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A LONG LEG BRACE FOR COMPREHENSIVE FORCE MEASUREMENT

Design and Experimental Studies

By

FREDERICK G. LIPPERT III

MUNKSGAARD
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This work is dedicated to my wife, Nona and our three children: Nina, Felicia and Todd.

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PREFACE

During my residency training 1966—1970 at the University of Vermont, Burlington, Vermont, the extent to which engineering principles could be applied to orthopaedic surgery became increasingly evident. This relationship was further emphasized by contacts with Professors Victor Frankel and Albert Burstein. Because of my engineering-oriented background I felt a strong desire to become more acquainted with this field by undertaking a biomechanical research study.

Following discussions with Professor and Chairman Frank Hoaglund it was recommended that I seek a biomechanical research fellowship with Professor Carl Hirsch, Chairman of the Department of Orthopaedics, Karolinska Hospital, Stockholm, Sweden. Due to his renowned reputation for basic research in this field, it was anticipated that I would obtain the best indoctrination possible in the experimental approach and measuring techniques necessary for biomechanical studies.

Shortly afterwards the author was challenged one morning on rounds by visiting Professor Arthur Hodgson from Hong Kong concerning the efficacy of weight-bearing braces. A preliminary search of the literature revealed an astonishing lack of quantitative data to answer this question. It was then that the idea to design a method for quantitating forces between the leg and brace occured. Further discussions with Dr Richard McLay of the Mechanical Engineering Department of the University of Vermont confirmed the feasibility of such a project.

Later the same year Professor Hirsch visited the University of Vermont and the idea was presented to him. He immediately recognized the possibilities and extended an invitation for me to come to Sweden and pursue the work.

Largely through the interest and support of Drs Frank Hoaglund, John Frymoyer, Phil Davis and other members of the Department of Orthopaedic Surgery at the University of Vermont, financial arrangements were made to help defray my fellowship expences.

The major part of this work was done in collaboration with the Instrumentation Department of The Aeronautical Research Institute of Sweden. Their expertise in the use of instrumentation techniques was well utilized in solving the measuring problems

of this test brace. I am most grateful to Department Head Knut Fristedt for his help in evolving the final brace design after many discussions and interchange of ideas. Through him I have learned valuable concepts about various instrumentation techniques. Electronics engineer Lennart Magnusson maintained the recording equipment in an operational status and offered many practical suggestions during our close association in the gait laboratory.

Ragnar Ranveg, Head of The Prosthetics-Orthotics Workshop, Department of Orthopaedic Surgery, Karolinska Hospital, and his staff did a fine job in constructing the plaster positive of my leg, the plastic laminated thigh shells and the shoe components.

My appreciation is extended to the visiting biomechanical engineers, Leon Kazarian, Albert Schultz and also Bo Klasson, Head, Prosthetic Research Laboratory Norrbacka Institute, for their stimulating criticisms of the engineering aspects of this study; and to Drs John Kenwright and Manhar Patel for their clinical discussions.

I also wish to thank Lennart Jansson for the photography, Ingrid Ledström for help with administrative details, Vivian Kenwright for help with data reduction and Kerstin Sanfridson for typing assistance.

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It has been my privilege as a citizen of the United States to publish this work as a thesis in Sweden, which was made possible by academic grants.

Frederick G. Lippert III
Stockholm in November 1971

INTRODUCTION

Many affections of the lower extremity are treated and protected in orthotic appliances. In the United States the ratio of prosthetic to orthotic needs is 10/1 (Rehabilitation Engineering 1971). The field of orthotics has not kept abreast with the advances in prosthetics and technology. It has only been within the past 10 years that significant results from orthotic research efforts have begun to emerge.

Despite the number of orthotic devices prescribed, the manner of formulating an orthotic prescription and selecting the appropriate device is ill-defined (McCollough, 1970). To merely state the need for an appliance is insufficient. The physician involved must be able to establish the biomechanical aspects of his patient's disability and then use them to translate his medical objectives into specific orthotic objectives. Finally he is faced with the problem of filling the prescription from a bewildering array of manufactured appliances which are poorly classified as to their specifications and indications (Henderson & Lamoreaus, 1969; Murdoch, 1970). It is no wonder that controversy exists over the functions a given brace can perform and that different prescriptions are written for the same problem.

Appliances are assumed to perform functions for which they are unqualified while their detrimental effects remain either unknown or unheeded. The traditional ambulatory treatment of Legg-Perthe's disease in an ischial weight-bearing brace presupposes that loads on the affected femoral head will be diminished (Glimcher et al., 1970). In fact, the additional forces needed to accelerate or decelerate the braced limb by muscles spanning the diseased joint exceed body weight (Frankel and Burstein, 1970). The assumption that femoral fractures treated in weight-bearing cast braces can be effectively unloaded has been seriously challenged by Mooney et al. (1970). That the fractures heal may in fact be more related to the amount of loading which does occur because of the brace's ineffectiveness.

Braces which span joints limit their normal intersegmental motions in the transverse, frontal and sagittal planes (Viel, 1968). These motions act to conserve energy requirements by facilitating the interchange of potential and kinetic energy and reducing center of gravity excursions (Inman, 1968). A locked knee increases the total energy cost by up to 25% (Viel, 1968; Inman, 1968). Malalignment of brace joints produces excessive wear on the appliance and potentially detrimental effects on the patient (Lechlitner, 1966; von Werssowetz, 1962).

Research in the field of lower extremity orthotics today is hampered by the lack of quantitative clinical data, no standardized means of acquiring this clinical data and the inability to correlate clinical findings with a particular brace. Consequently long range response predictions to orthoses can only be speculated. (Lehmann et al., 1970.)

Brace research and development is mostly based on past experience and clinical observations which results in overdesign; excessive safety margins, increase in weight and bulk (Lehneis, 1969). In order to develop an optimal brace configuration clinical evaluation must be supplemented with biomechanical analysis. This combined approach provides a broader insight into the dynamic behavioral characteristics of the leg-brace system, and enables responses to be predicted for which no directly applicable empirical data is available. Design and modification can then be accomplished with more confidence that the desired clinical result will be obtained.

Approach to the problem

Biomechanical analysis of lower extremity braces can be approached in several ways. One is to instrument and study standard appliances. This method is easier but limited in the information obtained because of the susceptibility of such flexible systems to large cross-sensitivity errors. Another approach and the one taken by the author is to design a test brace, simulate selected prescriptions and study force-time histories. This comprehensive approach helps to define broader force envelopes, yield more realistic response predictions, provide a greater understanding of the biomechanical nature of the interacting systems, facilitate the correlation of physiological and psychological responses with mechanical input and enable a more scientific approach to brace design.

Purpose and objectives

The purpose of this project was to study the interactions between the brace and leg by the analysis of force-time histories recorded from selected ischial weight-bearing brace combinations.

The project was divided into two basic parts:

I. The design of a test brace, compatible instrumentation and reliable data acquisition and analysis techniques.

II. The application of the test equipment and methods to a general investigation on selected ischial weight-bearing brace combinations.

Specific information was sought about the following:

1. The magnitude of forces and moments to which specific brace structures and the leg are subjected.

2. The magnitude and distribution of loading which occurs on the brace with different thigh shell configurations.

3. The degree of weight relief (or skeletal unloading) experienced by the leg with different brace combinations as a function of:

   a) the anterior wall in the seat shell and maintainance of ischial seat contact

   b) patten bottom vs. shoe attachments

   c) fixed and hinged ankle joints

4. The significance of the joint hinge location on calf band force patterns
   a) heel hinge
   b) ankle hinge

5. The effect of foot alignment on the calf band force patterns
   a) free ankle; internal, neutral, external rotation
   b) fixed ankle; dorsiflexion, neutral, plantarflexion

6. Rotational and angulatory stability imparted to the extremity within the thigh shell.

7. Brace loading caused by muscle forces.

8. Alteration of the gait pattern expressed in terms of the contact duration of each foot position and the timing of force maximums.

SURVEY OF THE LITERATURE

Organized research efforts in the field of lower extremity orthotics did not appear until 1956 when the experience gained in prosthetics from working with veteran amputees was found also applicable to the allied field of orthotics (Wilson, 1963).

The main stream of lower extremity orthotic research has been oriented toward qualitative improvements in design, wider application to more acute problems such as fracture treatment and administrative efforts to systematize orthotic prescription, component classification and prefabrication (Staros, 1971). Several recent and detailed reviews cover the current trends in orthotics (Lehneis, 1969; Lehmann, et al., 1970; Report of Seventh Workshop Panel, 1970). A representative survey is presented to distinguish this type of research from the quantitatively oriented nature of biomechanical force-measurement studies (Braces, Splints and Assistive Devices, 1969; Cochran, 1971).

Anderson and Bray (1964) and Strohm et al. (1966) reported a functional long-leg brace designed to permit ambulation without knee locks. Grynbaum et al. (1969) reported an adjustable ischial weight-bearing brace which featured an inflatable anterior wall lining to maintain a snug fit and reduce the number of brace sizes required in stock. Russek (1958) attached a wood quadrilateral socket to the proximal end of an ischial weight-bearing brace to provide better support. Recognizing that hip joint forces developed during ambulation often exceed body weight, and that this was partly due to the abductor muscle group activity, Glimcher et al. (1970), are developing an ischial weight-bearing brace used for the treatment of Legg-Perthe's disease which requires the patient to balance on the end of a single medial upright without any leg fixation thus keeping the abductor muscle group inactive.

Plastic materials which are lighter than steel or aluminum and cosmetically more appealing are being utilized in below knee orthoses such as the spiral helix drop-foot brace (Fifth Workshop Panel on Lower Extremity Orthotics) and corrugated polypropylene versions of different below and above knee orthoses (Engen, 1970). Attempts have been made to more closely follow anatomic joint motion. Polycentric joint designs which simulate the multiplanar motion of the knee are currently under evaluation (Staros, 1971).

Several authors have reported encouraging results using cast and conventional braces for the early ambulatory treatment of fractures (Mooney and Harvey, 1970) and delayed union (Werner, 1969). Because of the array of brace components available from different manufacturers with no standard classification system, Staros (1971) has suggested a system of component classification based on whether the device prevents, limits, resists or permits joint motion.

McCollough et al. (1970) has described a systematic approach to orthotic prescription which should help to standardize terminology and facilitate a more accurate asscessment of the disability.

By contrast to these numerous, diversely oriented and qualitative approches to orthotic research, quantitative studies devoted to force measurements are limited (Prosthetics and Orthotics, 1967).

Glimcher (1950) reported stress analysis studies conducted on an instrumented long-leg brace. Strain gages mounted on the medial upright and horizontal bands were used to study the effects of strap tightening and ambulation. However, applied forces could not be directly measured because of cross-sensitivity difficulties.

Desai and Henderson (1961) designed a short-leg brace with a double axis ankle which permited both ankle and subtalar motion.

Staros and Peizer (1965) evaluated the unweighting effectivness of the Veterans Administration Prosthetics Center Below-Knee, Weight-Bearing Brace by simultaneously measuring an instrumented stirrup assembly and force plate output signals. Maximal unweighting occured at heel strike and was different on the right and left sides.

Hodge (1967) statically loaded an instrumented aluminum ischial weight-bearing brace which was mounted in a jig. Loads up to 250 lbs were applied and stresses from lateral deflections at the ankle and knee levels were computed. He considered this type of loading more conducive to structural failure than axial loads. From the stresses calculated aluminum was considered a suitable material for weight-bearing braces.

Magora et al. (1968) and Robin et al. (1968) instrumented lower leg drop-foot braces and computed stresses from recorded strain output signals using different gait patterns on a normal subject. Stress patterns indicated that posterior spring-type braces were more susceptible to fatigue failure.

A series of biomechanical studies have been conducted at the University of Washington, Seattle in conjunction with the Departments of Mechanical Engineering and Physical Medicine and Rehabilitation (Kirkpatrick et al., 1969; Lehmann et al., 1969 and 1970).

By using sophisticated data acquistion and analysis techniques, instrumented brace and floor reaction output signals are synchronized, digitized, interfaced with a computer and subsequently printed out in the form of fully integrated force-time histories by a Calcomp plotter. This measuring technique has facilitated a series of force-measurement studies on the performance characteristics of standard braces.

From these studies the unloading effectiveness of ischial weight-bearing braces has been seriously challenged. Optimal effectiveness was found with a quadrilateral thigh shell, fixed ankle, metal sole plate extending to the metatarsal heads, rocker bottom attachment and skilfull loading by the wearer.

The effect of different instrumented below knee braces on knee stability was evaluated by measuring calf band forces exerted against the leg. The forces cause knee flexion and extension moments which substantiates quantitatively what has been known clinically about the effect of heel and toe lever arms.

Mooney and Harvey (1970 investigated the weight-bearing properties of total contact cast braces used for recent femoral and tibial plateau fratures. Static comparisons between forces at the knee and floor showed unloading to be inversely proportional to body weight. Dynamic studies were not conducted.

INSTRUMENTATION AND TEST PROCEDURE

The Design of a Measuring System

In order to design and construct a functional test brace, compatible instrumentation and related equipment the Instrumentation Dept of the Aeronautical Research Institute of Sweden and the Prosthetic-Orthotic Workshop of the Dept of Orthopedic surgery, Karolinska Hospital, Stockholm, were consulted. Throughout the project is was necessary to maintain a close working relationship with the two disciplines.

The primary objective was to measure forces during ambulation exerted on the leg-brace interfaces. These forces arise from a combination of muscular activity, body weight, inertial and ground reactions. The method used in principle was to attach representative brace interfaces to the leg and study the resulting dynamic interactions.

In practice this was accomplished by fastening the interfaces to a measuring system which was mounted on a frame of sufficient stiffness to prevent mutual interactions between the interface segments. A typical ischial weight-bearing brace is attached to the leg by essentially four segments which have the following functions:

1. ischial seat platform, lateral and anterior walls: provide weight bearing and rotational control
2. corset: provides alignment and partial weight bearing
3. calf band: provides alignment but no weight bearing
4. foot assembly: provides alignment and foundation for the brace structure.

A study of individual forces acting on each contacting segment will yield a clearer picture of the overall performance at different levels.

The interactions between the leg and brace are represented by a combination of moments and forces. In order to have a statically determinate system is was necessary to convert the forces and moments into three orthogonal force components at three different points on the two upper brace rings corresponding to the seat and corset areas. This was achieved by using universal couplings between the leg-brace interface assemblies and the measuring system.

In the lower segment the calf transducers were aligned so as to record only the transverse plane components X and Z; and the ankle transducers

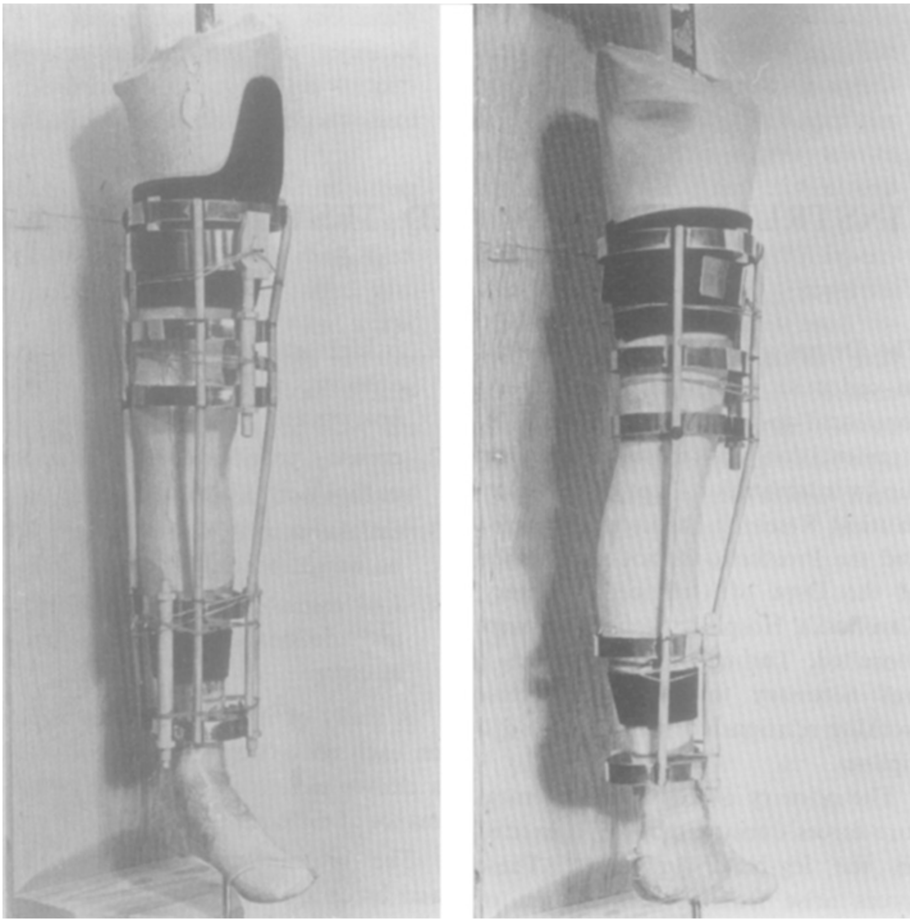


Figure 1. Positive plaster model and brace mockup under construction.

only the brace long axis component Y. Fig. 17 c shows the orthogonal force component system.

The force components applied to each transducer were predicted from estimates based on a known body weight of 61kp acting vertically downward on a brace system undergoing maximal inclinations from the vertical of approximately 14 degrees. At the slightly slower than normal walking speeds expected inertial forces were considered neglible.

Vertical floor reactions were correlated with the brace forces by utilizing the 5 meter electronic walkway (Rydell, 1966) in the gait laboratory of the Department of Orthopaedics, Karolinska Hospital. Since preliminary measurements, taken during ambulation, showed the maximal angle of the brace's long axis to be in the sagittal plane, 14 degrees, and in the frontal plane, 5 degrees, conversion of floor reaction forces to the brace axis as described by Lehmann et al. (1970) was not considered necessary. The cosine of small angles approaches 1, and the difference is too small to justify more complicated photographic methods of angle measurement.

