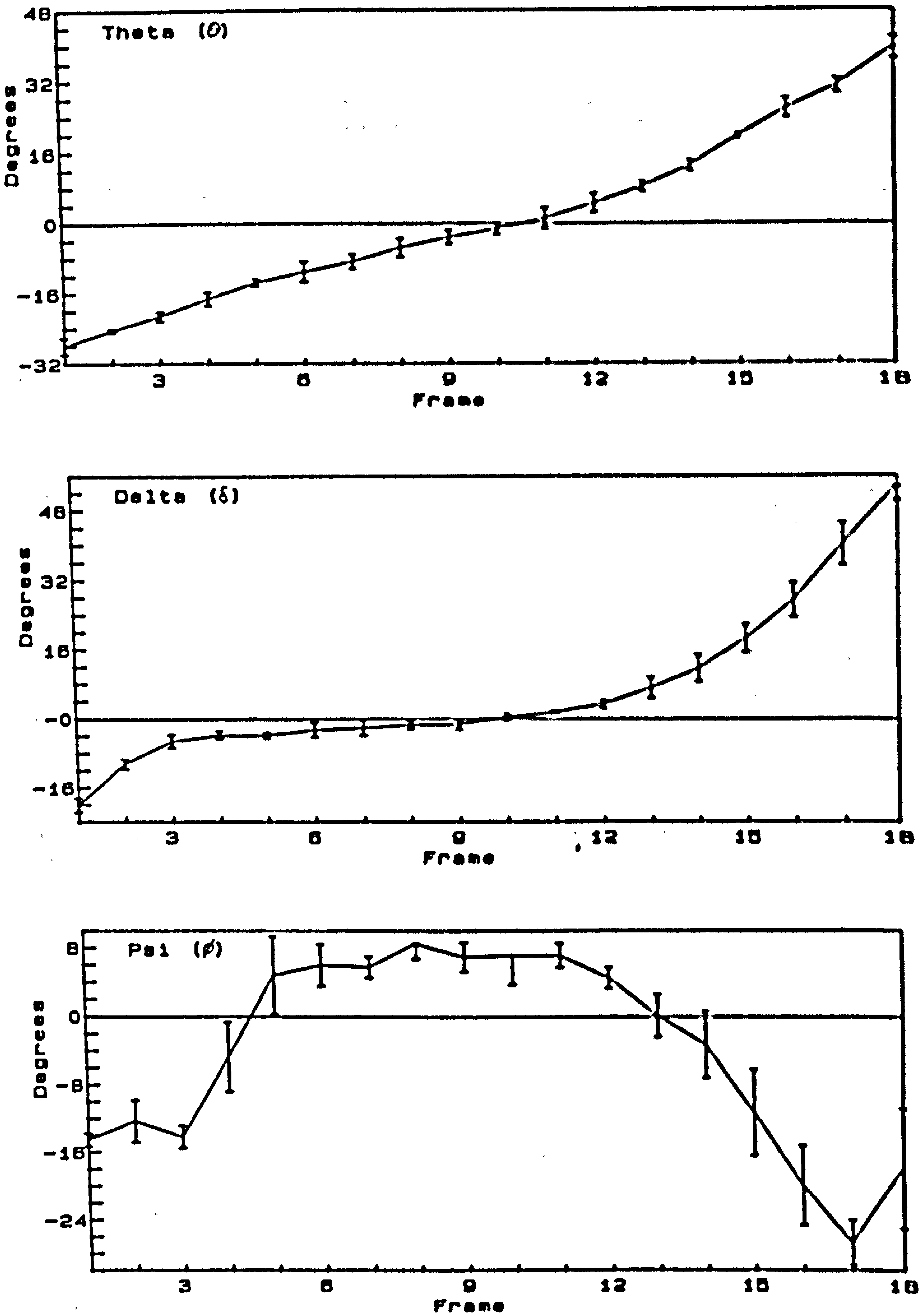


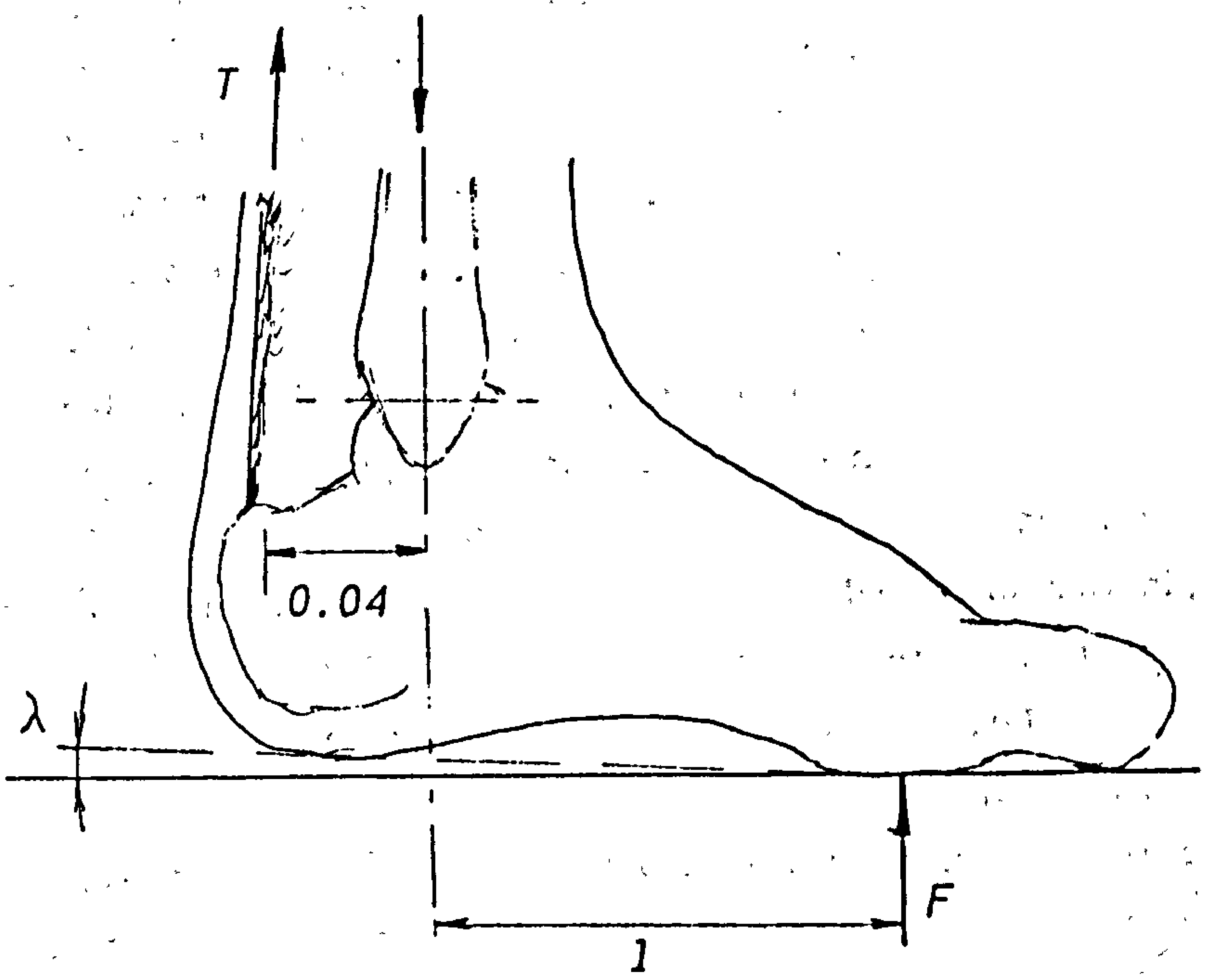
Figure 4.10 (cont): Mean and std deviations of the foot angles for the same subject tested over three different trials.

The two angles associated with the rear of the leg, α and γ, followed patterns that fit closely the expected behaviour. The leg angle, α, always remained positive so that the leg was inverted and this inversion increased toward midstance. This pattern would be necessary to maintain the centre of gravity above or near the support foot or feet base. The variation in this angle did not appear to be sex based. The calcaneal angle, γ, was interesting in that it maintained a vertical orientation during the majority of stance phase to provide a stable support base.

The difference between α and γ is the pro/supination angle of the rearfoot. The rearfoot was seen to be supinated up to 4° at heel strike and this supination increased to 8° during midstance. Toward late stance the ankle began to pronate and left the floor around 2° supinated. There was no apparent relation between static standing ankle pro/supination and the angles observed dynamically.

The forefoot angle, ε, was around 10° inverted at heel strike, decreasing to zero at foot flat and rising at push off to 4° eversion. These angular changes were found to be inversely correlated with time, the forces on the heel and the flexion angle of the toes. From the curves it is clear that the foot undergoes a steady pronating motion from heel strike to toe off.

Both the sagittal leg and foot angles, θ and δ respectively progressed as expected although it is interesting to note the point of inflexion in the graph of θ against time shortly after the leg is vertical. This 'pause' in the forward progression of the leg over the foot corresponds to the period in the gait cycle when the knee is extending so the progression of the hip joint remains steady in the forward direction and begins its upward progression for late stance and push off. The more anatomically significant graph is that of δ-θ (or ankle plantar/dorsiflexion) against time (fig 4.8g). This curve shows the foot plantarflexed after heel strike and then dorsiflexing up to 5° at 40% of the gait cycle before plantarflexing again to leave the floor at toe off 10° plantarflexed. The limited range of motion seen here is quite important when assessing the muscle behaviour. With reference to figure 4.11 it can be seen that up to 2.3° of the dorsiflexion motion can be accounted for by elastic stretch in tendo calcaneus. This leaves around 2.7° of motion to be absorbed by the triceps surae muscle group. This equates to a eccentric contraction of the muscles of around 2.8mm over a period of 0.3s. Under these circumstances it would be reasonable to



Young's Modulus for tendon (E) = 1.67 GPa (from Ker, 1981)
Cross sectional area of tendon = 0.585-0.89 cm^2 (from Procter, 1980;
Use A = 0.8cm^2 Ker et al, 1987)
Insertion of tendo achilles = 0.04m posterior to ankle joint
(from Procter, 1980)

If subject supports 1BW at the ball of the foot at heel rise then the ankle moment = F x l = (70 x 9.81)N x 0.13 = 89 Nm

Plantar flexing moment required to balance this => 89Nm
equivalent force in tendo achilles => T x 0.04 = 89Nm
T = 89/0.04 N = 2.22kN

Stress in tendo achilles (σ) = T/A
= 2220/80E-6 = 28 MPa

Extension in tendo achilles (ε) = σ/E
= 28E6/1.67E9 = 0.0166

Assuming tendo achilles length (L) is 100mm
Change in length of tendo achilles = εL
= 0.0166 x 100 = 1.7mm

Angular extension (λ) = $\tan^{-1}$(1.7/40) = 2.4°

Figure 4.11 The forces in the ankle plantarflexors near midstance and the resulting extensions in tendo calcaneus.

assume that the gastrocnemius and soleus are contracting isometrically.

Interestingly θ and δ were closely related to time (in the form of the frame number) and with contributions from the third metatarsal and the first metatarsal or heel included, the regression equation explained the majority of the variation (R^2=83-91). The toe angle is an unusual measure and was mostly correlated with θ and δ (R^2=60). It is interesting to note that the toes remained dorsiflexed (>9°) from heel strike until foot flat and become dorsiflexed again as soon as the heel left the ground reaching their maximal dorsiflexion angle of 25-30° in the last stages of push off as would be expected. There are at least two possibilities for the early dorsiflexion. Firstly it may simply be activity on the part of the toe extensors in support of tibialis anterior although this seems rather inefficient as none of these muscles have any major activity about the ankle joint, inserting as they do into the phalanges and the transverse ligament. The second possibility is based on the principle of the Hick's windlass mechanism acting on the plantar aponeurosis which would be tensioned by dorsiflexion of the toes. The aponeurosis would then be in a position to raise and tension the foot arches. The extent of the late stance toe dorsiflexion would be sufficient to similarly and more extensively tension the aponeurosis for support of the push off forces.

4.4.4 The Overall Pattern

For the present discussion it seems appropriate to look at the foot behaviour during the complete stance phase as a whole. This is best presented as two graphs which show the patterns of force and angle change based on the median values from all the feet (fig 4.12 & 4.13). It is in a way the 'typical' foot though it is important to realise that it is not the normal foot since, as has been presented 'normal' includes a wide range about these typical values. This section of the discussion will be continued in the final chapter when the results found here will be compared with those found in the muscular study.

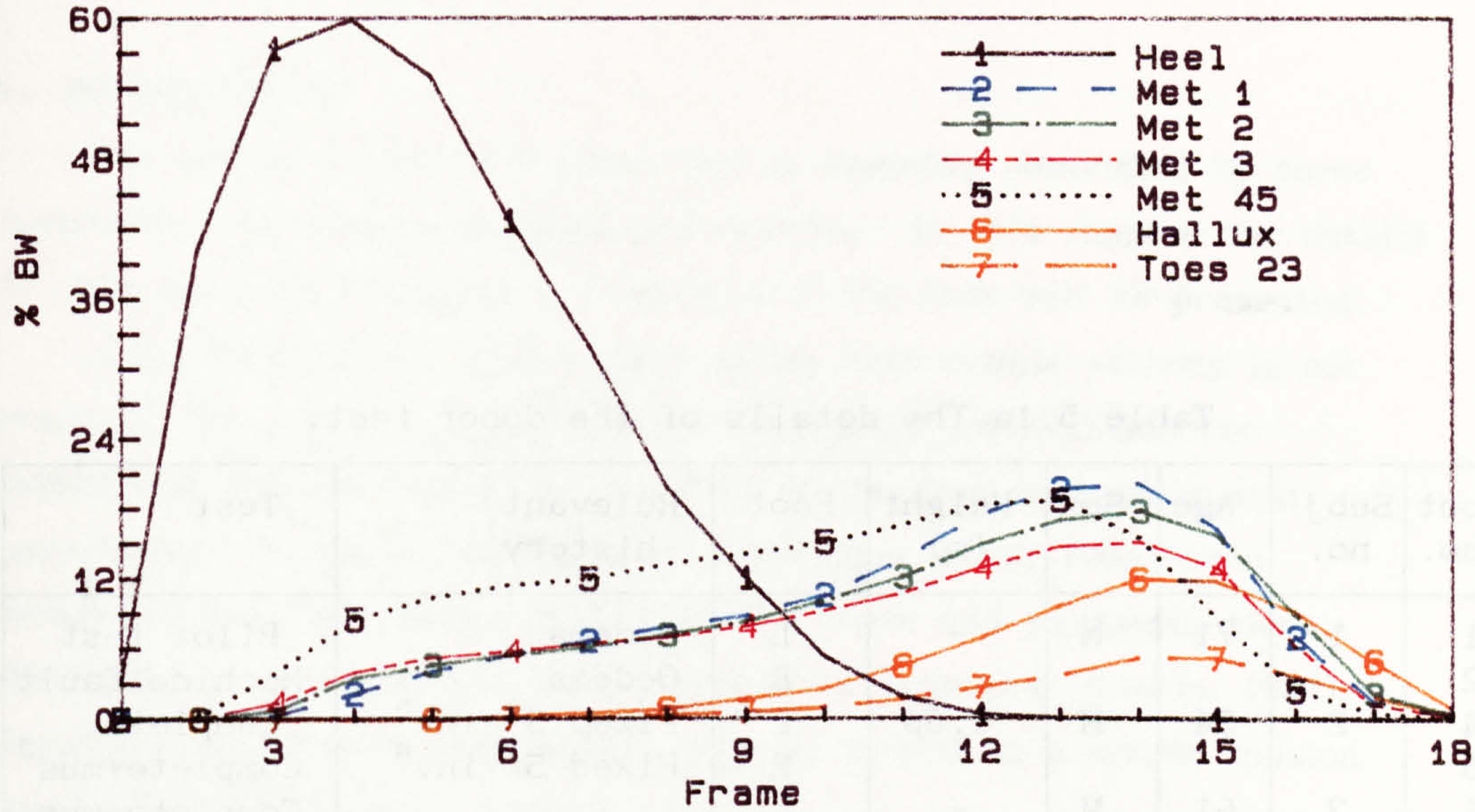


Figure 4.12 The median forces in the areas under the foot (N = 60).

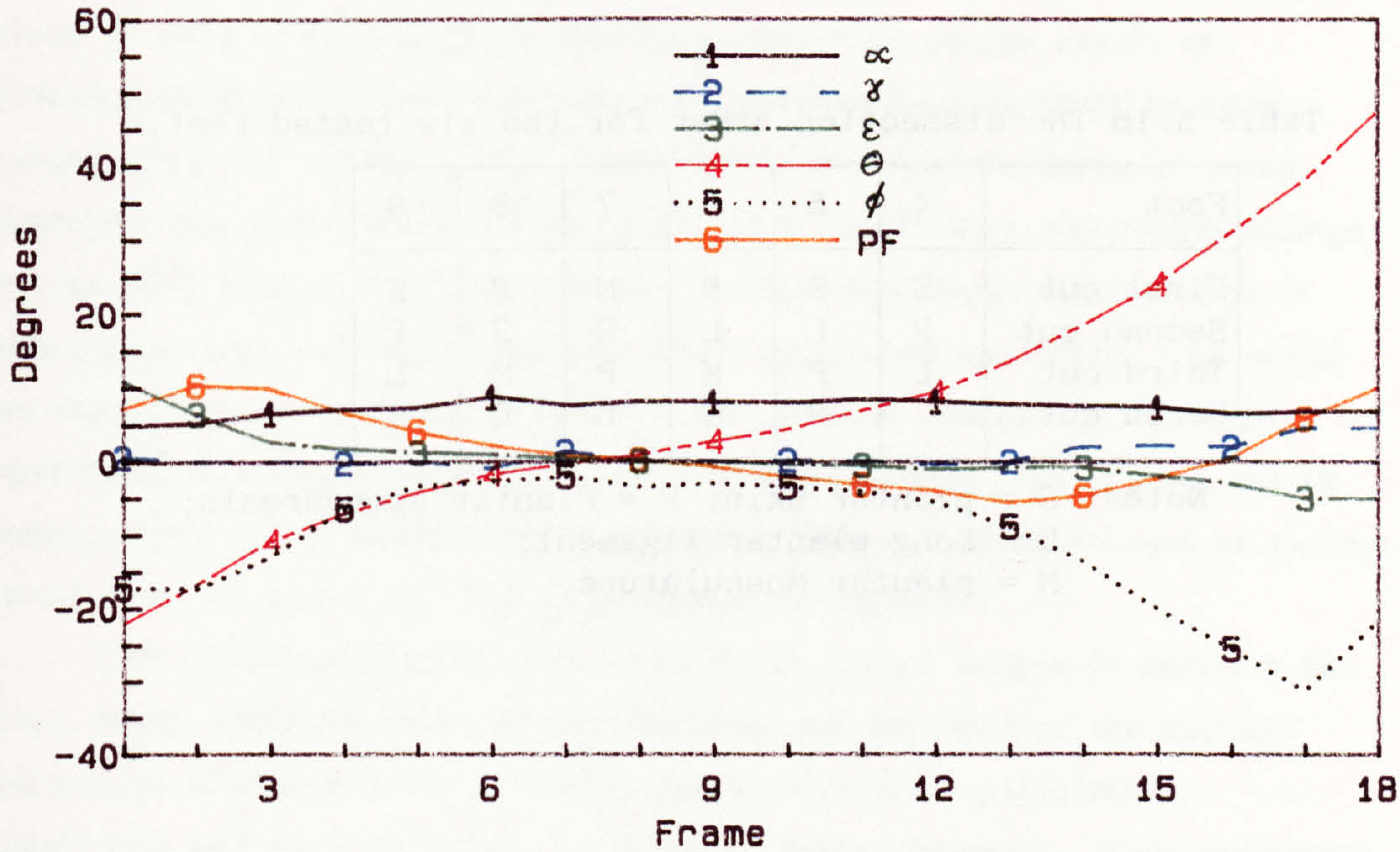


Figure 4.13 The median curves of six of the measured foot angles (N = 60).

Table 5.1a The details of the donor feet.

Foot no.	Subj. no.	Age	Sex	Height (m)	Foot	Relevant history[1]	Test
1	1	71	M	-	L	Oedema	Pilot test
2					R	Oedema	Machine fault
4	2	54	M	1.68	L	Fixed 5° inv^{n}	Complete
5					R	Fixed 5° inv^{n}	Complete+mus[2]
6	3	61	M	-	L		Complete+mus
7					R		Complete+mus
-	4	83	F	-	L&R	HV >30°	Not tested
8	5	90	F	-	L	HV 45-50°	Complete
9					R	HV 35°	Complete

Notes: [1] History provided from the medical records (where provided) or from visible signs. Inv^{n} = Inversion;
HV = Hallux Valgus (angle is the MTP angle)
[2] Complete+mus = Complete tests and tests on three extrinsic muscle tendons.

Table 5.1b The dissection order for the six tested feet.

Foot	4	5	6	7	8	9
First cut	S	S	P	M	P	S
Second cut	P	L	L	S	S	P
Third cut	L	P	M	P	M	L
Forth cut		M	S	L	L	M

Notes: S = plantar Skin; P = Plantar aponeurosis;
L = Long plantar ligament;
M = plantar Musculature.

CHAPTER 5 CADAVERIC TESTING OF THE LIGAMENTS OF THE HUMAN FOOT

5.1 INTRODUCTION

The action of the foot under load is arguably controlled by three structures: the bones, ligaments and muscles. In this chapter the details of tests made on the passive structures of the foot will be presented.

Since Basmajian & Stecko have shown that muscle activity is not necessary for static support there is reasonable justification for considering the load carrying capability of the foot without muscle involvement. Previous investigators (Ottevanger et al, 1989) have considered the load characteristics of the bones and ligaments alone, while others have modelled such a simplified structure (Simkin, 1982). Studies of the ligaments are often done by testing in a simple tension test out of their special support structure.

This experiment was undertaken to quantify the role of the passive soft tissue elements of the foot under dynamic loading. The foot was loaded through the tibia and fibula and the loads generated under the support areas of the foot were recorded.

5.2 MATERIAL

The work of Benick (1985) and others before him, has indicated that there is no serious change in specimen properties as the result of freezing fresh specimens and then subsequently thawing them to room temperature for testing. The preservation of cadaveric material using formalin and other preservation agents does result in a significant change to the soft tissue. For these tests the pathology department of one of the city's hospitals made available fresh amputated specimens. As would be expected, the securing of relative and medical consent to use such specimens in experimental work restricted the number, but during the 18 month tests ten specimens were obtained. Only a limited amount of patient detail was disclosed and this is presented in Table 5.1.

Two of the specimens were used in the initial stages to perform the pilot study using an older testing machine and two showed the distinct structural characteristics of hallux valgus and other pathological conditions and were therefore rejected as being abnormal. Four specimens were studied definitively as reasonably normal feet while two further feet that showed the clinical signs of hallux valgus were studied for a preliminary assessment of the changes resulting from the condition.

5.3 METHOD

5.3.1 Specimen Mounting

Following amputation each specimen was stored in a dedicated tissue freezer at -20°C. For the experiment the foot was thawed to room temperature. A brass support collar was manufactured for the experiments with an inner diameter of 60mm and a depth of 100mm. This collar was of a split tube construction (for releasing the specimen at the conclusion of the tests) with a support cover over the top end. The tibia and fibula were mounted in the support collar using bone cement (Acrulite Microtech® Type A).

5.3.2 Loading System

An Instron® 5010 Materials Testing Machine (MTM) was used for the tests with the specimen mounted below the crosshead. The speed of testing was an important consideration. If a normal walking style has a stride frequency of say 0.8Hz or greater then the period of stance phase is about 60% of the period (1.25s) or 0.75s. So the foot is loaded and then unloaded in under 1s. From the work of Simkin (1982) the vertical deflection of the talus under 500N (approximately 70% body weight) can be expected to be near 3.5mm. The resulting extension rate necessary to achieve the extension in 0.38s (half the period) is

$$\text{Deflection/time} = 3.5/0.38 \text{ mm/s} = 9.2 \text{ mm/s}.$$

In order to assess the effects of loading rate on the foot a series of extension rates around the "physiological" speeds were required. To maintain consistency between tests the system's computer control was programmed (Appendix 2) to provide the load regime. Three speeds were used during each test:

Stage 1	3 cycles (0- 600N)	at 4.16 mm/s	(250mm/min)
Stage 2	3 cycles (0- 800N)	at 8.32 mm/s	(500mm/min)
and Stage 3	3 cycles (0- 1000N)	at 13.33 mm/s	(800mm/min).

The increased load ranges were not intentional but a result of the inertial characteristics of the machine operating at the high speeds over a very short extension. The limits on the computer were successively reduced to contain the load levels to those shown above (less than 200% BW).

The specimen mounted in the support collar was attached to the special coupling of the Instron load cell. The collar support restricted tibial rotation and in/eversion so particular care was taken to place the

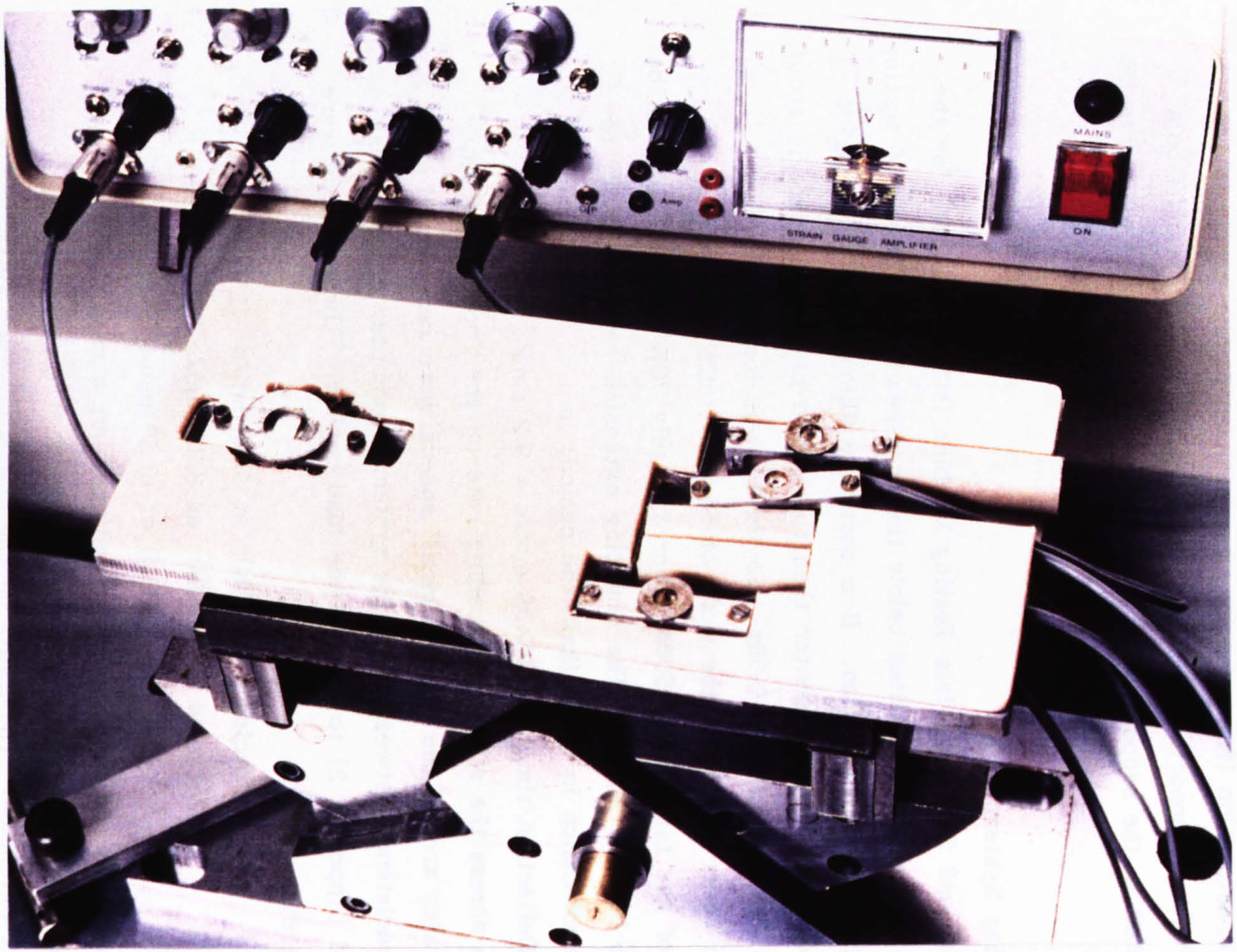


Figure 5.1 The support platform (inverted) with the four load cells in place and the amplifiers behind. (Fig 5.5 shows platform set for plantarflexion)

limb vertically under the support with neutral rotation. The plantar surface of the foot was supported on a specially constructed platform to allow the setting of plantar/dorsiflexed and in/everted angles.

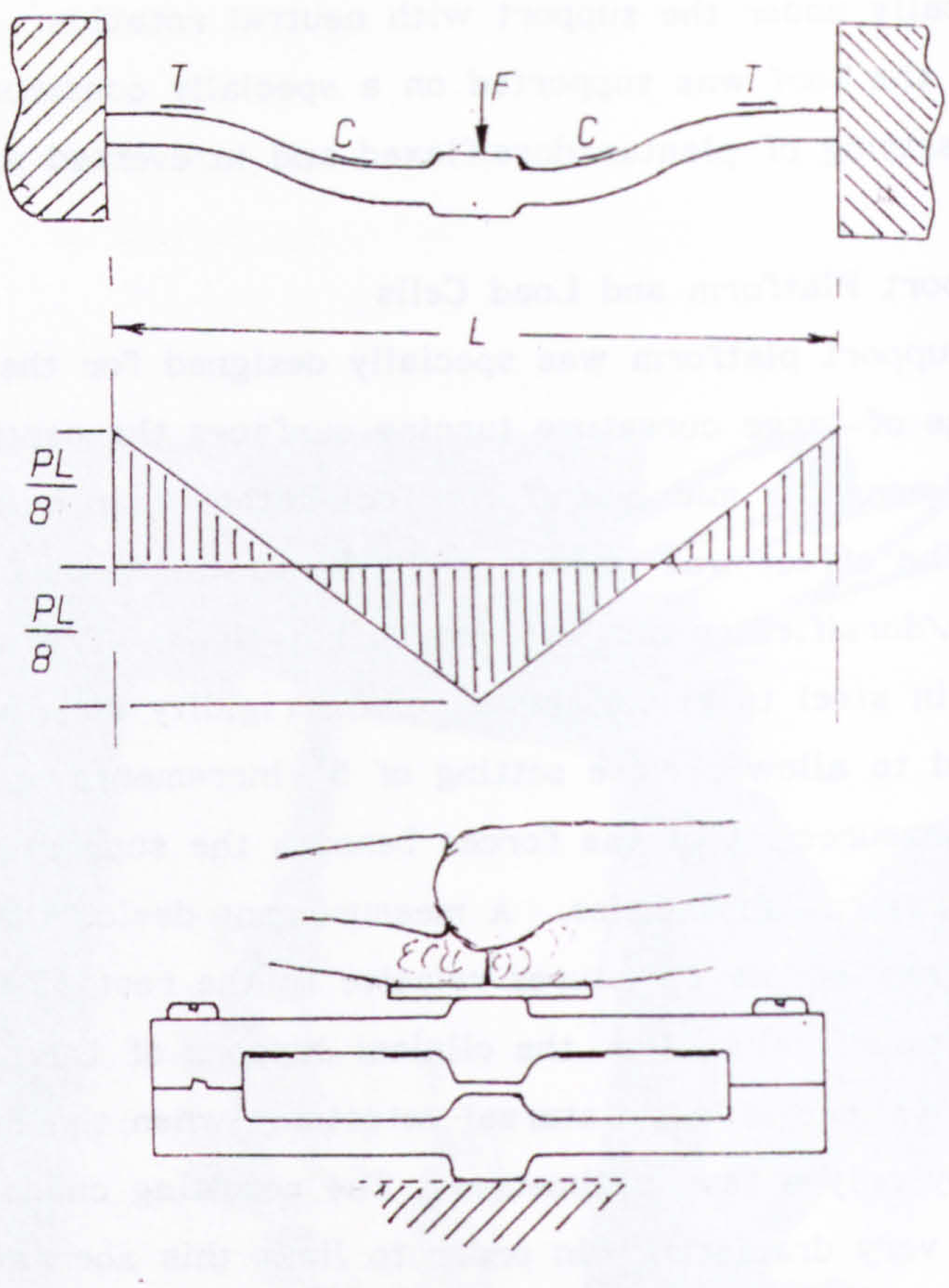
5.3.3. Support Platform and Load Cells

The support platform was specially designed for the tests (fig 5.1). Through use of large curvature turning surfaces the centre of curvature was placed near the mid line of the foot rather than below the platform surface. The effect was to reduce the linear displacement as a result of the plantar/dorsiflexion and in/eversion rotations. The assembly was fabricated in steel to give it strength and rigidity and location holes were drilled to allow precise setting of 5° increments about both axes.

The measurement of the forces beneath the support surfaces of the foot posed several difficulties. A measurement device that alters the position of any of the structures relative to the rest of the foot will not give 'typical' values (for the clinical support of this consider the case of the failure of a metatarsal osteotomy when the head may become depressed by only a few millimetres. The resulting change in the support pressure is very dramatic). In order to limit this aberration the gauges should be part of a support surface and be relatively rigid. The zones of interest beneath the foot can be as small as 64 mm^2 and the gauge should be capable of supporting this area without interfering with other zones. The forces transmitted by these zones was also quite variable from a maximal level of 1500N beneath the heel to less than 10N beneath the lateral metatarsal heads in certain positions.