Brace Construction

The design objective was to provide a structure of sufficient stiffness to prevent interactions from occuring between the leg-brace interface assemblies. Following the preliminary discussions, a positive plaster of paris model of the author's left leg was prepared (Fig. 1). This served as a basis on which to construct a wood mockup of the initial brace assembly, to locate the longitudinal structures and cross sections, to limit column length and determine the locations of transducer placement.

Early in the study it was concluded that the basic assembly must be divided into an upper and lower segment in order to facilitate entry and egress.

Construction material was:

Aluminum SM 6508 (weldable),
Svensk norm 515 4212-6
$E = 7000$ kp/mm^2
$\sigma_{0,2} = 25$ kp/mm^2

Each segment is constructed of circular ring sections and four longitudinal 10 mm solid tubular uprights (Fig. 2).

The individual segments were jigged, argon welded and mechanically fastened to each other. In the completed assembly applied loads were transferred to the longitudinal members which were stabilized by the circular cross sections. Cutouts were kept to a minimum to maximize the brace's stiffness and reduce the bending and torsional deflections during loading. The assembled configuration is in the form of a truss made of Duraluminum which provides a high degree of struc-

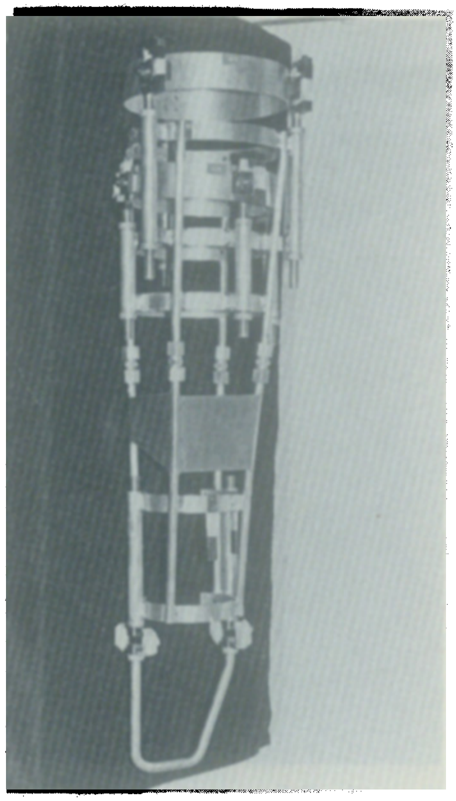


Figure 2. Complete brace without wiring. Couplings between the the upper and lower segments are seen above the stiffner plates.

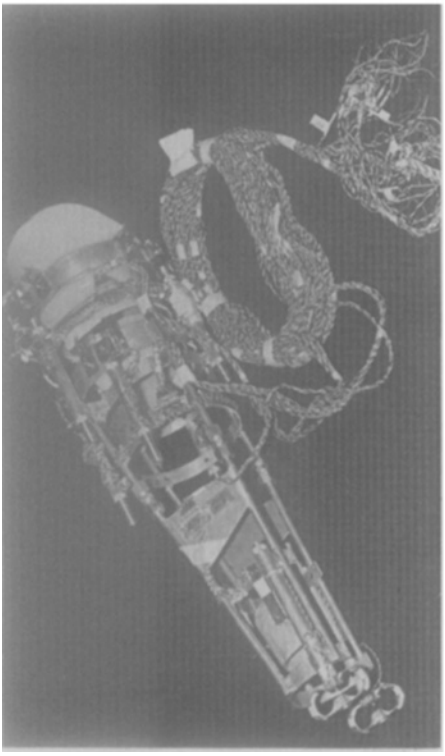


Figure 3. Fully assembled brace with shells, calfband and cabling.

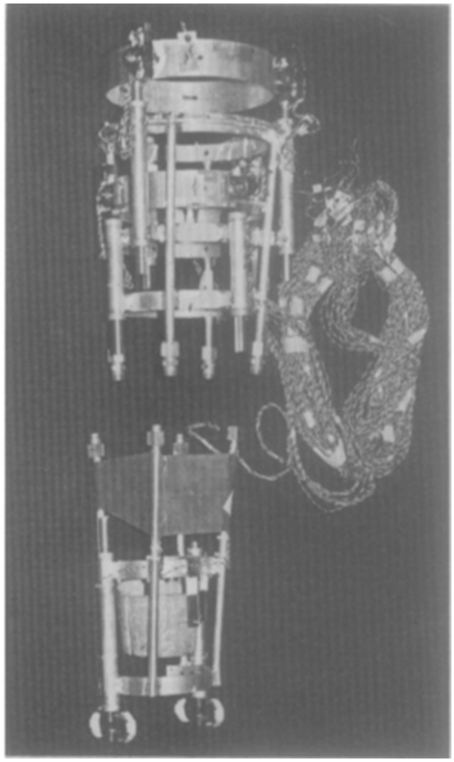


Figure 4. Upper and lower segments uncoupled.

tural rigidity with minimum weight (Fig. 3 & 4).

Onto the basic structural assembly were welded eight uniaxial and parallel keyed telescoping fixtures which provided a means of securely mounting, adjusting and locking the seat, corset and foot-ankle assemblies. The strain gage transducer bodies were mechanically fitted onto the telescoping fixtures and served as a mount to interface the sensing elements with the seat and corset and foot-ankle assemblies.

Transducers

The basic principle underlying the design of the transducers was that moments applied to a beam on which sensing elements are mounted produce greater strains per given force than direct compressive or tensile forces.

Six transducer bodies support the thigh segments and convert via universal couplings the forces and moments transmitted from the leg to the brace into six different reaction forces. Each transducer body has the ability to resolve its own force into three orthogonal components. The geometry of the force transducers allowed each force component to exert its own moment upon a strain gage bridge situated where the cross sectional area had been choosen to give a suitable strain for the predicted loads. The same bending beam principle was used to design the calf and ankle transducers. The ankle transducers were aligned to measure only the brace's long axis component (Y) and the calf transducers were aligned to measure only the transverse plane components (X & Z).

Transducer bodies were constructed from BOFORS RD-653 steel Svensk Norm 515 2225-5

$E = 21{,}000$ kp/mm^2

$\sigma_{0,2} = 70$ kp/mm^2

This material was chosen because of its high yield strength and the fact that steel is a good base material for strain gage bonding. Micro-Measurement foil gages were bonded with an epoxy cement and then heat treated. Details of construction and strain gage locations are shown for the upper segment in Fig. 5 a & b, and for the lower segment in Fig. 6 & 7.

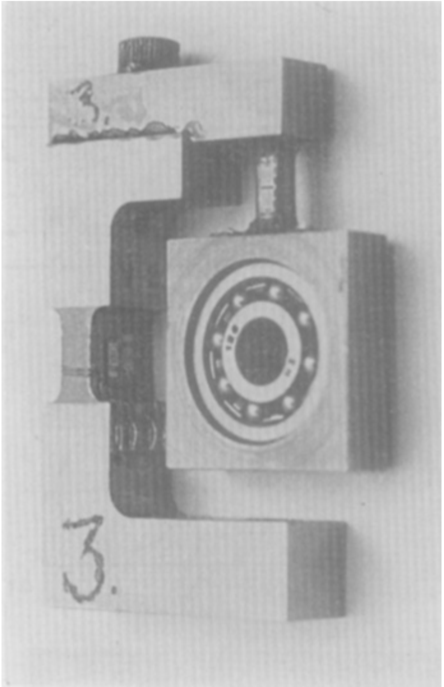
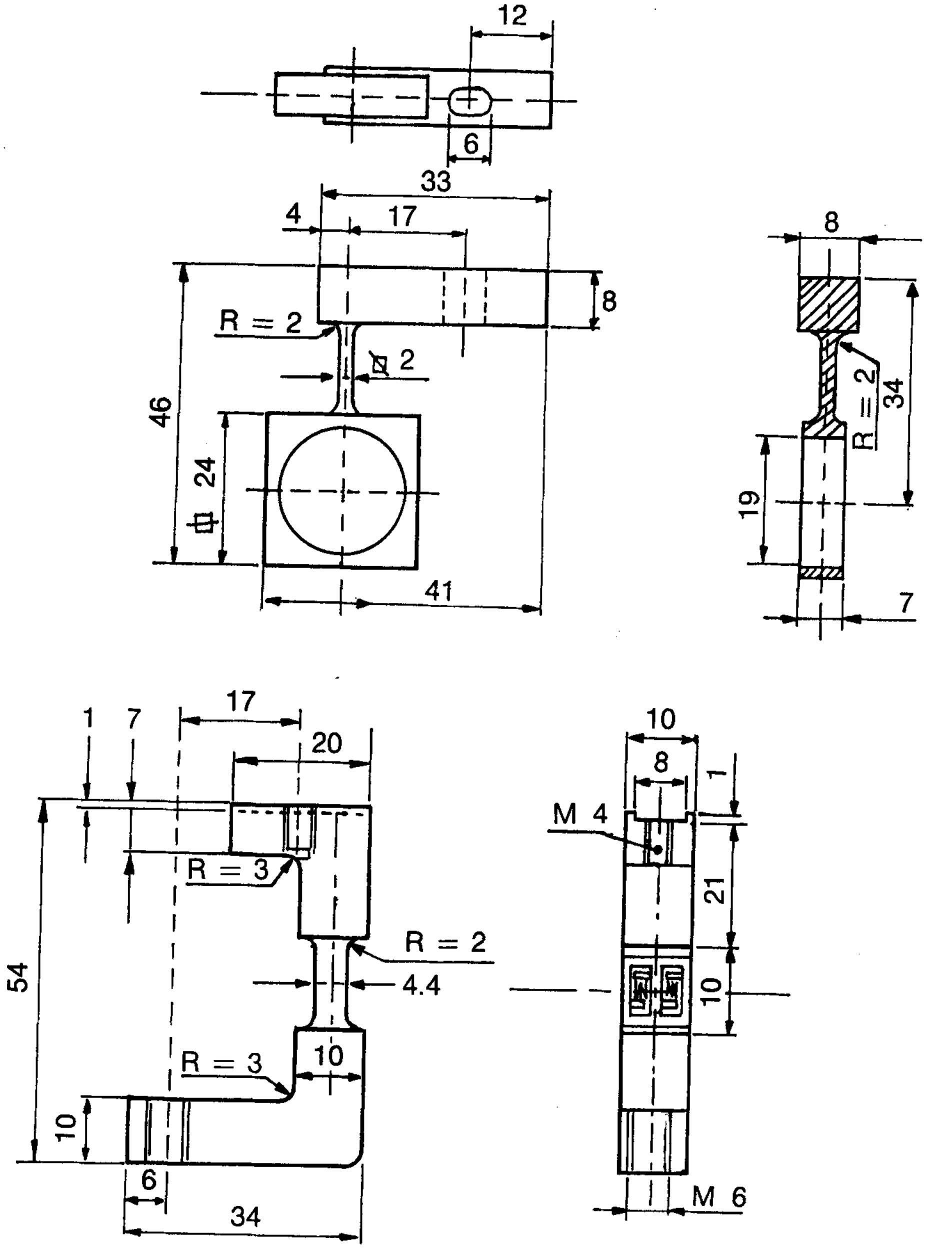


Figure 5 a. Typical upper segment transducer showing mounted foil strain gages and universal coupling.



BLUE PRINT OF TYPICAL
UPPER SEGMENT TRANSDUCER

Figure 5 b. Dimensions are in mm.

The calf transducer bodies consist of two beams 76 mm long, 12 mm wide and 3 mm thick bolted to the anterior and lateral aspect of the middle ring of the lower segment (Fig. 7 & 12). Connection between the calf band and the transducer bodies was established by an adjustable system of bolts and locknuts. Only the force components along the X and Z axis were measured at this level since the calf band has no weight-bearing function.

The ankle transducers were constructed using a ring design in order to measure only the force component along the Y axis. Because the transducer bodies also functioned as a coupling between the brace and foot assembly, it was desirable to maintain the distance separating them as small as possible. This was accomplished by using a flattened circle configuration (Fig. 6 & 7).

Interface Assemblies

Interface assemblies are composed of contour-fitted plastic shells, mounting rings and foot-ankle components.

The plastic contour shells act to transmit forces between the leg and brace and to assist in maintaining alignment within the brace.

The seat and corset were cut from a 3 mm thick basic thigh shell layed up on the plaster positive mold (Fig. 8). The shell was divided into two segments, an upper 1/3 seat shell and a lower 2/3 corset shell. The upper seat perimeter was anatomically contoured (Fig. 9). The anterior wall was

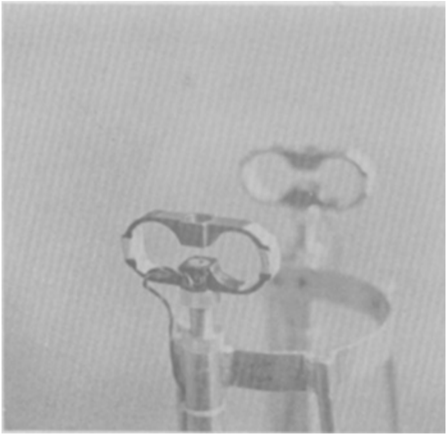


Figure 6. Ankle transducer without foot assembly.

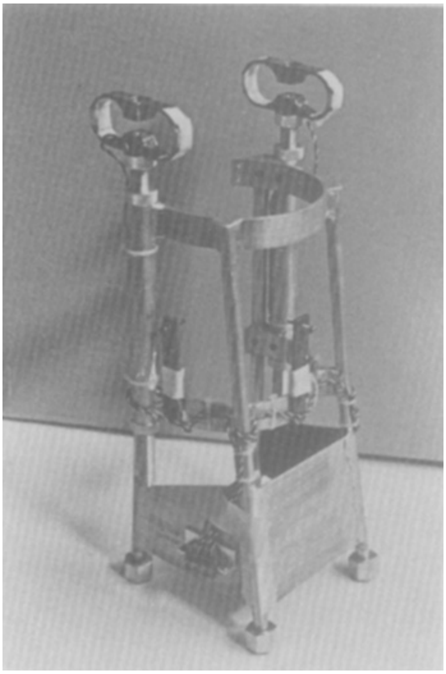


Figure 7. Lower segment upside down showing the ankle transducers, calf band beam transducers and couplings for attachment to the upper segment.

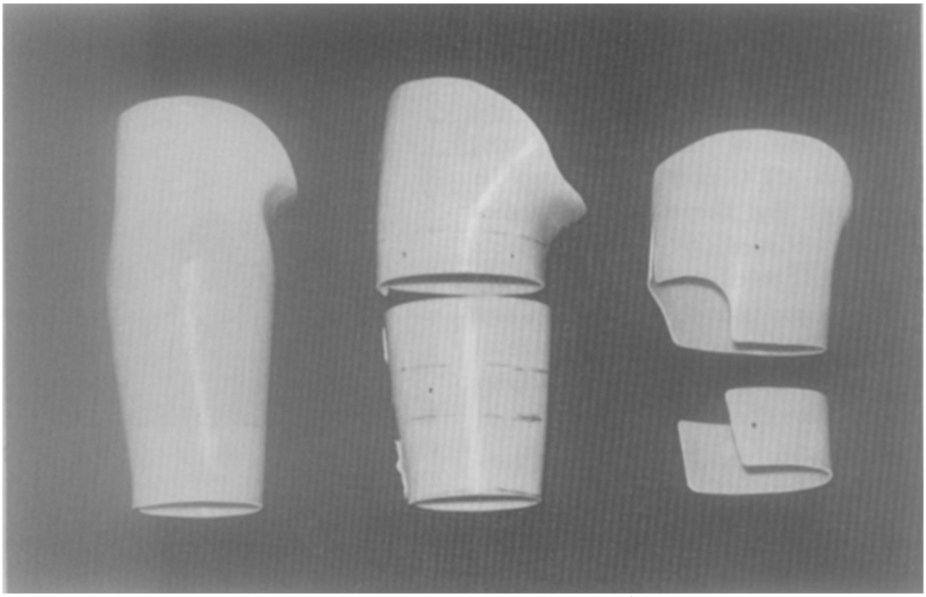


Figure 8. Basic thigh shell and segmented seat and corset shells.

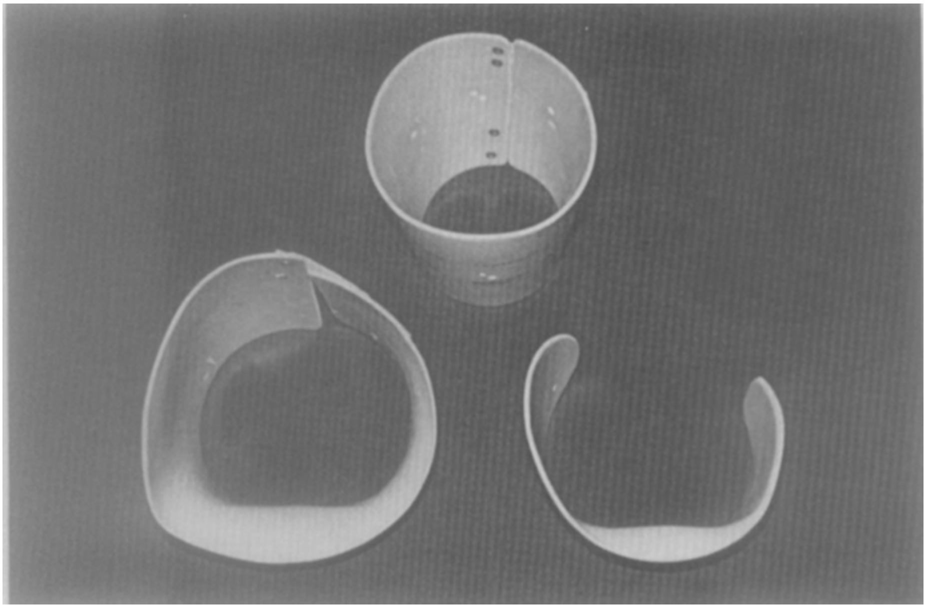


Figure 9. Above, closed corset shell. Left, closed seat shell. Right, open seat shell.

sectioned from one seat shell to access its importance in maintaining ischial contact. Corset shells were similarly divided. Entrance and egress was accomplished through anterior longitudinal seams held closed by Velcro fasteners. The shells were fastened to the seat and corset mounting rings by four steel angle iron brackets (Fig. 10 and 11).