All the instruments associated with these tests of human tissue must be thoroughly cleanable, normally in powerful bleaches and disinfectants. The testing conditions are not ideal with significant levels of movement occurring as well as some instances of overloading force (>2000N). Failure of a gauge during a test can seriously disrupt an experimental session. The number of specialist requirements outlined as well as cost limitations, made commercial units unsuitable for this application. It was decided to design and fabricate special load cells.

A strain gauged beam was considered the simplest solution to the above requirements. By constructing the beam into an arrangement where it was loaded by reverse bending (fig 5.2a) the transducer could satisfy the criteria for rigidity and sensitivity. The construction shown in figure 5.2b was calculated as being appropriate for the application and the use



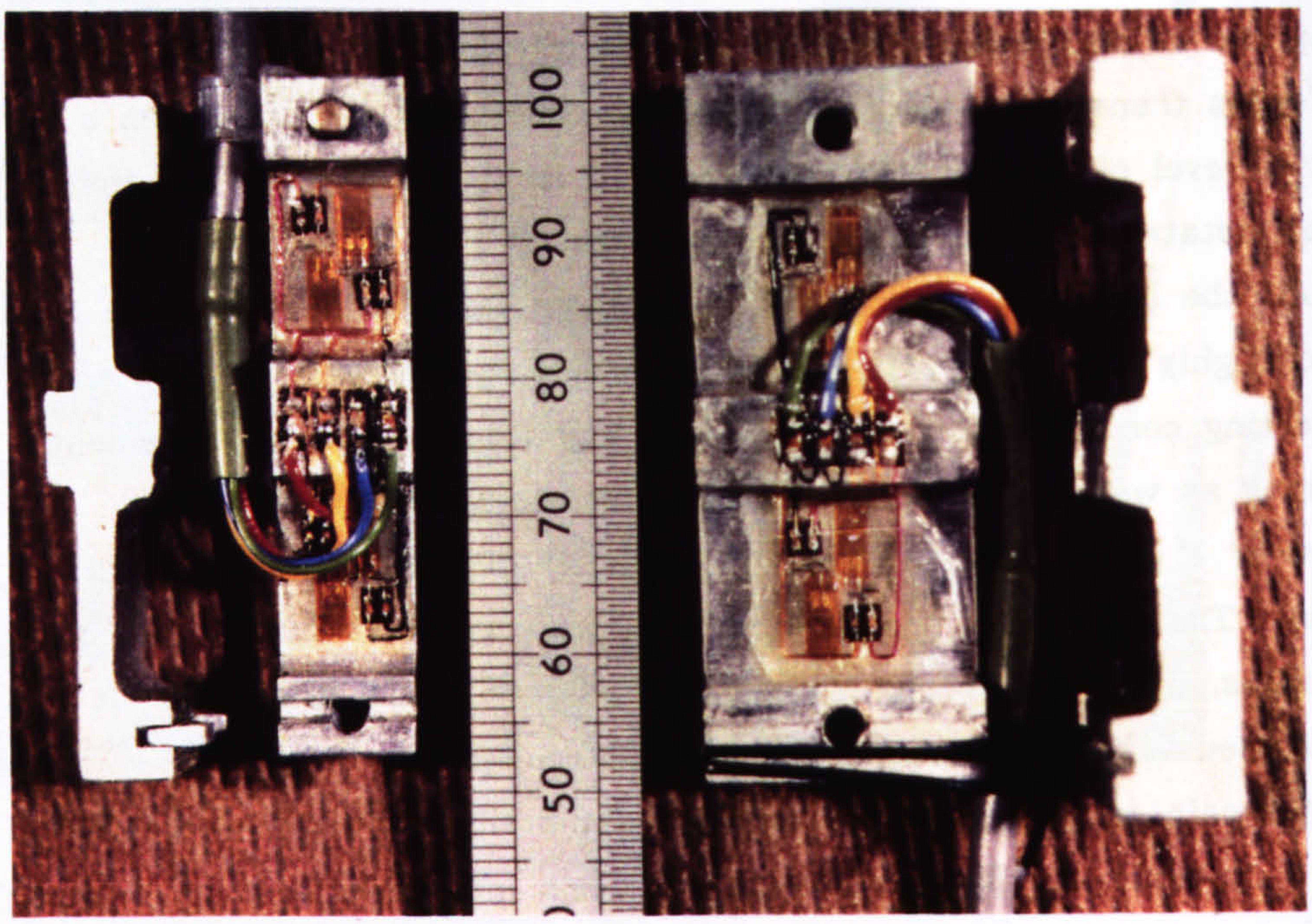


Figure 5.2a) (top) The principle of reverse bending.
b) (bottom) The application of reverse bending to the design of the load cells

of four strain gauges enabled a full bridge system to be wired that was sensitive to 4x the applied load. The reverse bending construction was inherently insensitive to temperature effects and to offset loadings (providing the load occurred within the raised centre region). The use of the central 'bulge' provided the overload protection, and calibration of the transducer indicated a linear and consistent response to vertical load (fig 5.3). The construction was the same for the heel transducer though section dimensions were adjusted to suit the higher loads. A sealing layer of polyurethane and silicone rubber proved to be sufficient to allow cleaning in the normal laboratory disinfectants.

Four load cells were constructed including one heel transducer. For the tests they were placed under the heel, and the first and fifth metatarsal heads. The fourth gauge was placed under the second metatarsal head for foot flat tests and under the hallux for plantarflexed positions. The gauges were movable on the support platform to cater for the range of foot sizes and polyurethane blocks were placed to provide support for those areas not carried by the load cells. The load cells were all connected to a full bridge amplifier with bridge voltage set to 3V and gain of 1-2k. The output from the amplifier and from the interface panel of the Instron MTM were connected (fig 5.4) to a PC computer mounted A/D conversion card and sampled in real time by in-house data acquisition software (Phillips, 1989). This separate computer digitally logged the four load cells under the foot, the Instron load cell (and so the load applied to the tibia and fibula) and the extension of the Instron crosshead.

All the load cells (with the exception of the Instron load cell) were calibrated using dead weights or the Instron load cell as a reference, before and after each series of tests. Throughout the study the gauges remained linear and consistent, though fatigue of the wiring of two gauges necessitated some repairs after which the proportionality constant was found to be slightly altered.

A tensioner was available on the loading collar to tension the tendo achilles for the plantarflexed positions. Dead weight pulley tensioners were also included around the support collar to provide near vertical tension on several of the long tendons from the foot (tibialis anterior, peroneus longus and extensor digitorum longus).

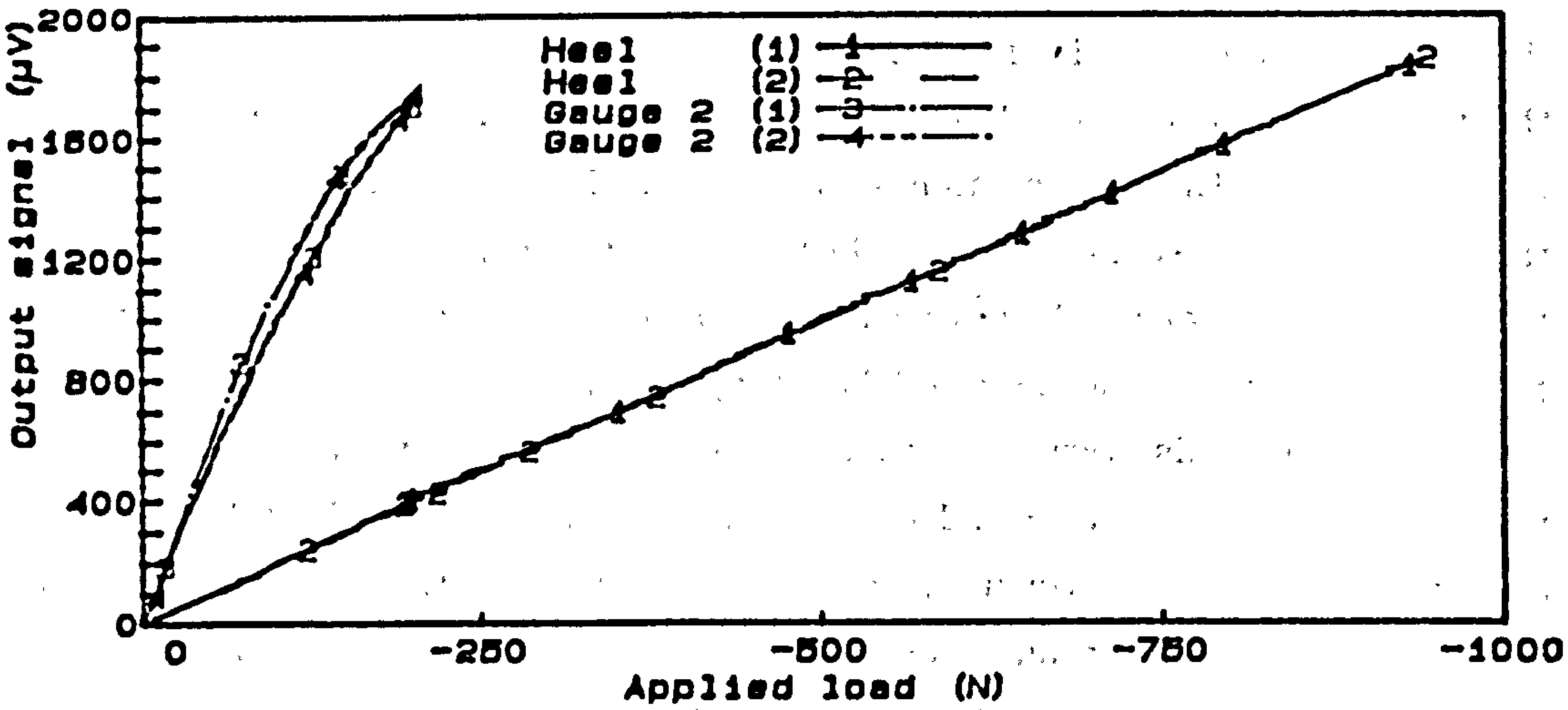


Figure 5.3 Calibration curves for two of the foot support transducers

5.3.4 Testing Procedure

Each foot was tested in five positions :- neutral, 10° dorsiflexed, 20° plantarflexed, 10° inverted and 10° everted. An extra test was applied to three of the feet (feet 5-7). In these three feet the tendons of tibialis anterior, peroneus longus and extensor digitorum brevis were tensioned in succession using the system of dead weights and pulleys with the foot in a neutral position (fig 5.5). The first stage under computer control (250mm/min) was run without tension in the tendon. During the pause associated with the change in testing speeds a 5kg weight was added to the weight system hanger, therefore applying the tension in the tendon during the subsequent stage (500mm/min). During the second pause between stages 2 and 3 the weight was removed and the final stage (800mm/min) tested the foot with no tension in the muscle tendon.

A method of sequential dissection was used to remove the plantar structures each in turn. The order of dissection was changed for each foot in an attempt to gain a better understanding of the in vivo situation (Table 5.1b). Specimens were maintained in a moist condition with plastic covers and an antibacterial spray (Dettox®).

5.3.5 Data Analysis

The analysis procedure was done immediately following a test series. During the testing procedure the acquisition program was only capable of producing a time tracing of the various voltage waveforms and these were used to detect saturation conditions and other data abnormalities.

Two Pascal programs were written for the postprocessing. Each test file contained the details from the transducers for the three speeds during the tests. The program CADPROC (Appendix 4) was developed to read one of these test files, calibrate the data to produce output in N or mm and extract the periods of interest out to files. The resulting files were called RECord files and were analysed using GRAPHER (Appendix 4). This program read in up to six REC files and displayed any data channel versus any other channel. The graphs could be scaled as required and mouse control was provided to determine the XY coordinates within the graph area and to mark straight line approximations with their numeric equations for the curves.

Two graphs in particular were found to be the most informative. The graph of the applied load (from the Instron Load Cell) versus the crosshead extension indicated the overall support behaviour of the foot.

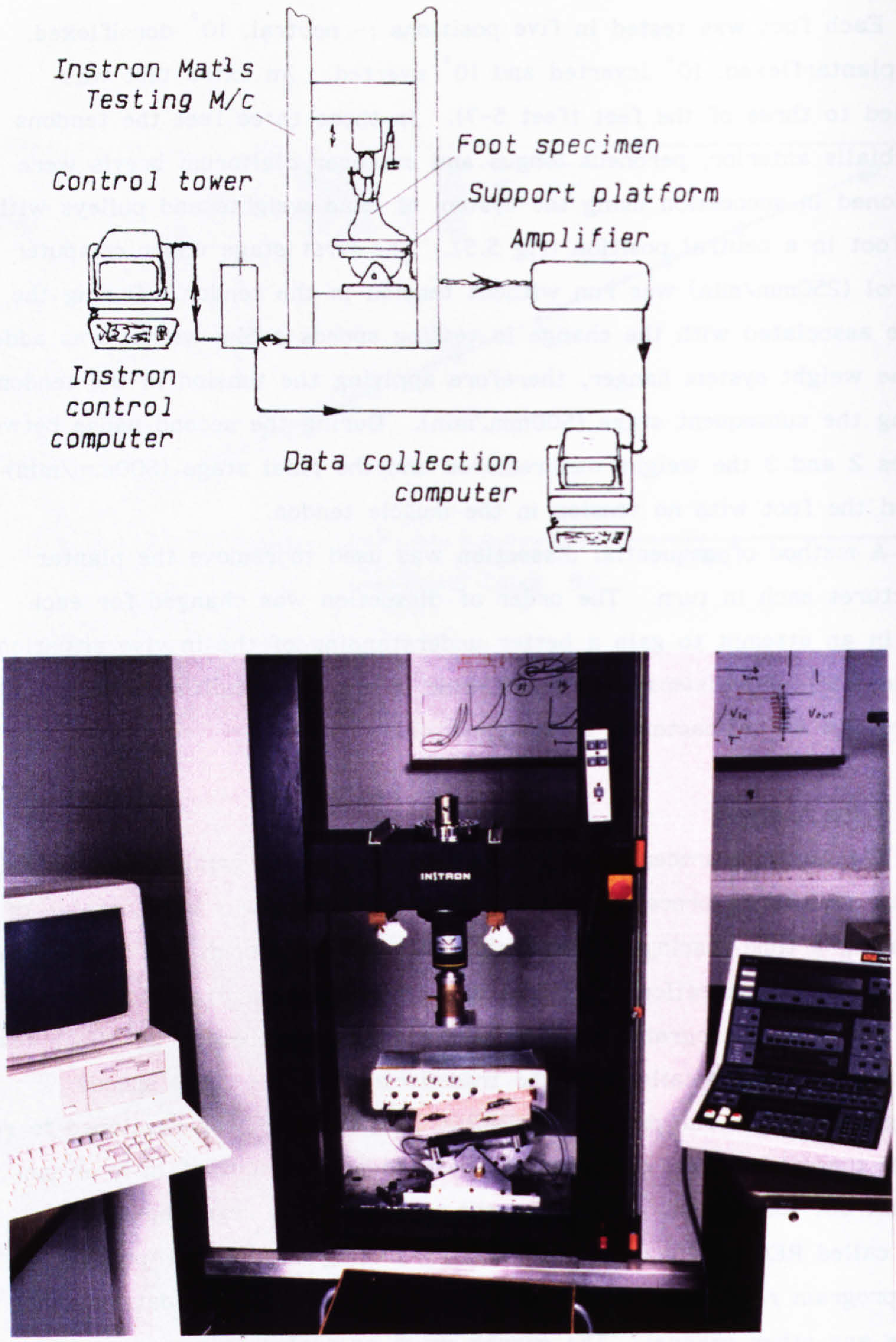


Figure 5.4 The in vitro testing system, (top) the schematic representation of data flow illustrating the two computers, (bottom) the test equipment showing the support collar and support platform in the centre with the control computer left and the system console right

Secondly the curve of the load beneath a support area versus the load applied to the tibia and fibula provided a clear indication of the load distribution during the loading regime. The slope of the load measured vs load applied curves (found using the straight line approximations in GRAPHER) was equivalent to the fraction of the total applied load that was supported within the area. The intercepts were disregarded as they were generally the result of the initial offsets applied to the signals to prevent them saturating the amplifier at maximum loads.

5.4 RESULTS

5.4.1 Introduction

Unfortunately due to the small number of specimens that were tested that could be regarded as normal it was not considered reasonable to apply statistical analysis to the data so the results for each specimen will be presented.

5.4.2 Applied Load versus Extension.

All the specimens showed typical nonlinear load-deformation behaviour (fig 5.6) in keeping with their biological nature. As expected the deformation required to produce considerable force when the heel is down was quite small. In the tests the specimens were placed in the testing position and the crosshead lowered until a nominal 10N compressive force was measured at which point the extension measurement was reset to zero. In such circumstances the first full load cycle is likely to settle the foot articulations after which the no load condition may occur when the crosshead is lower than the previously set zero point. For these reasons the first load cycle was eliminated from all results and the change in extension from no load condition to loaded was considered more reliable than an absolute value based on a preset zero.

The feet in all tests became stiffer as more of the plantar structures were removed. This is slightly misleading since the feet were also 'collapsing' steadily at the longitudinal arch so that relative to the initial loading position the extension may well have remained the same although the loading did not occur until the latter part of that extension. This increasing stiffness became a serious hindrance to the higher speed tests. At 800mm/min the crosshead will cover 4mm in about 0.3s. These tests were at the limits of the mechanical testing machine's capability and faster or more tightly controlled tests would certainly

Figure 5.5 A foot specimen mounted in the testing system. The tibia and fibula are fixed in the support collar and the base of the foot rests on the load cells on the support platform. In this case tendo achilles is tensioned to produce plantarflexion and a weight has been added to produce tension in one of the extrinsic muscle tendons. (The plastic cover is removed for the illustration)

require some form of servo hydraulic actuator.

There was also a noticeable level of hysteresis in the curves but the patterns remain elastic since they reload along the same line. The hysteresis is associated with energy loss and appears to be of the same levels found by Ker et al (1987).

5.4.3 Load Distribution

As mentioned previously the load distribution was assessed most effectively by graphing the load under a particular area against the load applied through the tibia and fibula. Nearly all these graphs were independent of the testing speeds (fig 5.7) and approximately linear (fig 5.8).. The lateral ray (metatarsal 5) did show a convex shape that was quite characteristic (fig 5.8). The slopes of the curves were calculated on the sections from 150 to 600N applied load. The actual slope values for the complete feet and the feet with all the major plantar structures removed (excluding the interosseus ligaments) are tabulated in Tables 5.2 and 5.3 respectively. As a result of the different orders of the sequential dissection the various load distributions after each ligament was removed are very foot specific. The results (Tables 5.4 to 5.6), therefore, are based on the *change* (in % applied load) from the previous loading pattern as a result of the dissection.

5.4.4 The Effect of Tensioning Three of the Extrinsic Foot Muscles

For these tests the differences between the load distributions on the feet were examined for the three stages of each test; the second stage containing the detail of the foot with tension applied to the appropriate tendon. The differences in the support load between stages 1 & 3 (tendon loose) and stage 2 (tendon tensioned) were calculated for each measurement area. The results provide an indication of the change in the load distribution due to tension in each of the tested tendons. The effects of tension in the tendons were found to be quite consistent irrespective of the degree of dissection and so the results from all tests on the tendons are grouped and those for the forefoot areas are presented in figure 5.9. The heel loads showed a very wide variation both within and between subjects and so no clear findings were available. The mean and associated standard deviations (in brackets) of the load changes under the heel for all cases was -0.8%(6.4%), 3.8%(9.3%) and 2.93%(3.55%) for tibialis anterior, peroneus longus and extensor digitorum longus respectively and

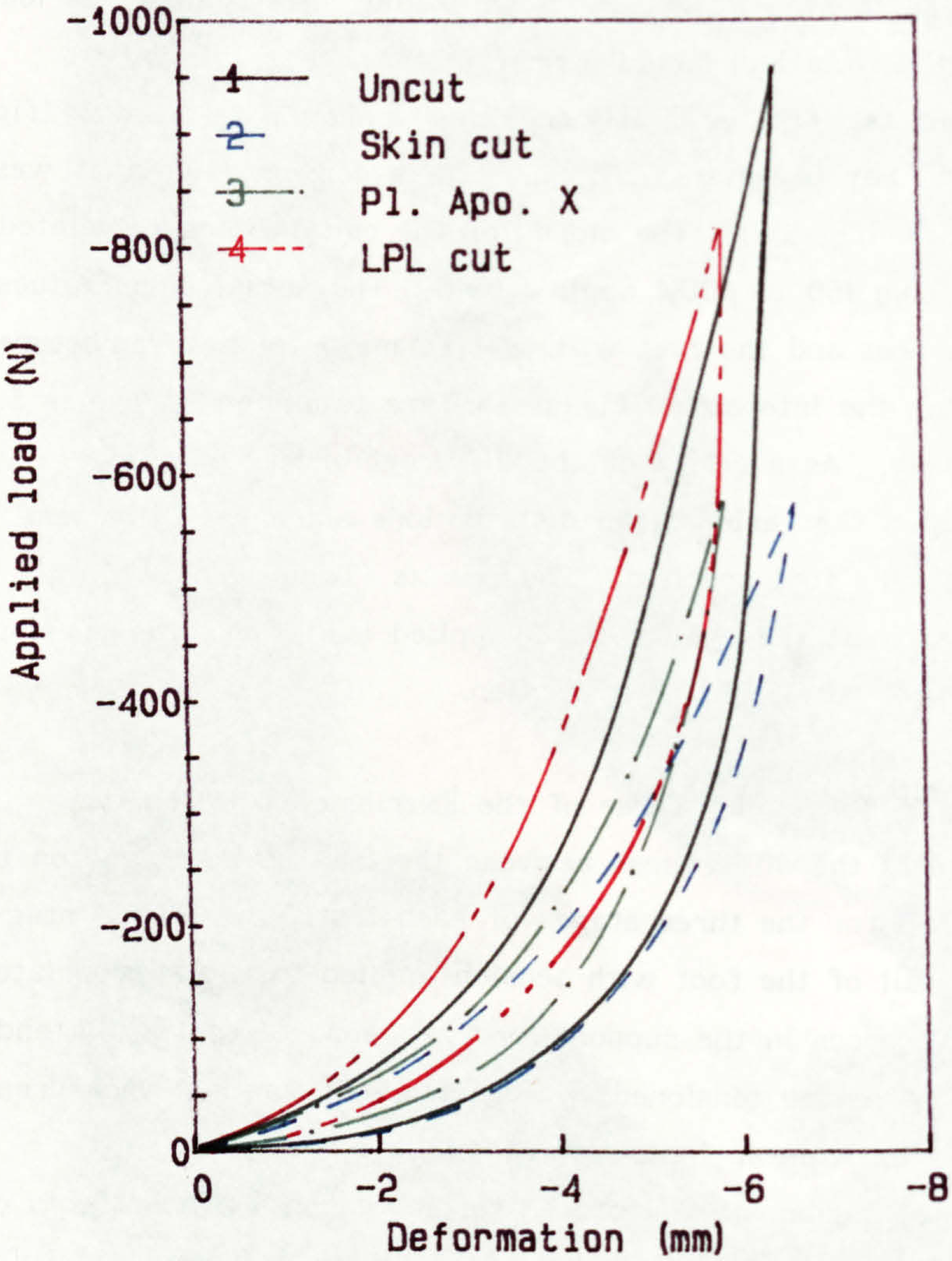


Figure 5.6 Typical load versus deformation behaviour of the feet indicating the changes as a result of the removal of the plantar tissues.

only the last mean value is significantly different from 0 ($p<0.05$).

5.4.5 The Two Feet with Hallux Valgus (feet 8 & 9)

The results for these two feet are presented in tables 5.7 to 5.11. Foot 9 has a dissection order almost the same as foot 4 although the subsequent discussion will consider the differences between the two hallux valgus feet and the other four feet as a group. In general the forces on the lateral aspect of the foot are higher for the feet with hallux valgus when compared with the four normal feet.

5.5 DISCUSSION

5.5.1 Introduction

In order to consider the various aspects of the loading patterns the four feet will be considered together firstly complete and then as different plantar structures are removed before finally considering the overall change after all structures are cut. The removal of the muscular tissue produced only limited and very variable loading changes under the foot and so was not considered separately.

5.5.2 The Complete Foot

The degree of variation associated with these tests is immediately apparent when considering the complete feet. There are several reasons for this variation such as movement of the load cells beneath the support areas, and attempts to eliminate it, unless very carefully conceived may severely constrain the foot behaviour.

The heel supported the highest proportion of the applied force throughout the tests. In the neutral position this level ranged from an unusually low 36% of applied load for foot 5 to 71% for foot 6. Levels between 55 and 70% of applied loadseem normal from other tests and would correspond to the force levels reported in Chapter 4 for normal bare foot walking at early to mid stance (Table 4.2). The metatarsals carried from 4 to 9% of the applied load between them. Feet 4,5 & 7 distributed the majority of the load on the lateral side of the foot with the first metatarsal supporting less than half that borne by the fifth. Foot 6 was the exact opposite of this although no reason was apparent. The records from normal bare foot walking (Table 4.2) indicate that the median distribution of supported force was 5%:5%:6% from the medial to the lateral side of the three zones of interest.

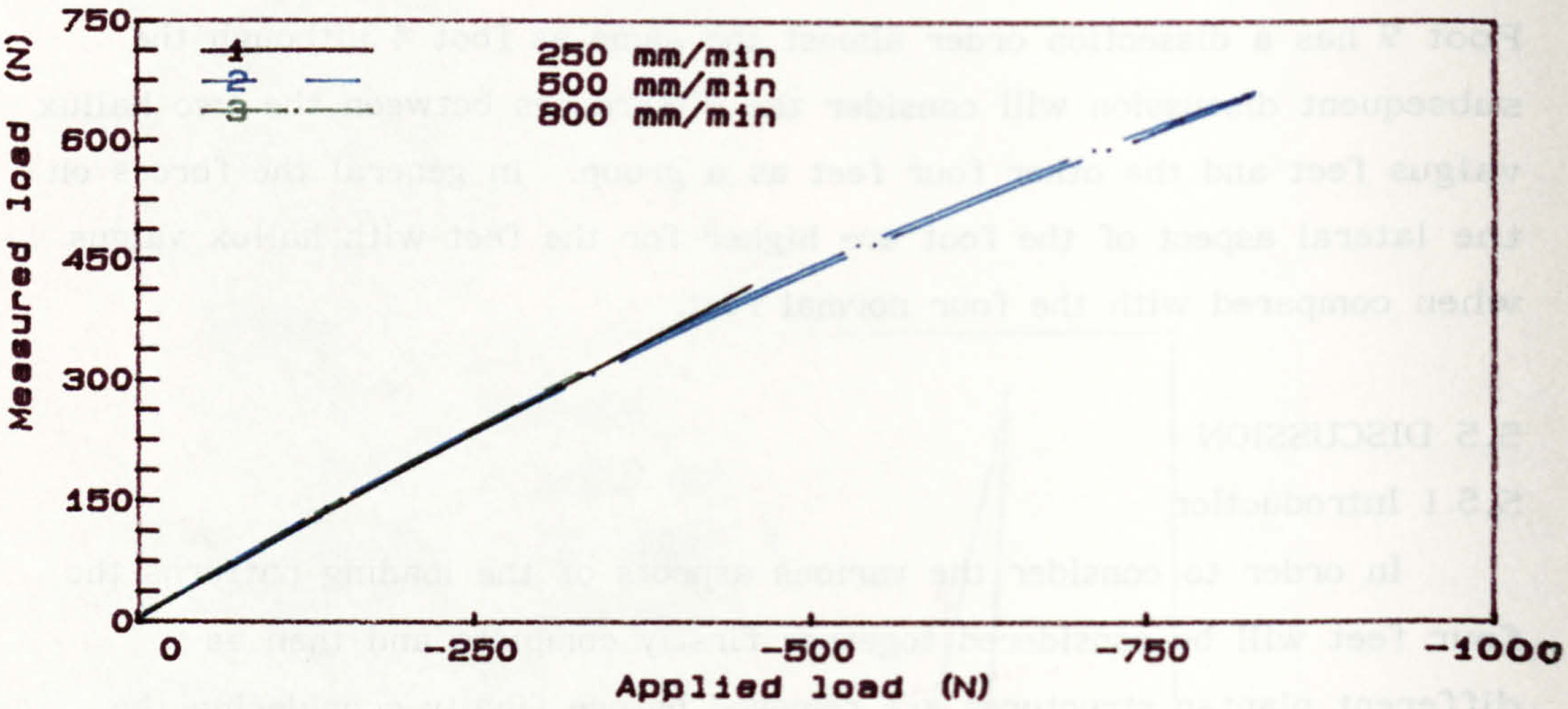


Figure 5.7 The loads measured under the heel in the inverted position at three different loading rates (250,500 & 800 mm/min)

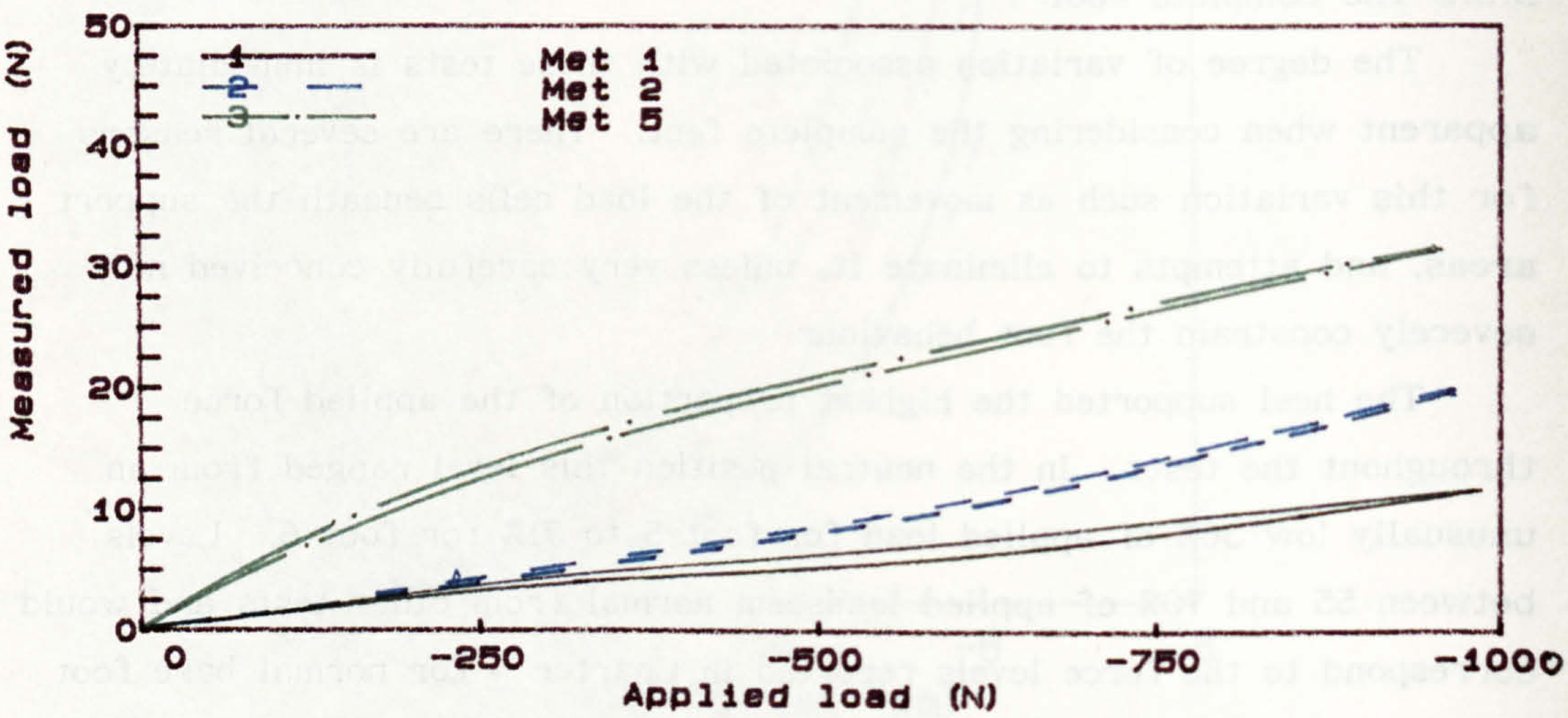


Figure 5.8 The loads measured under the metatarsal heads versus the load applied to the tibia and fibula in the inverted position.

When inverted 10°, all the feet showed a decrease in heel support except foot 5 which almost doubled its support (bringing it to a more typical level). As should be expected the medial side of the forefoot increased its load share (50% relatively) while there was only a slight or no decrease in the lateral rays. Foot 7 was the exception this time transferring load laterally. This variation can be due to a very loose forefoot (the intermetatarsal ligaments) that allows the metatarsals to roll on the sloping platform thus loading the lateral rays more than would be expected.

In eversion there was a slight increase in loading of the heel relative to the neutral position, except in foot 4. This increase may be due to the limited pronation that is generally found at the ankle. The medial ankle ligaments may tense earlier than their lateral counterparts and transmit a greater proportion of the load through the heel. There was a consistent shift of the forefoot forces toward the lateral border. The first metatarsal showed up to a 33% decrease in support while the fifth remained unchanged or increased up to 15% applied load.

Dorsiflexion resulted in a general rearward shift of the loads. The heel forces increased or remained the same (foot 6 fell however) while the forefoot forces decreased or shifted toward the lateral rays (foot 4 rose to 20% on metatarsal 5).

Tensioning tendo calcaneus and putting the foot into plantarflexion produced a wide variation in the load distribution on the forefoot. This was most likely due to the unrestrained nature of the talar joint in plantarflexion allowing more pro/supination than in foot flat positions. Forefoot forces ranged from 16 to 68% of the applied force and were generally higher toward the lateral rays (foot 6 was more highly loaded medially). The hallux supported between 3 and 26% of the loads.

5.5.3 Loss of Plantar Skin

The plantar skin is one of the structures that is often removed before tests begin on the foot and there is seldom much attention paid to its contribution to the overall characteristics. In the neutral position the loss of the skin resulted in a small increase in the load support of the heel and a similar net increase to the forefoot. The distribution was more variable with feet 4 & 5 transferring some load to the fifth metatarsal while foot 7 transferred load medially.

Table 5.2 The force measured under the areas of the normal feet uncut.

(expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 4				
Neutral	57.4	1.0	3.1	4.9
Inverted	50.2	1.4	1.6	2.3
Everted	43.3	0.0	0.3	4.0
Dorsiflexed	85.2	0.3	0.9	20.0
Plantarflexed	*	2.6	26.4	16.9
Foot 5				
Neutral	36.0	0.5	0.6	3.0
Inverted	71.4	2.1	1.8	2.5
Everted	58.3	1.2	0.0	15.0
Dorsiflexed	44.0	0.2	1.2	3.7
Plantarflexed	*	4.5	13.5	50.3
Foot 6				
Neutral	70.6	4.8	3.0	0.4
Inverted	65.3	6.9	2.6	0.5
Everted	75.0	2.6	2.2	1.2
Dorsiflexed	49.1	3.5	3.0	0.2
Plantarflexed	*	16.3	5.5	3 0
Foot 7				
Neutral	68.9	1	1.5	2
Inverted	46.8	0.4	0.8	3.5
Everted	70.5	0.6	1.1	2.5
Dorsiflexed	65.3	0.6	1.3	2.3
Plantarflexed	*	4.3	3.1	9.1

Table 5.3 The force measured under the areas of the feet with all plantar ligaments cut (expressed as % force applied to tibia/fibula).

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 4				
Neutral	61.3	1.1	1.8	7
Inverted	61.5	2.5	3.5	3.2
Everted	63.5	0.1	0.4	10.5
Dorsiflexed	100	0.2	2.7	20.4
Plantarflexed	*	1.1	3	2.9
Foot 5				
Neutral	43.3	1.6	2.2	4.8
Inverted	29.9	3.1	2.4	1.1
Everted	46.7	0.6	1.5	11.1
Dorsiflexed	36.4	2.1	2.7	2
Plantarflexed	*	5.5	7.2	0.3
Foot 6				
Neutral	48.5	5	1.5	3.8
Inverted	49	4.3	2.7	3.1
Everted	50.2	2.6	1.7	8.3
Dorsiflexed	57.8	3.9	2.2	3.7
Plantarflexed	*	38.5	7.4	2.1
Foot 7				
Neutral	62.1	2.8	3.4	3.1
Inverted	53.2	1.3	2.2	4.2
Everted	54.7	1.5	3	3.5
Dorsiflexed	63.3	1.8	2.5	3.5
Plantarflexed	*	9.2	3.9	13

Notes for both tables:

Met = Metatarsal head;

Met 2/Toe 1 : The gauge was placed under the second metatarsal head for all positions except plantarflexion when it was placed under Toe 1 (Hallux).

Inversion also resulted in higher loads on the heel and similar load transfer onto the lateral rays of the forefoot (except for foot 5). Eversion was even more marked (although for foot 7 there was a decrease in the heel support) with load transfer laterally. Dorsiflexion had differing effects on the heel support though there was a consistent shift of the forefoot loads toward the medial rays of the foot. This medial shift was also present in plantarflexion.

It is difficult to assess what the general effect of skin loss is on the foot loading. There does appear to be a transfer of load toward the front and the rear of the foot (perhaps from some mid or lateral support areas). In conditions where the forefoot is twisted the forces seem to transfer to the lateral side of the foot while in straight foot loading they move toward the centre or medial side of the forefoot. Certainly the effects of the skin on the loading characteristics are not minor and should not be ignored.

5.5.4 Loss of the Plantar Aponeurosis

There was a consistent decrease in heel support associated with loss of the plantar aponeurosis. In the neutral position all feet showed a decrease of heel support by 2 to 11% of applied load. Some of this load was transferred forward to the mid and lateral side of the foot although foot 4 showed a marked decrease in the loads through the fifth metatarsal.

During inversion the decrease in heel support was not as great (and foot 7 actually increased) and the load transfers on the forefoot were almost identical to the changes seen in the neutral position. Eversion produced a wide variety of changes for the four feet with two decreasing and the other two increasing the load on the heel. There was some indication of load transfer laterally on the forefoot though foot 7 showed a considerable medial transfer.

Dorsiflexion indicated a consistent lateral load shift on the forefoot and some decrease in the heel support. In plantarflexion there was a noticeable decrease of loads under the hallux (foot 4 increased!) and a consistent transfer of force to the lateral side of the foot.

In general loss of the plantar aponeurosis would be expected to affect the hallux and the three middle toes (since they are the bones of its insertion) and the loading patterns on the medial side of the foot. The results here do show a decrease of load on the heel consistent with a partial release of the longitudinal arch and in general transfer of loads

Table 5.4 The change in the force measured under the areas of the feet after removing the plantar skin. (expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe	Met 5
Foot 4				
Neutral	5.8	-1	-2.8	7.3
Inverted	17.2	0.1	0.9	1.8
Everted	21.8	0	0	7.2
Dorsiflexed	15	-0.1	-0.3	-11.5
Plantarflexed	*	0.7	-1.75	-5.5
Foot 5				
Neutral	2.3	0.2	-0.2	1.3
Inverted	0.2	0.9	0.3	-0.3
Everted	10.1	0.1	-0.3	4
Dorsiflexed	-15.7	0.3	0	-12.5
Plantarflexed	*	0.8	2.5	-18.3
Foot 6				
Neutral	-4.4	1.5	-0.7	0.9
Inverted	0	-0.4	0.4	1.7
Everted	3.6	0.6	0	3.5
Dorsiflexed	4.8	0.4	0.2	1.5
Plantarflexed	*	15.9	-7.9	-1.1
Foot 7				
Neutral	2.3	0.2	-0.2	1.3
Inverted	0.2	0.9	0.3	-0.3
Everted	10.1	0.1	-0.3	4
Dorsiflexed	-15.7	0.3	0	-12.5
Plantarflexed	*	0.8	2.5	-18.3

Table 5.5 The change in the force measured under the areas of the feet after removing the plantar aponeurosis (expressed as % force applied to tibia/fibula).

	Heel	Met 1	Met 2/Toe	Met 5
Foot 4				
Neutral	-1.8	0.8	1.5	-3.8
Inverted	-1.2	0.3	0.6	-1.1
Everted	3.6	0	-0.1	3.8
Dorsiflexed	*	*	*	*
Plantarflexed	*	1	12.6	-0.6
Foot 5				
Neutral	-5.3	0	-0.3	1.9
Inverted	-4.4	-1.2	-0.5	1.6
Everted	-4.7	0	-0.3	-0.4
Dorsiflexed	-6.3	0	0.4	12.9
Plantarflexed	*	-4	-19.1	13.9
Foot 6				
Neutral	-10.7	-2	0	1.6
Inverted	2.4	-3.1	-0.2	-0.9
Everted	-18.4	-1.22	-0.5	2.8
Dorsiflexed	10.8	-0.2	-0.7	1.2
Plantarflexed	*	-6.1	-1	3.5
Foot 7				
Neutral	-6	0.4	1.7	0.3
Inverted	22.7	0.8	-1	0.3
Everted	15.5	2.2	1.2	0.4
Dorsiflexed	-16.4	0.5	-0.8	2.1
Plantarflexed	*	0.5	-10.7	11.9

Notes for both tables:

Met = Metatarsal head;

Met 2/Toe 1 : The gauge was placed under the second metatarsal head for all positions except plantarflexion when it was placed under Toe 1 (Hallux).

away from the first metatarsal toward the fifth.

5.5.5 Loss of the Long Plantar Ligament

This ligament inserts into the lateral three or four rays of the foot and is responsible for supporting the lateral arch of the foot. In the neutral position its loss resulted in an increase in the loads on the heel, but in the three other foot flat conditions three of the feet showed decreases in the heel support.

Across the forefoot the first and second metatarsal generally sustained greater loading while the fifth metatarsal remained unchanged or slightly decreased. In the plantarflexed position there was a consistent trend of load increase and transfer onto the medial side of the foot.

The long plantar ligament showed a variable response depending on the foot orientation. In the plantarflexed position its removal was most noticeable with a shift in the force distribution toward the medial side, consistent with a fall in the lateral arch. In the foot flat positions the load transfer was more variable though there was a general decrease in the heel loads and an appropriate increase in the loads on either all the metatarsal heads or the medial side alone.

5.5.6 Loss of All the Plantar Structures

This condition was achieved as the last stage of all the tests when all the longer ligaments, all the muscular tissue and all the skin on the plantar side of the foot was removed. The only structures supporting the arched foot were the interosseus ligaments (including the calcaneo-navicular and cuneonavicular ligaments).

The data presented in Table 5.3 are the detail of the final load distributions on the feet as an illustration of how they supported load with only interosseus ligaments present. In the neutral position the heel supported loads of between 43 and 62% of applied load and represented a gain of 6% for foot 5 and a decrease of 22% supported load for foot 6. The load distributions over the forefoot were quite noticeably increased (up to 2.1% support load increase for certain areas) particularly on the lateral side. In the fully cut state the fifth metatarsal supported more force than the first in three of the four feet.

In inversion the load distribution was more even across the forefoot and the total load supported by the forefoot was also increased. This increase in force on the forefoot was also apparent during eversion

Table 5.6 The change in the force measured under the areas of the feet after removing the long plantar ligament. (expressed as % force applied to tibia/fibula).

	Heel	Met 1	Met 2/Toe	Met 5
Foot 4				
Neutral	-4.4	0.1	0	-0.7
Inverted	-3.2	-0.8	0.4	-0.6
Everted	2	2.4	0.2	-4.4
Dorsiflexed	10	0	2	12
Plantarflexed	*	-4.5	-10.5	-8.5
Foot 5				
Neutral	32	0.2	0.8	3.4
Inverted	2.7	1	0.8	-0.6
Everted	6.2	-1	0.7	-3.5
Dorsiflexed	10	0.9	0.1	0.1
Plantarflexed	*	7.7	20	-40
Foot 6				
Neutral	4	1.63	0	-0.7
Inverted	-9.3	0.5	0	-0.3
Everted	-18.8	0.6	0	0.5
Dorsiflexed	-22.5	-0.6	0	0.1
Plantarflexed	*	13.2	8.7	-0.7
Foot 7				
Neutral	10.1	1.2	-1	0.3
Inverted	-9.8	-0.1	1.4	-1
Everted	-8.3	-1.4	-0.2	-0.7
Dorsiflexed	4.1	0.4	0.7	-1.2
Plantarflexed	*	0.1	0.7	-0.6

Notes:

Met = Metatarsal head;

Met 2/Toe 1 : The gauge was placed under the second metatarsal head for all positions except plantarflexion when it was placed under Toe 1 (Hallux).

although there was still a higher proportion of the load borne on the lateral side. In the dorsiflexed position there was less change as a result of the loss of the plantar structures. Only the heel supports more load with the structures cut.

The plantarflexed position was more variable mostly as a result of the difficulty of holding the foot in a plantarflexed position after the arches were released. Under these circumstances the load distributions recorded are unreliable.

The overall effect, then of removing the entire plantar support structures was to transfer load onto the forefoot and in particular the lateral aspect.

5.5.7 Application of Tension to Three of the Extrinsic Muscle Tendons

The effects of tension in the tendons of tibialis anterior, peroneus longus and extensor digitorum brevis were surprisingly consistent and well illustrated in figure 5.9. Tibialis anterior had an inverting effect lifting the medial rays of the foot and passing a greater percentage of the support load to the lateral rays. It had an inconsistent pattern of action on the forefoot - heel load distribution. Peroneus longus produced an everting change in foot position, causing a medial shift of the forefoot loads, a finding similar to that of Jones (1941) although the load distributions are not numerically equivalent. There was a slight increase in the loads at the heel when peroneus longus was tensioned. Extensor digitorum brevis did not cause any overt in/eversion effects but instead lifted the middle rays of the foot and caused a slight transfer of load to the rear foot.

It is interesting to note that none of the tendons had a greater effect after the extensive dissection process. As a result of this finding it is clear that the action of tendons alone would not significantly alter changes seen in the foot as a result of the loss of the plantar structures, a result that is in agreement with the theories of Jones (1941).

5.5.8 The Effects of Hallux Valgus Deformity

The intact hallux valgus feet showed higher loading levels under the fifth metatarsal than the normal feet when in the neutral position and also higher loads under the first metatarsal head when inverted. These findings are consistent with the theory that the medial arch is supported

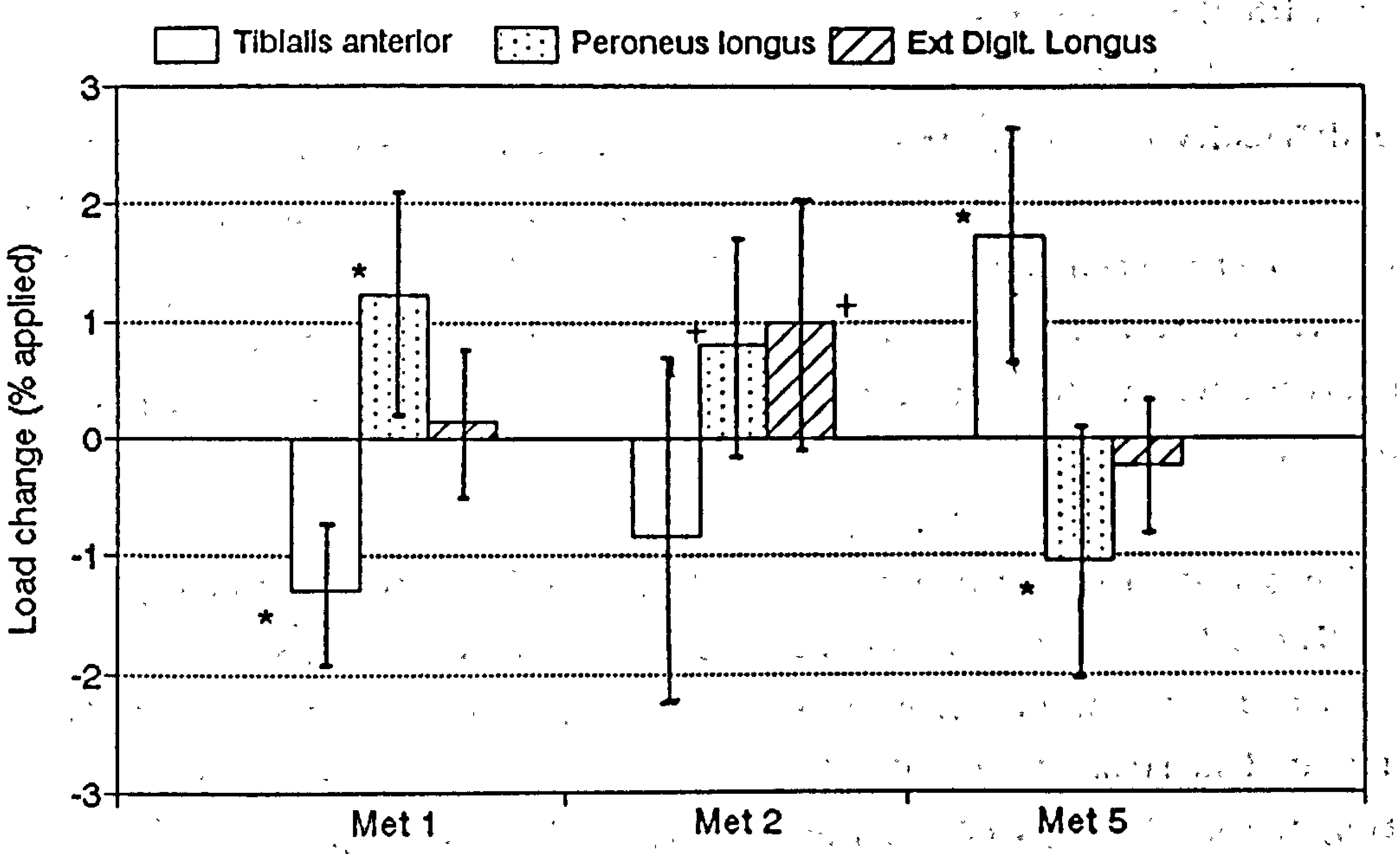


Figure 5.9 The effect on the loads under the forefoot of applying 50N tension to the tendons of the extrinsic muscles.
Test of mean <> 0 (* p<0.001; + p<0.01)

by the plantar aponeurosis. With its action severely reduced through hallux valgus the medial arches are depressed and loads are transferred laterally. When inverted however the arches are supported mostly by the interosseus ligaments which do not readily extend to transfer load onto neighbouring structures. The load distribution in the dorsiflexed position was more even across the forefoot by comparison with the tendency of the normal feet to transfer the loads to either medial or lateral rays. The force distributions during eversion and plantarflexion fall within the range of the normal feet.

The loss of the plantar skin produced greater changes under the first metatarsal in the feet with hallux valgus. There was also an increase in the forces under the hallux during plantarflexion, the opposite effect that the loss of the skin had for the normal group. It is possible that the loss of the skin releases the arches and tensions the aponeurosis applying more force through the hallux. In the normal foot the aponeurosis would already be tensioned and the loss of the skin may allow a straighter line of action thus releasing some tension.

In general the loss of the plantar aponeurosis would be expected to affect the mid and occasionally the lateral metatarsals to a greater extent than the first metatarsal since hallux valgus weakens the action of the plantar aponeurosis on the first metatarsal. These effects were confirmed when the aponeurosis was dissected with a notable decrease in the forces under the second and to a lesser extent the fifth metatarsal while the first metatarsal increased its load share. The loss of the aponeurosis did not affect the medial arch (which was already depressed) in this case but released the arches across the rest of the foot. In the plantarflexed position the changes in the loadings of the first metatarsal head and hallux were not consistent as the foot with the worse hallux valgus angle transferred force from the metatarsal head to the hallux while foot 8 showed the opposite response to aponeurosis release.

The loss of the long plantar ligament in the hallux valgus feet produced opposite patterns of load changes to the normal feet. The first metatarsal head showed a general decrease in loading with no consistent change apparent in the second metatarsal. There was increased loading on the fifth metatarsal in nearly all positions for foot 9 while for foot 8 only plantar and dorsiflexed positions produced an increase in load. In the plantarflexed position the loss of the long plantar ligament resulted in an increase in the forefoot loads for foot 8 and both hallux valgus

Table 5.7 The hallux valgus feet uncut (details as for Table 5.2).
(expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 8				
Neutral	70.7	3.6	3.4	6.8
Inverted	62.7	6.3	3.1	1.8
Everted	73.5	2.6	2.7	3.5
Dorsiflexed	73	2.2	2.1	2.7
Plantarflexed	*	6.1	12.5	14.2
Foot 9				
Neutral	74.7	2.2	2.6	7.6
Inverted	73.2	6.3	3.2	4.1
Everted	74.9	1.2	4.5	9.7
Dorsiflexed	75.9	1	1.9	3.4
Plantarflexed	*	11.3	8.4	45.6

Table 5.8 The hallux valgus feet after all plantar tissues were cut
(details as for Table 5.3) (expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 8				
Neutral	69.3	4.8	0.2	2.5
Inverted	70.9	8.5	3.1	2.7
Everted	69.3	3.7	2	2.9
Dorsiflexed	61.5	8.2	0.3	2.8
Plantarflexed	*	12.5	19.5	28.1
Foot 9				
Neutral	59.2	1.7	1.2	4.8
Inverted	52.7	4.5	1.8	5.3
Everted	74.3	4.9	4.2	7.3
Dorsiflexed	58.5	4.1	1.4	2.6
Plantarflexed	*	25.5	5.4	18.5

feet showed a load transfer toward the medial side of the forefoot. The unusual load transfers as a result of the loss of the long plantar ligament in the hallux valgus feet may again be the result of weakened medial arches. If the medial interosseus ligaments are already stretched through overloading, further loading due to a decrease in the lateral arches may continue to be borne by the stronger lateral interosseus ligaments.

With all the plantar tissues removed the feet with hallux valgus showed higher loads both at the heel and under the first metatarsal head when compared with the normal feet. The forces under the first metatarsal head were in some instances over twice those recorded for the normal feet. This suggests that the lack of plantar aponeurosis activity in support of the medial arches may have resulted in long term damage to the interosseus and associated ligaments causing a permanent lowering of the medial arch. With this arch lowered the foot will tend to 'evert' and subsequent loading will be preferentially transferred to the medial side further overloading the medial arches.

5.6 CONCLUSIONS

The experimental tests, although based on only four normal specimens do offer support for the accepted function of several of the plantar soft tissue structures. The plantar aponeurosis was primarily responsible for the support of the medial arches while the long plantar ligament supported the lateral arches.

A new aspect of this work was the consideration of the plantar skin which was found to affect the foot loading characteristics. Although the loss of the plantar skin in the neutral position had only a limited effect, the twisting of the foot into either inversion or eversion produced an increase in loading on the lateral metatarsals when the skin was removed. It would seem that the skin acted as a 'binding' structure limiting the ability of the metatarsals to act independently.

The changes measured in the load distributions as a result of the application of tension to several of the extrinsic tendons were surprisingly consistent for the three feet tested. The changes appeared to be independent of the extent of plantar structure loss and this finding alone would indicate that, as has been suggested before (Jones, 1941) the extrinsic muscles do not change their action as a result of the loss of ligamentous arch support and their ability to maintain or raise the arches

Table 5.9 The change in forces under the hallux valgus feet after the plantar skin was removed (details as for Table 5.4)
(expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 8				
Neutral	3.5	2.7	1.5	2
Inverted	4.5	1.6	0.7	3.3
Everted	6.1	-0.7	1.1	3
Dorsiflexed	13.9	-0.2	0.3	2
Plantarflexed	*	-5.4	-6.3	20.5
Foot 9				
Neutral	-32.8	1.1	0.8	-4.7
Inverted	-0.3	2	-0.1	-3.6
Everted	3.2	0.7	-0.5	-6.2
Dorsiflexed	-17.4	3.2	1.8	-0.8
Plantarflexed	*	32.1	19.3	-45.4

Table 5.10 The change in the forces under the hallux valgus feet after the plantar aponeurosis was removed (details as for Table 5.5)
(expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 8				
Neutral	-1.9	3	-3.2	-5.4
Inverted	4.3	2.4	-2.8	-0.8
Everted	-5.2	2.3	-2.4	-3.1
Dorsiflexed	-21	1.1	-1.2	-2.7
Plantarflexed	*	5.2	-6.7	20.7
Foot 9				
Neutral	17.4	1.5	-2.5	-1.6
Inverted	-5.5	-1.8	-1.9	3.1
Everted	-30	0.4	-3	0.3
Dorsiflexed	-0.5	-2.2	-1.8	-2
Plantarflexed	*	-23.4	6	20

Table 5.11 The change in the forces under the hallux valgus feet after the long plantar ligament was removed (details as for Table 5.6)
(expressed as % force applied to tibia/fibula)

	Heel	Met 1	Met 2/Toe 1	Met 5
Foot 8				
Neutral	-3	-4.5	-1.5	-0.9
Inverted	-0.6	-1.7	2.1	-1.6
Everted	-5.1	-0.5	0.6	-0.5
Dorsiflexed	-4.4	5.1	-0.9	1.1
Plantarflexed	*	6.6	13.1	17.9
Foot 9				
Neutral	-0.1	-3.1	0.3	3.5
Inverted	-14.6	-2	0.6	1.7
Everted	28.4	2.5	3.2	3.5
Dorsiflexed	0.5	2.1	-0.5	2
Plantarflexed	*	5.5	-28.3	-1.7

is the same for both normal and flatfooted subjects. Tibialis anterior produced an inverting effect with associated transfer of load laterally while peroneus longus produced eversion with an associated medial load transfer. Extensor digitorum longus does not appear to noticeably change the orientation of the foot although it does raise the middle (2,3 & 4) rays of the foot and produce some load transfer to the heel.

The investigation of two feet with a hallux valgus deformity confirmed the biomechanical changes as a result of the loss of the 'windlass' effect of the plantar aponeurosis around the first metatarsal head. Loads were generally transferred to the neighbouring and lateral metatarsals. Interestingly when all the plantar tissues were removed the bare foot, supported only by the interosseus and related ligaments, bore more load on the medial side in the hallux valgus condition when compared with the normal feet. This would suggest that the medial arches have fallen and due to the resulting eversion greater loads are placed on the medial metatarsal rays. As it is unsupported high loads that stretch the ligaments, this condition would be likely to deteriorate.

Although there was considerable variation between the loading patterns of the specimens, the major effects of the loss of the plantar structures became most apparent when the foot was placed in an inverted or everted position. The other important finding was the strength in the foot even without all these major plantar structures. The interosseus, calcaneocuboid and calcaneonavicular ligaments were able to maintain the normal foot structure (although considerably flattened) and support several times body weight through the talus. In the only specimen tested to failure, loads reached 3000N before one of the metatarsals failed. The complete structure, both ligamentous and osseus is well designed to support heavy load.

CHAPTER 6. ELECTROMYOGRAPHIC (EMG) ANALYSIS OF LOWER LEG AND FOOT MUSCLES

6.1 INTRODUCTION

This thesis has considered several elements of the foot as they relate to normal foot function but until now there has been one overriding simplification, the inactivity of muscles. Justification has been rightly provided that the muscles are generally seen to be inactive during static activity and other elements appear to form the foundation of the foot structure. In considering the dynamic behaviour and behaviour under higher static loads, the muscles cannot be ignored. To that end it was decided to undertake experiments to clarify the role of the foot muscles (both intrinsic and extrinsic) during the gait cycle. Furthermore a study of the muscle properties, such as fatigue resistance, seemed an appropriate addition to the tests.

As has been more fully stated in the literature review earlier only limited research has been conducted on the foot and lower leg muscles. Several researchers have studied the muscle behaviour in the lower leg during walking (Mann & Inman, 1964; Gray & Basmajian, 1968; Ericson et al, 1986; Winter & Yack, 1987) using methods ranging from surface to implanted techniques. Very few studies though have investigated more than temporal EMG action but the studies of Dubo et al (1976) and Ericson et al (1986) have used a normalized EMG signal as a good indicator of the level of muscle activity. Winter & Yack (1987) normalised their EMG signals to a mean activity level. This type of normalisation results in curves that show lower variation bounds but contain no indication of the true level of activity. These three studies appear to be the only ones to provide EMG amplitude data for the muscles in the lower leg and to the author's knowledge no studies have investigated the EMG amplitude of the intrinsic muscles of the foot. Neither to the author's knowledge, has the fatigue phenomenon of the foot muscles under sustained or heavy contraction been previously quantified.

It is also pertinent to be aware of the fatigue characteristics of these muscles as these changes will affect the ability of muscles to maintain load support. Fatigue characteristics of muscles as indicated by the change in the EMG frequency spectrum have been investigated by Lindström & Magnusson (1977), Kadefors (1976) and others (De Luca, 1986). The aims of the present study were to describe the fatigue characteristics

Table 6.1: Subject Details for the EMG Tests

Subject number	Age	Height (m)	Mass (kg)	Length of Leg (mm)	Length of Foot (mm)
1	37	1.83	75	390	270
2	34	1.75	63	370	270
3	37	1.89	76	410	280
4	26	1.99	85	420	270
5	25	1.76	73	400	250
6	25	1.75	68	420	265
MEAN	30.6	1.83	73.3	400	266
Std Dev	5.95	0.09	7.5	19	10

of the intrinsic and extrinsic muscles of the foot and lower leg utilizing the Mean Power Frequency (MPF; the mean frequency of the frequency spectrum) of the EMG signal.

6.2 TEST SUBJECTS

Six healthy subjects (Table 6.1) volunteered for the study. No subject had a history of orthopaedic or neurological pathology and each gave their written consent to the tests after a clear explanation of the aims and procedure as recommended by the Helsinki code. The tests were undertaken at the Karolinska Hospital biomechanics laboratory (Stockholm, Sweden) and were approved by the Karolinska hospital ethical committee.

6.3 METHOD

6.3.1 Muscles Studied

The muscles were studied in two groups during the tests:

the extrinsic muscles

a) gastrocnemius, medial and lateral heads, peroneus longus, tibialis anterior and

the intrinsic muscles

b) flexor hallucis brevis, abductor hallucis, flexor digitorum brevis and extensor digitorum brevis.

6.3.2 Equipment and Recording Method

6.3.2.1 Fitting the electrodes

a) The extrinsic muscles

External Ag/AgCl electrodes (D-05-VS, Medicotest a/s, Denmark) were used for all the extrinsic muscles and the extensor digitorum brevis. The site for electrode placement was chosen over the muscle bulk and the area prepared by shaving, abrading and cleaning with an ether/ethanol preparation. A proprietary electrode gel was used to reduce impedance and the electrodes were placed 20mm apart in line with the muscle fibres. Once placed the electrodes were checked with an impedance meter to ensure that the impedance between them was less than 5kΩ. The electrodes were taped on to the skin surface and the shielded, connection leads taped down to limit motion artefacts.

b) The intrinsic muscles

The extensor hallucis brevis was the only muscle of the intrinsic

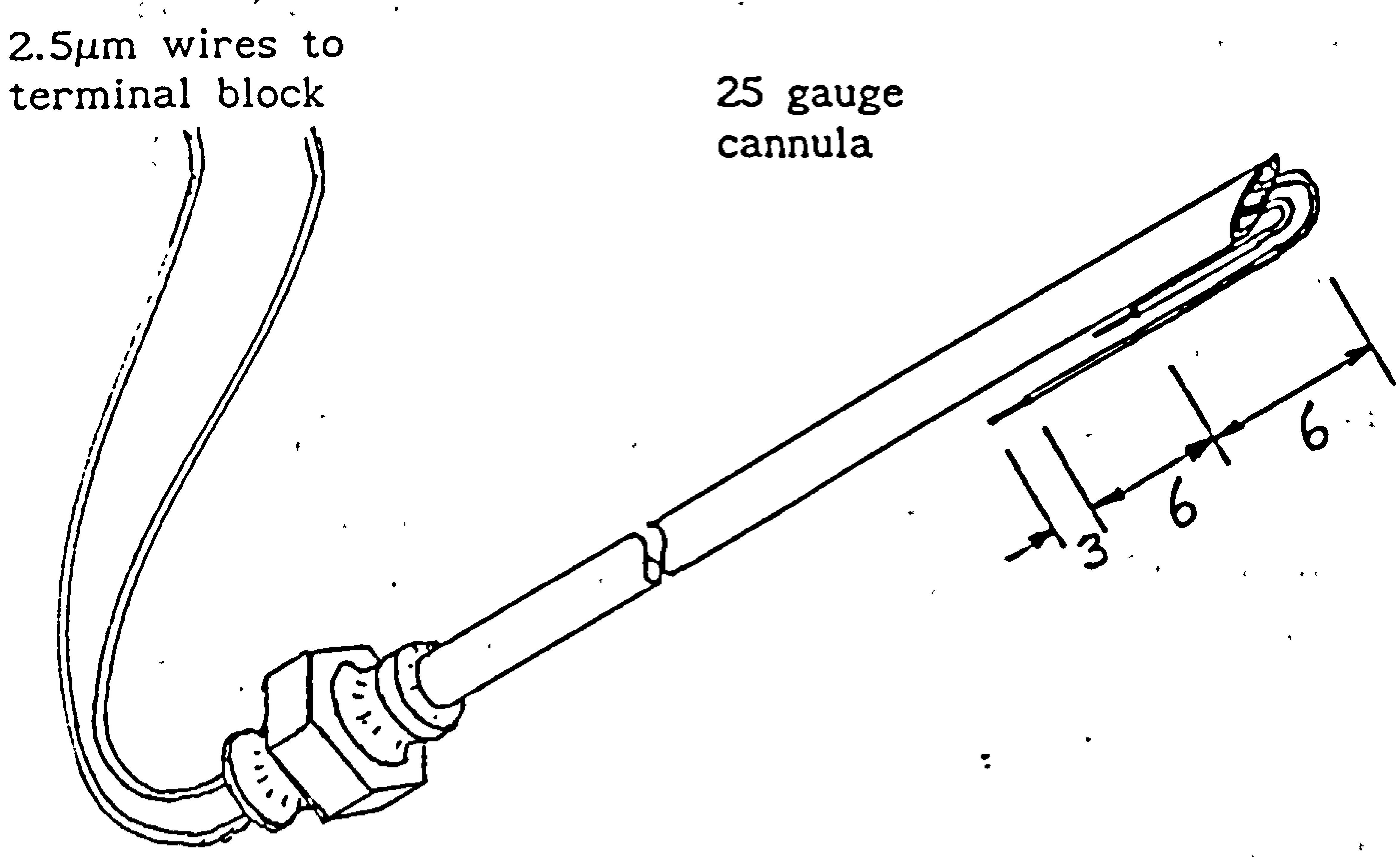


Figure 6.1 Illustration of the prepared bipolar, fine wire implanted electrodes.