The mounting rings serve as attachments for the body-brace interface components and the structure by which the whole assembly is connected to the transducers. They are circular 3 mm × 40 mm aluminum argon welded units. Fastened to the outer ring surface are mounting bolts in positions corresponding to the transducers on the main brace frame. Slots were cut to receive the shell mounting bracket bolts and permit adjustments in alignment.

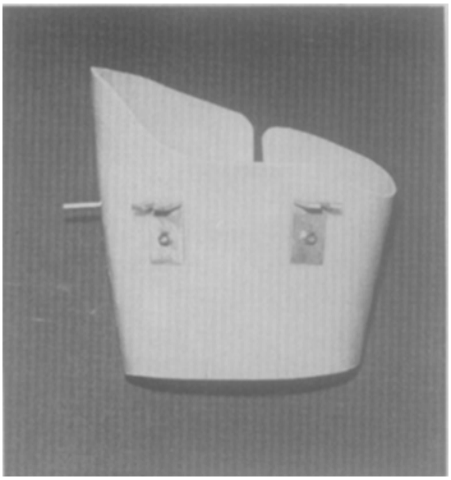


Figure 10. Seat shell with brackets ready to mount on the interface assembly ring.

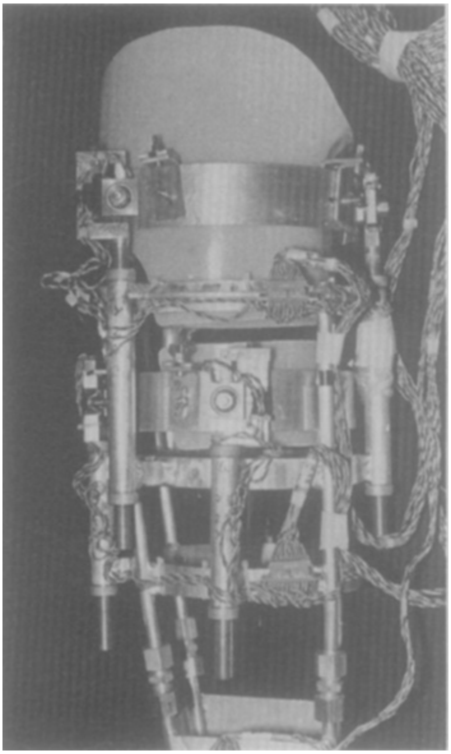


Figure 11. Seat shell mounted on the interface assembly ring.

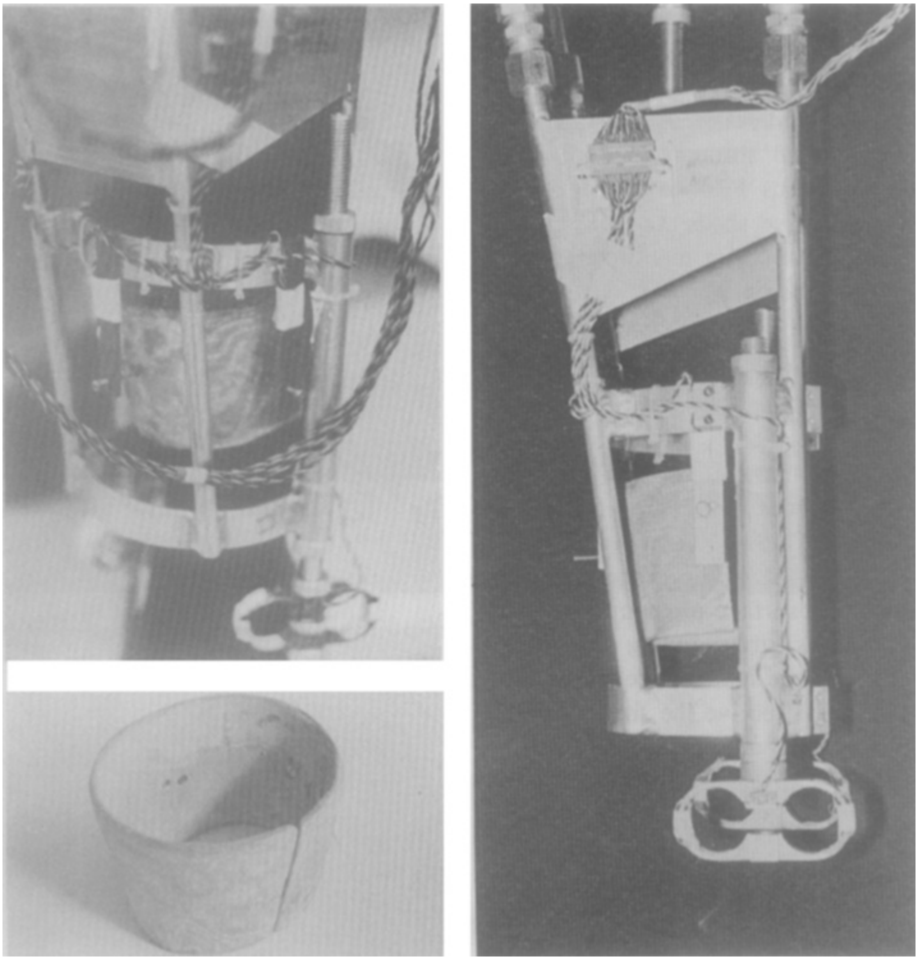


Figure 12. Details of calfband and mounting position.

The calf band consisted of a circumferential plastic laminated layup, 3 mm thick and 70 mm high with a posterior-medial entry cut, mechanically fastened to the beam transducers by two bolts (Fig. 12).

The shoe assembly was fabricated from a standard low cut oxford shoe with a leather sole and rubber heel. There were no orthopaedic modifications. A steel plate, 3 mm thick, was firmly riveted in place on the under

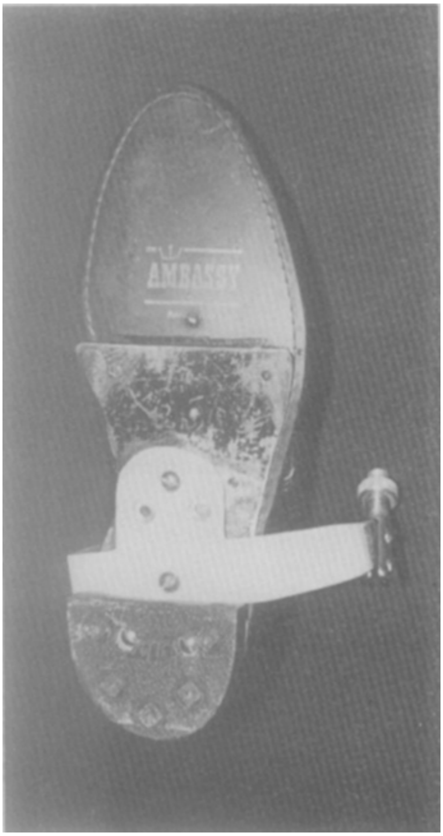


Figure 13. Shoe with steel sole plate, stirrup and rubber heel.

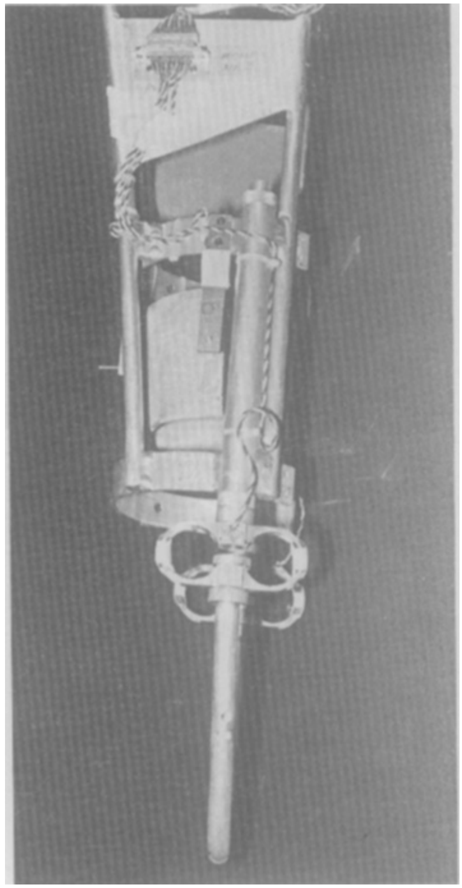


Figure 14. The caliper frame as a patten bottom.

surface of the sole and extended from the heel to the metatarsal heads to provide a rigid mounting platform for the stirrup and caliper attachments. To this plate were mechanically fastened the caliper or stirrup assemblies by means of an array of tapped holes which permitted different mounting positions. The rubber heel was fastened in place over the selected assembly (Fig. 13).

The caliper frame consisted of a 10 mm solid tubular aluminum U shaped structure which was threaded at each end for insertion into the ankle transducer bodies. Alone, it functioned as a patten bottom. In conjunction with a channel iron fastened to the sole plate it functioned as a caliper with heel hinge (Fig. 14 and 15).

By attaching stiffner struts between the sole plate and caliper frame arms the ankle could be fixed in selected positions of dorsiflexion neutral and plantarflexion (Fig. 15).

The stirrup assembly consisted of

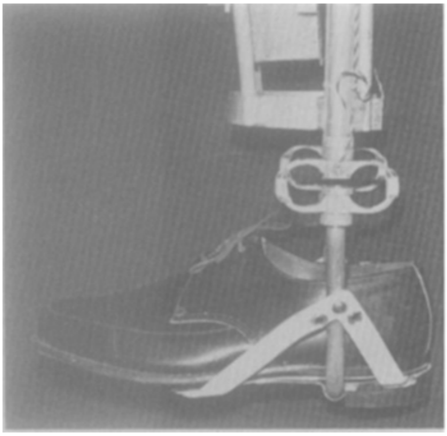


Figure 15. Shoe with caliper frame mounted in the channel iron and fixed by angle struts.

Figure 16. Shoe with stirrup assembly attached.

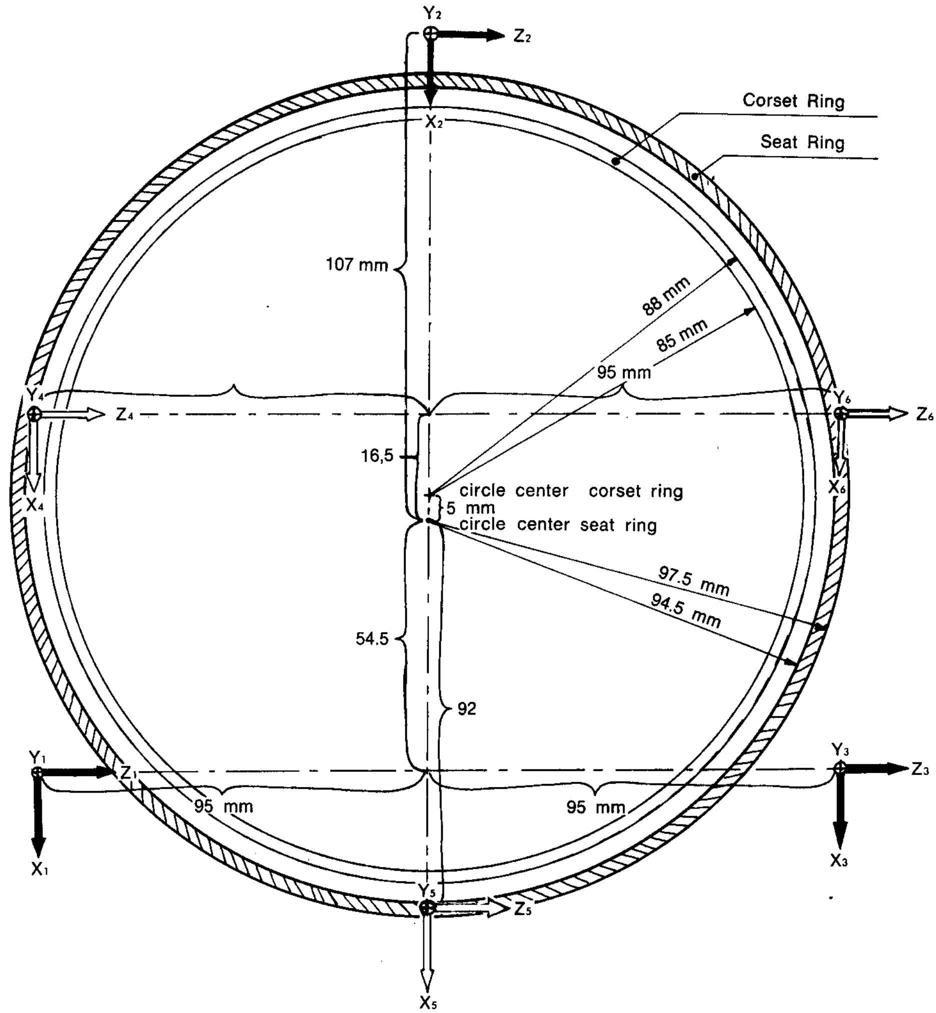
a stainless steel U shaped base plate and two threaded mounting bolts connected by a box-type hinge. The base plate was fastened to the sole plate and the mounting bolts fastened to the ankle transducer bodies (Fig. 13 and 16).

The assembled brace lacks contouring and from a survey of ischial weight-bearing appliances in this institution, is 2—3 kg heavier. Weight without cabling is 5 kg (upper segment 3 kg, lower segment 2 kg with shoe) which was considered acceptible if the inertial forces from rapid extremity movements during gait were avoided.

There were no restrictions on leg movements other than would normally be expected in a standard long-leg brace.

Alignment of Components

The leg with thigh shell applied was placed inside the brace frame, alignment adjusted and the sites of attachment to the interface assembly rings marked. The shell was then divided as prevously described. A margin of 5 mm was left between apposing edges to prevent mutual contact. Sufficient mounting bolts, spacers and fillers were used to insure good fixation and uniform force transmission to the rings. In the standing position a plumbline passed from the middle of the ischial seat to the heel's midline and the shoe sole was flat on the floor. Fig. 17 a shows the relationship of the upper and lower measuring ring centers to each other.



TOP VIEW RELATIVE ORIENTATION OF SEAT AND CORSET RING CENTERS

Figure 17 a.

For computation purposes the corset ring transducer distances to the X, Y and Z axis were measured using the upper ring center as a reference (Fig. 17 b).

An over head trolly carried the electrical cable, relieving the walker of its weight and obstructing influence.

Recording Equipment

26 transducer output signals from the brace and walkway were fed into 3 recording oscillographs. Details concerning the wiring of the transducers, bridge conditioning units, galvanometer specifications and the integrated systems diagram are shown in Fig. 18, 19, 20, 21. The whole system was synchronized by a foot switch circuit which marked the analogues directly at the onset and for the duration of each foot position, heel strike, foot flat, heel off and toe off (Fig. 22).

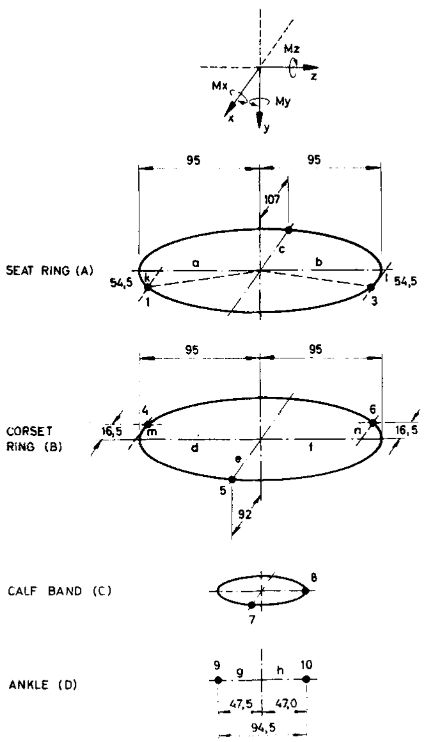


Figure 17 b. Alignment of the seat and corset rings. Dimensions are in mm.

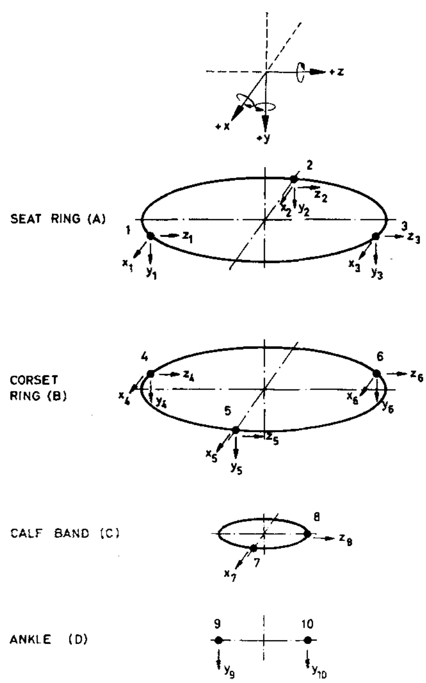


Figure 17 c. Brace measuring rings, transducer locations and orientation of force vectors.