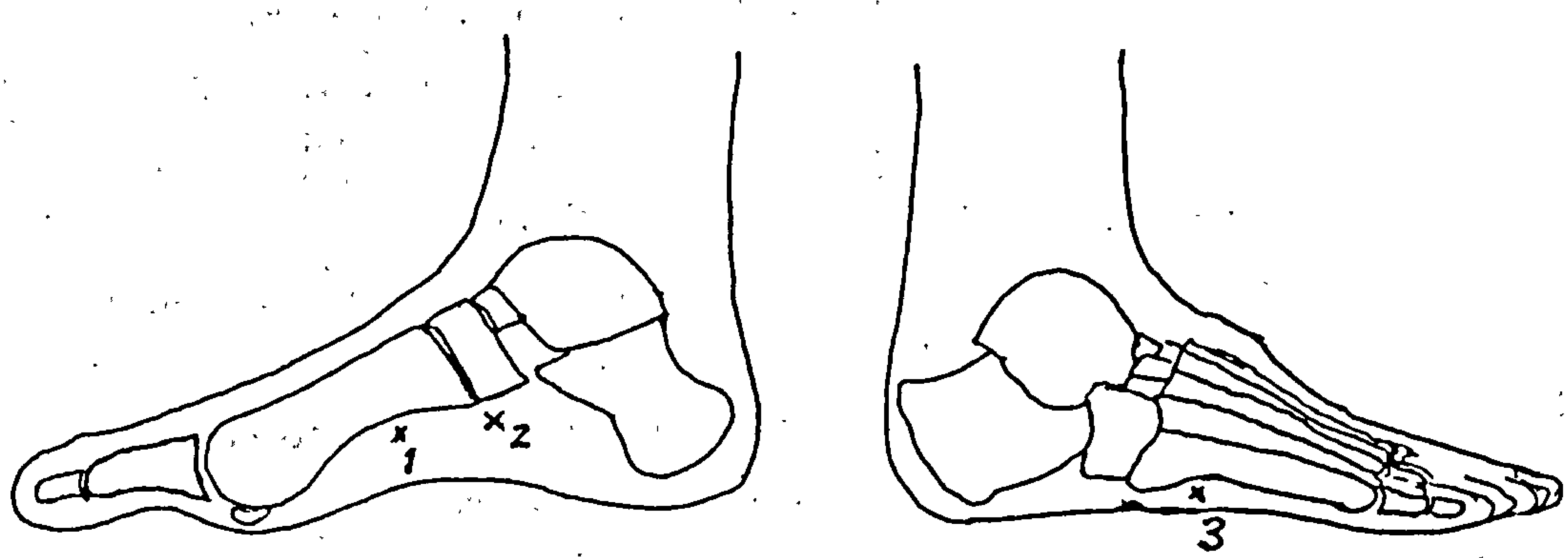


Figure 6.2 The points of insertion for the cannulas carrying the fine wire electrodes: 1) flexor hallucis brevis; 2) abductor hallucis; 3) flexor digitorum brevis.

group that was readily accessible using surface electrodes. The method of placement was the same as for the extrinsic muscles except that the inter-electrode distance was reduced to 10mm.

Bipolar, fine-wire electrodes were implanted under local anaesthetic into the bellies of the three remaining intrinsic muscles. The electrodes were prepared as suggested by Basmajian & Stecko (1962). Two 2.5μm polyurethane coated, platinum wires (HP Reid Co Inc, Neptune, NY, USA) were inserted into a 25 gauge cannula. The two ends emerging from the tip of the cannula were stripped of 3mm of their insulation and bent back upon themselves to form a barb (fig 6.1). Four electrodes were prepared in this way and placed into a stainless steel sterilization pack. The complete arrangement was then sent for steam sterilization.

Before the placement of the intrinsic electrodes the surface of the foot was cleaned with the ether/ethanol preparation and the surgeon responsible for inserting the cannula put on sterile gloves and mask to minimize the risk of infection. To insert the electrodes a local anaesthetic (Prilocain) was infused into the region of electrode entry. The effects of this anaesthetic normally subsided within 15 minutes and as EMG tests were not made until at least 30 minutes after electrode insertion the effects of the anaesthetic on the EMG signals were considered negligible. Three entry points were used for the three electrodes (fig 6.2). The procedure for each was as follows:

Flexor hallucis brevis : the subject was asked to resist passive dorsiflexion of the toe, making FHB palpable over the medial/inferior region of the first metatarsal. A cannula containing the electrode wires was inserted through the region just anterior to the base of the first metatarsal, from the medial, dorsal aspect. The cannula was angled toward the centre of the muscle belly and inserted until the passage through the muscle septum was felt and then inserted only a few millimetres more.

Abductor hallucis : this muscle was the easiest to place being quite near the medial border of the foot and readily palpable when the subject resisted passive adduction of the hallux. The electrodes in their cannula were inserted through the medial side of the foot at the centre of the palpated muscle belly. Only 10mm or so was required to penetrate the muscle septum and position the barbs near the centre of the muscle.

Flexor digitorum brevis : insertion through the lateral skin passing immediately dorsal to flexor digiti minimi brevis, 15mm distal to the base of the fifth metatarsal bone. The cannula was angled upward toward the

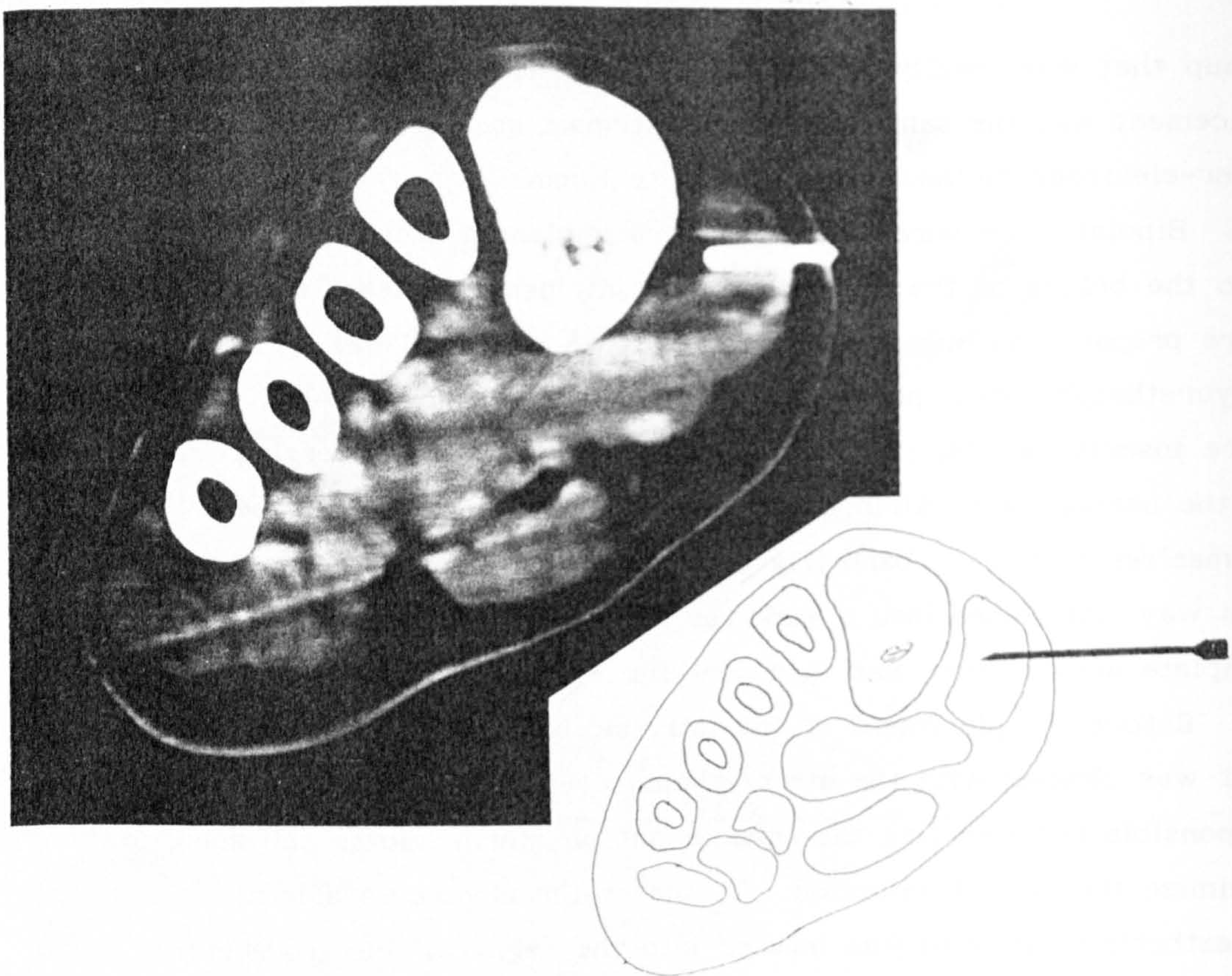
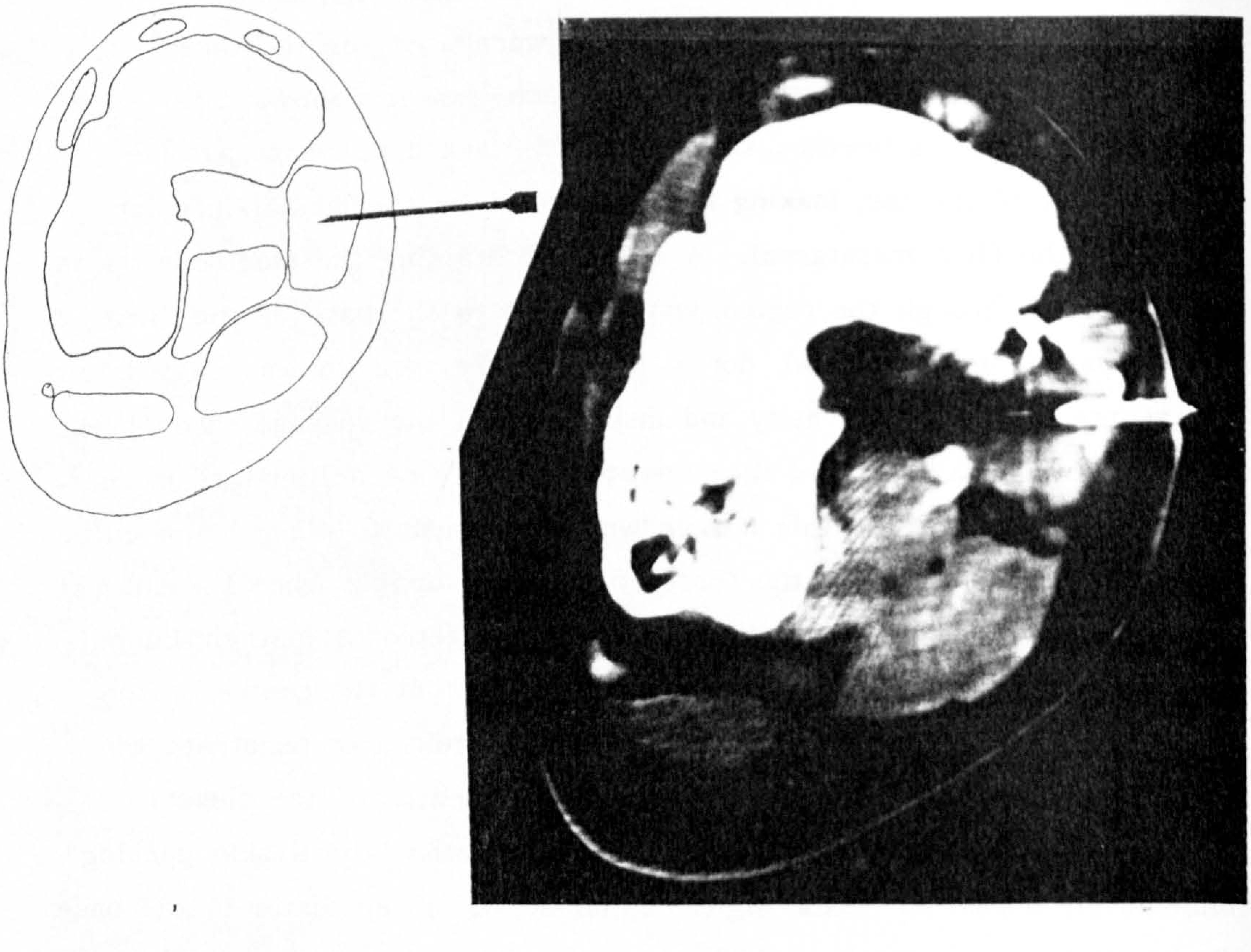


Figure 6.3 CAT scans of the foot indicating the placement of the cannulas introducing the fine-wire electrodes into: a) flexor hallucis brevis (frontal section); b) abductor hallucis (frontal section); (cont over)



medial cuneiform in order to avoid the tendons of the lateral side of the foot and the plantar aponeurosis by passing above it. This was the most difficult electrode to place as the position of the muscle bulk was difficult to assess during insertion. Insertion ceased about 3mm after the septum of the muscle had been pierced. A check was made on the insertion angle and depth with the use of a duplicate cannula placed along the plantar aspect of the foot in line with the one inserted.

As a test of the accuracy of placement, one of the six subjects consented to a CT scan of the foot with the electrodes inserted. The results are shown in figure 6.3 and offer endorsement of the technique used.

To withdraw the cannula from each muscle, care had to be taken that the electrode was not moved during the process. It was found that this could be achieved by supporting the "flying" leads and keeping them loose while withdrawing the cannula. Once the cannula had been removed pressure was applied to staunch bleeding and sterile tape was placed over the site of insertion. The fine leads were taped to the leg avoiding any application of stress. They followed the contours of the leg toward the knee where they were connected to a specially constructed terminal block (figure 6.4). The impedance meter was inappropriate for testing the intrinsic muscles and the use of stimulation to check electrode placement has previously been questioned (Basmajian, 1985).

c) Earth electrode

A single large size, passive earth electrode (VL-10-D, Medicotest A/S, Denmark) was used to limit external interference. For such an electrode to be effective it must be placed on a region of low electrical activity. In these tests the electrode was placed over the bony region of the proximal anterior crest of the tibia.

6.3.2.2 Signal transmission and storage

Four channels of an FM telemetry system were available for use. The transmitters (IC-600-EMG, Medinik AB, Sweden, Frequency Bandwidth 10-1500Hz) were strapped to the subject's waist and the earth electrode then the appropriate leads from the surface electrodes or the terminal block were connected. Failure to connect the earth electrode first was found to result in uncomfortable stimulation of the intrinsic muscles from random noise.

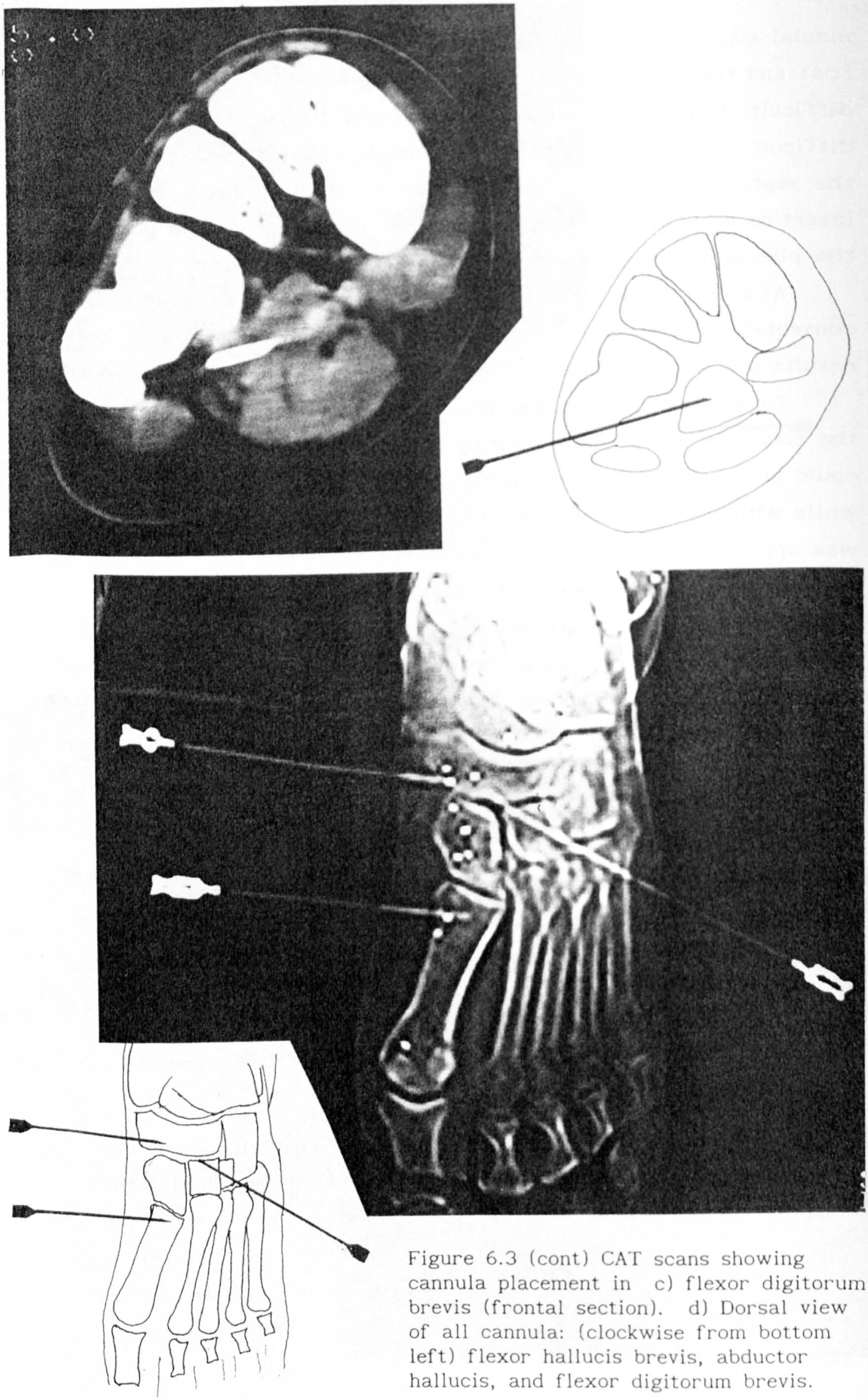


Figure 6.3 (cont) CAT scans showing cannula placement in c) flexor digitorum brevis (frontal section). d) Dorsal view of all cannula: (clockwise from bottom left) flexor hallucis brevis, abductor hallucis, and flexor digitorum brevis.

The signals were transmitted to base stations (IC-800-K, Medinik AB, Sweden) and amplified to provide an output signal for maximum contraction of +/- 0.5 V (normally from 0.5-2k gain). The signals were recorded, unfiltered on an eight channel FM, cassette data recorder (TEAC R-71, TEAC, Japan) with a frequency response of DC to 1250Hz. To provide an indication of the muscle activity in real time the signals were displayed on an oscilloscope (SS-661, Iwatsu, Japan) and on an eight channel thermal chart recorder (WR 7700, Graphtech Corp, Japan) with a bandwidth of DC to 1kHz minimum. The signals were also RMS integrated (NL-705, Neurolog Digitimer, UK, τ = 100ms) and the output plotted on the chart recorder. At this time it was possible to check the signals for 50/60Hz or random interference. Reasons for poor quality signals were found to include:

- broken wire
- poor connection at intrinsic wire interface
- low transmitter battery
- an intrinsic electrode not in muscle
- proximity of fine wire electrode leads to other (surface) electrodes.

During recording, details of the test situations were kept in three forms:

memo records on the tape audio channel
written notes on the chart record and
a detailed log (including tape count) in a separate book.

6.3.3 The Test Methods

6.3.3.1 Preliminary tests

Before the activity tests, it was necessary to have an indication of some predefined levels, notably the background EMG activity and the activity that resulted from maximal voluntary contraction of each muscle. The test of EMG activity at maximum voluntary contraction also served to indicate the level of cross-talk between muscles, a matter of particular concern for the intrinsic muscles.

To assess the background activity the leads for the muscles of interest were connected and the subject was asked to lie or sit in a comfortable position. The subject was then asked to completely relax and a recording of the activity levels was made for 5 seconds.

The activity level at maximum voluntary contraction was recorded during tests for each individual muscle. The method of achieving the maximal contraction level for each muscle was :

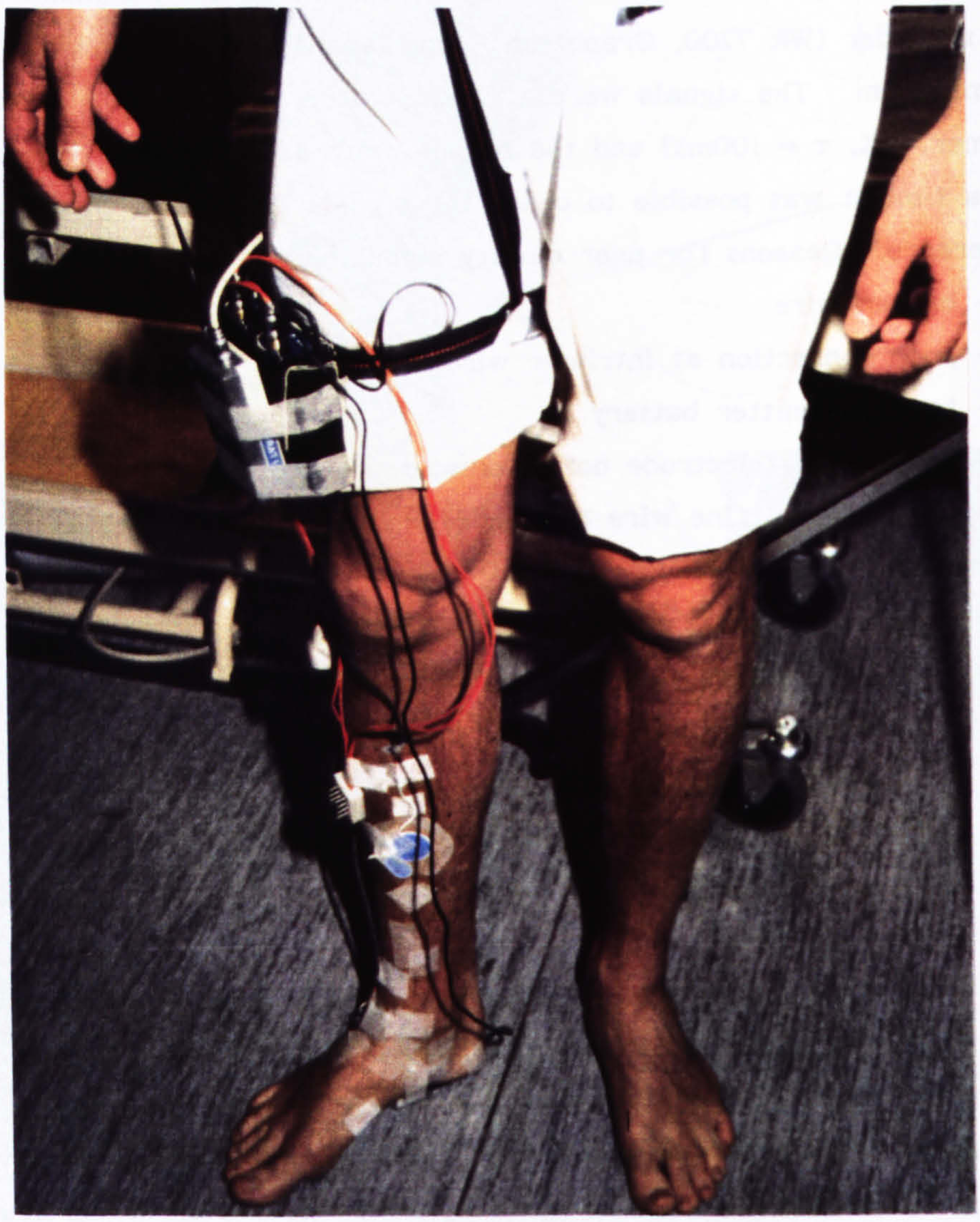


Figure 6.4 A subject prepared for a test of the intrinsic muscles with the heel switch in place. Note the telemetry transmitters above the knee, the white, fine wire electrode interface on the anterior aspect of the lower leg, and the blue earth electrode over the medial, bony region of the tibia.

Gastrocnemius (both heads) - the subject, standing upright, applied a maximal lift to a structure fixed to the floor. They were asked to lift on their forefoot so raising their heel just off the floor.

Peroneus longus - maximal resistance to applied dorsiflexing and supinating force.

Tibialis anterior - maximal resistance to applied plantar flexing force.

Flexor hallucis brevis - maximal resistance to applied extension of the hallux.

Abductor hallucis - maximal resistance to applied adduction and/or extension of the hallux.

Flexor digitorum brevis - maximal resistance to applied extension of the second, third and fourth toes.

Extensor digitorum brevis - maximal resistance to applied plantarflexion and supination of the foot.

The forces applied during these tests were produced manually by the experimenters with the joints in a neutral and static position.

6.3.3.2 The gait analysis and forefoot support tests

The muscles were studied in two groups during the gait analysis tests; the intrinsic group and extrinsic group. The two groups were studied separately although the tests used were identical. Some supplementary tests combined muscles from these two groups mainly to clarify timing aspects.

The first test was a simple walk with bare feet. Each subject was asked to walk at a comfortable pace (stride frequency 0.65 - 0.85Hz) until ten complete strides were recorded. The length of walkway (5m) required the subjects to turn 180° after five or six complete strides (which was marked on the chart record), so the strides immediately preceding and following this turn were discounted for the tests of normal gait. The test was then repeated first with a heel switch mounted in a moulded epoxy heel plate (4mm thickness) and then with the subject's normal shoes (these were all soft, flexible casual shoes).

The next tests involved the subjects standing on toes and forefoot. While recording was in progress each subject was asked to steadily rise onto their toes then return to foot flat. They then rose again onto their toes and upon request (which was marked on the signal) transferred their

TABLE 6.2: The time (s) subjects maintained the different fatigue contractions. NC = no contraction of the investigated muscle

	Standing on forefoot with attempted max toe flexion			Standing on forefoot with heel elevated 20-30 mm			Standing on heels	
Subj. no	Flex hall brevis	Abductor hallucis	Flex dig brevis	Peroneus longus	Gastroc. lat head	Gastroc. med head	Ext dig brevis	Tibialis anterior
1	180	180	180	180	180	180	180	180
2	180	180	180	180	180	180	180	180
3	180	180	180	180	180	180	75	75
4	120	120	120	135	NC	135	180	180
5	45	60	45	NC	180	180	45	75
6	105	NC	120	NC	180	180	NC	NC

foot loading to the medial side then the lateral side, then medially, laterally and then return to foot flat. This test was again repeated with the subject wearing their own shoes. Unfortunately no force platform was available in the laboratory so no record was made of the forces beneath the feet during these tests.

6.3.3.3 The fatigue test

In order to produce a meaningful indication of muscle fatigue it is necessary to load the muscles sufficiently to cause fatiguing behaviour after a reasonable length of time (3 minutes). It is also helpful to produce the loading simultaneously in as few muscles as possible so a more individual behaviour is discerned. Three types of contraction were proposed to satisfy the above criteria as closely as possible.

1. Standing on the forefoot with the heel elevated 20-30mm above the floor with attempted maximum flexion of the toes

2. The same as number one but without flexion of the toes

3. Standing on the heels with the forefoot everted and elevated just off the floor.

Subjects were instructed to maintain the contraction for 3 minutes or until they experienced exhaustion (Table 6.2). Some subjects did not show a contraction in all the investigated muscles using the above contraction methods (Table 6.2). Muscles can take quite a long period to recover from fatigue so in these tests at least 10 minutes rest was allowed between the contractions.

Because of the nature of the contractions the muscle groups were altered for these tests. During fatigue contraction 1, flexor hallucis brevis, abductor hallucis, and flexor digitorum brevis were recorded. During contraction 2, EMG activity was recorded from both heads of gastrocnemius and peroneus longus and for contraction 3, recordings were made from tibialis anterior and extensor digitorum brevis.

6.3.4 The Analysis

6.3.4.1 Amplitude analysis

The raw, unfiltered EMG signal was replayed from the FM cassette data recorder and passed through an RMS integrator (τ= 50ms). The output signal was sampled with an IBM computer (80386 processor) based A/D converter card (PC-30, United Electronic Ind, USA, max freq. = 6000Hz) controlled by application software (ADVI300) written by one of the Swedish coexperimenters (Walker et al, 1991). The effective sampling rate was 50Hz. The sampling software utilized the background and peak EMG levels recorded during the first part of the tests to calibrate the EMG signal. The digital records so formed were stored on disk for subsequent processing.

Further software was written (Walker et al, 1991) to allow the visual analysis of the EMG records (AMPA950F). Following the sampling process (ADVI300) the program (AMPA950F) was used to select two representative strides for each subject in each test. Manual methods were used to minimise motion artefacts detected from the shift in the baseline of the raw EMG signal . These two records were the basis of all further tests.

The routine STDEMGGF was written by the author (Appendix 4) to take the 12 records produced (6 subjects x 2 representative records/subject) for each test, normalize them to 100% full cycle, and display them simultaneously on the screen. Routines were implemented to allow individual traces to be time shifted as necessary to correct for errors in determining the heel strike timing. When the traces had been set at their correct timings a separate routine calculated the mean and standard deviation of all 12 curves at intervals of 1% of the gait cycle. (Due to the small sample size it was felt that a detailed statistical analysis was inappropriate.) The mean and standard deviation records for the full gait cycle were written to disk and from there passed to a graphical suite (GCGRAPH, Appendix 4) for output.

6.3.4.2. Forefoot support activity

The results from the forefoot loading exercises were analysed from the EMG signals recorded on the chart recorder. Activity levels as indicated by the RMS integrated signal level were manually measured from the chart records for each muscle during the different stages of the forefoot support. The levels were normalised to the recorded maximum levels and the means and standard deviations calculated for the subjects standing symmetrically. Several methods were tested to determine a reliable measure of activity change as the load support was transferred to the medial or lateral side of the foot. The results were subsequently calculated as an absolute change in the symmetrical activity levels and graphed using commercial software (Quattro Pro, V5, Borland International).

BAREFOOT

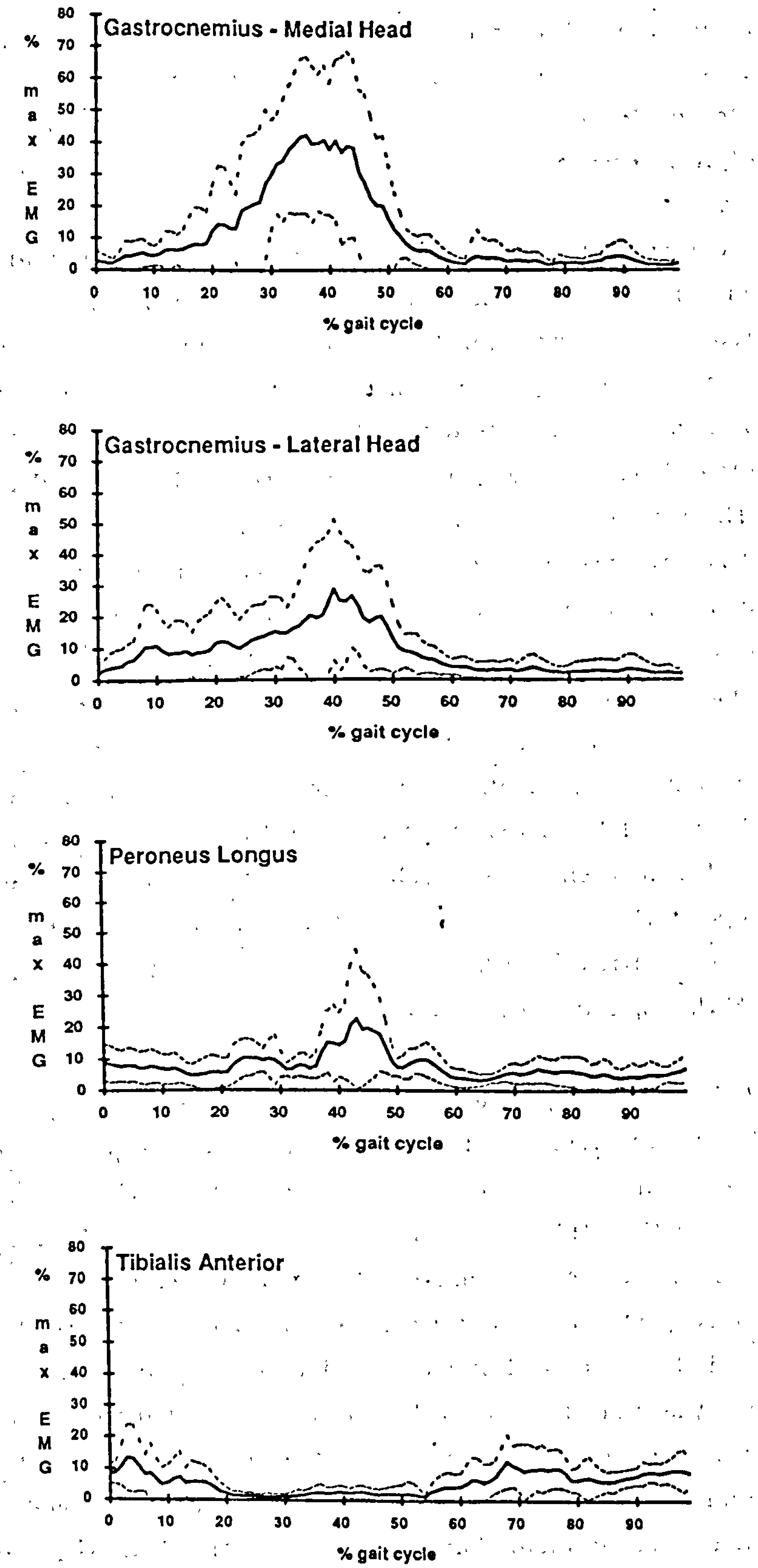


Figure 6.5 Mean EMG activity (solid line) +/- 1SD from the extrinsic foot muscles recorded while subjects were walking barefoot. (N = 6)

6.3.4.3. Fatigue analysis

For the fatigue analysis the raw EMG signal was replayed from the cassette data recorder and sampled with the computer A/D convertor card. The signal was not passed through the RMS rectifiers. A Fast Fourier Transform algorithm was implemented in sampling software (EMGFFT; Németh et al, 1991) and the Mean Power Frequency (MPF) of the signal was displayed for each second. The record of MPF values was filed on disk. The routine STATSTOR (Appendix 4) was written to take the six MPF records for a particular muscle and display them graphically on the screen against time. A pointer driven routine was then used to digitize typical values from each curve at 15s intervals. The individual MPF values were stored as well as the mean and standard deviation for each 15 s interval. Further investigation showed that while there was wide variation in the MPF values for each subject the overall trend in change of frequency was consistent for a particular muscle. In order to illustrate this a further routine (NORMFFT; Appendix 4) was written to take the output files from STATSTOR and normalize all the curves to 100% initial frequency and then recalculate the means and standard deviations based on the new values. Both the STATSTOR MPF files and the normalized NORMFFT output files were passed to the graphical package for output.