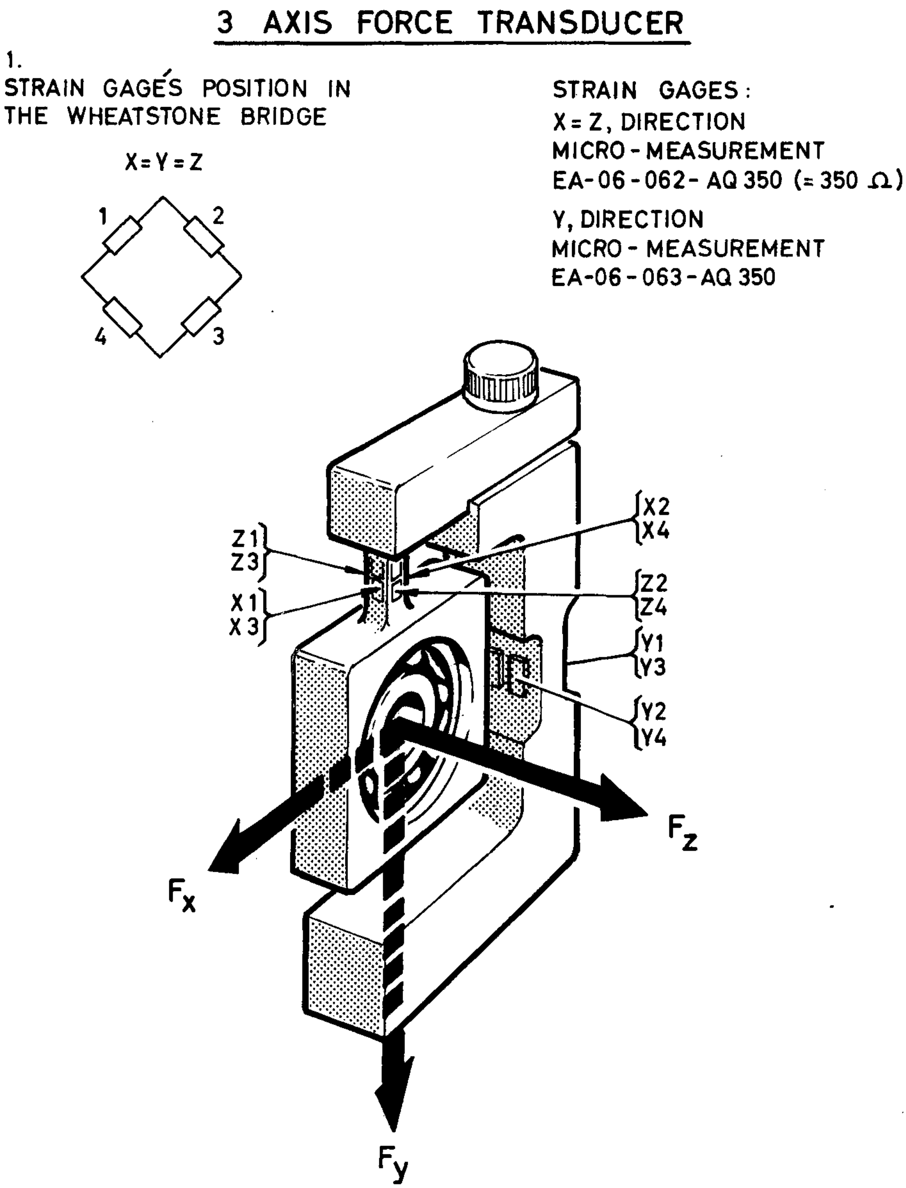

3 AXIS FORCE TRANSDUCER
1.
STRAIN GAGE'S POSITION IN
THE WHEATSTONE BRIDGE
X = Y = Z
1
2
4
3
STRAIN GAGES:
X = Z, DIRECTION
MICRO - MEASUREMENT
EA-06-062-AQ 350 (= 350 Ω)
Y, DIRECTION
MICRO - MEASUREMENT
EA-06-063-AQ 350
Z1
Z3
X1
X3
X2
X4
Z2
Z4
Y1
Y3
Y2
Y4
F_z
F_x
F_y

Figure 18.

Block diagram for one of three measured force directions (X, Y, Z).

2, 3, 4, 5 and 6.
Bridge condition units: Two 6 channels one 10 channels model FFA with stab. DC volt.

7.
Oscillographs, two. One ABEM Ultralette 10 channels. One ABEM Ultragraph 19 channels.
Galvanometers with natural frequency min 100 Hz.
Flat frequency interval 0—60 Hz with

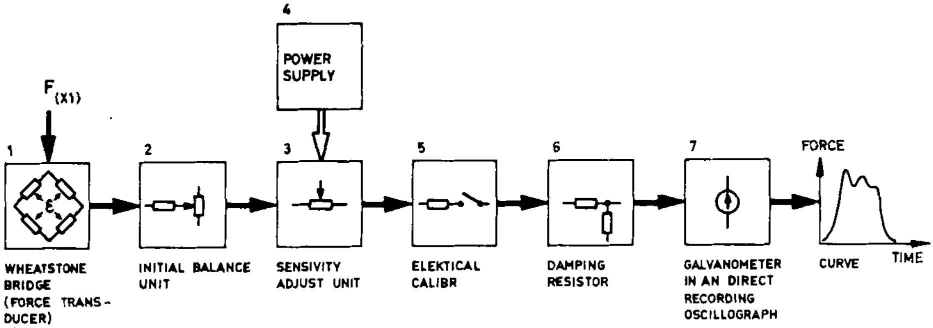
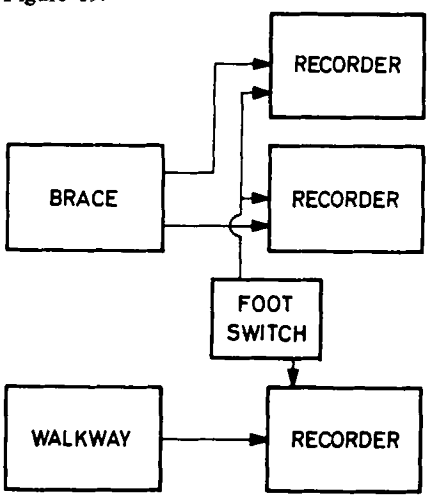


Figure 19.



BLOCKDIAGRAM OF BRACE, WALKWAY, FOOT SWITCH AND RECORDING SYSTEM

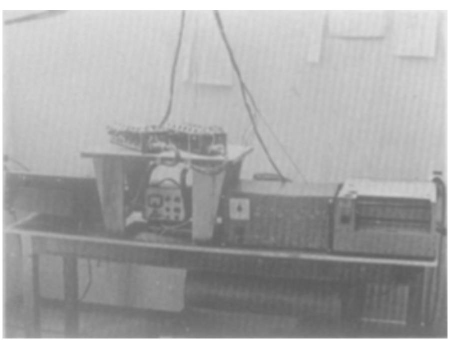
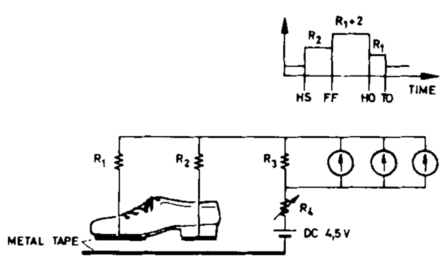


Figure 20.



FOOT SWITCH CIRCUIT SHOWING METAL TAPE, RESISTORS, DC POWER SUPPLY AND CONNECTIONS TO THE THREE RECORDER GALVANOMETERS

Figure 22.

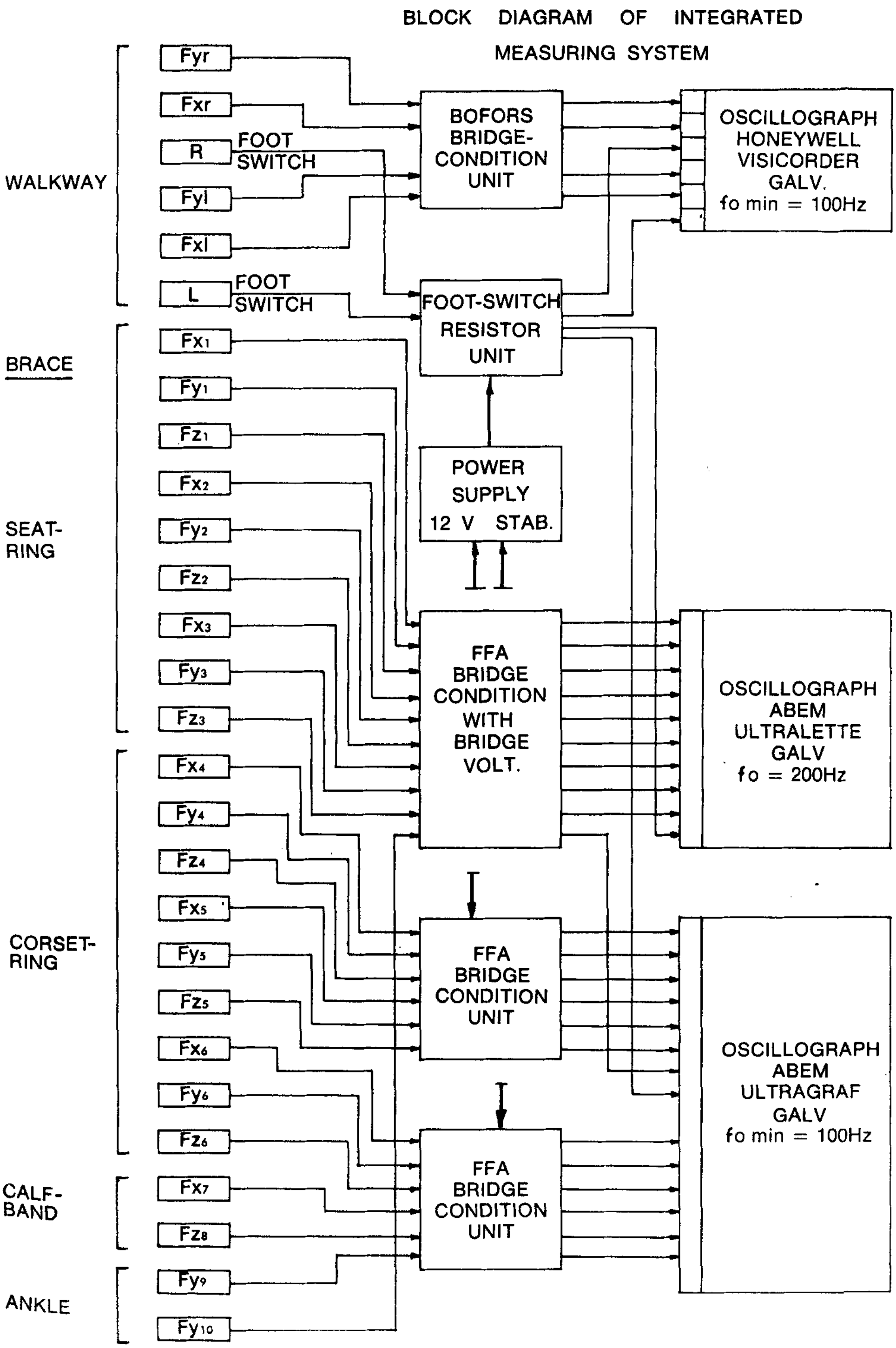

BLOCK DIAGRAM OF INTEGRATED MEASURING SYSTEM
WALKWAY
Fyr
Fxr
R
FOOT SWITCH
Fyl
Fxl
L
FOOT SWITCH
BRACE
Fx1
Fy1
Fz1
SEAT-RING
Fx2
Fy2
Fz2
Fx3
Fy3
Fz3
CORSET-RING
Fx4
Fy4
Fz4
Fx5
Fy5
Fz5
Fx6
Fy6
Fz6
CALF-BAND
Fx7
Fz8
ANKLE
Fy9
Fy10
BOFORS BRIDGE-CONDITION UNIT
FOOT-SWITCH RESISTOR UNIT
POWER SUPPLY 12 V STAB.
FFA BRIDGE CONDITION WITH BRIDGE VOLT.
FFA BRIDGE CONDITION UNIT
FFA BRIDGE CONDITION UNIT
OSCILLOGRAPH HONEYWELL VISICORDER GALV.
fo min = 100Hz
OSCILLOGRAPH ABEM ULTRALETTE GALV
fo = 200Hz
OSCILLOGRAPH ABEM ULTRAGRAF GALV
fo min = 100Hz

Figure 21.

Foot Switch Circuit

The foot switch circuit consisted of metal tape, attached to the walkway plates and heels and soles of each shoe, connected across a DC power supply, three different resistors and a galvanometer in each recorder. A step pattern of different height for each foot position appeared on the recording (Fig. 25).

Calibration

Calibration refers to the application of known loads along a specified axis of a measuring system and recording the output signals produced. From the established relationships, unknown forces applied to the system in the same direction can be quantitated. During calibration strains may be registered along axes other than the one loaded. This phenomenon is called cross sensitivity and is produced within the system by externally applied loads. The accuracy of the measurements and ability to define the applied forces requires that the cross sensitivity be small. Factors which increase cross sensitivity are related to improper transducer design, inexact mounting of the strain gages, and distortions of the mounting frame producing misorientation of the strain gages under loading.

Compared to standard braces which have been directly instrumented, the cross sensitivity error on this brace should be greatly reduced by the careful design, accurate alignment of strain gages and the rigid framework on which the transducer's parallel alignment is maintained.

Individual transducers were calibrated statically in a jig by applying loads up to 4 kp along the X and Z axis and up to 14 kp along the Y axis (Fig. 23 a). Fig. 23 b shows typical calibration curves for one transducer. Cross sensitivity was less than 5%.

The upper and lower segments were separately jigged and loaded to access the accuracy of the integrated thigh, calf and ankle measuring systems (Fig. 24). The error was found to be 5%. The two segments were not coupled and calibrated as a unit because it was assumed that each segment's stiffness properties would preclude a significant effect from being introduced as a result of the connection.

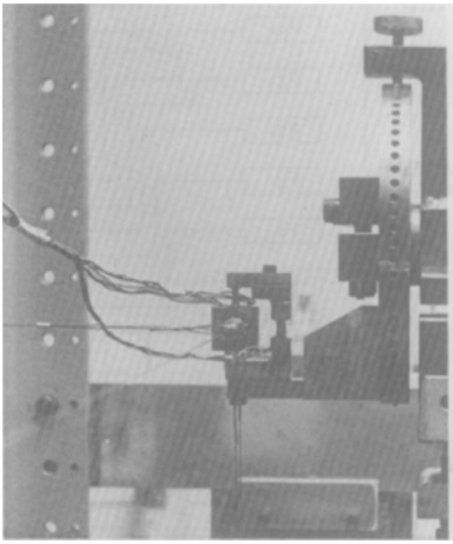
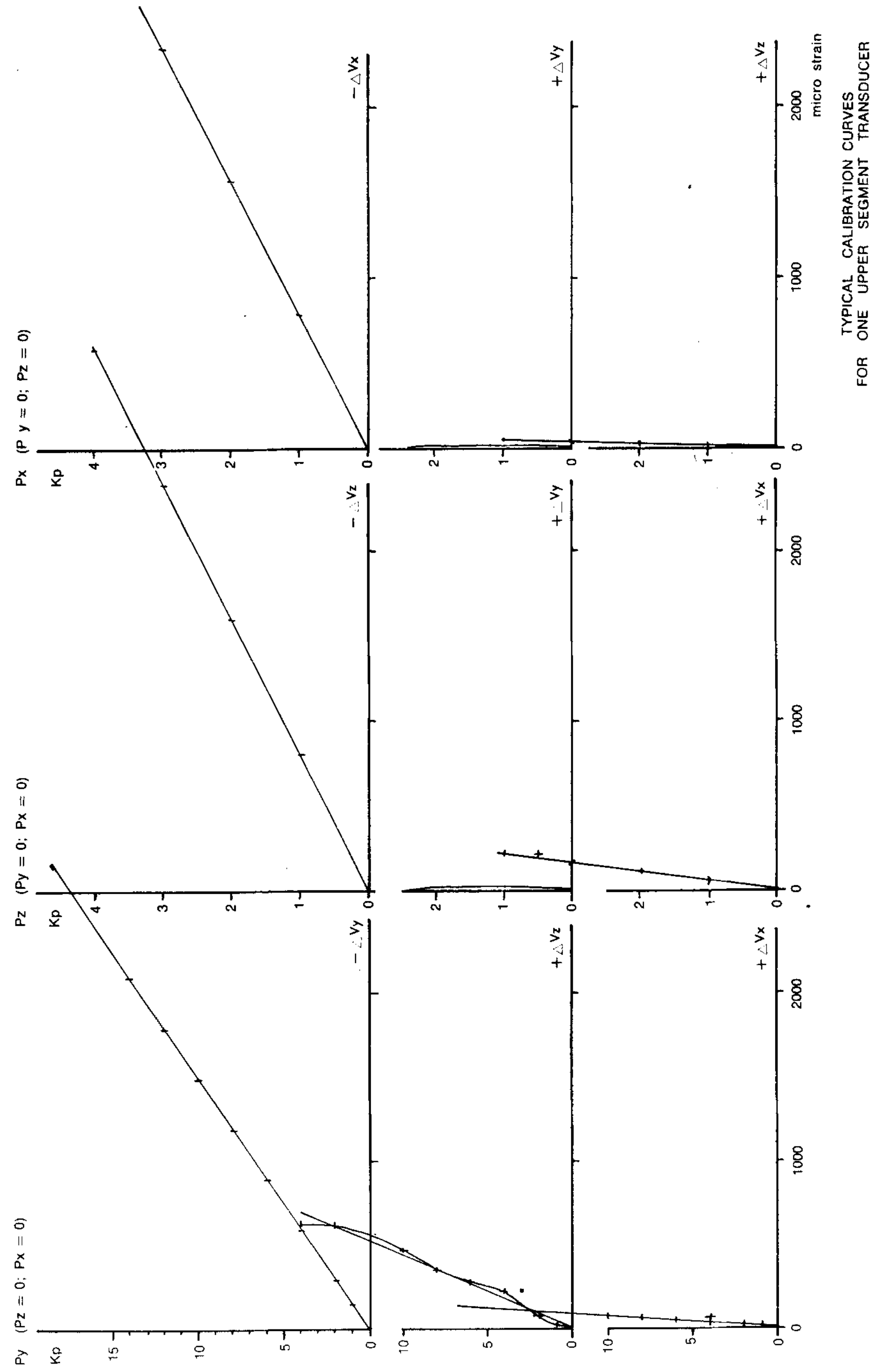


Figure 23 a. An upper segment force transducer, jigged and undergoing calibration.



TYPICAL CALIBRATION CURVES
FOR ONE UPPER SEGMENT TRANSDUCER
Py (Pz = 0; Px = 0)
Pz (Py = 0; Px = 0)
Px (P y = 0; Pz = 0)
Kp
micro strain
$-\Delta Vy$
$+\Delta Vz$
$+\Delta Vx$
$-\Delta Vz$
$+\Delta Vy$
$-\Delta Vx$

Figure 23 b.

Test-Walking Procedure

Prior to conducting test walks, the recording equipment was allowed to warm up and then all brace and walkway channels were calibrated with a calibration resistor. The brace was applied over a single layer of stockinette and after sufficent walking to settle well into the thigh shell segments it was established that good contact existed between the ischium and seat platform by finger palpation (Orthopaedic Appliance Atlas, 1952) and that the heel was not in contact with the inner sole of the shoe.

The test walk was started two steps before contact with the walkway and ended two to three steps beyond the end of the walkway to minimize the effects of acceleration and deceleration on the force-time histories.

Walking speed was gaged to a previously established time reference from many averaged walks in the brace. This walking speed was about half the normal unbraced walking speed of the same subject. No test walk was accepted unless within 0.5 seconds of the 10 second reference and unaccompanied by accidental lurches, scrapes or scuffs.

The walk was observed by an orthopaedic surgeon, orthotist, the instrument technician and the author himself by means of a mirror situated at the far end of the walkway.

Adequate rest periods were taken between walks to eliminate fatigue variabilities on the gait pattern.

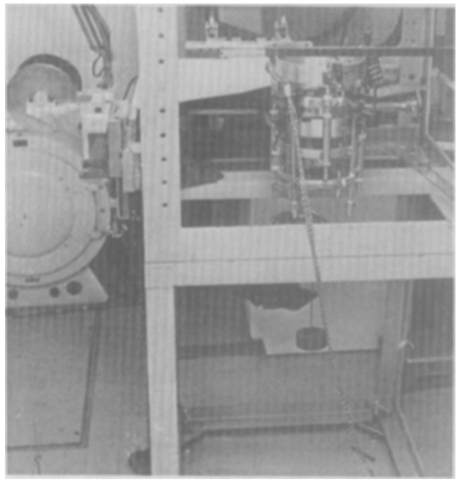


Figure 24. Assembled upper segment mounted in calibration jig. The seat ring is axially loaded to check the integrated system's response.

DATA REDUCTION AND ANALYSIS

Manual reduction

Oscillograph recordings were inspected for calibration marks and then base-lines for all channels were drawn. Three steps in the middle of the walk closest in time duration and force pattern were segmented into foot position and force maximum intervals (Fig. 25).

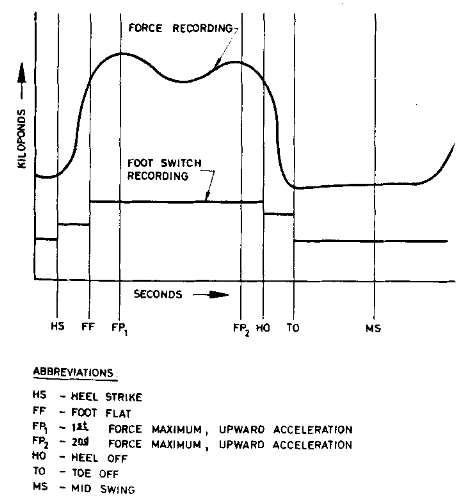
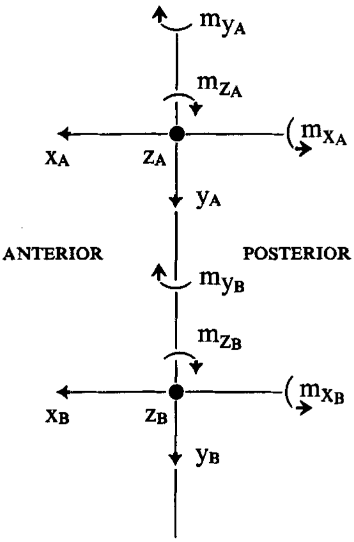


Figure 25. Segmentation of a typical force curve for digital conversion using the foot switch recording.

Superimposed tracings of the channels were then drawn to confirm their similarity and facilitate comparison with other walks. One step, closest to the average of the three, was converted into digital form at each interval. Using the numerical data, equations of equilibrium Fig. 26, the ring dimensions in Fig. 17 b and the sign conventions in Fig. 17 c, individual forces at the ring, calf band, and ankle units were summed and moments computed. Allowance was made for the slight offset between the corset ring center and seat ring center (Fig. 17 a).

Walkway recordings were similarly segmented and digitized. Foot positions were converted to percentages of total stance phase time. The relationship of each force maximum was determined with respect to a specific foot position. Different runs and each run, braced and unbraced sides, were compared.

The profile used for further analysis of each walk consisted of three analog force tracings, summed forces and moments and the foot-position force-peak relationships.



$$x_A = x_1 + x_2 + x_3$$
$$y_A = y_1 + y_2 + y_3$$
$$z_A = z_1 + z_2 + z_3$$
$$m_{x_A} = b\,(y_3 - y_1)$$
$$m_{y_A} = b\,(x_3 - x_1) - a\,(z_1 + z_3) + c \cdot z_2$$
$$m_{z_A} = c \cdot y_2 - k\,(y_1 + y_3)$$

$$x_B = x_4 + x_5 + x_6$$
$$y_B = y_4 + y_5 + y_6$$
$$z_B = z_4 + z_5 + z_6$$
$$m_{x_B} = f\,(y_6 - y_4)$$
$$m_{y_B} = f\,(x_6 - x_4) - e \cdot z_5 + m\,(z_4 + z_6)$$
$$m_{z_B} = m\,(y_4 + y_6) - e \cdot y_5$$

Figure 26. Equations of equilibrium and lateral view of the free body diagram for the upper segment.

Reliability of the equipment and experimental method

The overall error can be divided into instrumentation, manual reduction and experimental method errors. These are outlined below.

1. Instrumentation error
   a) cross-sentivity error ±3%
   b) calibration error ±5%
   c) recorder drift ±2%
   d) recorder errors within the flat frequency range used ±2%
2. Manual reduction error ±3%
3. Variability in the experimental method ±5%

Apart from the brace performance, force variations between walks can occur from a change in body weight, fatigue, removal and application times for the apparatus, normal step to step and day to day diurnal differences in gait pattern. Appoldt & Bennett (1968) studied the effect of these variables during investigations of AK stump-socket pressures and found that they could cause a scatter of ±10% about the mean values. This variability was minimized when test runs were conducted on the same day.

During this project the subject's weight remained constant at 61 kp, adequate rest periods were observed, the apparatus on and off time was never more than 15 minutes, most of the test runs were conducted on the same day and the remainder within the next 9 days. Because of these precautions the scatter effect is estimated to be less than 5%. It is unlikely that all errors will combine simultaneously in the same direction to produce an error as large as 20%. A realistic estimate of the measuring error is considered to be ±10% (Hetenyi, 1963).

RESULTS

General interpretation

Data representing different brace variables listed in Table 1 was evaluated.

The walks can be divided into essentially three similar groups:

1. closed seat shell-closed corset
2. open seat shell-closed corset
3. closed seat shell-open corset

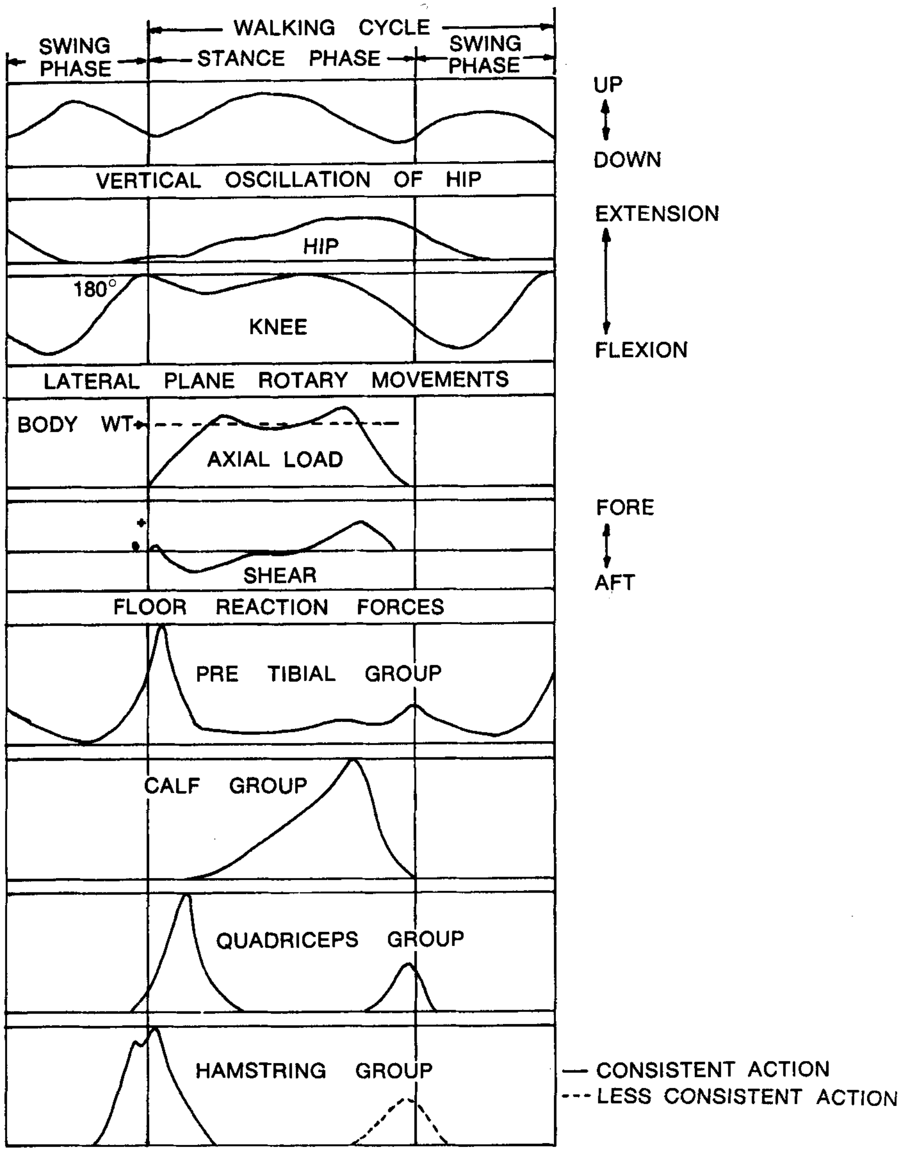
The first two groups were tested on the same day and the third group 9 days later. There were greater differences in the general results between the third and first two groups which is attributed to diurnal variables previously discussed and the greater dynamic alignment changes permitted by the reduced corset area.

Since one of the major purposes of this project was to elicit broad force envelopes it was decided to pool all runs together and compute force means and ranges. Where interesting differences were noted further amplification is presented. Foot-assembly evaluations are compared on an individual basis.

Figures 17 b and 17 c show the in-

Table 1. Walks conducted in different brace combinations.

Walk number	*Seat shell*	*Corset shell*	*Foot assembly*
1.	closed	closed	patten bottom
2.	closed	closed	shoe, fixed ankle, neutral
3.	closed	closed	shoe, heel hinge
4.	closed	closed	shoe, free ankle, internal rotation
5.	closed	closed	shoe, free ankle neutral
6.	closed	closed	shoe, free ankle, external rotation
7.	open	closed	shoe, free ankle, external rotation
8.	open	closed	shoe, free ankle, neutral
9.	open	closed	shoe, free ankle, internal rotation
10.	open	closed	shoe, heel hinge
11.	open	closed	shoe, fixed ankle, neutral
12.	open	closed	patten bottom
13.	closed	none	shoe, free ankle, external rotation
14.	closed	open	shoe, fixed ankle, neutral
15.	closed	open	shoe, free ankle, external rotation
16.	closed	open	shoe, fixed ankle, plantar flexion
17.	closed	open	shoe, fixed ankle dorsiflexion
18.	closed	open	shoe, fixed ankle, neutral
19.	closed	open	shoe, free ankle, external rotation



IDEALIZED MUSCLE ACTION CURVES (95 STEPS/MIN)
CORRELATION OF PHASIC MUSCLE ACTIVITY WITH JOINT DISPLACEMENTS AND FLOOR REACTION FORCES

FROM UNIVERSITY OF CALIFORNIA PROSTHETIC DEVICES RESEARCH 1947

Figure 27.

dividual transducer locations, mounting ring positions, important dimensions and the sign conventions used. All forces are expressed in kp and moments in kp-mm. Unless otherwise stated forces are interpreted as acting on the brace. Reactions on the leg are equal and opposite in direction.

General concepts used in the analysis of results are illustrated in the consideration of axial loading during stance and swing phase. During stance phase axial forces registered at the ankle transducers indicate the combined effect of body weight, brace weight above the ankle, muscular activity within the brace and ground reaction. In order to separate body-weight forces from muscle-activity forces it is necessary to look at that part of the cycle where the other's contribution is minimal. The brace weight is assumed constant. From EMG studies reported (Eberhart et al., 1947) muscle activity is greatest during late stance-early swing and late swing-early stance corresponding to the periods of acceleration, deceleration and weight-acceptance stability (Fig. 27).

During the middle part of stance phase the body is riding foward on the brace and weight-bearing loads are maximal.

Toe clearance is normally accomplished by hip and knee flexion and ankle dorsiflexion. In the test brace with a locked knee, toe clearance depended on circumduction, hiking of the pelvis and ankle dorsiflexion, when the ankle was free. During swing phase the ankle transducer forces reflect the brace weight not suspended from the upper thigh, shoe assembly wt. 0.8 kg, upward accelerating forces from the foot if distal migration of the brace has occured at toe off and forces caused by any attempted hip, knee or foot motion against the confines of the apparatus.

Muscle forces are expected to exert their greatest effect on the system during late stance-early swing and late swing-early stance, while maximal weight bearing forces occur during mid stance when the upward accelerations are taking place (FP1, FP2). This means that the best part of stance phase to evaluate a test brace's weight-relieving effectiveness, irrespective of muscle loading, is between foot flat and heel off.

The axial loading patterns recorded throughout the gait cycle are a result of these interactions. By similar analogy somewhat lesser forces appear along the X and Z axis.

Abbreviations used in the text

HS = heel strike
FF = foot flat
FP1 = 1st vertical floor maximum
FP2 = 2nd vertical floor maximum
HO = heel off
TO = toe off
MS = mid swing

Gait Pattern

The subject's, i.e. the author's, gait was evaluated by an orthopaedic surgeon, the orthotist in charge of the brace workshop and the author who could observe himself in a mirror at

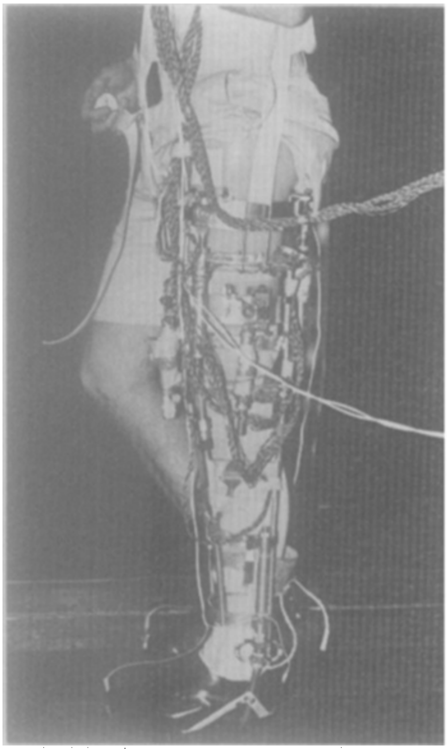


Figure 28. *Lateral view.*
Author in the brace during mid stance phase. Note foot-switch circuit wiring attached to the shoes.

Figure 29. *Front view.*
Just before heel strike. Note pelvic hiking on the normal side.

the end of the walkway. The typical gait pattern displayed slight vaulting during stance phase on the braced side. Toe clearance during swing phase was accomplished by ankle dorsiflexion and circumduction on the braced side and pelvic hiking on the opposite side. Viewed from in front, the stance phase was abducted 0—5 deg., and from the side the long axis of the brace was inclined posteriorly from the vertical 14 deg. at heel strike decreasing to 7 deg. at toe off (Fig. 28, 29, 30).

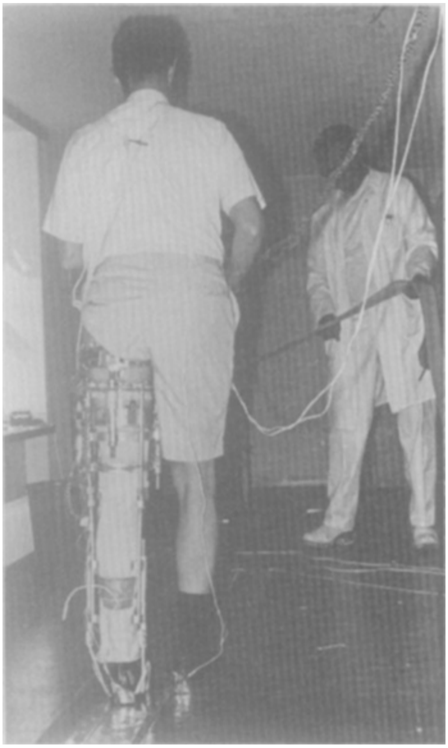


Figure 30. *Posterior view.*
Heel strike on the braced side. Metal tape is on the walkway plates and shoe heels completing the foot-switch circuit.

MEAN LOADING TRENDS

Axial loading (Y) Table 2, Fig. 31

Axial forces were exerted throughout the gait cycle, including swing phase, except with a patten bottom frame. This is explained by the fact that brace combinations with a shoe were loaded by forces transmitted from foot movements within the shoe.

In the patten bottom brace forces could not be transmitted to the brace because the foot was unattached to the patten frame.

Anterior-posterior loading (X)

During the period heel-strike-to-first-force maximum the largest forces were recorded in the seat shell and were directed posteriorly. The thigh shell resultant is posteriorly directed and probably reflects the acceleration of the upper thigh against the posterior wall (Appoldt & Bennett, 1968). At the second force maximum the resultant became anterior and then posterior at heel off and mid swing. The AP thigh shell and calf band force distribution during swing phase correlates with the angular moment necessary to accelerate the leg in the line of progression.

Table 2. Range of forces (kp) during ambulation. Body weight is 61 kp.