For statistical analysis of the changes in frequency the routine TTEST (Appendix 4) was written to implement a standard two tailed student's T-test. This program with the appropriate statistical tables enabled the analysis of the differences between means. A p value < 0.05 was regarded as significant for marking the time varying MPF.

6.4 RESULTS.

6.4.1 The Amplitude Analysis

The results of the complete series of tests are shown in figures 6.5 to 6.10. Of interest is the level and timing of the EMG signal as it provides an indication of the degree of muscle activity. To aid the reader appreciate the detail in the figures a short summary for each muscle is appropriate. For brevity, the references to the epoxy heel plate with the heel switch will be referred to as the heel plate.

BAREFOOT WITH HEELMARKER

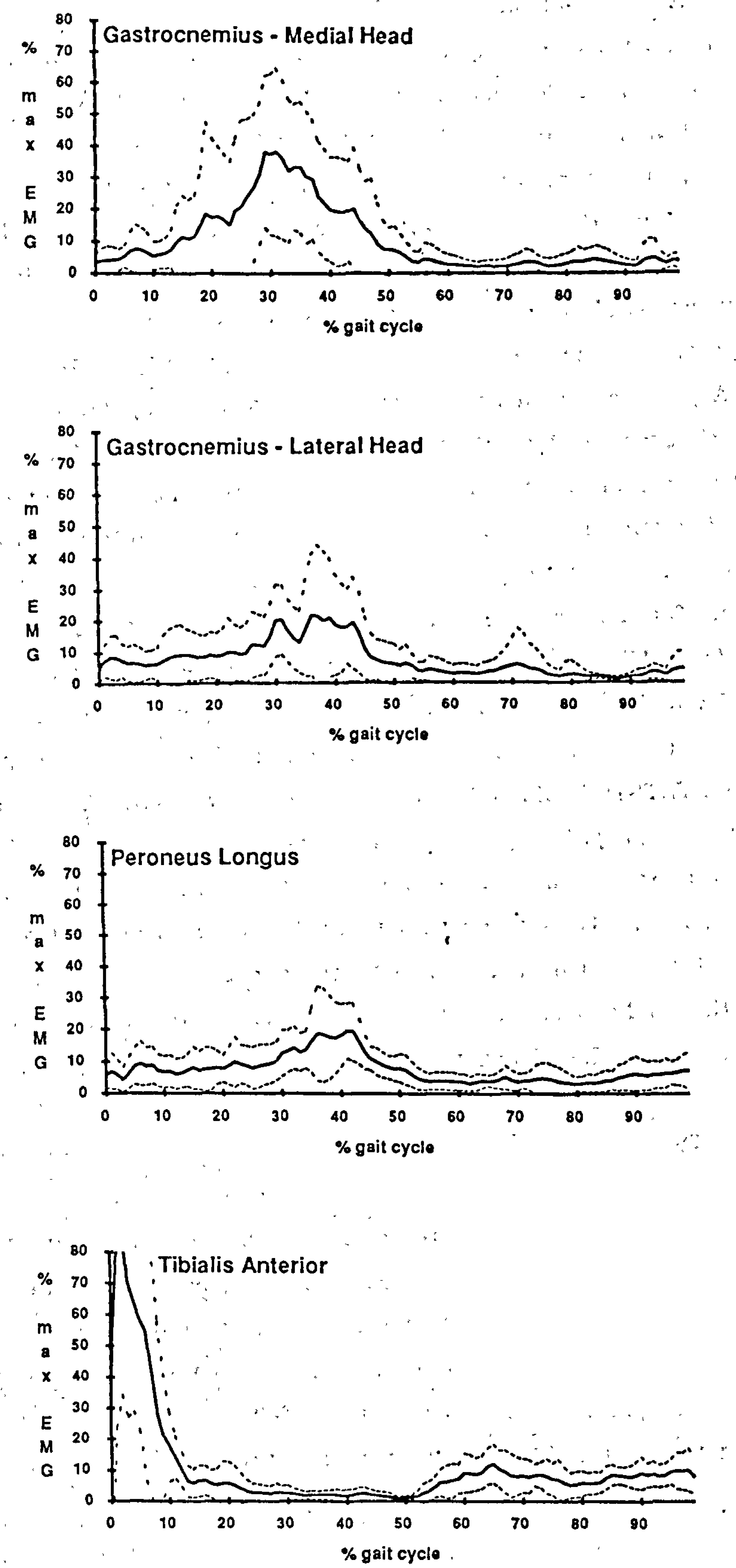


Figure 6.6 Mean EMG activity (solid line) +/- 1SD from the extrinsic foot muscles recorded while subjects were walking barefoot with an epoxy moulded heel plate containing the heel switch. (N = 6)

SHOES

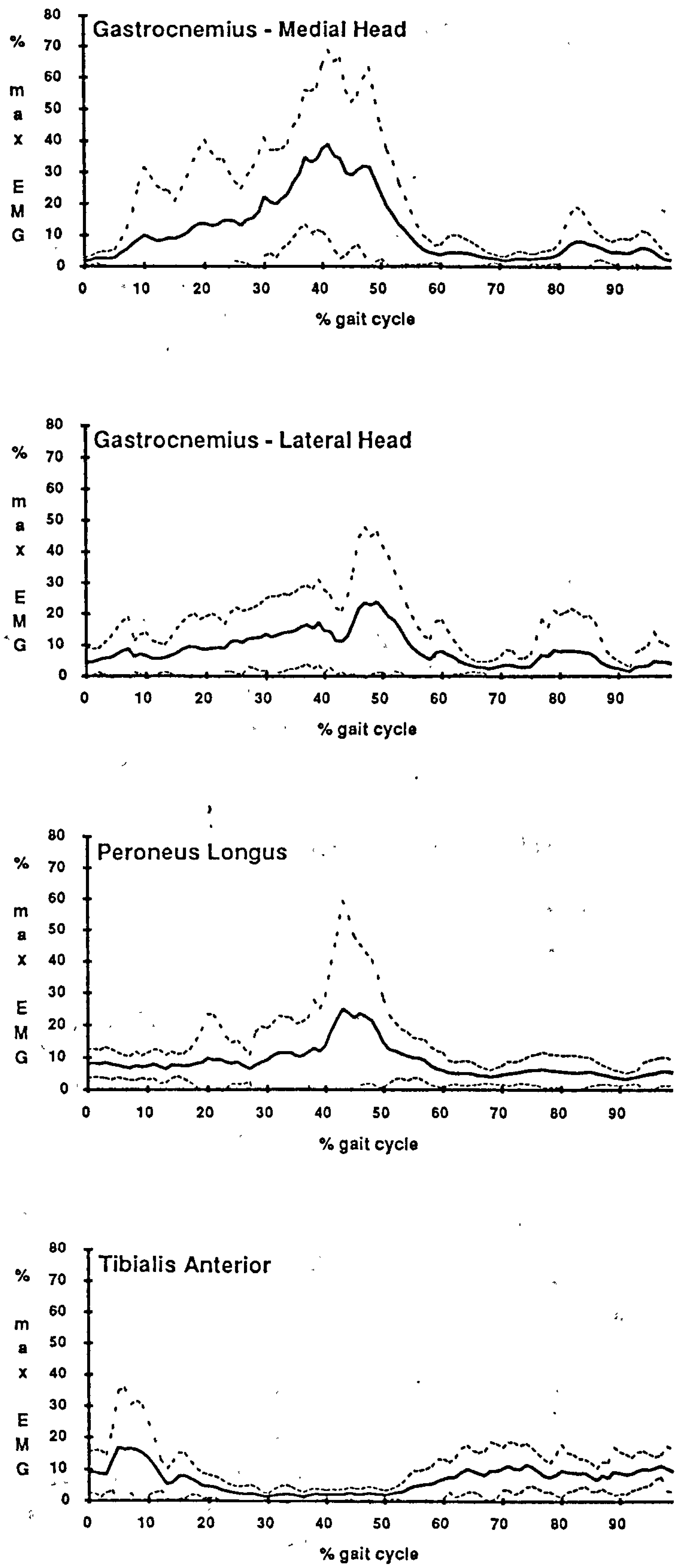


Figure 6.7 Mean EMG activity (solid line) +/- 1SD from the extrinsic foot muscles recorded while subjects were walking wearing their normal shoes . (N = 6)

BAREFOOT

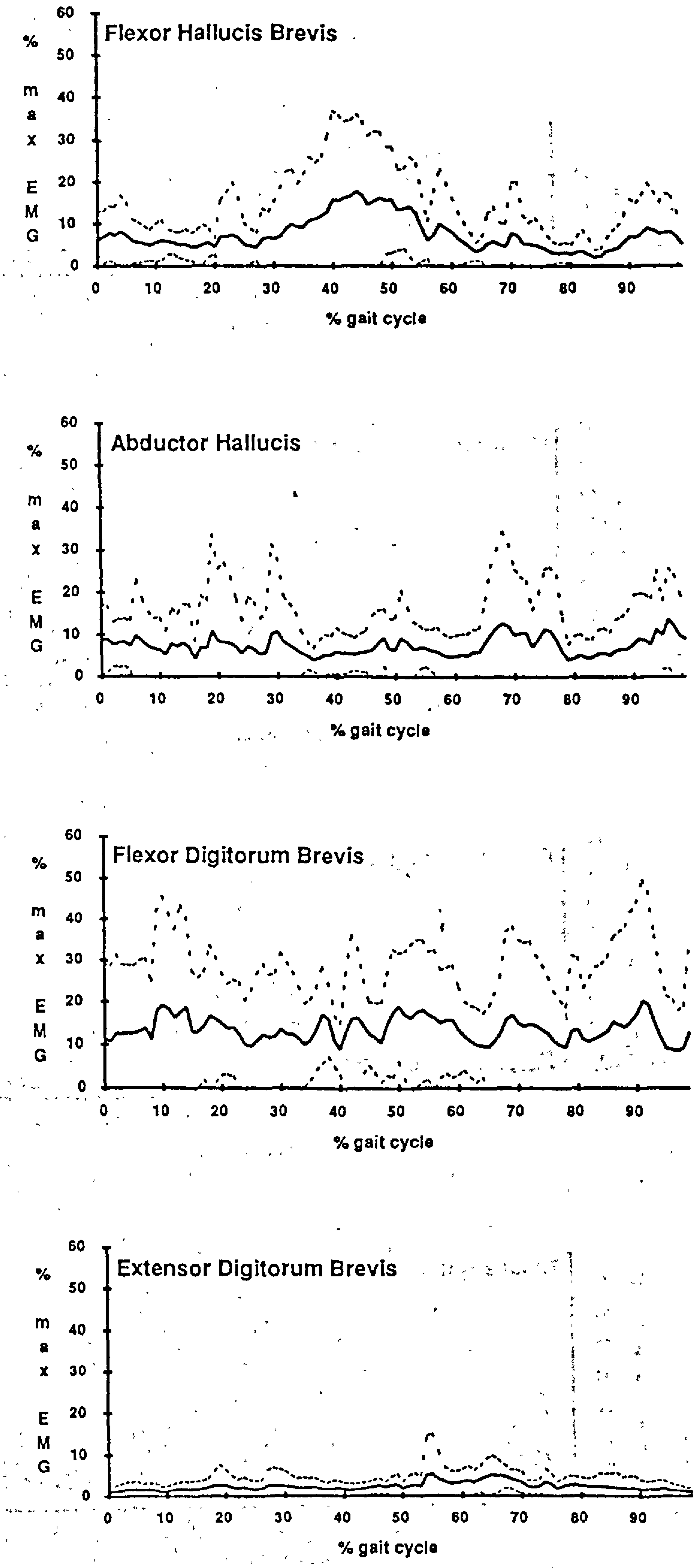


Figure 6.8 Mean EMG activity (solid line) +/- 1SD from the intrinsic foot muscles recorded while subjects were walking barefoot. (N = 6)

Medial gastrocnemius rose steadily to 43% EMG at 35% gait cycle from a 2% level at heel strike. A rapid fall in signal followed after the 45% gait cycle mark was reached, to 2% EMG at toe off and it remained low during the swing phase. Wearing the heel plate appeared to advance and the shoes to retard the peak activity by about 5% from the bare foot peak.

Lateral gastrocnemius indicated a higher activity level at heel strike (5% EMG) but rose to only 30% EMG at 40% gait cycle. The signal decreased rapidly to 4% EMG at toe off where it remained for the rest of the cycle. Wearing the heel plate introduced an unusual dip in the signal at 35% gait cycle and precipitated a rapid drop of 20% EMG from 45 to 50% gait cycle. Shoe walking drew a similar response to use of the heel plate during stance though the muscle relaxation at toe off was not as severe.

Peroneus longus was active at heel strike (8% EMG) and decreased to 5% EMG at 20% gait cycle before rising to 10% EMG at 23% gait cycle. A trough showed from 30-36% gait cycle and the signal then rose to 20% EMG at 42% gait cycle before reducing to 5% EMG at 51% gait cycle. Apart from slight fluctuations the muscle activity remained at 5% EMG until it began to rise before heel strike. Use of the heel plate produced higher levels of contraction during mid stance with less fluctuation. The peak value was not higher though the relaxation at 50% gait cycle was earlier. With shoes on the peroneus had a very stable signal level (8% EMG) during early to mid stance, rising to 21% EMG at 40% gait cycle and then decreasing steadily in the next 20% of the gait cycle.

Tibialis anterior, the dorsiflexor of the foot, was active at heel strike and increased to 13% EMG before 4% gait cycle. The muscle partially relaxed to 6% EMG at 10% gait cycle and then held until 20% gait cycle before relaxing almost completely (2% EMG). At the end of stance phase the activity rose to be 16% EMG at 70% gait cycle which was generally maintained for the next 10% of the gait cycle and the levels then dropped to 7% EMG at 80% gait cycle. When wearing the heel plate, the switch masked much of the early stance activity of tibialis anterior but activity was reduced to 2% EMG by 30% gait cycle. A rapid rise occured at the end of the stance phase but the peak (12% EMG) was not quite as high as the bare foot. Wearing the shoe again elicited strong contraction, on this occasion a step rise to 17% EMG at 5% gait cycle. Relaxation was delayed as in the bare foot case and a level of 2% EMG was not reached until 30% gait cycle. Activity returned earlier at 52% gait cycle and stabilized to 10% EMG for the remainder of the cycle.

BAREFOOT WITH HEELMARKER

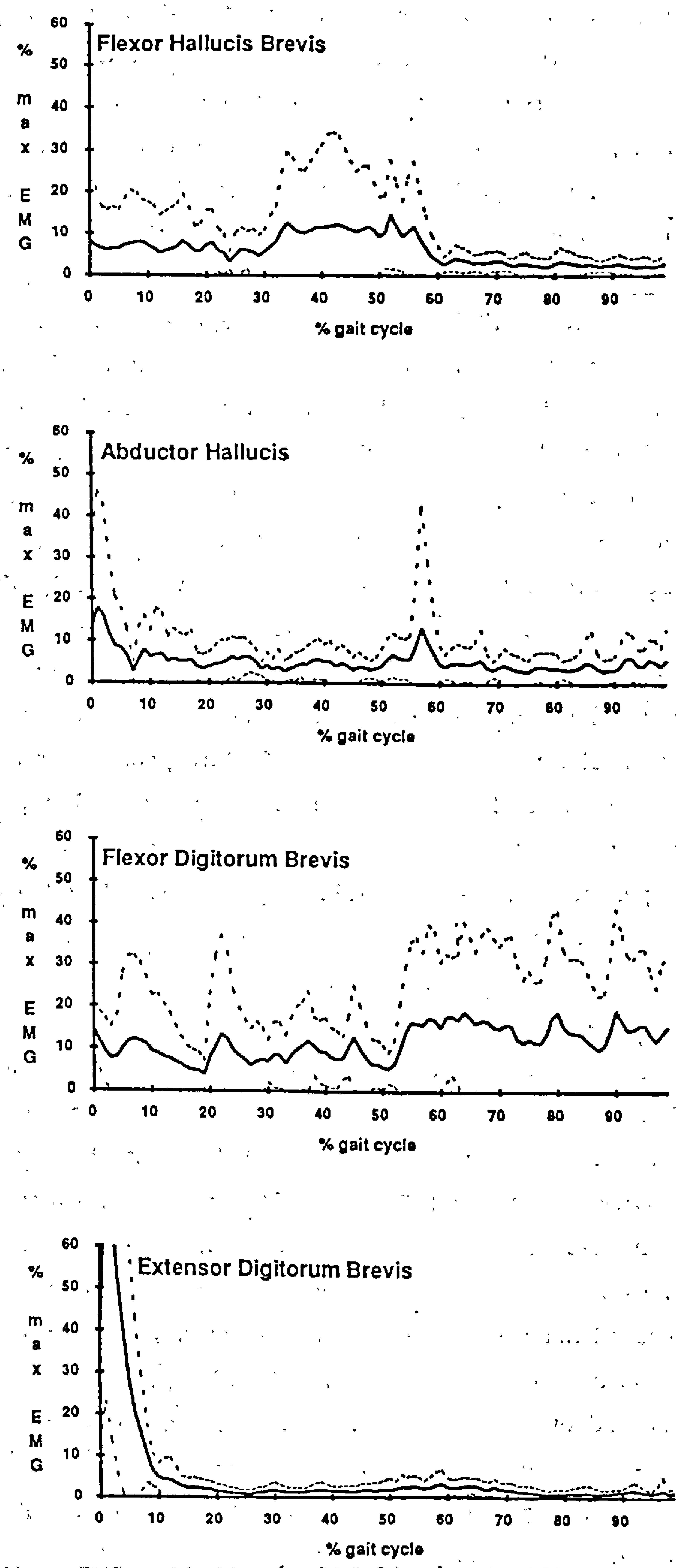


Figure 6.9 Mean EMG activity (solid line) +/- 1SD from the intrinsic foot muscles recorded while subjects were walking barefoot with the heel plate containing the heel switch. (N = 6)

SHOES

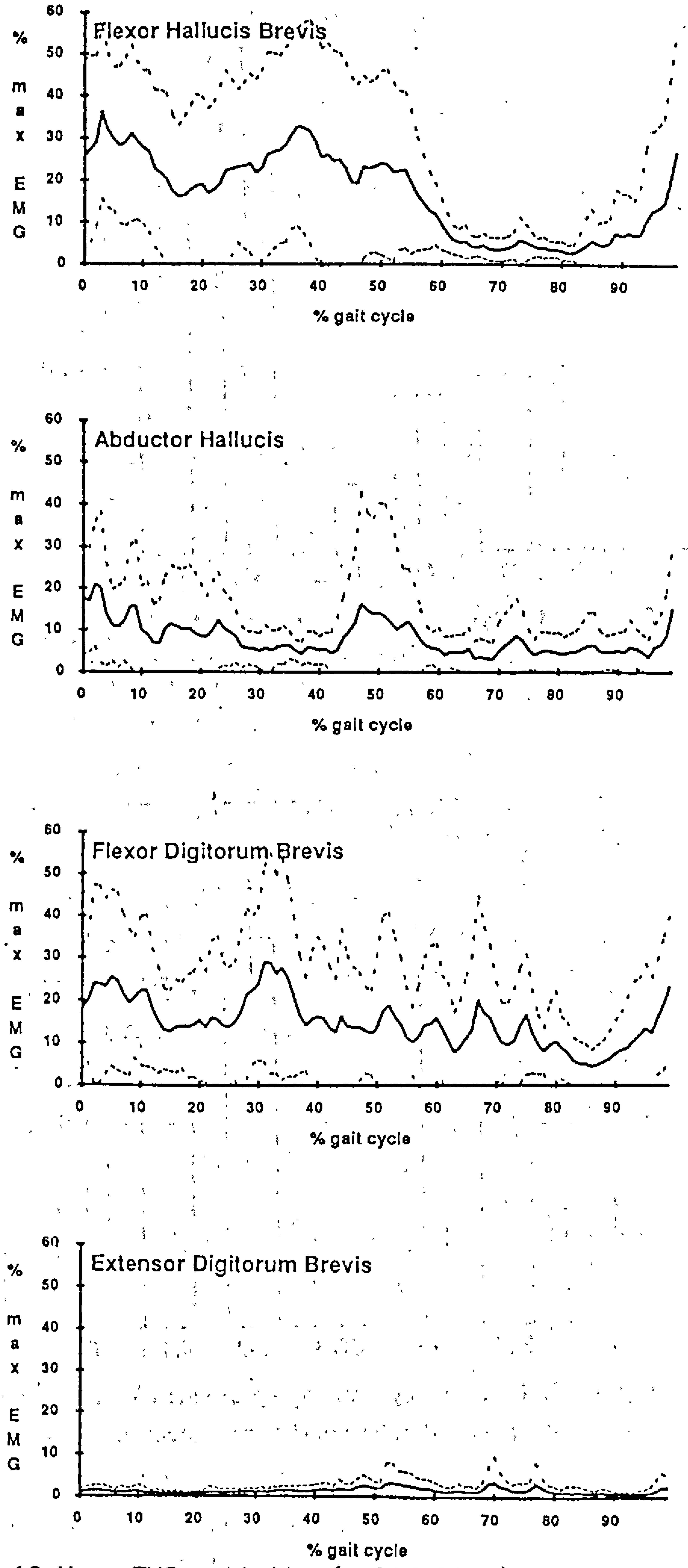


Figure 6.10 Mean EMG activity (solid line) +/- 1SD from the intrinsic foot muscles recorded while subjects were walking wearing their normal shoes. (N = 6)

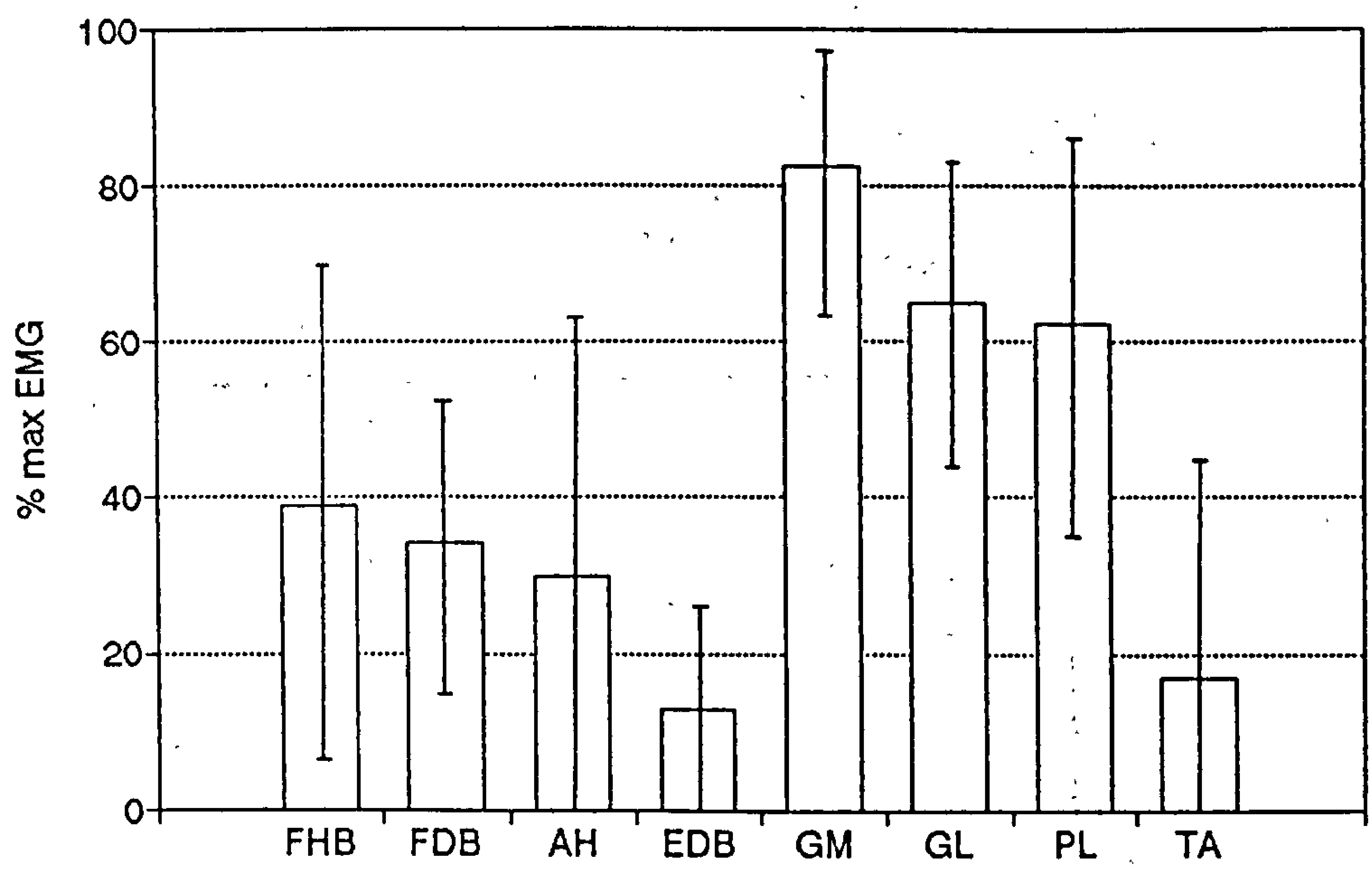


Figure 6.11 EMG activity standing symmetrically on barefeet
Mean with standard deviations shown (n=6)

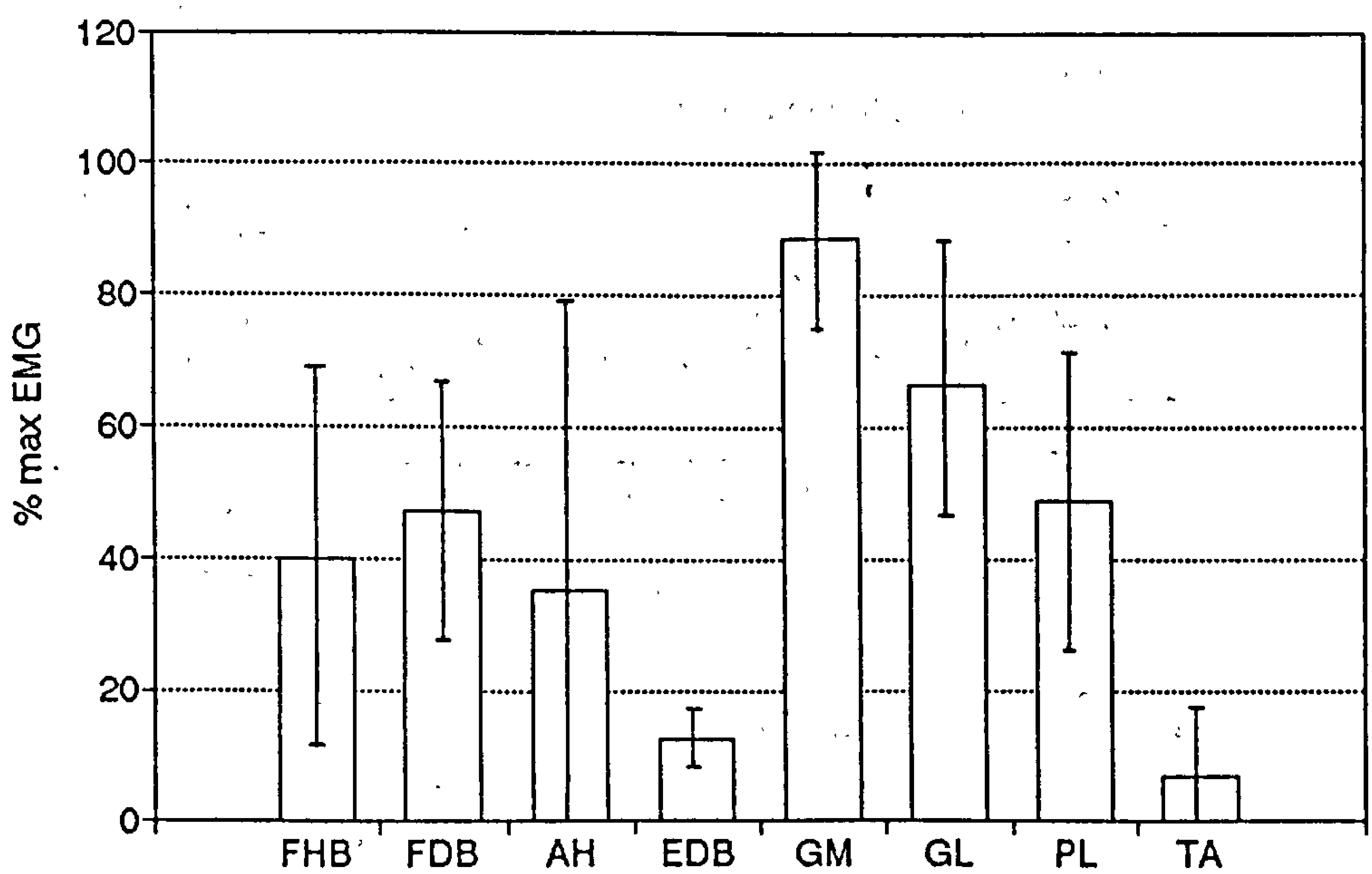


Figure 6.12 EMG activity standing symmetrically on shod feet
Mean with standard deviations shown (n=6)

Flexor hallucis brevis activity was seen throughout the gait cycle though at 30% gait cycle the levels rose significantly to 15-20% EMG at 45% gait cycle (corresponding to the normal heel off point) and then decreased to 5% EMG at the toe off point (60% gait cycle). Use of the heel plate appeared to result in a lower contraction level. Shoe walking dramatically changed the initial stages of stance phase with an increase in both the mean heel strike level (to 30% EMG) and the variation between subjects. The period after 40% gait cycle followed a similar pattern to the bare foot situation.

Abductor hallucis showed a wide degree of variation between and within subjects. Activity levels remained low (< 10% EMG) with only slight rises apparent at 55% gait cycle and at the end of the swing phase. Wearing the heel plate induced a sharp contraction (25% EMG) at heel strike in some subjects. Walking in shoes also evoked higher activity levels (20% EMG at heel strike and 18% EMG at 50% gait cycle) but not during the swing phase.

Flexor digitorum brevis showed a higher "background" level of activity (10% EMG) and no significant variation in activity level during the full cycle. There was a slight rise to 15% EMG at 50% gait cycle with some oscillation (+/- 5% EMG) during the swing phase. Use of the heel plate again changed the activity pattern with an increase of 10% EMG recorded during the swing phase. A higher contraction level was also seen when shoes were worn (25% EMG at heel strike). The peak level of 30% EMG was reached at 30% gait cycle and activity decreased to oscillate at +/- 5% EMG about 15% EMG for the remainder of the cycle.

Extensor digitorum brevis showed relatively low activity in most of the tests. In gait there was a gradual rise to 5% EMG at 65% gait cycle and a subsequent decrease to 1% during swing. The use of the heel plate had very little effect while walking in shoes tended to advance the time of peak contraction by 10% gait cycle.

6.4.2 The Forefoot Support Tests

Activity of the muscles when standing symmetrically on the bare forefeet is presented graphically in figure 6.11. The wide standard deviation bounds were typical of this activity. Figure 6.12 presents the the EMG activity for the same test while wearing shoes. Figures 6.13 and 6.14 contain the details of changes as a result of loading the foot on the lateral or medial side for barefoot and shod foot conditions respectively.