	UP-DOWN		MEDIAL-LATERAL		POSTERIOR-ANTERIOR	
seat	— 1.0	+39.0	—2.5	+10.5	—20.0	+ 4.2
corset	—10.5	+15.5	—6.7	+ 0.8	— 6.7	+ 9.0
calf	force not measured		—6.0	+10.0	— 7.0	+10.0
ankle	0.0	+70	forces not measured			
floor	0.0	+77	forces not measured			

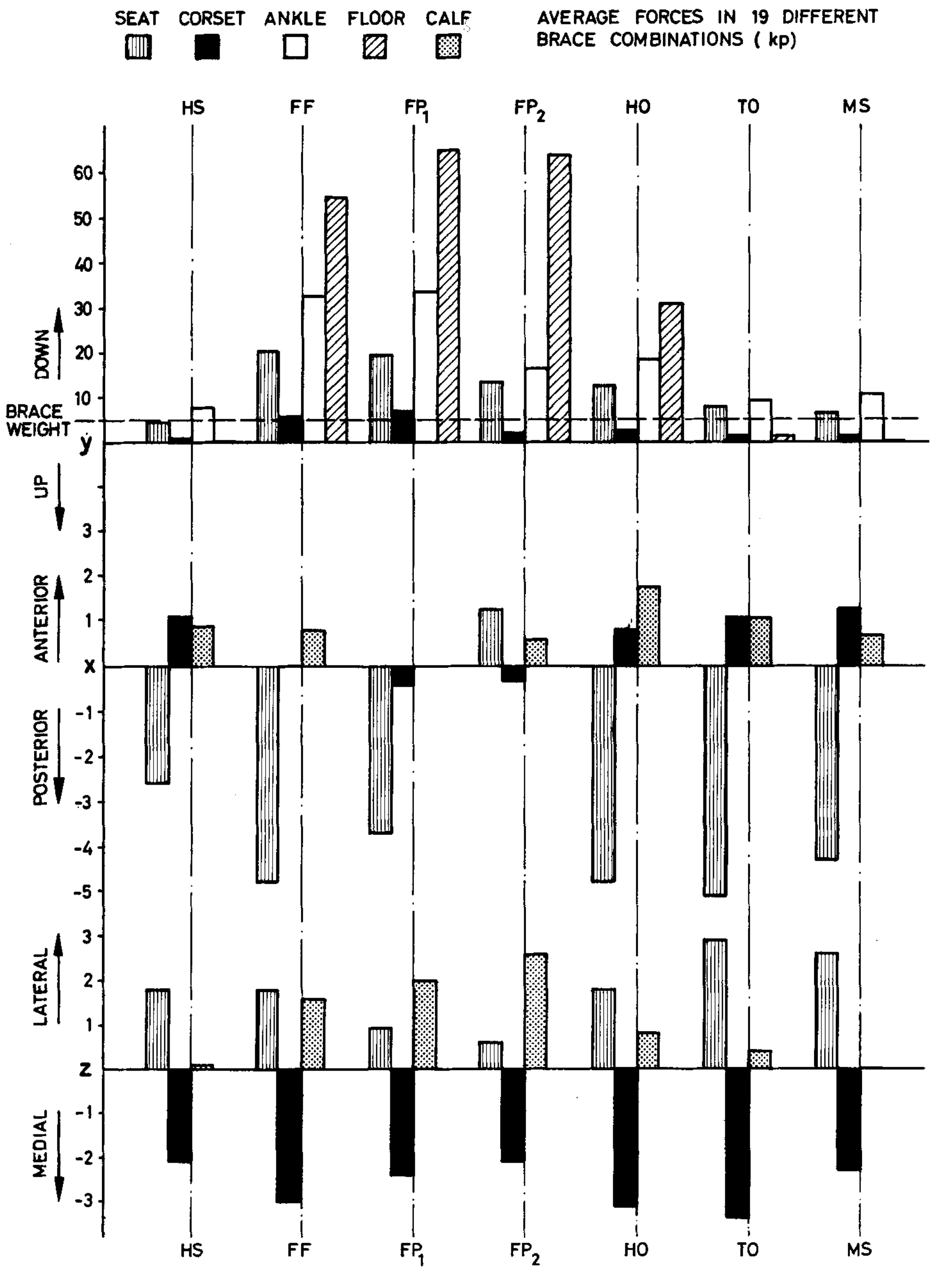

SEAT
CORSET
ANKLE
FLOOR
CALF
AVERAGE FORCES IN 19 DIFFERENT
BRACE COMBINATIONS (kp)
HS
FF
FP1
FP2
HO
TO
MS
DOWN
BRACE
WEIGHT
y
UP
ANTERIOR
x
POSTERIOR
LATERAL
z
MEDIAL
60
50
40
30
20
10
3
2
1
-1
-2
-3
-4
-5
3
2
1
-1
-2
-3
HS
FF
FP1
FP2
HO
TO
MS

Figure 31.

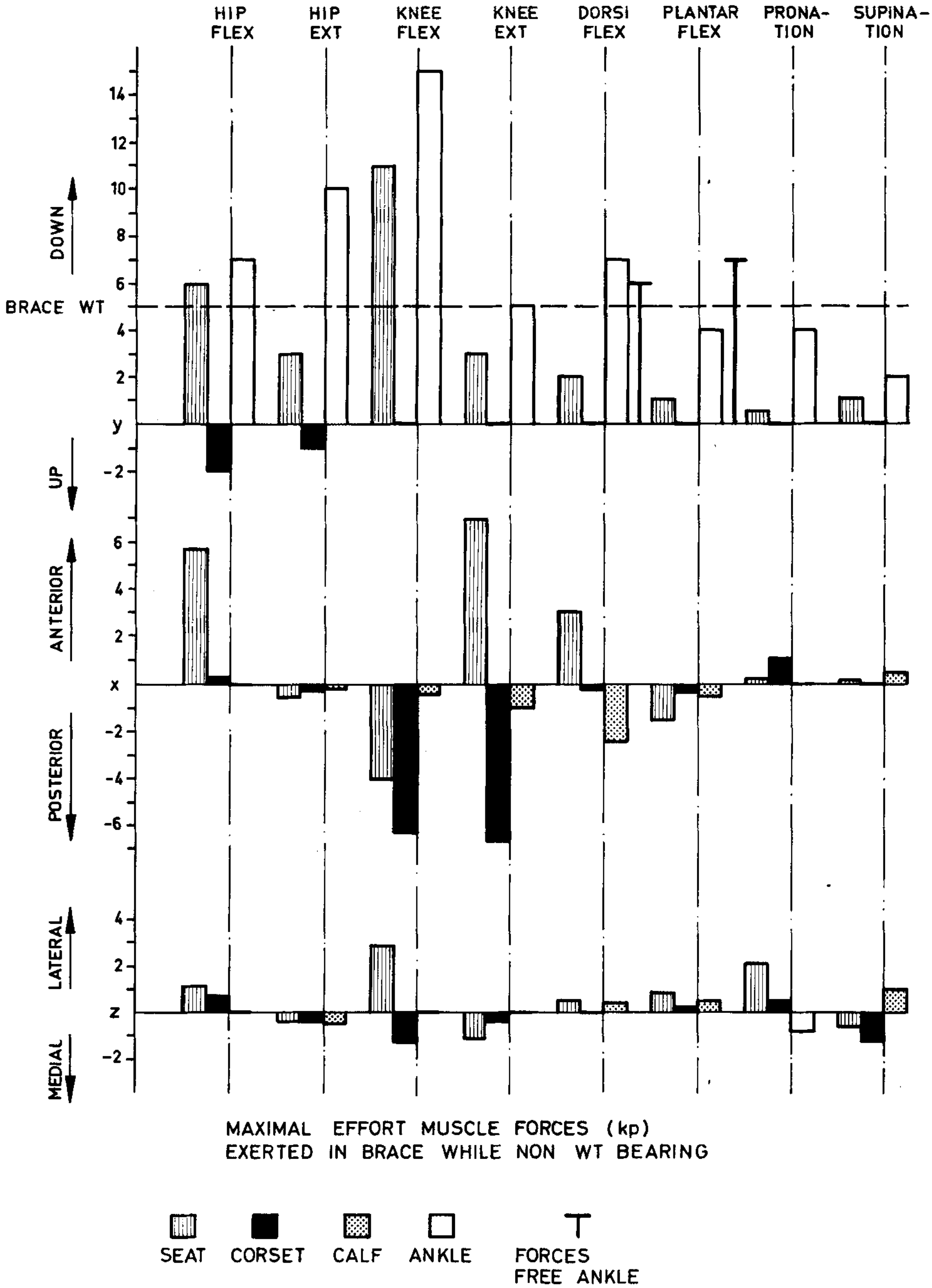

HIP FLEX
HIP EXT
KNEE FLEX
KNEE EXT
DORSI FLEX
PLANTAR FLEX
PRONA-TION
SUPINA-TION
14
12
10
8
6
BRACE WT
4
2
y
-2
DOWN
UP
6
4
2
x
-2
-4
-6
ANTERIOR
POSTERIOR
4
2
z
-2
LATERAL
MEDIAL
MAXIMAL EFFORT MUSCLE FORCES (kp)
EXERTED IN BRACE WHILE NON WT BEARING
SEAT
CORSET
CALF
ANKLE
FORCES
FREE ANKLE

Figure 32.

Medial-lateral loading (Z)

Seat and calf forces were lateral and opposite to the corset forces and produced a medial bending effect against the brace and shearing forces against the leg. Possible causes of this phenomenon are the normal valgus thrust of the knee (Perry 1967) and vertical loading of the brace in abduction.

Loading From Muscle Forces

Loading was produced on the brace by attempted hip, knee and foot motion while standing non weight bearing (Fig. 32). Since the brace is rigid and attached to both ends of the limb, axial shortening or lengthening loads caused by attempted joint motion will produce opposite tensile and compressive loads on the brace and leg.

Non weight-bearing forces along the brace's axis were recorded before and during the applied forces from attempted joint motion. Longitudinal before-loading values ranged from 2—7 kp indicating the weight of the brace plus the presence of unavoidable muscle activity. Plantar flexion produced a compressive instead of the expected tensile effect which can be explained by a slightly anterior alignment of the ankle joint with respect to the caliper frame (Table 3).

Mean Moments

Moments are expressed in kp-mm units (Fig. 33) and interpreted as being applied to the brace.

Torsional moments

Torsional moments about the longitudinal brace axis (Y) were internally directed from heel strike to the first force maximum. They became external at the second force maximum and then internal again during swing phase. This sequence correlates with the pattern of transverse intersegmental moments generated by an unbraced limb, but is smaller in magnitude (Murdoch, 1970).

Table 3. Loading (kp) due to attempted joint motion while standing non weight bearing.

Attempted motion		*Before loading*	*during loading*	*muscle force*	*net effect brace*	*net effect limb*
Hip	flexion	6	7	+1	compression	tension
	extension	6	10	+4	compression	tension
knee	flexion	6	15	+9	compression	tension
	extension	7	5	—2	compression	tension
ankle	dorsiflexion					
	plantarflexion	4	7	+3	compression	tension
	fixed	3	4	+1	compression	tension
	free	3	7	+4	compression	tension
	supination	2	2	0	none	none
	pronation	2	4	+2	compression	tension

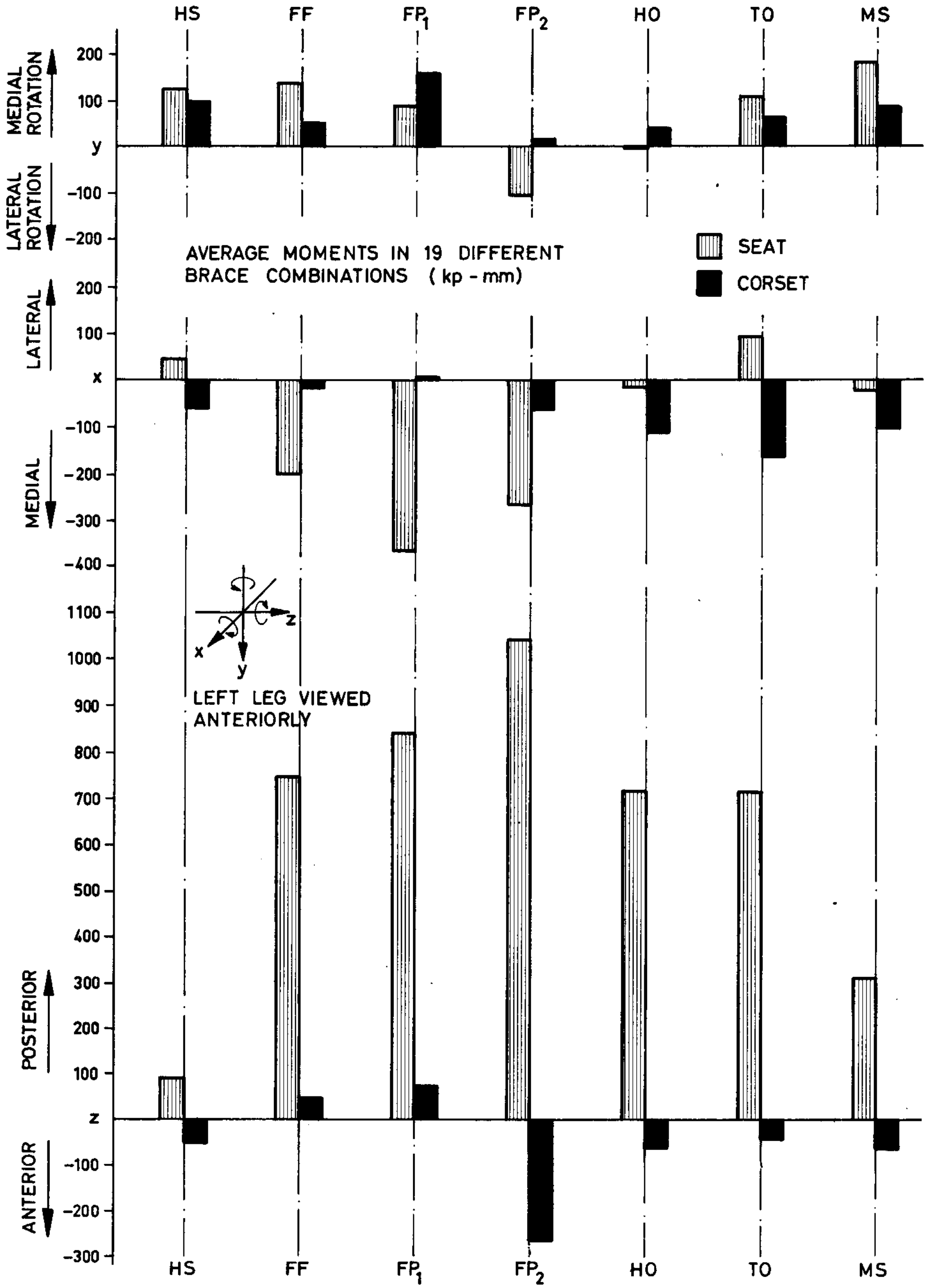

HS FF FP1 FP2 HO TO MS
MEDIAL ROTATION
LATERAL ROTATION
200
100
y
-100
-200
AVERAGE MOMENTS IN 19 DIFFERENT
BRACE COMBINATIONS (kp - mm)
SEAT
CORSET
LATERAL
MEDIAL
x
-300
-400
1100
1000
900
800
700
600
500
400
300
LEFT LEG VIEWED
ANTERIORLY
POSTERIOR
ANTERIOR
z
-300

Figure 33.

Anterior-posterior moments

The large posterior moments exerted on the seat shell were caused by body weight forces applied to the ischial platform, which is posterior to the medial-lateral (Z) axis of rotation. In addition, a small downward component is provided from forward flexion of the pelvis during early swing phase (Strohm et al., 1963).

Medial-lateral moments

During mid stance medial moments are exerted primarily on the seat shell, which correlates with the greater axial loading on the medial side of the brace. At heel strike and toe off seat moments became lateral possibly reflecting the preparatory movements of the body to meet these transitional phases.

Table 4.

Moments about ring axis closed seat and corset shells		HS	FF	FP1	FP2	HO	TO	MS
				Σmy (kp-mm)				
Patten bottom	Seat	167		628	483		1346	1218
	Corset	168		167	224		148	231
Fixed ankle	Seat	145	29	72	—275	—26	31	107
	Corset	432	83	1661	73	101	148	149
Caliper heel hinge	Seat	191	116	72		—28	34	209
	Corset	178	45	101		101	119	178
Ankle hinge	Seat	104	—18	—31		4	94	145
	Corset	120	101	119		52	109	176
				Σmx (kp-mm)				
Patten bottom	Seat	—95		—570	—760		570	190
	Corset	—238		—380	—380		—475	—285
Fixed ankle	Seat	95	—95	—475	—380	—95	143	95
	Corset	0	—95	—570	—285	—238	—285	—95
Caliper heel hinge	Seat	48	—285	—380		10	143	48
	Corset	—95	—48	—48		—238	—475	—95
Ankle hinge	Seat	—95	—285	—665		—95	—48	—190
	Corset	—285	—285	—171		—475	—380	—380
				Σmz (kp-mm)				
Patten bottom	Seat	—30		2023	2239		105	107
	Corset	—100		93	139		17	—17
Fixed ankle	Seat	265	1329	1541	525	584	558	156
	Corset	289	37	1695	—168	—130	—155	—76
Caliper heel hinge	Seat	132	901	953		638	560	292
	Corset	17	—143	—126		146	—109	—76
Ankle hinge	Seat	—59	685	790		693	774	210
	Corset	17	182	234		17	33	33

Interpretation of moments

Table 4 lists the moments about the X, Y, Z axes of the upper segment's measuring rings in four closed seat shell braces with different foot assemblies. Where no moments are listed under FP2 it means that no definite FP2 existed on the ankle force tracing. Patten bottom braces do not have a FF or HO phase.

Pooled trends have been shown in Fig. 33. It is evident that the moments are different in each segment and vary during the gait cycle. The greatest moments were exerted when the forces along the Y axis were largest. Moments were largest in patten bottom combinations and occured primarily on the seat shell. With shoe assemblies the moments were less because part of the weight bearing forces responsible for the moments shunted through the limb directly to the ground. During mid stance phase more loading was transferred to the corset and the moments acting on this segment increased.

LOADING DISTRIBUTION WITHIN THE BRACE

Weight relief refers to the amount of body weight transfered to the brace. By comparing the ankle transducer recordings with the body weight of 61 kp it is possible to assess the unweighting effectiveness of a particular test brace. Fig. 34 compares 8 different brace combinations; closed and open seat shell types, each with 4 different foot assemblies. Weight transfer was complete in the patten bottom braces which is expected since there is no foot-ground contact. The other combinations displayed a progressively lower FP2 loading in the fixed, heel-hinge and free-ankle combinations respectively which suggests that the skeletal loading increased as the subject rolled forward into his metatarsal heads. All combinations except the patten bottom types displayed a progressive decrease in unweighting toward the latter part of stance phase.

Fig. 35 compares the three groups with different thigh-shell configurations in free and fixed ankle assemblies. Although the group with an open corset was tested with a nonabducted gait, the marked difference in loading recorded is primarily attributed to the small corset segment which is unable to accept any weight transfer.

Comparative load distribution in the medial and lateral uprights

Comparative loading of the medial and lateral uprights was studied at the ankle level for abducted and non abducted gait patterns (Fig. 35). The lateral upright loading was smaller and of a different pattern than the medial upright. During the loading period foot-flat-to-force-maximum-two, abducted gait produced 3—7 times greater axial forces on the medial upright. This asymmetry was not present during the same period in non-abducted walking.

Level of weight transfer

The closed seat and open seat shell

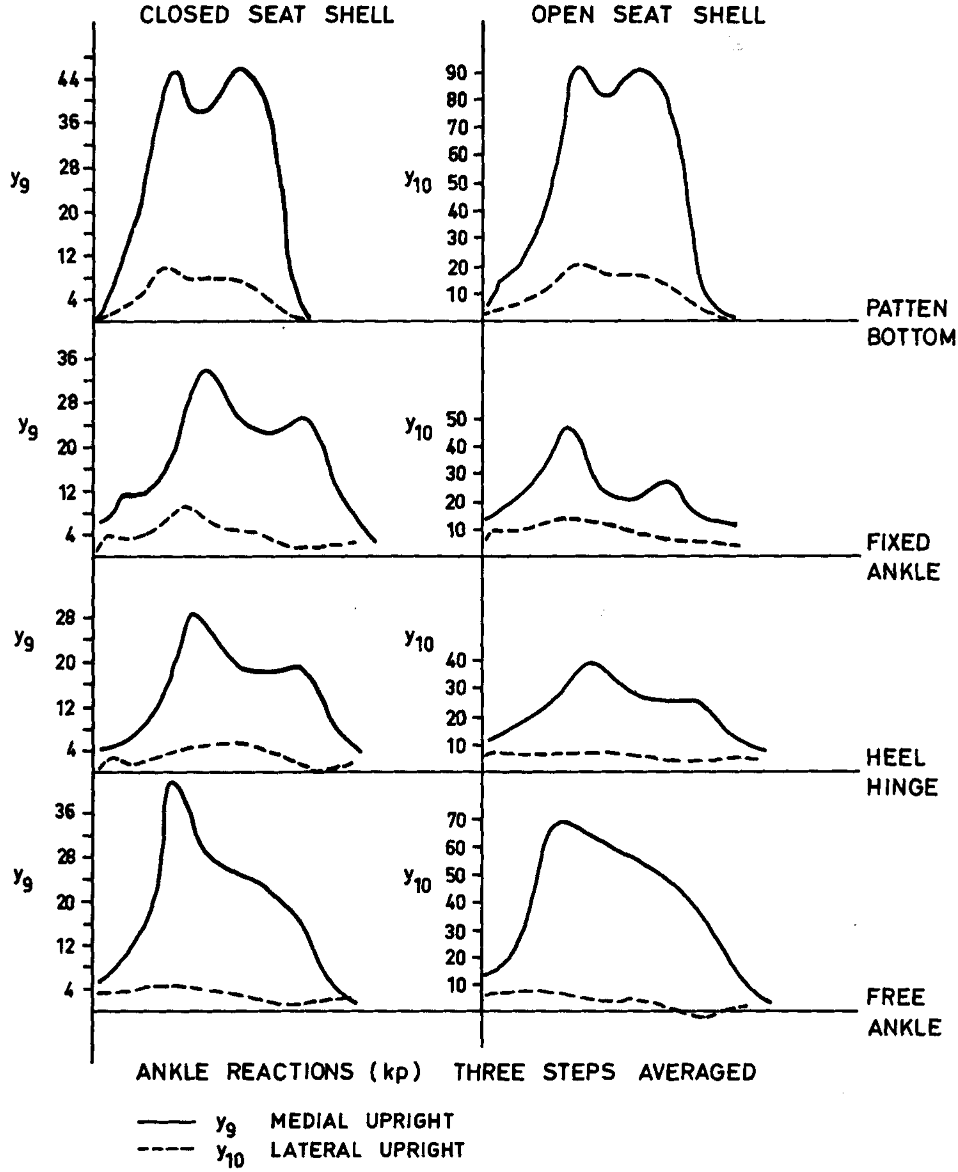

CLOSED SEAT SHELL
OPEN SEAT SHELL
y_9
y_{10}
PATTEN BOTTOM
FIXED ANKLE
HEEL HINGE
FREE ANKLE
ANKLE REACTIONS (kp) THREE STEPS AVERAGED
y_9 MEDIAL UPRIGHT
y_{10} LATERAL UPRIGHT

Figure 34.

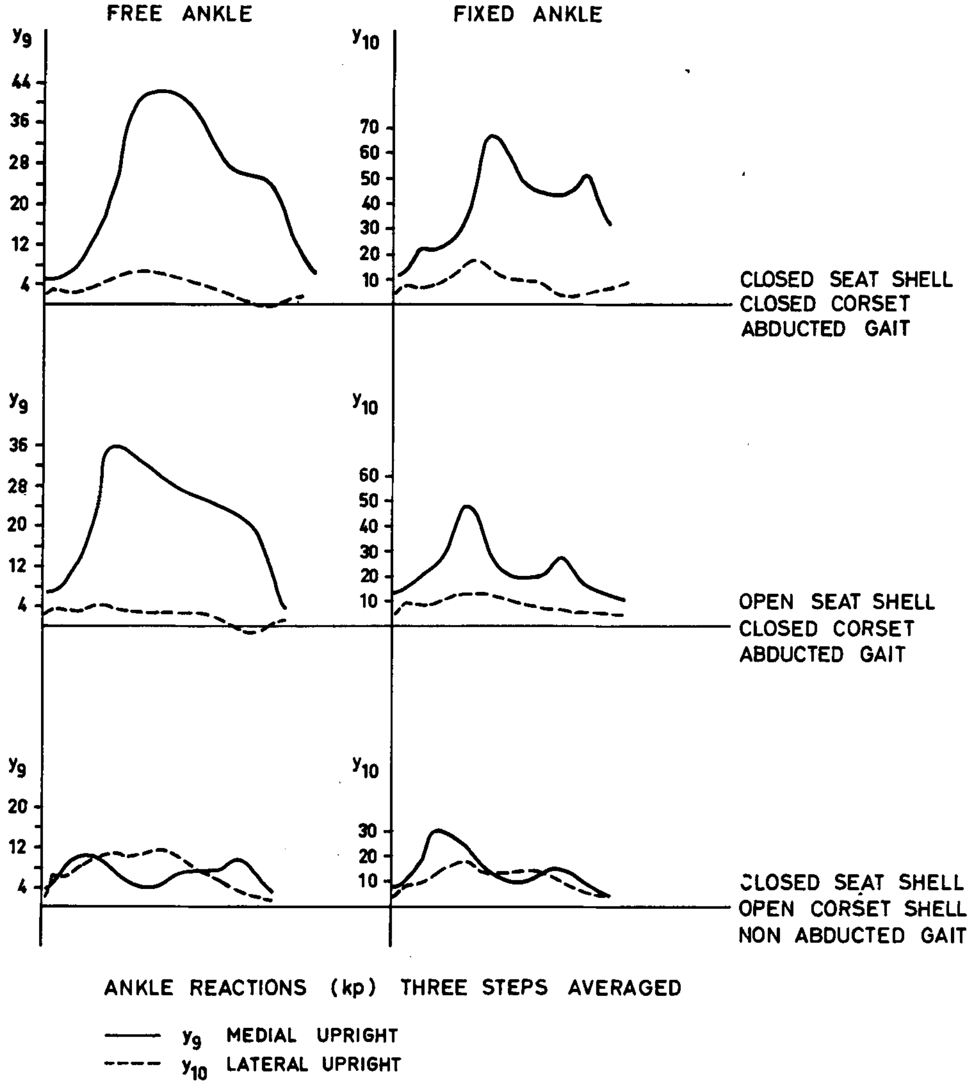

FREE ANKLE
FIXED ANKLE
y9
y10
44
36
28
20
12
4
70
60
50
40
30
20
10
CLOSED SEAT SHELL
CLOSED CORSET
ABDUCTED GAIT
OPEN SEAT SHELL
CLOSED CORSET
ABDUCTED GAIT
CLOSED SEAT SHELL
OPEN CORSET SHELL
NON ABDUCTED GAIT
ANKLE REACTIONS (kp) THREE STEPS AVERAGED
y9 MEDIAL UPRIGHT
y10 LATERAL UPRIGHT

Figure 35.

brace combinations were pooled, the average seat and corset forces computed and then expressed as a percentage of the total thigh shell load (Fig. 36).

In all cases the greatest percentage of the total thigh shell load was borne by the seat shell. However, in the open seat shell types without an anterior retaining wall, ischial contact was decreased resulting in more weight being transferred to the corset. This is shown by the corresponding increase in load percentage on the corset.

Calf band forces

Calf band forces along the X and Z axis were recorded for selected runs and their patterns compared with different foot assemblies for different alignments. Although signs sometimes differ between the test-brace patterns, the overall force trends are similar for similar braces. Group III differed more than groups I and II and is most likely explained by the changes in leg alignment permitted by the diminitive corset segment during dynamic loading.

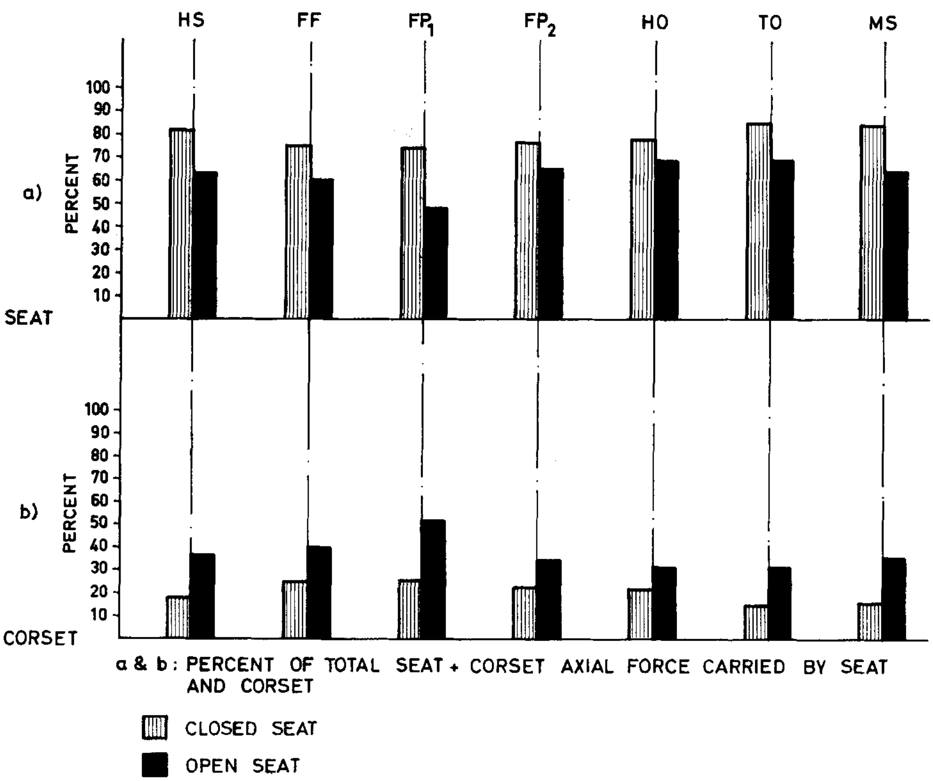


Figure 36.

Free ankle — Rotational Alignment

Free ankle force patterns are shown in Fig. 34, 35 and 37. Fig. 37 compares the effect of changing rotational alignment of the stirrup and illustrates the general force trends. At heel strike the leg is exerting anterior forces against the calf band. This pattern became progressively posterior and lateral peaking at the first vertical floor maximum. As the leg begins to pivot forward it produces more anterior calf band forces again which peak just before toe off. Different rotational alignments showed essentially similar force trends.

Free and fixed ankle combinations

Free and fixed ankle combinations were compared in the test brace having a closed seat shell and open corset shell. The essential differences be-

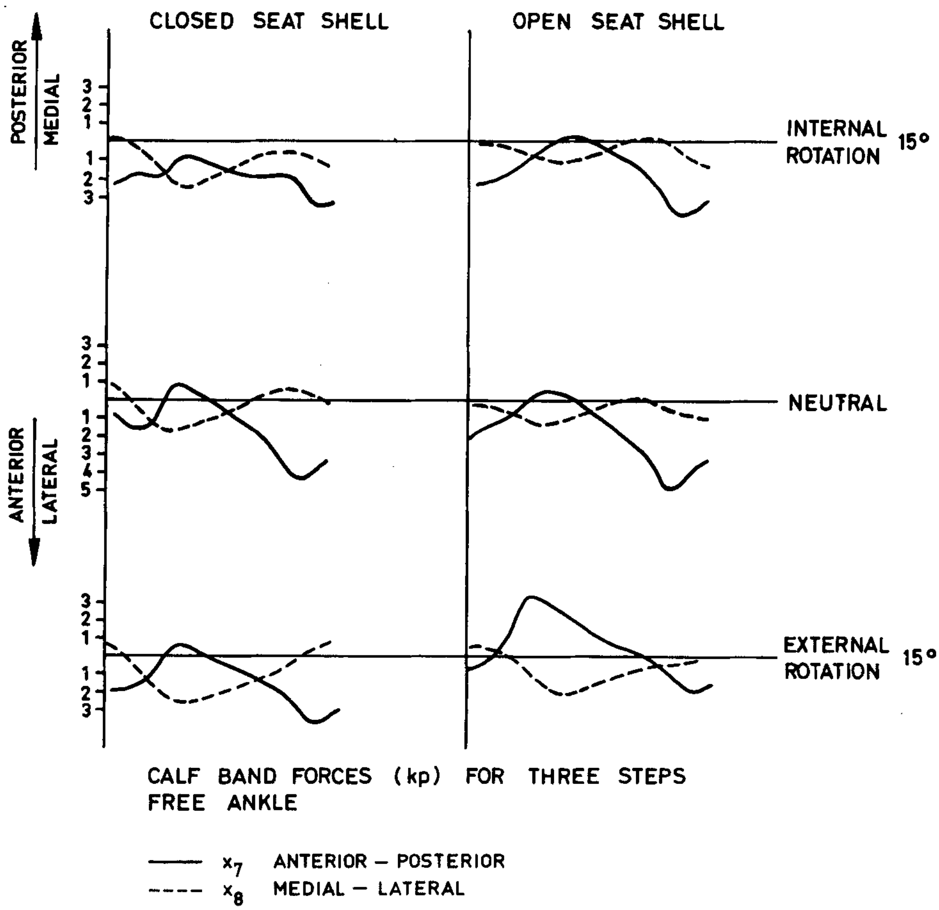
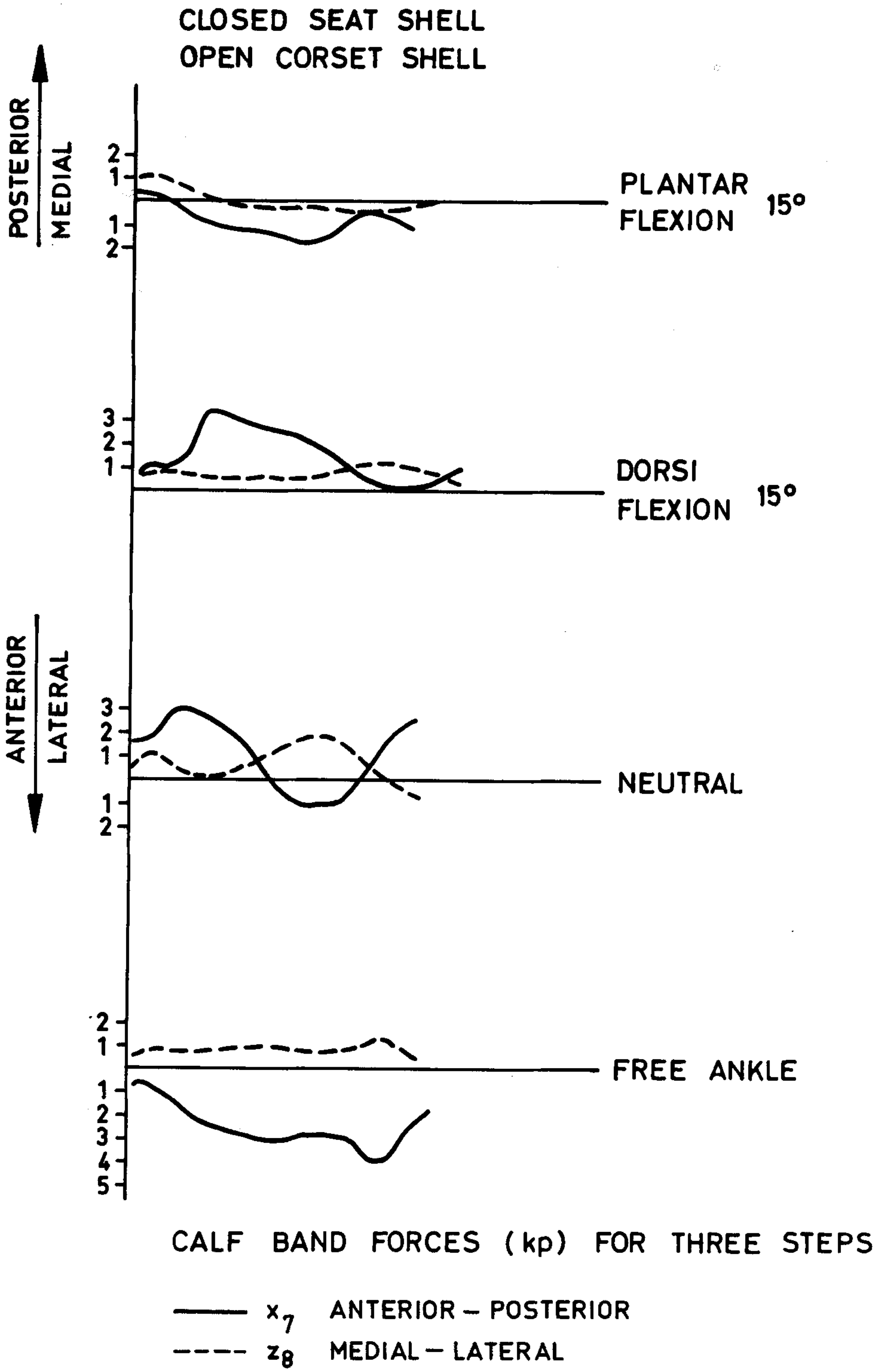
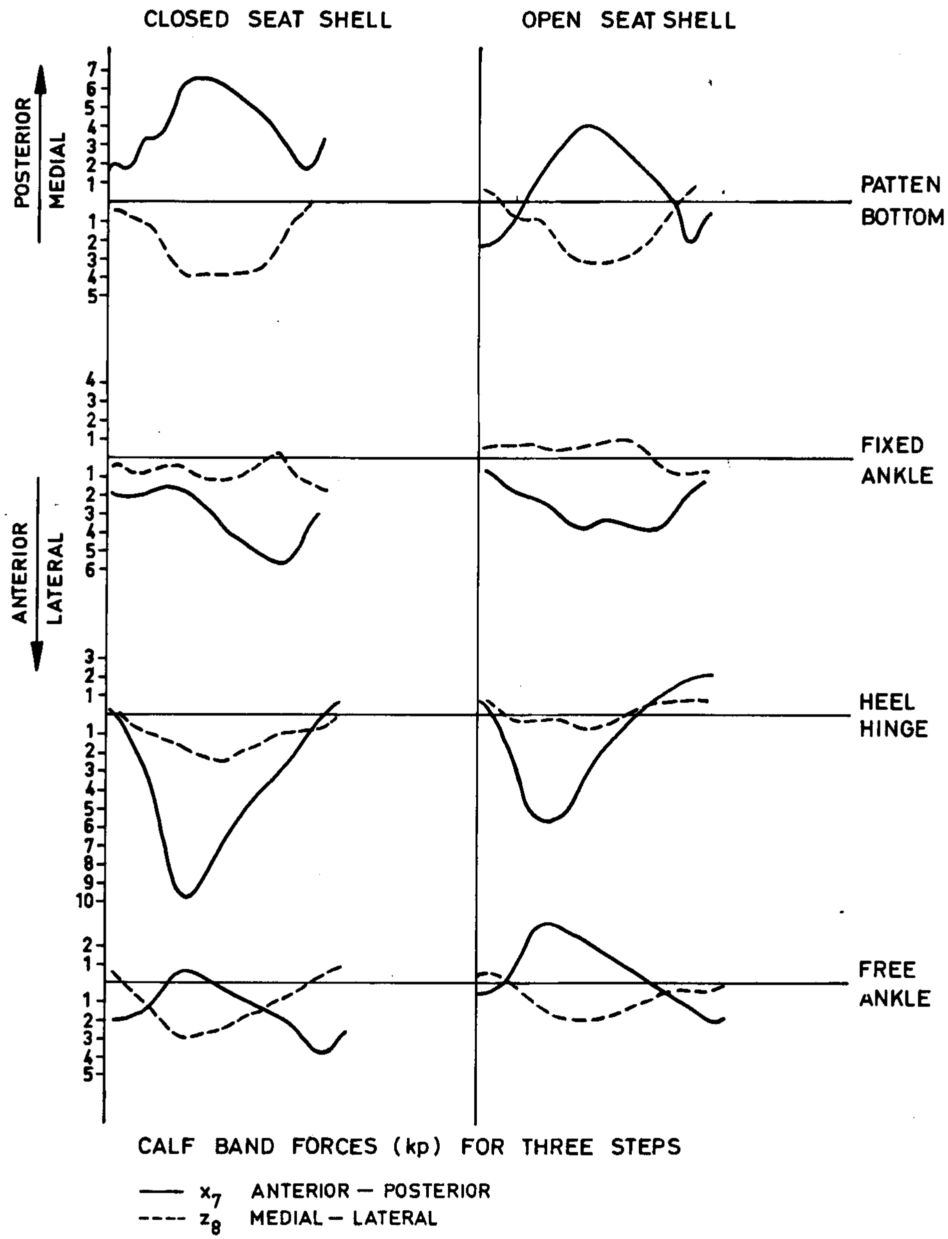


Figure 37.



CLOSED SEAT SHELL
OPEN CORSET SHELL
POSTERIOR
MEDIAL
ANTERIOR
LATERAL
PLANTAR FLEXION 15°
DORSI FLEXION 15°
NEUTRAL
FREE ANKLE
CALF BAND FORCES (kp) FOR THREE STEPS
x_7 ANTERIOR – POSTERIOR
z_8 MEDIAL – LATERAL

Figure 38.



CLOSED SEAT SHELL
OPEN SEAT SHELL
POSTERIOR
MEDIAL
ANTERIOR
LATERAL
PATTEN BOTTOM
FIXED ANKLE
HEEL HINGE
FREE ANKLE
CALF BAND FORCES (kp) FOR THREE STEPS
— x_7 ANTERIOR — POSTERIOR
---- z_8 MEDIAL — LATERAL

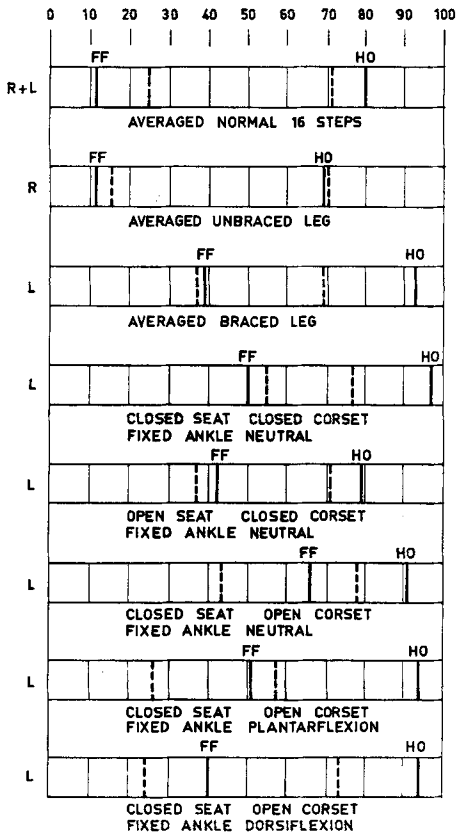
Figure 39.

tween fixed and free ankles are the lack of plantar flexion at heel strike and dorsiflexion (toe clearance) at toe off. This problem is less for the fixed plantar flexed position at heel strike and the fixed dorsiflexed position at toe off. Hence one would expect to see similar force trends, compared to a free ankle, at the beginning and end of stance phase with plantar flexed and dorsiflexed positions respectively. These predictions are substantiated in Fig. 38.

Comparison of four standard foot assembles

Different foot assemblies consisting of a patten bottom, fixed ankle (neutral), caliper heel hinge and a free ankle (15 deg. of ext rotation) were compared with the two basic thigh-shell configurations (Fig. 39). The free and fixed ankle patterns have already been described. A heel caliper hinge is eccentrically located compared to the anatomic ankle joint. During plantarflexion the leg presses forward against the calf band, while dorsiflexion produces an opposite posterior force. This effect is demonstrated by a 10 kp sharp anterior force peak occuring with plantarflexion and just before the first force maximum. As the leg pivots forward and ankle dorsiflexion begins, these forces reverse but are not as large. The 10 kp reaction force exerted posteriorly against the anterior tibial border would produce an extension moment about the knee.

In the patten bottom brace type distal alignment of the leg was maintained by the calf band alone. The posterior force peak corresponds to the first force maximum. As the leg pivots forward the posterior force decreases and in the open seat became positive near toe off.



FOOT POSITION CONTACT PERIODS AND FORCE MAXIMUMS
PERCENTAGE TOTAL STANCE PHASE

---- (FORCE MAXIMUM) FP_1, FP_2

Figure 40.

FOOT-POSITION AND FLOOR MAXIMUM RELATIONSHIPS

Floor-contact periods for each foot position and the timing of FP1 and FP2 for all steps in each walk (except the first and last) were determined, averaged and expressed as a percentage of the total stance phase time. The results were compared against averaged walks without the brace and between the braced and unbraced sides for each walk (Fig. 40). Compared to unbraced gait, with the left leg braced, the right side showed an earlier FP1, longer HO-TO, and FP2 closer or within HO-TO. The braced leg demonstrated a longer HS-FF, FP1 close to or within HS-FF, shorter HO-TO and earlier FP2 in FF-HO. The greatest changes occured with fixed ankle combinations where HS-FF was prolonged with FP1 usually occuring well within this period indicating that the full impact of weight acceptance was borne on the heel.

DISCUSSION

The data presented was obtained from the force-time histories recorded with an instrumented test brace capable of simulating selected ischial weight-bearing brace combinations. The appliance was 2—3 kg heavier than standard designs, less flexible and lacked contouring. These differences arose from the construction details which were necessary to achieve a lower instrumentation error and a greater versatility. The instrumentation and test methods proved to be a suitable tool for investigating the variables outlined. Because one normal subject was used his biomechanical interactions with the brace are not considered necessarily representative of other normal subjects or patients requiring orthoses. However, the force patterns and envelopes described have helped to take a closer look at the effectiveness of certain general bracing techniques and learn more about the load distribution within brace structures. The net effect on the leg is determined from an analysis of these findings.

Floor contact patterns

Small changes in the duration of the foot positions and force-peak timing, not readily observable, were detectable by the foot switch circuit.

A normal gait without the brace, but at the same test speed showed the first force peak to occur in early Foot-Flat-to-Heel-Off phase, permitting the upward acceleration of the trunk to be applied gradually and distributed more evenly over the foot. With fixed ankle assemblies, the Heel-Strike-To-Foot-Flat period was prolonged and contained the first force maximum which indicates that the full impact was borne on the heel.

Loading distribution in the brace

Weight relief means the transference of body weight to the brace's structures. The location of this transference determines the level of skeletal unloading. Thigh shells with an anterior wall in the seat segment carried a more proximal distribution of the total load. This indicates that an anterior wall is important in order to maintain optimal ischial contact with the seat platform.

On the braces with different foot assemblies which were evaluated, the patten bottom types provided complete weight transference during stance phase, however they are uncomfortable and present instability problems (Lehmann et al., 1970). The braces with shoe combinations demonstrated lesser degrees of weight-relieving effectiveness. Although the fixed ankle with metal sole plate is considered necessary to reduce metatarsal head loading during late stance phase, in this study it actually was associated with greater loading than with the free ankle. Variability between runs could account for this apparent paradox but another contributing factor may be the unseating effect produced by a greater jolt at heel strike.

Swing phase loading patterns were recorded on the brace and suggested a contribution from muscle activity. The same interpretation has been reported by other workers (Kirkpatrick et al., 1969 and Magora et al., 1968). Consequently, investigations were undertaken to evaluate this possibility. Loading against the confines of the apparatus showed that the largest forces were produced with knee flexion, hip flexion-extension and ankle dorsiflexion plantarflexion in that order. That muscle forces of some magnitude can occur in a partially immobilized limb is supported by Schenck's et al. (1969) studies on instrumented short-leg plaster casts. He found that stresses were often higher during swing phase than stance phase.

It is concluded that the axial swing phase loading resulted from muscle activity plus any brace weight above the transducers not suspended by interface friction with the leg. Swing phase axial loading was not seen in the patten bottom types where ankle motion was independent of the brace. Forces were exerted on the braces with a shoe assembly connected. Assuming equal suspension forces at the thigh shell interfaces, and the presence of patterned ankle dorsiflexion to provide toe clearance, this loading must have been caused by muscle forces transmitted from foot movements within the shoe.

Unequal loading

The greater loading of the medial upright in a slightly abducted gait suggests that lighter designs might be possible using a single medial upright or lighter lateral upright.

The abducted gait and normal valgus thrust of the knee (Perry 1967) are considered to be the main causes of the bending loads recorded by the brace.

Comparison of calf band force patterns produced with a caliper and ankle hinge showed a marked difference. Calipers are normally inserted in the heel region which is poorly aligned with the anatomic ankle joint. During ambulation in a short leg brace this malalignment is usually not a problem because the eccentric rotations are converted to a compensatory sliding motion up and down the calf. In the long-leg test brace this compensatory motion was not possible. Instead, the lower leg was pressed anteriorly against the calf band with forces ranging up to 10 kp during plantar flexion. Since the reaction against the leg is posterior, a concomitant extension moment at the knee is produced. The possibility of using this moment to dynamically stabilize a weak knee in early stance phase is suggested.

That moment differences occured between the seat and shell segments is partly related to the dynamic interactions influencing the leg and whole brace, partly to the changing force distributions which occured between the shell segments during the gait cycle and partly to the additional forces applied above the hip by the pelvis to the seat-shell extensions. The significance of these differences clinically with respect to the effects on the contained thigh segment is not entirely clear. Further, shell forces applied to the thigh's soft tissues are not necessarily transmitted wholly to the underlying femur. If rotational stability is desired, such as to protect a femoral fracture, then torque applied to or by the braced leg should be exerted proximally (in order to by-pass the fracture).

From a comparison of torque distributions in the seat and corset shell segments of different test brace combinations it is apparent that the largest and most proximally exerted torques were produced when brace loading was greatest at the seat. Thus, in order to concentrate the applied torques proximal to a femoral shaft fracture, thereby enhancing rotational stability, a long-leg brace used for ambulatory fracture treatment should provide ischial weight-bearing capabilities.

Inadvertant swiveling of the braced leg on its heel with the attendant torque effect can be minimized by shortening the Heel-Strike-To-Foot-Flat period. Since fixed ankles prolong this susceptible period a free ankle should be used when torque effects must be avoided.

The effects of a disparity in anterior-posterior or medial-lateral moments is more difficult to evaluate. Because the leg and brace are assumed to be in equilibrium one moment must be counterbalanced by another. The large posterior moment demonstrated at the seat must be counterbalanced somewhere more distally on the thigh. For a recent fracture this could have either a stabilizing or disrupting influence depending on the fracture level. If dynamic stabilization of fractures is possible it will be accomplished by providing the proper moment and force interactions against the fragments.

SUMMARY AND CONCLUSIONS

1. Research in the field of lower extremity orthotics is hampered by a lack of quantitative clinical data which is necessary to correlate clinical findings with a particular brace.

2. Clinical evaluation of a lower extremity orthotic device used during ambulation must be supplemented by a quantitative analysis to understand the biomechanical nature of the interacting systems.

3. The test brace and experimental method herein described has proved to be a useful research tool in acquiring force-time histories on testwalks representing ischial weight-bearing brace combinations worn by the author.

4. Body weight transmission to the brace was complete only in the brace types with a patten bottom. Brace combinations with shoe attachments failed to completely unload the leg and became progressively more ineffective toward the latter part of stance phase. This resulted partly from incomplete transmission of bodyweight to the brace and partly from patterned muscle activity against the confines of the apparatus.

5. An anterior (closed) seat-shell wall is necessary to maintain effective ischial platform contact. Closed seat thigh shells demonstrated more proximal and more effective weight relief than the open seat thigh shell brace types tested.

6. Muscle forces voluntarily exerted against the confines of the brace with a shoe attachment were documented. Their role in producing swing phase loading was established. From patterned gait habits these muscle forces act to produce tensile limb loading during swing phase.

7. Braces with fixed ankle assemblies showed less weight relief than free ankle types. This may have been related to an unseating influence caused by the greater jolt effect felt during heel strike. Because of the prolonged heel strike period, full weight acceptance is borne on the heel instead of the whole foot as occured with free ankle assemblies.

8. The caliper heel hinge because of its axis eccentricity with respect to the anatomic ankle joint caused a sharp posteriorly directed force build up to 10 kp against the leg at the calf band during plantar flexion, just before full weight acceptance. Since the effect on the knee would be an extension moment, it should be possible to use this force for dynamic stabilization of a weak knee during that critical period when other factors are causing flexion moments.
9. Axial loading of the medial upright was increased up to 7 times greater than the lateral upright when walking in slight abduction. The design implications are that when this gait pattern is present, the use of a lighter lateral upright or a stronger single medial upright may be possible.
10. The best rotational stability within the mid thigh-shell area was achieved when applied torsional moments were exerted proximally. From a comparison of the torsional moments in different combinations, this condition was optimized with a closed seat shell and patten bottom brace. Thus when optimal torsional stability is required a long-leg brace with ischial weight-bearing features is desirable.
11. Anterior-posterior and medial-lateral moments computed for the corset and seat shell rings were compared and considered. Their relevance to dynamic fracture stabilization is discussed.
12. Different rotational alignments of the ankle stirrup assemblies produced no distinguishing force patterns. The calf band loading trends recorded for different fixed ankle positions varied between anterior forces up to 2 kp for the plantar flexed position to posterior forces up to 2 kp for the dorsiflexed position. The fixed neutral ankle cycled between these ranges. Patten bottom, free and heel hinge assemblies produced higher forces ranging between 7 kp of posterior force for a patten bottom to 10 kp of anterior force with a caliper heel. These larger values are attributed to the less restricted walking patterns and the magnifying effect ankle motion produces on hinge malalignment.
13. A foot-position floor contact switch is described which synchronizes force maximum timing and foot positions with separate recorders by marking the analogs simultaneously.

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