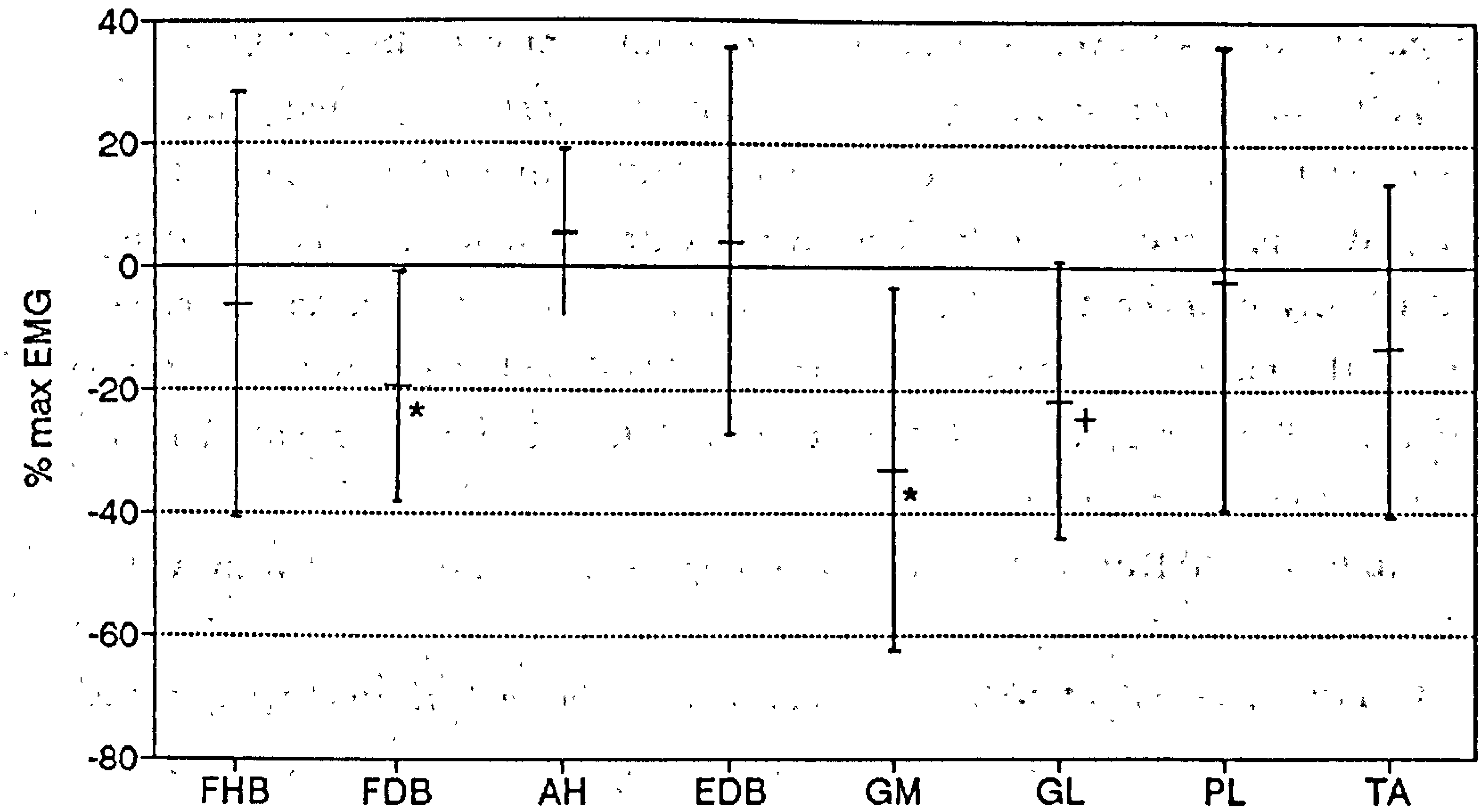


Figure 6.13a The changes in muscle activity levels in bare feet as a result of forefoot force transfer medially (n=6).
(Test for mean <> 0 ;* p<0.05, + p<0.1)

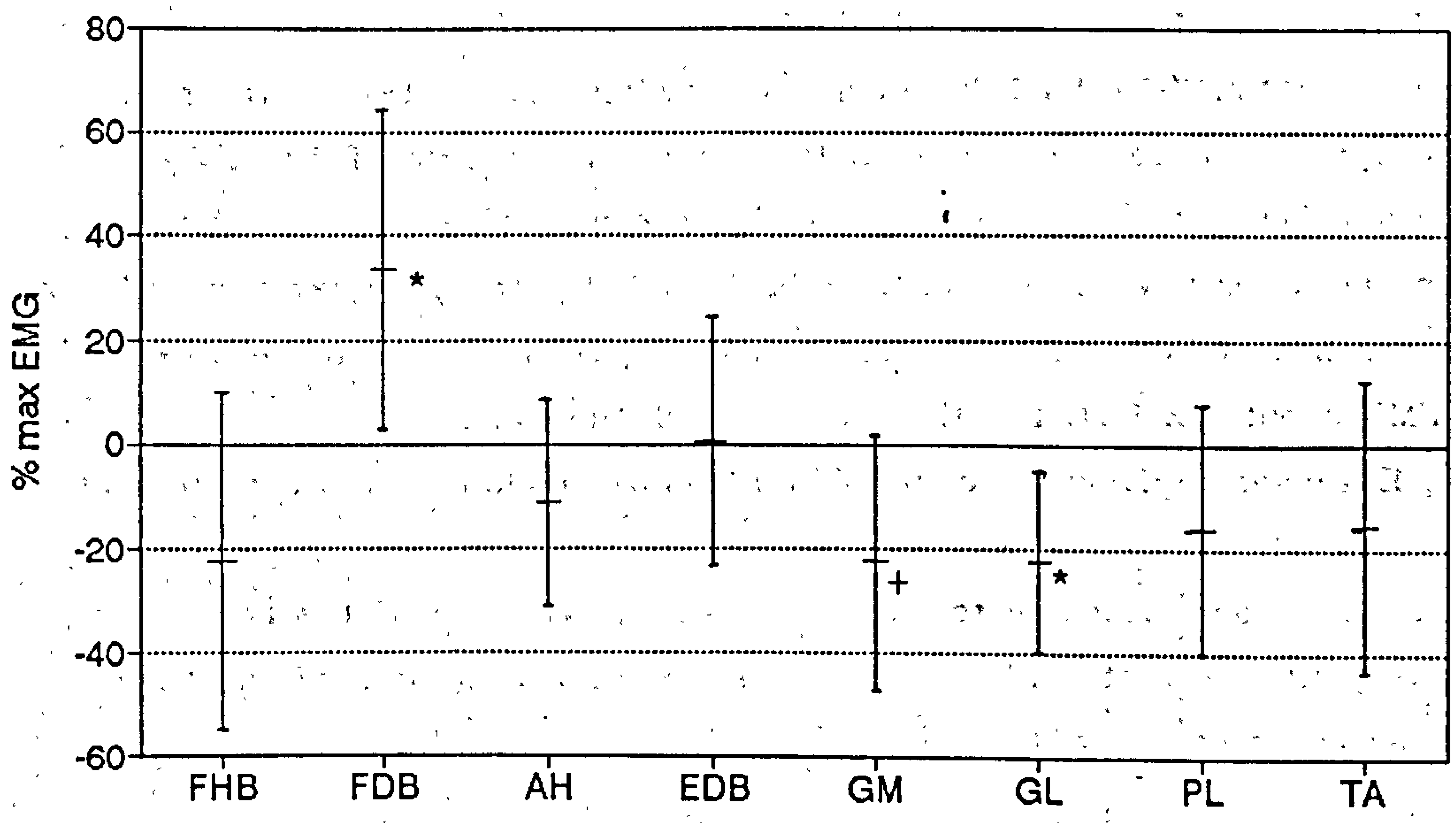


Figure 6.13b The changes in muscle activity levels in bare feet as a result of forefoot force transfer laterally (n=6).
(Test for mean <> 0 ;* p<0.05, + p<0.1)

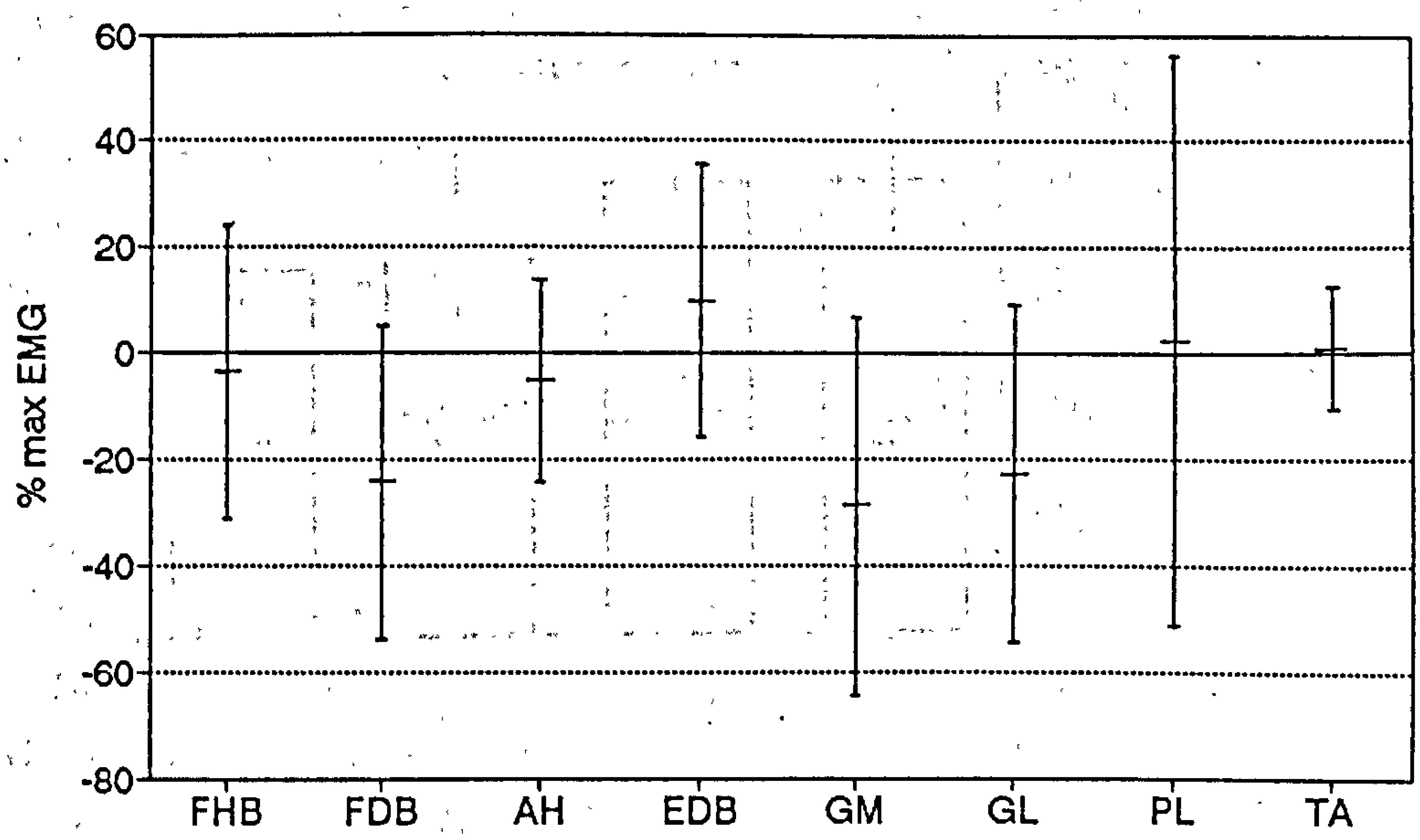


Figure 6.14a The changes in muscle activity in shod feet as a result of forefoot force transfer medially. (n=6)

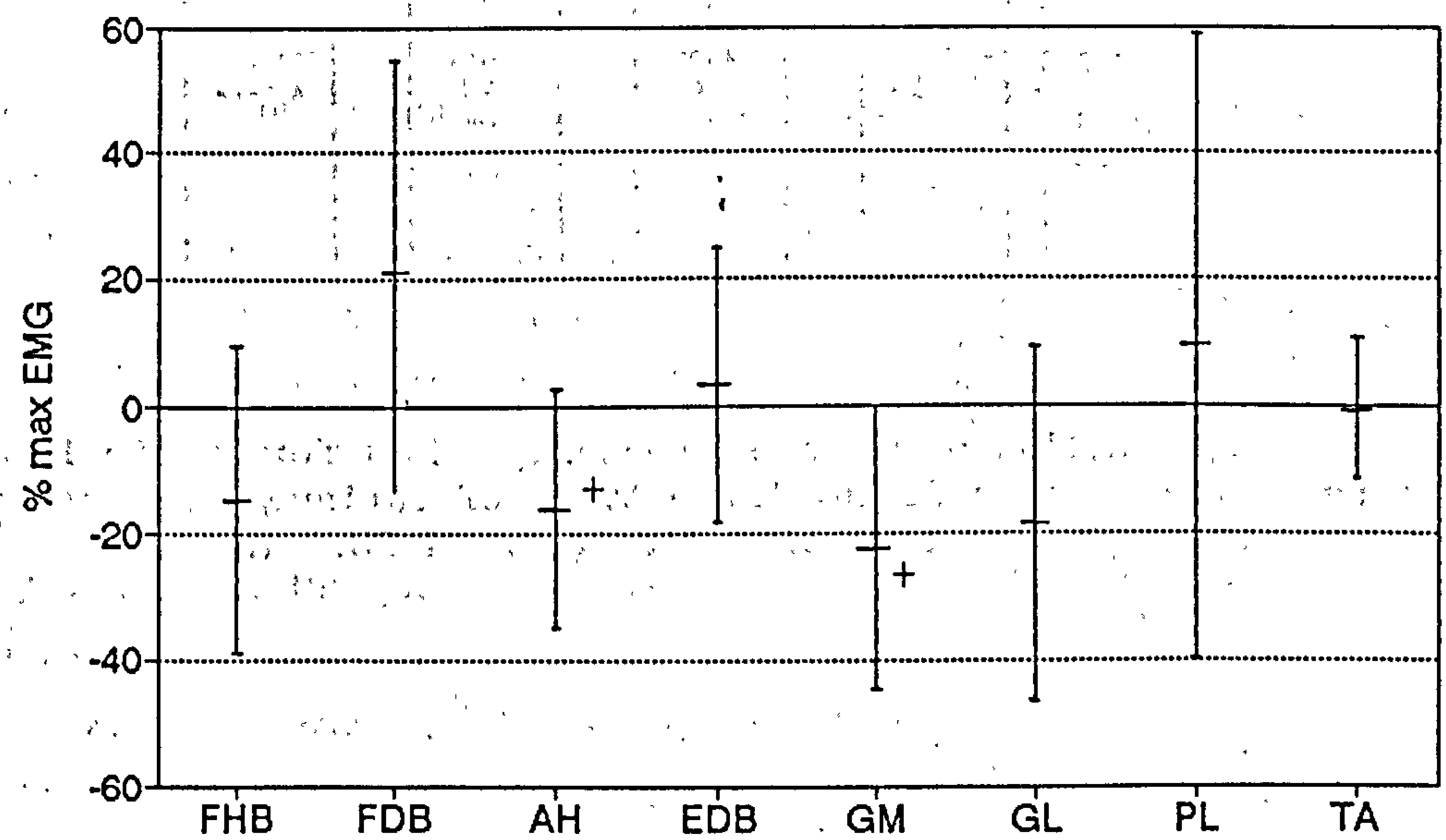


Figure 6.14b The changes in muscle activity in shod feet as a result of forefoot force transfer laterally. (n=6)
(Test for mean <>0; + p<0.1)

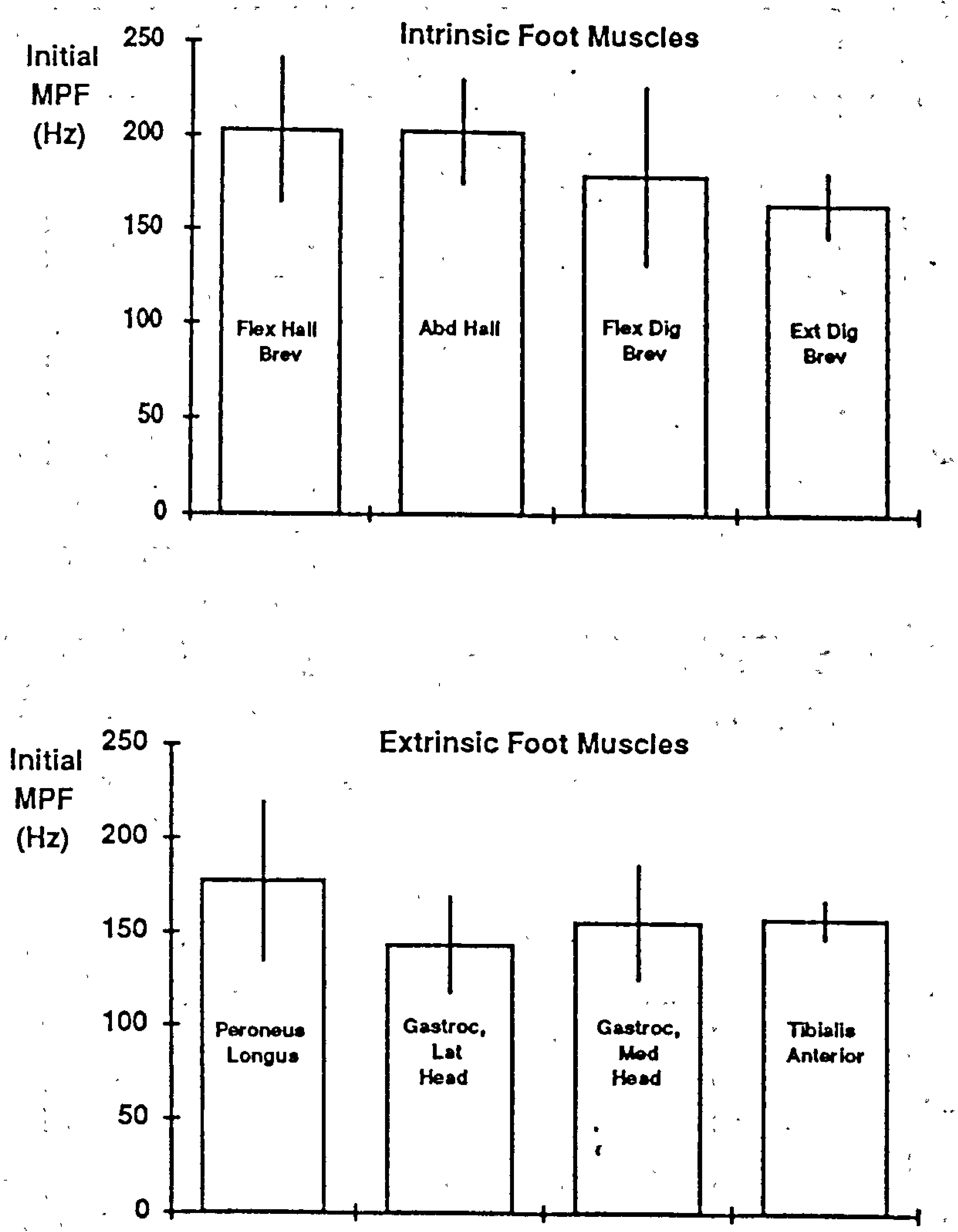


Figure 6.15 The initial mean power frequency (MPF) for intrinsic and extrinsic foot muscles during three types of fatigue contraction. Bars either side of mean indicate +/- 1 SD. (N = 6)

Figures 6.13 and 6.14 also include an indication of changes that are significantly different from zero.

6.4.2 The Fatigue Analysis

The frequencies recorded for each of the muscles in the isometric contraction are shown with their deviation bands in figure 6.15. Figures 6.16 to 6.18 present the normalised time history for each muscle during the three different contraction methods.

6.5 DISCUSSION

6.5.1 The Amplitude Analysis

The results for the lower leg muscles are in general agreement with previous studies (Mann & Inman, 1964; Dubo et al, 1976; Ericson et al, 1986; Winter & Yack, 1987). As Ericson et al (1986) found, the results presented here for the EMG activity of gastrocnemius and tibialis anterior for walking with shoes show a similar pattern to that found by Dubo et al (1976) but is significantly lower in terms of amplitude. The EMG activity presented by Winter & Yack (1987) also follows a similar pattern to all the presented results although the amplitude of Winter & Yack's curves is often quite misleading. Although I am unaware of studies on the amplitude of EMG signals of the intrinsic muscles, Mann & Inman's temporal analysis shows similar timing aspects.

Both heads of gastrocnemius were active as expected in the later half of stance phase though as Ericson et al (1986) have previously shown the lateral head was less active (30% EMG) than the medial head (43%) at their peaks (detail that was obscured by the normalisation process used by Winter & Yack, 1987). The peak activity of the lateral head was also delayed to 5% of the gait cycle after the medial head. Peroneus longus exhibited synergetic activity with the lateral gastrocnemius during late stance with some subjects showing more activity in peroneus longus than the lateral gastrocnemius. Peroneus longus was also active at heel strike possibly to aid ankle stability when tibialis anterior is active (this is supported by the in vitro muscle tests presented in figure 5.7 that indicate that these two muscles have equal but opposite in/eversion actions on the forefoot).

The changes associated with the use of shoes and the heel plate are difficult to explain. It is unusual that one should retard and the other advance activity (respectively) in gastrocnemius when both result in a

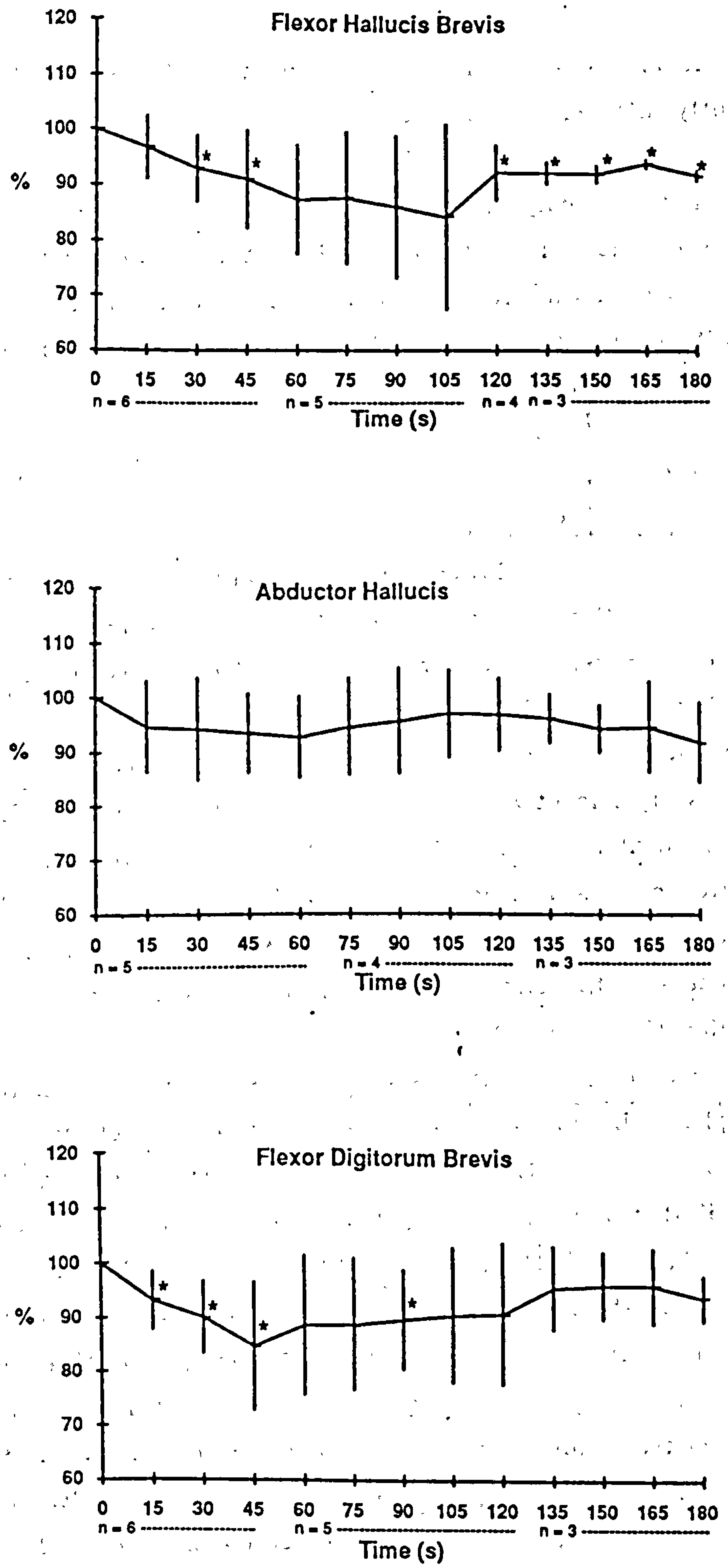


Figure 6.16 MPF normalized to the first reading in the investigated muscles. This fatigue contraction : standing on the forefoot with attempted maximal toe flexion. Bars either side of mean are +/- 1 SD. * indicates a significant decrease (p<0.05) from initial MPF value. n = number of subjects recorded.

slightly raised heel. The effect of the heel shape would be negligible since the foot is in a foot flat position with the reaction force vector passing anteriorly to the ankle when gastrocnemius activity begins. The stable peroneus signal for shod feet suggests greater stability at the ankle.

Tibialis anterior activity was very much as expected and corresponds in pattern with previous studies (Dubo et al, 1976; Ericson et al, 1986; Winter & Yack, 1987) although in the present study the magnitude of activity was noticeably lower than these other studies. The high signal (13% EMG) after heel strike necessary to balance plantar flexing moments fell to 6% EMG when the metatarsal heads made floor contact. The maintenance of this level until 20% gait cycle (1/3 stance phase) may be related to the higher loadings on the lateral side of the foot seen during early to mid stance (fig 4.6) since the in vitro muscle tests (fig 5.7) indicate that tibialis anterior is capable of causing such a load transfer. The wearing of shoes may extend this action later in stance. It is interesting to note the delay in the activity at toe off in tibialis anterior to the beginning of swing phase - a period when dorsiflexing action is necessary to provide ground clearance. This pattern of activity is the same as that reported by Dubo et al (1976), Ericson et al (1986), and Winter & Yack (1987).

Flexor hallucis brevis and flexor digitorum brevis both showed relatively high activity levels throughout the gait cycle but each peaked around 15-20% EMG during late stance. The effect of the use of shoes was to increase both the variation between subjects and also the maximum EMG levels (to 30 % EMG peak).

Abductor hallucis showed perhaps the widest variation in its activity of all the muscles tested. Activity was generally less than 10% EMG except around late stance. The use of both shoes and the heel plate induced higher peak levels of activity.

Extensor digitorum brevis was relatively subdued in all tests. Signal levels only rose to 5% EMG at 65% gait cycle.

6.5.4. The Forefoot Support Tests

In these tests it is not surprising that the most active muscles were the ankle plantar flexors. Most active was the medial head of gastrocnemius (82.5% EMG) with both the lateral head and peroneus longus the next most active at 65% and 62% EMG respectively. Tibialis anterior

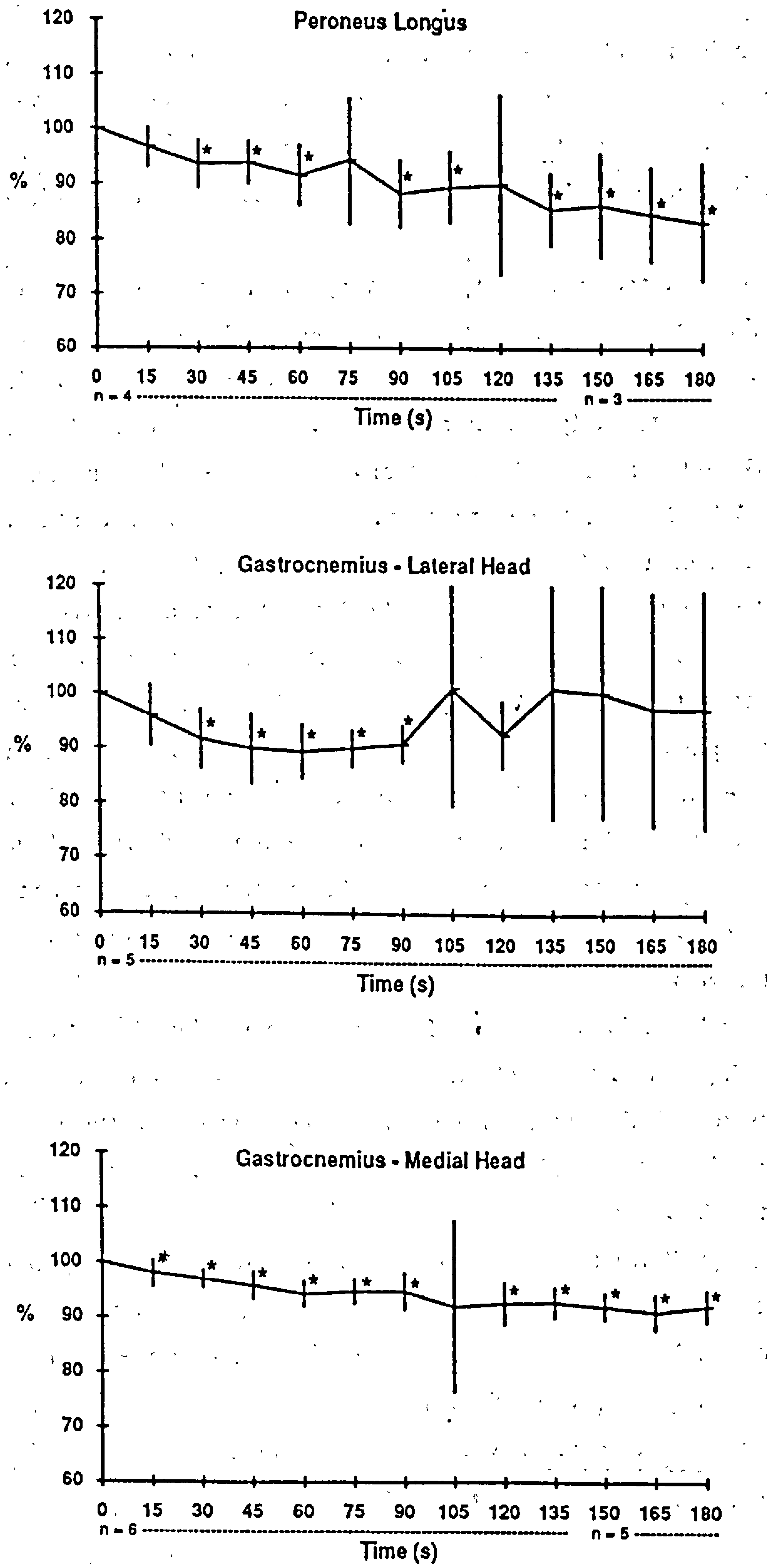


Figure 6.17 MPF normalized to the first reading in the investigated muscles. This fatigue contraction : standing on the forefoot with the heel elevated 20-30 mm. Bars either side of mean are +/- 1 SD. * indicates a significant decrease ($p<0.05$) from initial MPF value. n = number of subjects recorded.

and extensor digitorum brevis were the least active muscles at 17% and 13% EMG respectively.

The changes in activity as a result of the subjects wearing shoes were masked by the wide variation between subjects and, to a lesser extent, shoes. Only peroneus longus (which decreased on average 13.5% EMG) and flexor digitorum brevis (which increased 13.2% EMG) showed any notable changes.

Transferring the loads onto the medial aspect of the foot resulted in a decrease of flexor digitorum brevis and gastrocnemius activity and to a slight extent the activity of tibialis anterior. When wearing shoes the increased variation in the EMG activity masks the changes in the mean levels. The change in muscle activity is similar to that seen in the barefoot case, although none of the changes were significant at the $p < 0.1$ level.

Lateral loading of the foot resulted in changes in the EMG activity levels of several of the muscles. Significant decreases were observed in gastrocnemius activity while the activity of flexor digitorum brevis significantly increased. Activity changes were more variable when the subject wore shoes and peroneus longus and tibialis anterior now indicated *greater* EMG activity levels than when standing symmetrically.

From the tests it appears that the two heads of gastrocnemius act synergistically during forefoot stance although they possess a limited ability to function with a degree of independence. The decreased muscular activity as a result of loading the lateral side of the foot is most likely related to the ability of the muscles to sustain the foot in the two positions. The lateral four rays are all acted on by the same muscles while the first metatarsal and hallux have their own muscular control. As a result standing on the medial side of the foot is equivalent to support almost entirely by the first ray with the associated instability of this position. Lateral loading would be supported on metatarsals two to five since all these bones are plantarflexed by flexores digitorum longus and brevis. Clearly the second loading position is the more stable and also requires less ankle plantar flexion moments since the lateral rays are closer to the ankle joint than the medial rays (this is the basis of the Bojsen-Møller (1978) theory of the two axis foot).

The variation in the muscle EMG activity when wearing shoes during these tests was increased. The only notable change in EMG activity levels when wearing shoes was a slight increase in the activity of tibialis

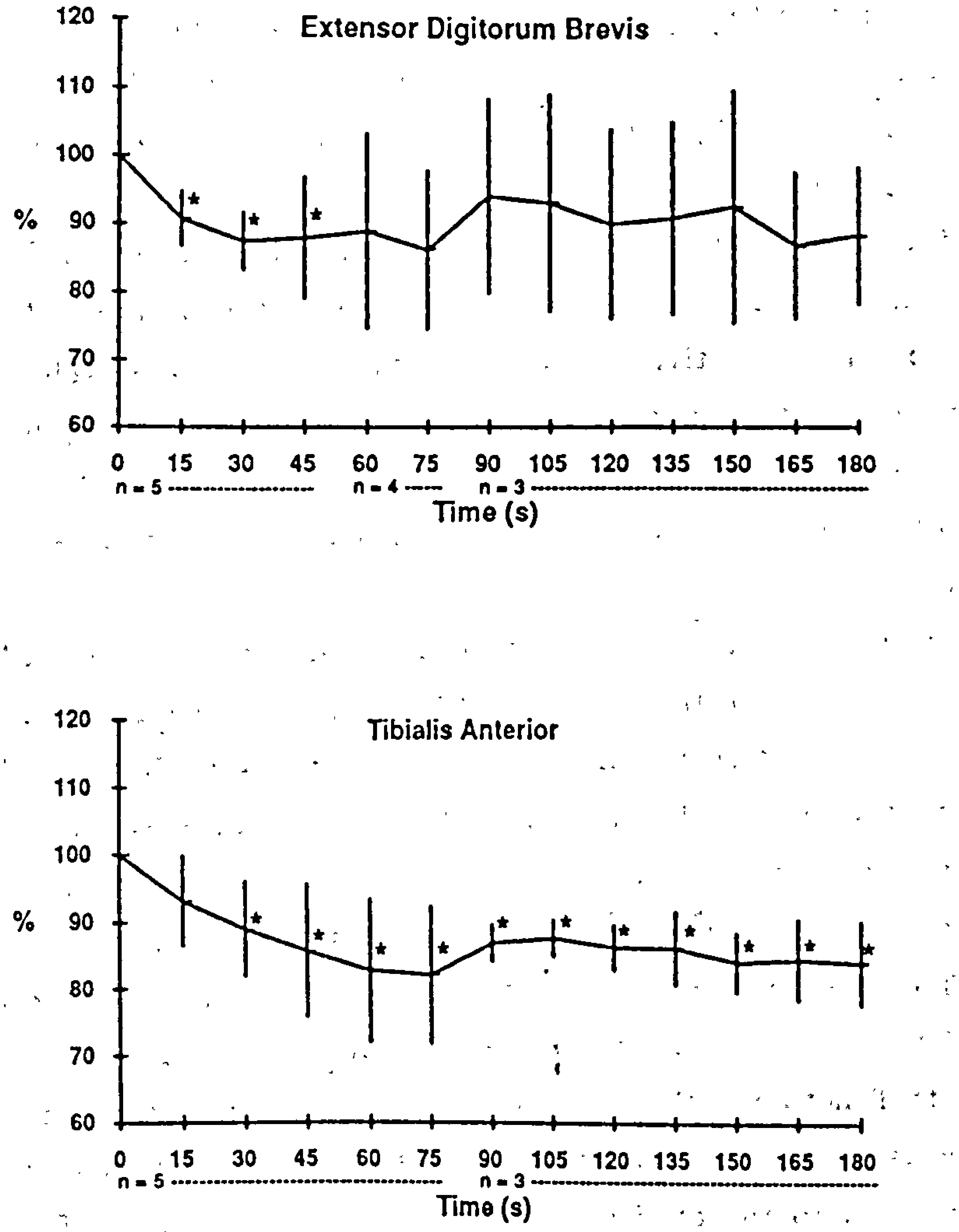


Figure 6.18 MPF normalized to the first reading in the investigated muscles. This fatigue contraction : standing on the heels and everting foot. Bars either side of mean are +/- 1 SD.
* indicates a significant decrease ($p<0.05$) from initial MPF value.
n = number of subjects recorded.

anterior and peroneus longus possibly to provide greater stability at the ankle.

6.5.3. The Fatigue Analysis

One of the most striking results was the relatively high MPF of both flexor hallucis brevis and abductor hallucis when compared with the other muscles. These two are significantly higher (at $p < 0.05$) than all the extrinsic muscles except peroneus longus. The technique of normalizing to the initial MPF allowed a clearer understanding of the frequency changes. Figures 6.16-6.18 indicate those levels that are significantly lower than the initial value (at $p < 0.05$). The figures also indicate the changing numbers of subjects during the tests and in some cases these decreasing numbers were responsible for the changes seen in the curves (eg flexor hallucis brevis and tibialis anterior).

Of particular note from the fatigue curves was that all muscles except abductor hallucis, showed significant changes in frequency within 30s. (Abductor hallucis showed a significant change [$p < 0.1$] at 45 s) This change continued generally up to one minute after which the frequency changes were inconsistent. The mean frequency of some of the muscles began to rise (eg flexor and extensor digitorum brevis and lateral gastrocnemius) while others continued to fall (eg peroneus longus and medial gastrocnemius). In general, accepting the effects due to loss of subjects for flexor hallucis brevis and tibialis anterior and the insignificant changes (at $p < 0.05$) to the frequency of abductor hallucis, the trend after an initial decrease appeared to be a continued decrease for the extrinsic muscle frequencies and a stable or slightly increasing frequency for the intrinsic muscle frequencies. The finding for the extrinsic muscles is in agreement with the frequency changes noted for muscle fatigue (de Luca, 1986). The behaviour of the intrinsic muscles after the first minute is, to the author's knowledge, previously unreported and difficult to explain. It is possibly an indication of an ability of these muscles to cycle the load support between themselves and their multiple heads (allowing contraction and relaxation on a duty cycle) and so maintain longer periods of contraction.

6.6. CONCLUSIONS

Considerable variation is observed in this study, in the recruitment patterns of the various subjects as might be expected from muscles adapted to many different loading patterns. The extrinsic lower leg muscle activity reported here is in agreement with previous reports. Activity levels for the intrinsic muscles were lower (<25% EMG) than those of the extrinsic muscles. The wearing of shoes did result in noticeable changes particularly to the intrinsic muscles.

The tests on the forefoot standing activity indicated support for some of the theories proposed by Bojsen-Møller (1978). There was also extensive activity in the intrinsic muscles during some parts of the loading regime suggesting their added support for the ligamentous structures under heavy loads.

There does appear from the experimental tests to be some functional difference between the intrinsic and the extrinsic musculature. The study of the Mean Power Frequency (MPF) has shown significant decreases in frequency for all muscles other than abductor hallucis, within 30s of beginning a fatiguing contraction. After one minute of contraction, the frequency of the intrinsic muscles stabilised or rose while the frequency of the extrinsic muscles continued to decrease. The frequencies of both flexor and abductor hallucis were significantly higher than the majority of muscles beginning contraction at a mean of 203Hz. The reasons and consequences for these differences in behaviour will require further investigation before definitive conclusions can be drawn.

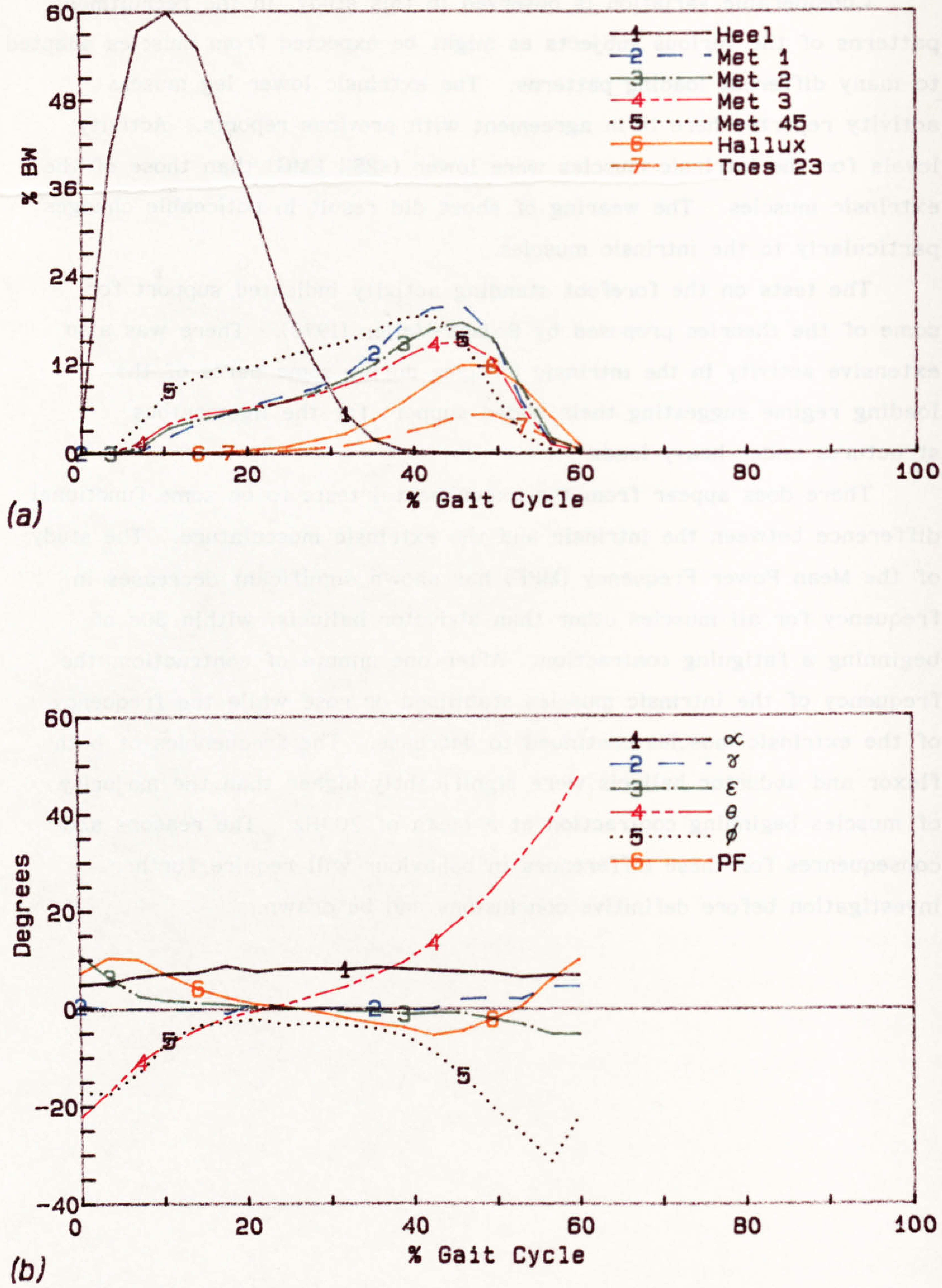


Figure 7.1 Median curves for a) forces and b) foot angles during walking (from Ch 4), normalised to the full gait cycle.

CHAPTER 7 SUMMARY AND CONCLUSIONS

7.1 REVIEW

The work presented in this thesis has followed three distinct lines of enquiry.

Firstly Chapter 4 considered the foot and ankle in vivo to determine the applied forces and the resulting kinematic motions during natural walking. The results were presented in figure 4.10 at the end of that chapter in the form of 'median' curves of:

a) force under seven discrete areas and

b) seven angles that describe the motion of the foot and lower leg.

In Chapter 4 these curves were all plotted against frame number based on the pedobarographic records. In this case frames 1 and 18 corresponded to heel strike and toe off respectively. In figure 7.1 these curves are redrawn against the full gait cycle from heel strike to heel strike of the same foot.

In Chapter 5 the results were presented of tests conducted on the foot in vitro to assess the behaviour of the soft tissues during dynamic loading. Due to the limited availability of suitable test material no statistical conclusions could be drawn though several aspects of normal and pathological behaviour were apparent. The effects of applying tension to three of the extrinsic foot muscles were also quantified.

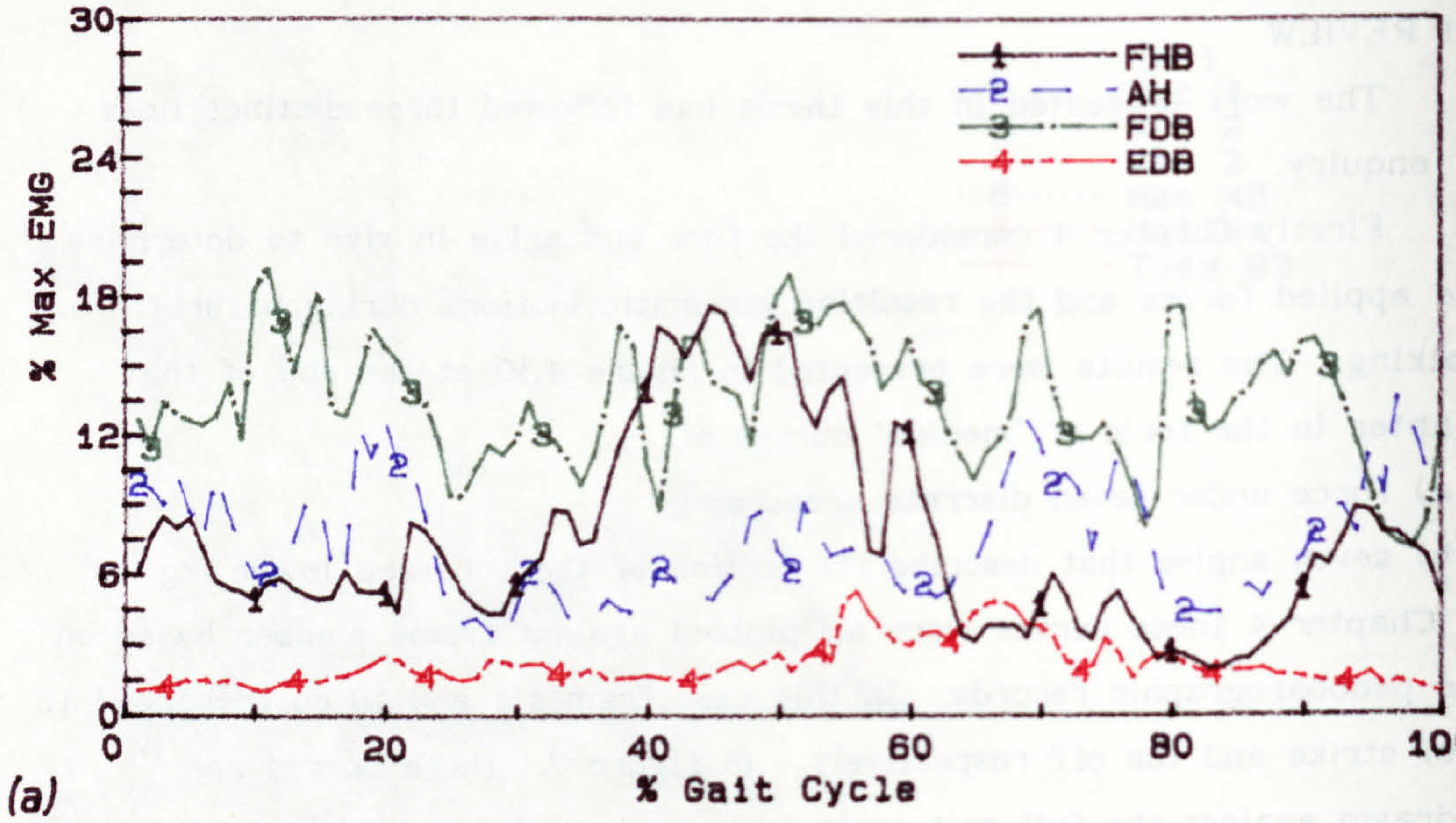
Finally in Chapter 6 the role of the muscles, both extrinsic and intrinsic to the foot, was investigated. Several activities were studied and firm conclusions drawn. The study of the EMG activity levels during natural barefoot walking produced a series of curves normalised to a full gait cycle and these are reproduced in figure 7.2 for the extrinsic and intrinsic muscles respectively.

The following section will consider all these studies together and how they relate to the foot action during normal walking.

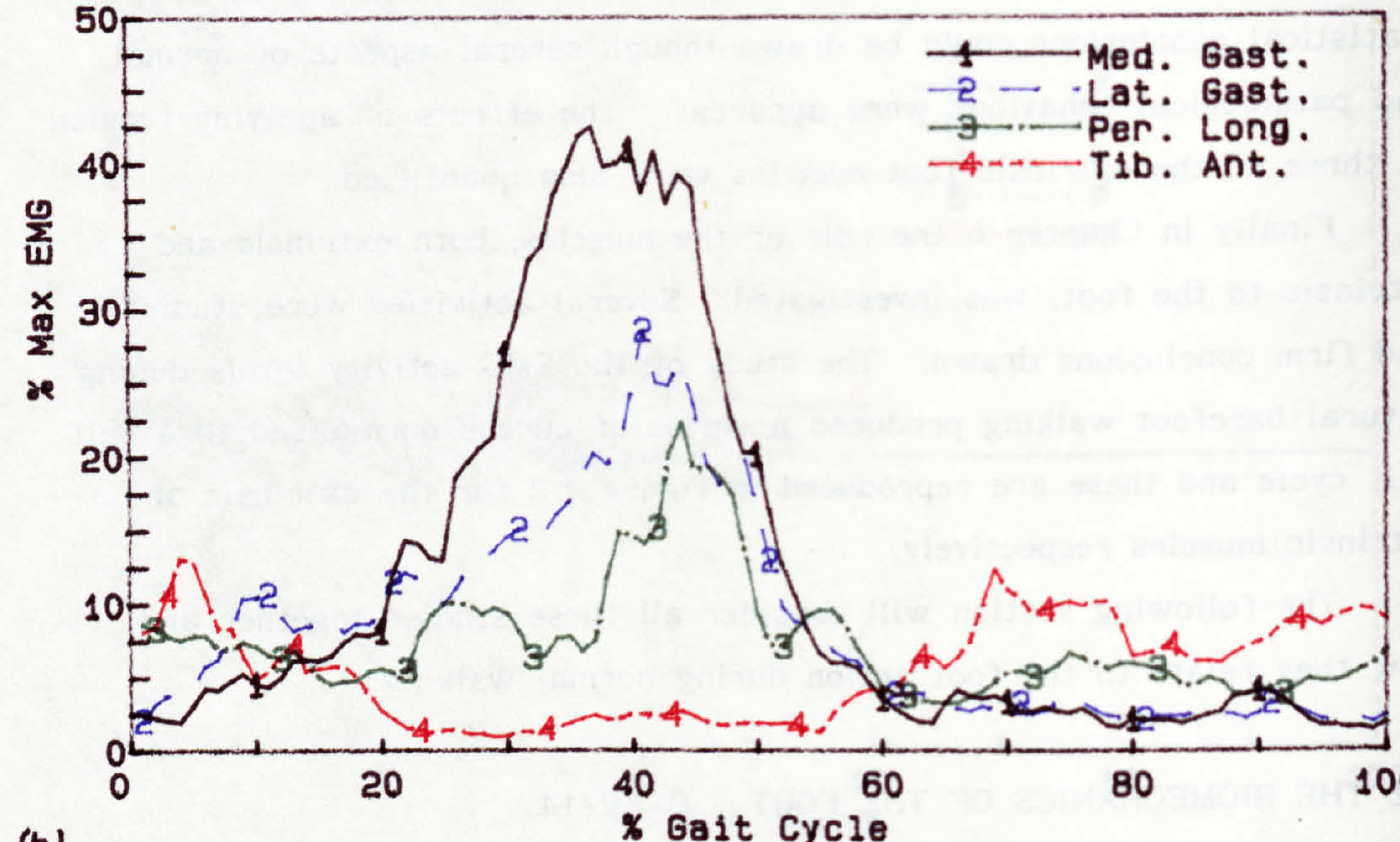
7.2 THE BIOMECHANICS OF THE FOOT - OVERALL

7.2.1 Introduction

The mechanics of the behaviour of the foot during normal walking is summarised by the curves presented in figures 7.1 and 7.2. These curves represent *typical* behaviour and the reader is referred to the earlier chapters for the detail of variation. All subsequent comments in this section relate to this typical behaviour as seen in figures 7.1 and 7.2.



(a)



(b)

Figure 7.2: A summary of the mean EMG levels (N=12) for a) intrinsic and b) extrinsic foot muscles during walking barefoot.

In order to provide a brief overview of the behaviour the gait cycle (GC) will be divided into five typical phases (Rodgers, 1988). These phases are separated by five events associated with stance phase: heel-strike (HS), foot flat (FF), mid-stance (MS), heel rise (HR) and toe-off (TO).

7.2.2 Heel-Strike to Foot Flat

Heel-strike was accompanied by activity in tibialis anterior, peroneus longus and to an extent all of the toe flexors. The two extrinsic muscles act as antagonists and would stabilize the ankle during this contact phase. The foot was slightly plantar flexed and pronated at the ankle, while the forefoot was 10° inverted with the toes dorsiflexed 15°.

The heel force reached its peak shortly after 10% of the cycle (see 4.3.1 for details of limitations of heel strike measurements). It is unclear why forefoot contact occurred before the accepted time for FF (20% of the gait cycle in Rodgers, 1988), for even the angular records indicate that the lateral foot angle (δ) was zero at 13% of the cycle. All the muscles exhibited a change in EMG activity during this period with the exception of extensor digitorum brevis which remained relatively silent. The toe flexor activity would control the degree of toe dorsiflexion (fig 7.1b) and produce antagonistic contraction to the extensors which would be assisting in the resistance of the ankle plantar flexion moment. The dorsiflexed toes would also tension the plantar aponeurosis and so raise the longitudinal arch in preparation for FF.

7.2.3 Foot Flat to Mid-Stance

This period is characterised by little kinematic change but a steady kinetic change.

After the forefoot (ε) and toe (ϕ) angles had returned to neutral positions the angles of the foot remained stable about zero. The ankle continued to dorsiflex passing through neutral at 20% GC. The lower leg was up to 10° inverted (ie the distal end was more medial than the proximal end) and this would allow the pelvis (and more importantly the centre of mass) to pass above the foot (in the coronal plane) as closely as possible. The lower leg slowed in its progression over the foot (θ) at a time that corresponded with knee extension (Mann & Hagy, 1980).

The forces progressed forward under the foot and during this time it was the lateral side of the foot that supported the majority of the forefoot force. The toes made contact at 20% GC and began to increase their force support although at MS the force levels were minimal.

The extrinsic muscles also produced a steady change in activity during this phase in line with the changes in support force. The plantar flexors increased activity to all produce 10-15% max EMG at MS while tibialis anterior decreased activity to a minimum after 20% GC. The toe flexors all showed moderately low activity (4-8% max EMG) although flexor digitorum brevis rose to 12% max EMG between 15 and 25% GC.

7.2.4 Mid-Stance to Heel Rise

The HR event was difficult to define since the force decreased to zero at 40% GC but the foot angle to the ground (δ) indicated movement shortly after MS. This may be associated with soft tissue compression and so HS will be marked as the time when the supported force fell below 2% BW at 38% GC. As with FF, HR occurs well before the time indicated by the literature (Rodgers, 1988) although the literature appears to associate the event with the beginning of ankle plantar flexion which does not begin until after 40% GC (PF; fig 7.1b).

The force is entirely transferred to the forefoot during this interval and there is also more force borne on the medial metatarsal heads than the lateral two heads.

The forward progression of the leg (θ; fig 7.1b) increased after the slowing in the previous stage and the foot itself began to follow this motion. This meant that the ankle ceased to dorsiflex and after a short period when there was no change in angle, it began to plantar flex. The progression of the foot over the metatarsal heads resulted in a corresponding dorsiflexion of the toes producing the windlass effect in the plantar aponeurosis and strengthening the longitudinal arch.

Due to the small motion at the ankle ($< 5^\circ$) during this phase the extrinsic musculature could be considered to have sustained isometric contraction (see fig 4.11 for the basis of the following comments). The stretch of tendo calcaneus can be shown to account for up to 50% of the mid stance ankle motion. The remaining 50% would result in a change in muscle length of only 2.6mm. Both heads of gastrocnemius showed higher activity during this stage and the medial head dramatically rose to over 30% max EMG shortly after MS.

7.2.5 Heel Rise to Toe-Off

As has already been stated this stage of stance is not clearly distinguished from the previous stage. The lateral two metatarsal heads were the first to sustain their peak force soon after 40% GC. The medial metatarsal head force peaked at 45% GC while the toes attained maximum support at 50% GC. During these peaks over 60% BW was supported by the metatarsal heads and the first three toes carried over 15% BW. There was a rapid decrease in support force during the last part of stance phase (50-60% GC) as the load was transferred entirely onto the toes before TO.

The ankle began plantar flexing at HO and this continued through to push off. It is interesting to see that the ankle did not pass the neutral position until after 50% GC. The ankle tended to supinate while the forefoot tended to evert as load was transferred into the toes during the final stages of stance. The toes continued to dorsiflex until 56% GC when they were 24° dorsiflexed.

The muscular activity reached a peak during this stage in all the flexor muscles. The medial head of gastrocnemius reached 46% max EMG just prior to 40% GC and the lateral head and peroneus longus attained peak activity shortly after 40% GC. The timing delay of the lateral leg extrinsic muscles corresponds closely with the load transfer, first toward the lateral side of the foot (medial gastrocnemius having an inverting effect) then toward the medial metatarsals (as both lateral gastrocnemius and peroneus longus have an everting action balancing medial gastrocnemius). The everting action of these muscles (lateral head of gastrocnemius and peroneus longus) on the foot would explain the everted forefoot angle measured at TO.

The toe flexors were active throughout this phase (particularly flexor digitorum brevis) although the activity in flexor hallucis brevis and abductor hallucis exhibited a particular peak around 50% GC. The sudden onset of activity in all these intrinsic muscles (as opposed to a steadier increase in the extrinsic ankle flexors) at 40% GC was closely linked with the HO event and the continued increase of force on the forefoot. It is possible that a threshold was reached whereby the ligamentous and soft tissue structures 'recruited' muscular support for the arches as previously proposed by Basmajian and Stecko (1963) in a series of static tests. Furthermore, the increase in activity of flexor digitorum brevis before the other two toe flexors may be due to the higher loads supported earlier by the lateral side of the foot. This would suggest that the lateral supporting ligaments (the long plantar ligament primarily) also contain 'stretch' sensors that recruit the lateral flexors.

Between 40% and 60% GC extensor digitorum brevis sustained a steady increase in EMG activity. Again this may be associated with the medial shift in the forefoot forces and corresponds to the eversion of the forefoot prior to TO.

7.2.6 Toe Off to Heel Strike (Swing Phase)

No angle data is available for the swing phase due to the speed of progression. Of the extrinsic muscles only tibialis anterior showed any major activity during swing phase. Surprisingly this activity did not commence until after TO but would have been maximal prior to mid swing. The combined effects of tibialis anterior, extensor digitorum brevis and presumably the other toe extensors, would have dorsiflexed the ankle sufficiently to allow the toes to clear the floor during swing through. There was some activity in peroneus longus that may be antagonistic to tibialis anterior to prevent excessive inversion of the foot both during swing and more importantly at HS.

There was moderate intrinsic toe flexor activity during the swing phase. Flexor digitorum brevis maintained the activity associated with push off until 70% GC. Abductor hallucis showed heightened activity prior to heel strike almost to peak levels. The reasons for these activity patterns during swing phase are not clear although they may be antagonistic contractions to toe extensor muscles.

7.2.7 Overall Effects of the Wearing of Shoes

The wearing of shoes resulted in a general increase in the activity of the intrinsic muscles during stance phase. This may be the result of an antagonistic contraction to oppose passive dorsiflexion of the toes due to impingement of the front of the shoe (dorsiflexing the MTP joint and plantar flexing the first PIP joint). The increase in variation seen in these muscles when wearing shoes may be related to the movement of the foot within the shoe (associated with the firmness of fit) accentuating the above effect on the toes.

7.3 SUMMARY OF OTHER FINDINGS

The previous section has shown how the many elements of the foot act together during normal human walking. The ligamentous contribution to this may have seemed to be negligible but it is important to bear in mind the role of several of the plantar ligaments in providing the first level of support for the arches and then in 'regulating' the muscular action necessary to provide greater load support. One of the most interesting results of the in vitro experiments (Chapter 5) was the speed independence of the ligamentous system which confirms the findings of Ker et al (1987). The results also indicated a relatively constant load sharing arrangement among the passive plantar structures independent of the applied load. This suggests that all the elements of the foot are involved in load support to maintain an appropriate load distribution under the foot. Chapter 5 also showed that many of the plantar structures contributed particularly when the foot was inverted or everted, such as on uneven walking surfaces which were not used during these tests. The changes in EMG activity seen when subjects stood on their toes (Chapter 6) further strengthen the theory of the intrinsic muscles acting as additional arch supports.

The in vitro tests on the feet with a hallux valgus deformity confirmed the effect of the condition on the action of the plantar aponeurosis in supporting the medial arches. Loads were in general higher on the lateral rays in the feet with this condition when compared with the normal feet. With all the plantar structures removed the feet with hallux valgus were found to have depressed medial arches suggesting permanent damage to the interosseus and associated ligaments. In this condition the load was higher on the medial side, a situation that would continue to damage these ligaments.

The fatigue study on the muscles (Chapter 6) highlighted the different EMG activity associated with the intrinsic muscles. Two of the muscles produced a higher Mean Power Frequency than has been shown for trunk and arm muscles (de Luca, 1986) and these differences were not apparently associated with electrode filtering effects. Although the extrinsic and intrinsic muscles showed quite similar fatigue patterns the intrinsic muscles were more individual in their overall responses (abductor hallucis showed no significant fatigue over 3 minutes).

7.4 FUTURE STUDY

This work has provided several new insights into the intricacies of normal foot biomechanics. Several factors need further investigation.

The study of the kinetics and kinematics of the foot needs to be continued and extended. Two areas are worthy of further investigation: normal foot behaviour on uneven (eg sloped) surfaces and the changes associated with several of the orthopaedic foot disorders (eg hallux valgus, hammer toe etc). By using the combined pedobarograph/video system as presented here the results should prove more valuable in providing the data necessary for various dynamic models.

Clearly the in vitro study presented in chapter 5 needs to be continued and extended to further clarify and quantify the role of the plantar structures both in normal and pathological feet. Further consideration needs to be given to the placement of the transducers. It is a difficult measurement problem, when dealing with tissue, of accurately assessing forces and displacements without altering them simply through the measurement process. The tests conducted for this thesis have highlighted the amount of motion that occurs in the foot during loading (eg rolling and extension of the arches) and test schemes that seek to limit this motion may produce very 'good' results simply by restricting normal behaviour artificially. Some account also needs to be made of the moments associated with internal and external rotation as this is a mechanism active through the subtalar joint in altering the inversion and eversion positions of the foot.

The EMG assessment of the foot muscles is also open to further study. The work can be continued by assessing further subjects to produce more definitive results for the general population, but more importantly this work can be extended to consider the muscular activity associated with the forces under the foot (measured using a force plate or pedobarograph), walking on uneven surfaces, or walking in high heeled shoes for example (the effects shown here of simply wearing soft shoes should prompt further studies on the changes associated with wearing shoes). The results reporting the changes in EMG as a result of fatigue in the lower leg and foot muscles also raise several questions. The MPF of several of the muscles was significantly higher than the MPF reported for trunk and upper limb muscles although no reason is apparent. There remains the puzzling fatigue behaviour of the intrinsic muscles which show such varied changes in the EMG MPF.

Finally study needs to continue on the aspect that all this work is directed toward and for which it is presented as a foundation: modelling. The cohesive description of a system is vital to the successful modelling of that system. The work presented here and hopefully in the future serves as the basis for all these models. The current capacity of computer modelling technology makes lifelike animation possible and several groups around the world are focusing increasingly on the computer's ability to handle and describe thousands if not millions of individual effects all responsible for the final behaviour of a body. With this capability it should not be too long before the details described here can be reproduced on the computer screen. When that occurs the understanding of the biomechanics of the foot will have entered a new era.

7.5 CONCLUSIONS

This work was originally conceived as an investigation into the general biomechanics of the foot. This general area of study is vast and so the scope of investigation was restricted to consider normal walking and related activities. In keeping with the aims of the project several new aspects of foot biomechanics during walking were studied.

Three parallel lines of enquiry were pursued :

-in vivo kinematics and kinetics,

-in vitro structural behaviour,

and -in vivo muscle activity.

These parallel experiments allowed the inclusion of preliminary findings from one test in the development of another, so ensuring that the final results would offer a comprehensive foundation for understanding the biomechanics of walking and, to an extent, foot biomechanics in general. The development of experimental conditions and the presentation of results has been intended to facilitate the interpretation of the findings for both clinical and general biomechanical use.

The foot has proved to be an intricate and carefully balanced arrangement of components each acting in concert with the others. The kinetic study has reported a consistent and smooth transfer of force from the heel to the forefoot and finally to the toes at the end of stance phase. Kinematically the angles of the joints of the foot changed smoothly and several of the these angles proved to be related to the force distribution under the foot. As an example, the ankle plantar flexion was inversely related to time and directly related to the forces under the first and third metatarsal heads. It is these forces that the ankle plantar flexion muscles must overcome and increasing plantar flexion in the late stance phase corresponded with the transfer of the forces onto the forefoot.

If the muscle EMG study is considered in relation to these kinematic and kinetic results (see section 7.2) certain factors emerge:

1. The amount of ankle plantar/dorsiflexion during mid-stance was small ($<5^\circ$). The corresponding contraction of the two heads of gastrocnemius muscle could be considered isometric.

2. Peroneus longus and the two heads of gastrocnemius demonstrated (in their EMG signals) a degree of independence in their action. It is suggested that these muscles may be partially responsible for the distribution of load on the forefoot as well as their accepted plantar flexion role.

3. Intrinsic toe plantar flexors appeared to have a 'trigger' level of forefoot load at which their activity dramatically increased. It is proposed that the plantar ligaments are the mediating structures containing these stretch sensors (supporting the suggestions of Jacob [1989]). The intrinsic muscles were also sensitive to the wearing of shoes.

4. The two dorsiflexors of the foot and toes measured during these tests (tibialis anterior and extensor digitorum brevis) acted separately and progressively around toe off to dorsiflex the foot and toes during swing phase for ground clearance.

The fatigue properties, as indicated by the EMG signal, of the intrinsic muscles were significantly different from the extrinsic muscles and other postural muscles. It may be possible that the intrinsic muscles are able to cycle loads among their separate heads or between different muscles in order to limit fatiguing effects.

The tests of the ligaments of the normal foot were not sufficiently extensive to draw statistically valid conclusions although general trends were found:

1. The foot behaviour was generally independent of the frequency of the applied loads, indicating the predominantly elastic behaviour of the plantar structures and further confirming their importance in energy storage as noted by Ker et al (1987).

2. The force distributions under the foot were relatively independent of the applied loads but strongly dependent on the foot orientation. These findings suggest that the ligament system of the foot is particularly involved in maintaining an optimum load distribution under the foot and controlling the foot's adaptation to varied surfaces.

3. The loss of the plantar aponeurosis or long plantar ligament caused the heel loads to diminish and the forefoot loads to partially transfer away from the metatarsals supported by the cut ligament. The plantar skin also affected the load distributions although the effects were less well defined.

4. All the changes associated with the loss of the plantar structures were greatest when the feet were inverted or everted suggesting that it is in these positions that the plantar structures are most important.

Preliminary tests on two feet with hallux valgus deformity would suggest that the condition does reduce the strength of the medial arches with a resulting increase of load on the lateral metatarsals. There was also some indication of permanent deformation of the interosseus and deep plantar ligaments (eg calcaneonavicular). Under high loads the medial arch would collapse without the effective tensioning effect of the plantar aponeurosis. The aponeurosis (if it does contain 'stretch' sensors) would not reach the threshold loads required to activate the strengthening intrinsic musculature until the arch was excessively depressed. In this situation the intrinsic muscles would not be as effective in arch support due to their poorer line of action (closer to the tarsal joints). Tests on three of the extrinsic muscles of the foot confirm the findings of Jones (1941) that extrinsic muscles are not well placed to support fallen arches but can significantly alter the distribution of the forces under the forefoot.

Finally the skeleton of the foot with only interosseus and deep plantar ligaments was capable of supporting several times body weight without sudden failure. Only one specimen did fail under these high loadings and then as the result of a fracture in one of the metatarsal bones.

In summary, the findings of this thesis suggest a general subdivision of load carrying between the structures. During stance phase of normal walking, the muscles act to support ligaments and limit abnormal motions at the joints (such as excessive pro/supination at the ankle) while limiting their energy expenditure by sustaining isometric contractions and utilizing the elastic properties of tendons and ligaments to allow the limited joint motions necessary for steady progression over the foot. In many ways the foot acts like a well controlled sprung lever. After foot flat the body imparts an increasing level of energy to the ankle plantar flexors and their tendons (particularly tendo calcaneus) and also to the plantar ligaments of the foot and the intrinsic musculature and their associated tendons. Toward the end of stance phase much of this stored energy is returned to give the 'lift' and 'push' associated with push off. The idea of "a spring in your step" appears to be very close to the biomechanical truth.

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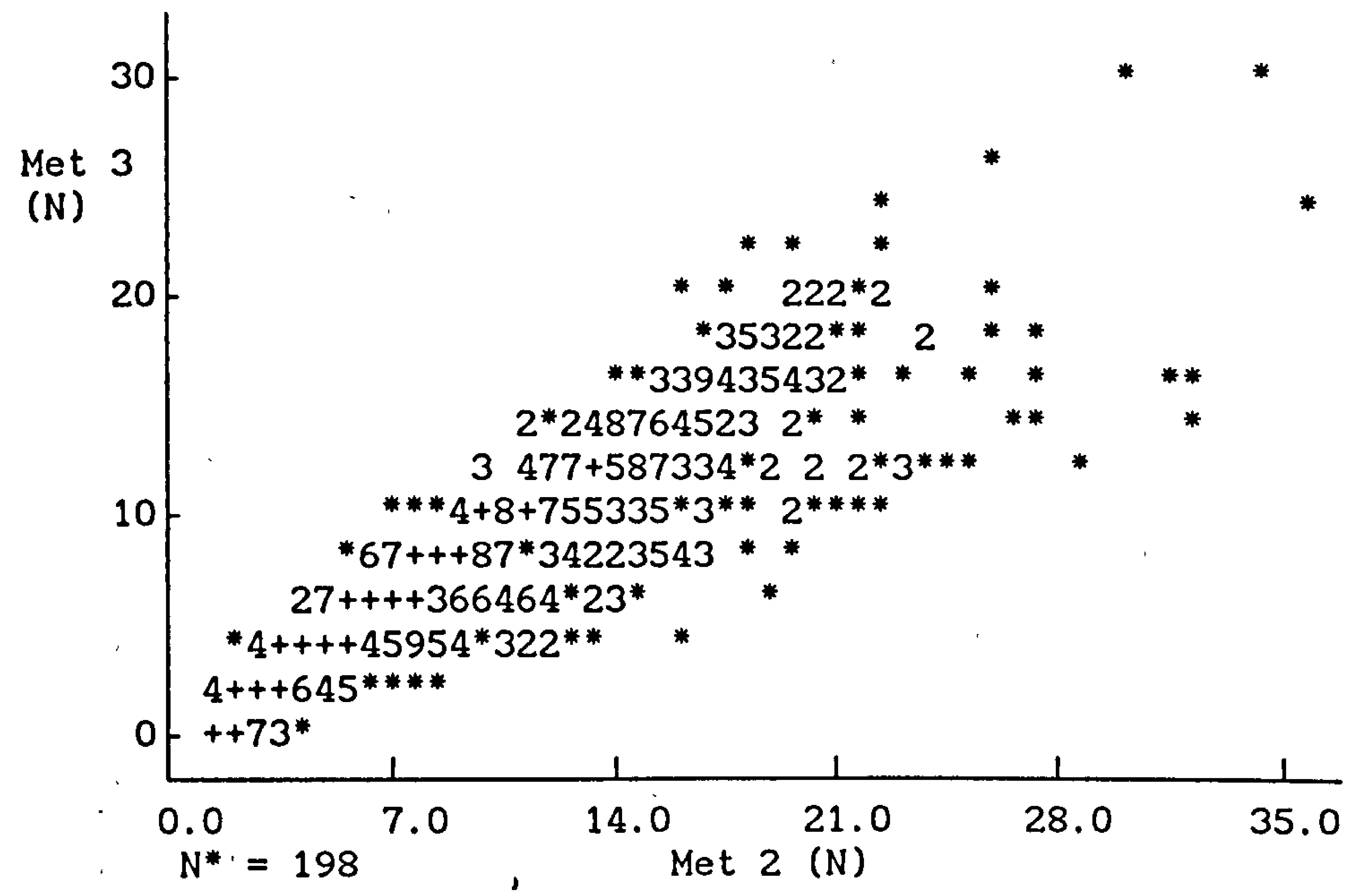
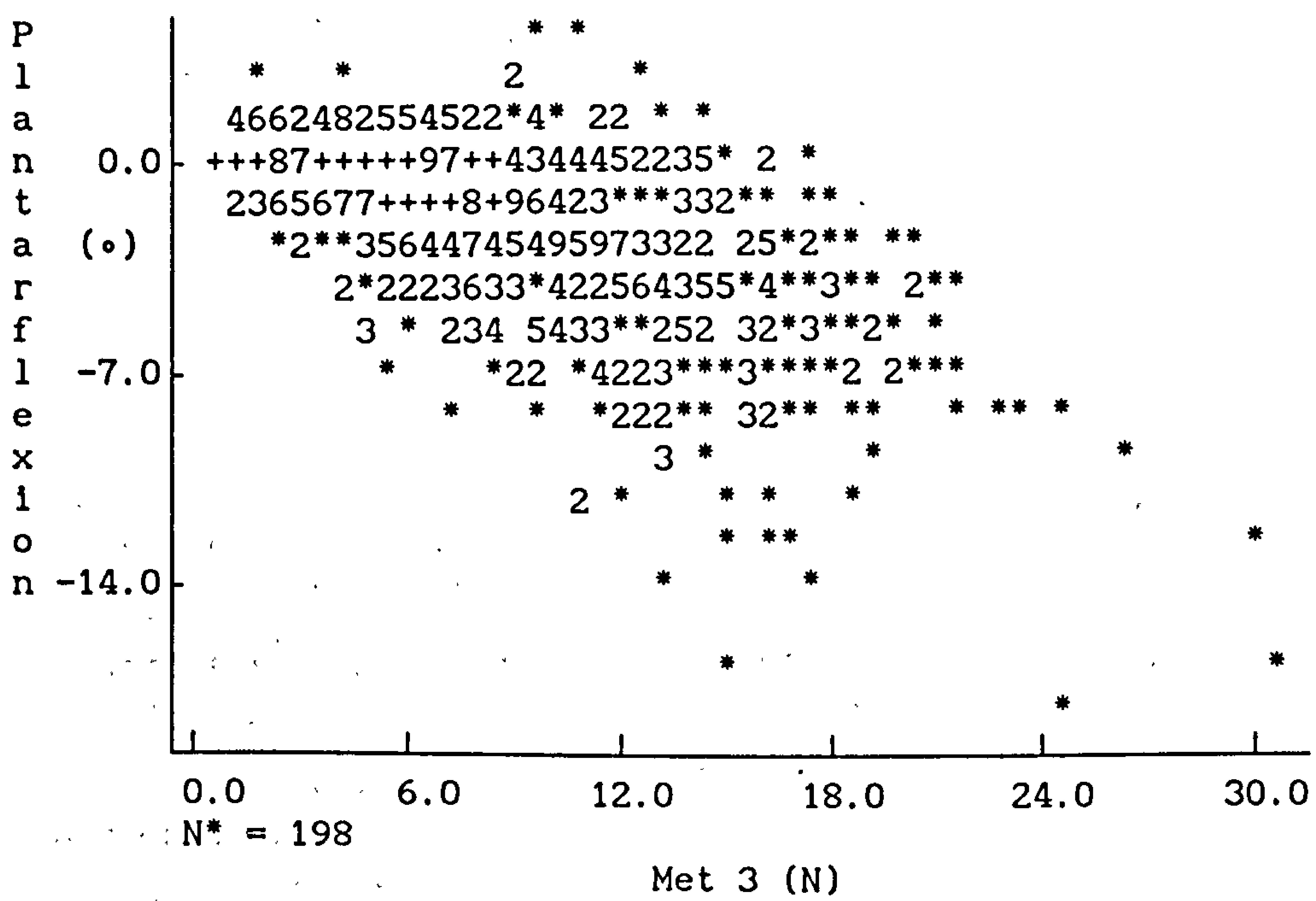
This appendix presents the statistical details produced by the Minitab program (version 7.2, Minitab, Inc, 1989, VAX/VMS operating system) to a regression command. As an introduction, the following scatterplots provide an indication of some of the relationships between the variables. They are based on the data from the LA feet files.





```
           -
        45+
           -
  Met 1    -                                                          *  *
           -
           -                         **                         *   *
        30+                    **23*             *
           -                   *4   ** 2 2*    ***
           -               * 2222    **   **22*     *
           -              **22 2 **3* 42**  * * *
           -        *     *2234*354733324***
        15+            *34554436562655222 **     *          **
           -          *336754+74876455*3222*2 **     2 *
           -       *324++++8756334*2** *3*    **
           -       28+++++++44*42**22**       *
           -    27+++++9722***2   **
        0+ ++++43***   *
            +---------+---------+---------+---------+---------+-----
           0.0       7.0      14.0      21.0      28.0      35.0
           N* = 198                  Met 2
```

<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<>>>>>>>>>>>>>>>>>>>>>>>>>>>>>>>>>>

The remaining part of this appendix presents calculated regression equations and their associated statistics for the variables. The variables that were chosen for the regression were based on results from a Step wise regression analysis made on all predicting variables.

The details calculated in the regression analysis are based on typical regression statistics. Immediately following the regression equation itself are the number of cases used in the calculations (note missing cases are not used in the determination of the regression equation).

-StDev is the standard deviations in the coefficients

-t-ratio is an indication of the weighting of the coefficient; a large t-ratio indicates a coefficient that is significantly different from zero

-p is the significance limits for the coefficients (a p level less than 0.05 would indicate the coefficient is significant)

-s is the standard deviation of the data about the regression line (a smaller s indicates a closer fit)

-R^2 is an indication of the proportion of variation in the dependent variable described by the regression equation (100% indicates that all data lies on the regression line)

Under the section on analysis of variance the test of the calculated regression equation is further detailed:

Regression is the details associated with the regression equation (explained variation), Error is the details of the error from the equation (unexplained variation) and Total is the sum of both.

-DF is the number of degrees of freedom associated with each component.

-SS is the sum of squares of variation.

-MS is the mean of the sum of squares (MS = SS / DF)

-F is the F ratio and gives an indication of the amount of variation in the data explained by the regression equation ($F = MS_{regr} / MS_{error}$). A large value for F indicates most of the variation in the data is explained by the regression equation.

-p is thesignificance limit that the regression equation is valid in describing the variation in the data. ($p < 0.05$ is considered significant)

In this first section the data is based on all four views of the foot (Left & Right, Toward & Away). The abbrieviation Met is used throughout for metatarsal head and the angles are as defined in figure 4.4. 'Frame' represents the frame number and is a representation of 'time'

<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<<>>

The regression equation is

Heel = 61.9 - 4.80 Frame - 0.687 ε + 0.385 ϕ

3456 cases used 918 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	61.8793	0.8007	77.28	0.000
Frame	-4.80283	0.09021	-53.24	0.000
ε	-0.68655	0.02872	-23.90	0.000
ϕ	0.38463	0.02716	14.16	0.000

s = 15.00　　R-sq = 56.2%　　R-sq(adj) = 56.1%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	995085	331695	1474.46	0.000
Error	3452	776561	225		
Total	3455	1771646			

****

The regression equation is

Met 1 = 0.555 + 0.816 Met 2 + 0.661 Hallux - 0.960 Toes 23

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	0.5553	0.1001	5.55	0.000
Met 2	0.81601	0.01359	60.06	0.000
Hallux	0.66068	0.01939	34.08	0.000
Toes 23	-0.95971	0.04938	-19.43	0.000

s = 4.037　　R-sq = 73.9%　　R-sq(adj) = 73.9%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	173876	57959	3555.91	0.000
Error	3758	61253	16		
Total	3761	235128			

****

The regression equation is
Met 2 = 0.0188 + 0.779 Met 3 + 0.194 Met 1 + 0.153 Hallux

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	0.01883	0.05825	0.32	0.747
Met 3	0.779138	0.009747	79.93	0.000
Met 1	0.194038	0.008107	23.94	0.000
Hallux	0.152546	0.008428	18.10	0.000

s = 2.257 R-sq = 87.6% R-sq(adj) = 87.6%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	135397	45132	8863.04	0.000
Error	3758	19137	5		
Total	3761	154534			

****

The regression equation is
Met 3 = 0.0110 + 0.575 Met 2 + 0.241 Met 45 + 0.149 Toes 23

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	0.01102	0.04959	0.22	0.824
Met 2	0.575497	0.007149	80.50	0.000
Met 45	0.241169	0.004875	49.47	0.000
Toes 23	0.14924	0.01682	8.87	0.000

s = 1.840 R-sq = 88.4% R-sq(adj) = 88.4%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	97090	32363	9560.89	0.000
Error	3758	12721	3		
Total	3761	109810			

****

The regression equation is
Met 45 = 2.62 + 1.47 Met 3 - 0.429 Hallux - 0.238 Met 2

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	2.6224	0.1188	22.08	0.000
Met 3	1.46987	0.03271	44.93	0.000
Hallux	-0.42880	0.01642	-26.11	0.000
Met 2	-0.23802	0.03108	-7.66	0.000

s = 4.615 R-sq = 62.3% R-sq(adj) = 62.3%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	132371	44124	2071.86	0.000
Error	3758	80033	21		
Total	3761	212404			

****

The regression equation is
Met 45 = 4.07 + 1.13 Met 3 - 0.414 Hallux + 0.130 ϕ

3456 cases used 918 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	4.0724	0.1453	28.03	0.000
Met 3	1.13300	0.01772	63.95	0.000
Hallux	-0.41395	0.01472	-28.12	0.000
ϕ	0.130351	0.007101	18.36	0.000

s = 4.411 R-sq = 65.4% R-sq(adj) = 65.4%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	126975	42325	2175.19	0.000
Error	3452	67169	19		
Total	3455	194144			

****

The regression equation is
Hallux = 0.648 + 1.56 Toes 23 + 0.351 Met 1 - 0.171 Met 45

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	0.64838	0.07459	8.69	0.000
Toes 23	1.56050	0.02279	68.48	0.000
Met 1	0.350648	0.007175	48.87	0.000
Met 45	-0.171327	0.006448	-26.57	0.000

s = 2.758 R-sq = 79.5% R-sq(adj) = 79.5%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	111047	37016	4867.11	0.000
Error	3758	28581	8		
Total	3761	139628			

****

The regression equation is
Toes 23 = - 0.0923 + 0.331 Hallux + 0.138 Met 3 - 0.0876 Met 1

3762 cases used 612 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	-0.09232	0.03245	-2.84	0.005
Hallux	0.331030	0.004695	70.50	0.000
Met 3	0.138364	0.005430	25.48	0.000
Met 1	-0.087557	0.004516	-19.39	0.000

s = 1.257 R-sq = 71.1% R-sq(adj) = 71.1%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	14643.7	4881.2	3088.35	0.000
Error	3758	5939.6	1.6		
Total	3761	20583.3			

****

The regression equation is
θ = - 26.1 + 3.75 Frame - 0.570 Met 3 + 0.0834 Heel

3456 cases used 918 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	-26.0663	0.3702	-70.41	0.000
Frame	3.75067	0.02660	141.03	0.000
Met 3	-0.57021	0.01983	-28.75	0.000
Heel	0.083365	0.006138	13.58	0.000

s = 5.743 R-sq = 90.2% R-sq(adj) = 90.2%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	1046847	348949	10581.74	0.000
Error	3452	113835	33		
Total	3455	1160682			

****

The regression equation is
δ = - 15.7 + 3.50 Frame - 1.00 Met 3 - 0.414 Met 1

3456 cases used 918 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	-15.7457	0.2801	-56.21	0.000
Frame	3.50346	0.02697	129.91	0.000
Met 3	-1.00034	0.03436	-29.11	0.000
Met 1	-0.41386	0.02369	-17.47	0.000

s = 7.565 R-sq = 83.1% R-sq(adj) = 83.1%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	974849	324950	5677.45	0.000
Error	3452	197576	57		
Total	3455	1172425			

****

The regression equation is
ankle plantarflexion = 22.0 - 3.54 Frame + 1.13 Met 1 + 0.492 Met 3

3456 cases used 918 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	22.0019	0.2166	101.58	0.000
Frame	-3.54282	0.02085	-169.90	0.000
Met 1	1.13248	0.01832	61.83	0.000
Met 3	0.49189	0.02657	18.51	0.000

s = 5.850 R-sq = 89.8% R-sq(adj) = 89.8%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	1039895	346632	10129.52	0.000
Error	3452	118127	34		
Total	3455	1158022			

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This next section to consider the rear leg angles is based on the two records taken from behind the leg (left and right, away).

The regression equation is
α = 5.63 + 0.0793 Met 45 + 0.0750 Met 2

1602 cases used 594 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	5.6330	0.1131	49.81	0.000
Met 45	0.07932	0.01129	7.02	0.000
Met 2	0.07498	0.01283	5.84	0.000

s = 2.762 R-sq = 10.2% R-sq(adj) = 10.1%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	2	1381.34	690.67	90.56	0.000
Error	1599	12195.43	7.63		
Total	1601	13576.77			

****

The regression equation is

$\gamma = 1.29 - 0.0626\ \phi - 0.157\ \text{Met 1} - 0.0299\ \text{Heel}$

1602 cases used 594 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	1.2936	0.3005	4.30	0.000
ϕ	-0.06259	0.01220	-5.13	0.000
Met 1	-0.15748	0.01841	-8.55	0.000
Heel	-0.029949	0.006416	-4.67	0.000

s = 4.982 R-sq = 9.2% R-sq(adj) = 9.0%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	4004.6	1334.9	53.78	0.000
Error	1598	39660.7	24.8		
Total	1601	43665.3			

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The only remaining angle is ε, which is best assessed when the subject is walking toward the camera so the following equation is based on left and right feet toward the camera.

The regression equation is

$\varepsilon = 11.5 - 0.997\ \text{Frame} - 0.0760\ \text{Heel} - 0.106\ \phi$

1854 cases used 324 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	11.5495	0.3387	34.10	0.000
Frame	-0.99729	0.02571	-38.79	0.000
Heel	-0.075967	0.005747	-13.22	0.000
ϕ	-0.105893	0.008160	-12.98	0.000

s = 4.002 R-sq = 49.9% R-sq(adj) = 49.9%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	29562.1	9854.0	615.34	0.000
Error	1850	29625.8	16.0		
Total	1853	59188.0			

****

The following is a scatter plot for the last regression equation for ε. For a perfect fit the points should fall on a straight line, slope = 1.

C17= 11.5 - Frame - 0.076*Heel - 0.106*ϕ

```
  8.0+                                                  *      *           *
     |                                       *      3   *42**     * *    *
 C17 |                                       *2 *3222939645*32332***  *   *
     |                                       **25 7+++++9+5*44****2** *  2
     |                              *       *3* +8+++++++++5725 *2   *
  0.0+                               *    *    *227++++++6684**
     |                                        **3258+++++++8*2
     |                               *    2 *36 6+8+++++++743*
     |                                   **34253++66++6+6*4*
     |                         * *  2 2 *23563344853532 323
 -8.0+                          * * 2*3234235224554****
     |                        *** 2****32* 44 *4 22              *
     |                       ** *  3**33223272 4 32*  *         *
     |                   *     ****2**324334233*3* 2** *
     |                    *    *    *    **   *
-16.0+
     |
     +---------+---------+---------+---------+---------+---------+
             -30       -20       -10         0        10        20
                                           ε (°)
```

Notes: A total number of 1854 cases of which (N) missing = 324
* = 1 point; 2-9=number of coincident points;
+ = > 9 points.

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The final aspect of the analysis was to consider the effect of normalising the area forces to body weight. Only the heel record had an improved regression equation and θ proved to also have a reasonable regression equation though not particularly different from the unnormalised equation.

The regression equation is
Heel = 94.0 - 4.46 Frame - 1.43 Met 45 - 0.864 Met 1

848 cases used 128 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	94.0366	0.8049	116.82	0.000
Frame	-4.46250	0.09564	-46.66	0.000
Met 45	-1.43494	0.04558	-31.48	0.000
Met 1	-0.86448	0.04696	-18.41	0.000

s = 9.263 R-sq = 91.1% R-sq(adj) = 91.1%

Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	743607	247869	2888.99	0.000
Error	844	72413	86		
Total	847	816021			

****

The regression equation is
θ = - 13.9 + 1.74 Frame + 0.455 δ + 0.133 Met 1

752 cases used 224 cases contain missing values

Predictor	Coef	Stdev	t-ratio	p
Constant	-13.9135	0.4465	-31.16	0.000
Frame	1.73650	0.06392	27.17	0.000
δ	0.45506	0.01827	24.91	0.000
Met 1	0.13330	0.02181	6.11	0.000

s = 4.170 R-sq = 92.3% R-sq(adj) = 92.3%

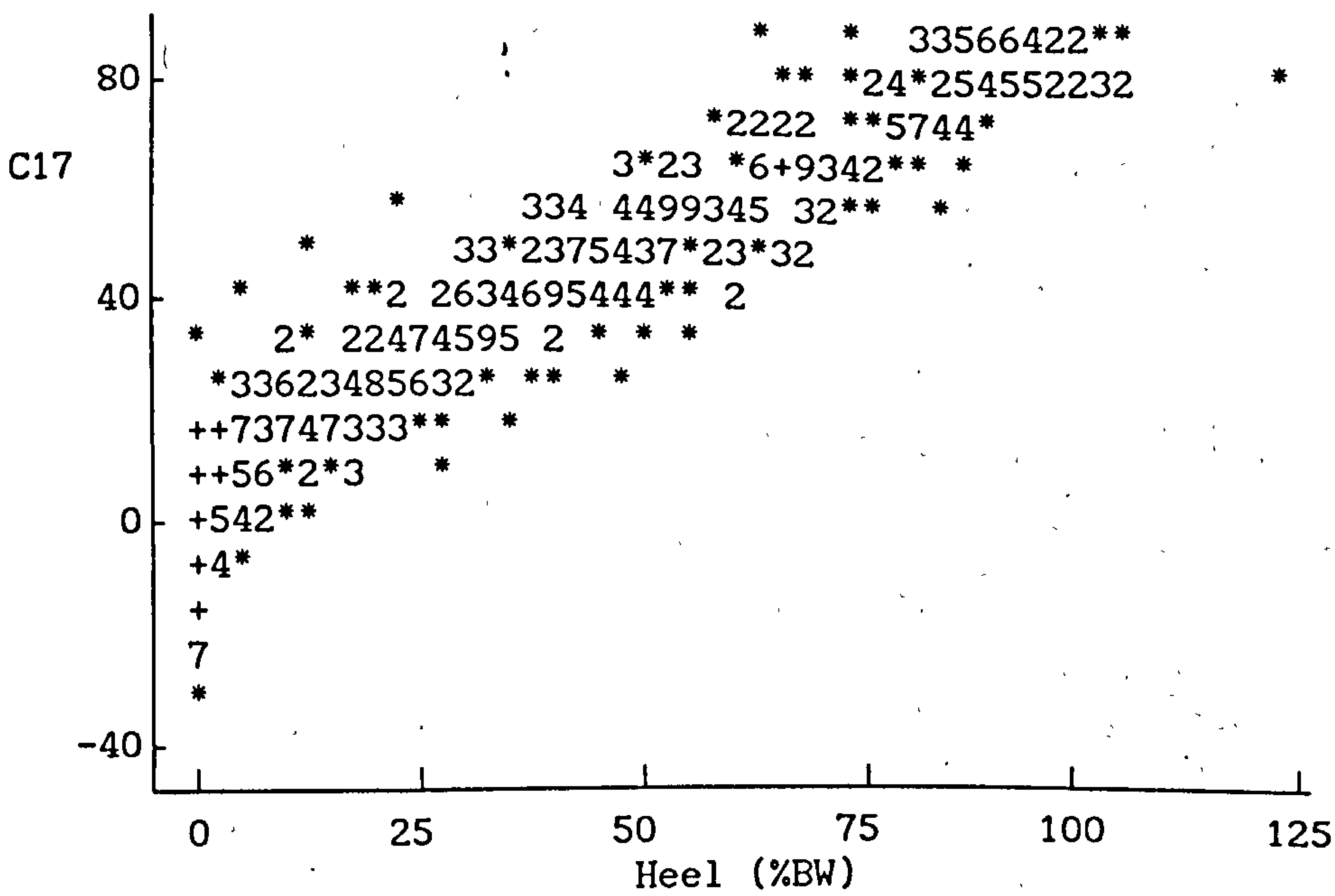
Analysis of Variance

SOURCE	DF	SS	MS	F	p
Regression	3	155832	51944	2986.99	0.000
Error	748	13008	17		
Total	751	168840			

****

The following scatterplot is based on the normalised equation for the heel record. As before for a perfect fit the points would lie along a line with slope = 1.

C17= 94-4.46*Frame-1.43*Met 45 -0.864*Met 1



Notes: Based on 848 cases of which 128 were incomplete
* = 1 point; 2-9=number of coincident points;
+ = > 9 points.

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APPENDIX 2 INSTRON CONTROL FILE AND LOAD CELL CONSTRUCTION

Table A2.1 A typical Instron control parameter file for the cadaveric foot tests.

Batch Information	
Batch descriptor 1	CYCLIC COMPRESSION TRIAL
Batch descriptor 2	FOOT
Batch descriptor 3	Load control; 3 speeds
Gauge length	1.0 mm
Parameter file	a:\FOOT2.CTP
Date read	31 July 1991
Time read	13:24:41

Sequence Setup	
Number of sequence repetitions	1
Number of blocks defined	3
Number of markers in sequence	3
Sequence order	
Marker 1	Block 1 : Cyclic
Marker 2	Block 2 : Cyclic
Marker 3	Block 3 : Cyclic

Test Control Data	
Block Number	1
Select block type	Cyclic
Control Mode	Displacement
Limit type	Load
First level	-10.0 N
Second level	-750.0 N
Crosshead speed	250.0 mm/min
End test/Break detect	No action
Number of transitions	8
Dwell time	0.0 seconds
Crosshead action at end of block	No action/Next block

Test Control Data	
Block Number	2
Select block type	Cyclic
Control Mode	Displacement
Limit type	Load
First level	-10.0 N
Second level	-500.0 N
Crosshead speed	500.0 mm/min
End test/Break detect	No action
Number of transitions	6
Dwell time	0.0 seconds
Crosshead action at end of block	No action/Next block

Test Control Data	
Block Number	3
Select block type	Cyclic
Control Mode	Displacement
Limit type	Load
First level	-20.0 N
Second level	-250.0 N
Crosshead speed	800.0 mm/min
End test/Break detect	No action
Number of transitions	5
Dwell time	0.0 seconds
Crosshead action at end of block	Crosshead return

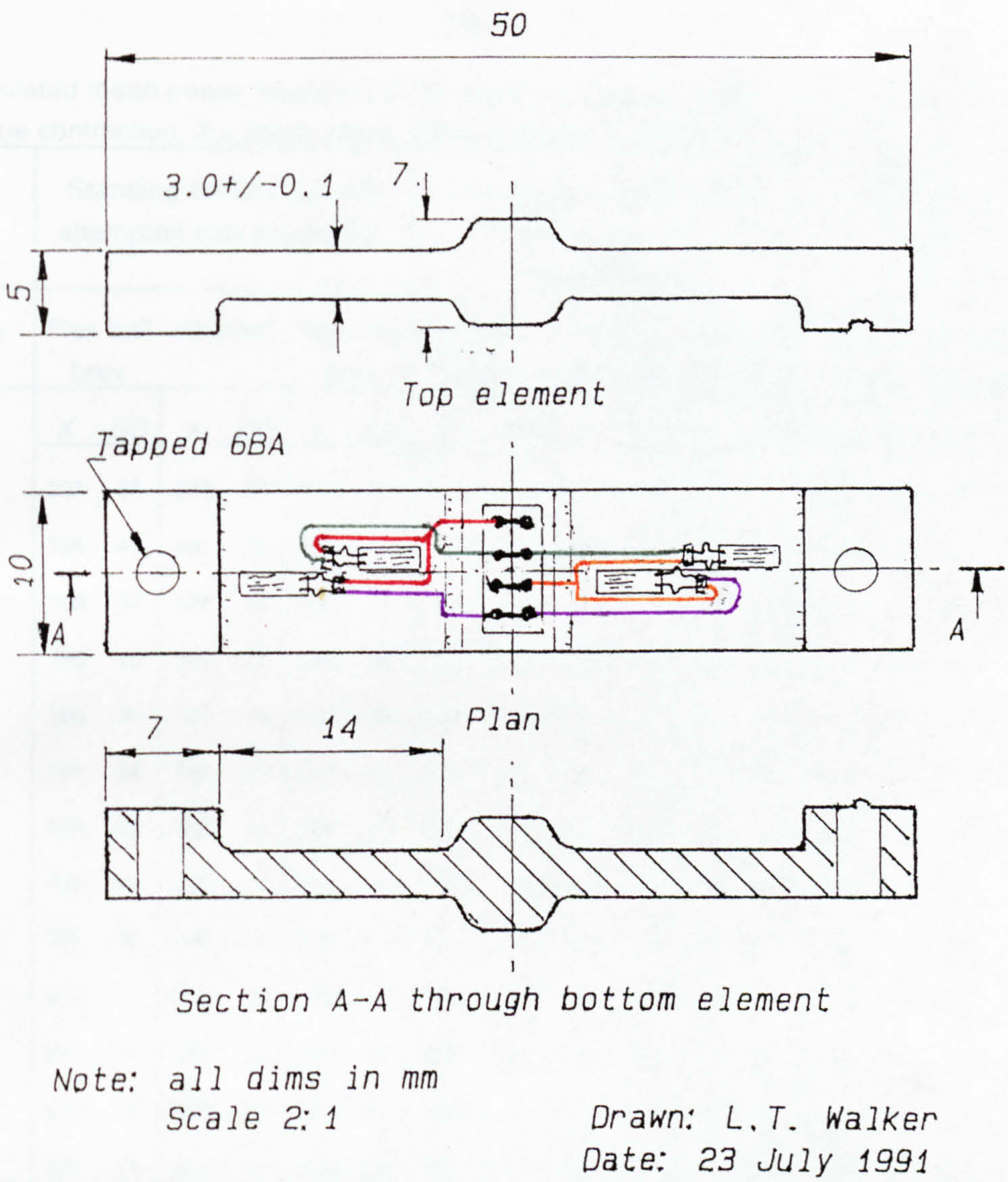

Figure A2.1 The technical drawings of the 250N load cell used in the cadaveric tests. Note the wiring of the gauges for a full bridge configuration.

APPENDIX 3 EMG MPF TABLES FOR THE MUSCULAR FATIGUE

Table A3.1

Calculated mean power frequency in the eight investigated muscles during sustained fatigue contraction. X = mean value. SD = 1 standard deviation

	Standing on forefoot with attempted max toe flexion:						Standing on forefoot with heel elevated 2-3 centimetres:						Standing on heels:			
Time [s]	Flex hall brev		Abd hall		Flex dig brev		Peroneus longus		Gastroc, lat head		Gastroc, m head		Ext dig brev		Tibialis anterior	
	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD
0	203	38	203	27	180	47	177	42	143	25	155	30	165	17	157	10
15	196	41	191	24	169	50	168	33	138	30	152	29	149	16	146	14
30	188	37	190	24	163	47	164	30	132	29	151	30	144	15	139	15
45	185	40	189	27	152	46	165	30	129	29	149	28	145	19	134	19
60	190	32	187	24	152	46	160	24	128	25	147	28	142	17	129	19
75	191	38	193	22	153	45	162	26	129	25	147	28	138	14	128	18
90	188	42	195	24	154	43	154	29	130	24	147	28	145	18	136	9
105	184	49	198	23	155	42	156	25	144	32	150	31	143	18	137	10
120	205	25	198	24	153	40	154	23	132	23	143	25	139	17	135	9
135	214	7	212	19	166	49	156	20	143	33	144	26	140	17	134	6
150	213	9	208	16	168	52	158	16	142	28	144	29	142	22	131	7
165	218	10	208	22	168	51	156	19	138	30	142	27	134	15	132	5
180	213	11	202	19	166	59	151	17	137	27	144	29	132	14	131	7

Table A3.2

Mean power frequency normalized to the first reading (=MPF-value at 0 s in table 3) in the eight investigated muscles during fatigue contraction. X = mean value, SD = 1 standard deviation

	Standing on forefoot with attempted max toe flexion:						Standing on forefoot with heel elevated 2-3 centimetres:						Standing on heels:			
Time [s]	Flex hall brev		Abd hall		Flex dig brev		Peroneus longus		Gastroc, lat head		Gastroc, m head		Ext dig brev		Tibialis anterior	
	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD	X	SD
0	100	0	100	0	100	0	100	0	100	0	100	0	100	0	100	0
15	97	6	95	8	93	5	96	3	96	5	98	2	91	4	93	7
30	93	6	94	9	90	7	93	4	92	5	97	2	88	4	89	7
45	91	9	94	7	85	12	94	4	90	6	96	2	88	9	86	10
60	87	10	93	7	88	13	91	5	89	5	94	2	89	14	83	11
75	88	12	95	9	89	12	94	11	90	3	95	2	86	12	82	10
90	86	13	96	10	90	9	88	6	91	3	95	3	94	14	87	3
105	84	17	97	8	90	12	89	6	101	21	92	17	93	15	87	3
120	92	5	97	6	87	15	90	16	93	6	93	4	90	14	86	3
135	92	2	96	4	95	8	85	6	101	24	93	3	91	14	86	5
150	92	1	94	4	96	6	86	9	100	23	92	2	93	17	84	4
165	94	1	94	8	96	7	85	8	97	21	91	3	87	11	84	6
180	92	1	92	7	94	4	83	11	97	22	92	3	88	10	84	6

APPENDIX 4 COMPUTER PROGRAMS FOR THE THREE EXPERIMENTAL CHAPTERS

(contained on Microfiche on back cover)

INDEX: