"KNEE MECHANISM PERFORMANCE IN AMPUTEE ACTIVITY"

by

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"The object should be to remove what is useless, in order to replace it by what is useful"

H. Bigg (1885)

SUMMARY

A strain-gauged pylon has been constructed which can be incorporated into the shank of an artificial leg for a patient with a below knee or higher levels of amputation. Together with a goniometer at the artificial knee joint, this enabled sufficient information to be recorded to calculate the force actions exerted between the socket and the amputee. A computer programme was written to carry out the necessary computation and plot out the results in graphical form.

An above knee amputee was fitted with a quadrilateral suction socket and the strain-gauged dynometer was attached to a SACH (solid ankle cushion heel) foot. Various knee mechanisms designed to aid stability were incorporated in the leg and the performance of the amputee assessed by measuring the force actions developed. Tests were carried out over a range of activities and the devices compared for their effectiveness.

A statistical ananlysis of level walking indicated that the Lammers poly-centric knee was particularly beneficial to this amputee, although in overall performance the Blatchford stabilized knee was found to be the most satisfactory device investigated.

This dynamic work and earlier research into the static characteristics of artificial legs were compared. The static testing of knee mechanisms has limited application in a limb-fitting clinic but can provide important information on their characteristics. The problem of correct prescription of

artificial legs in connection with an objective assessment of the amputee's capability was discussed.

The different requirements were outlined and lines of development indicated for a clinical tool and research instrument based on the strain-gauged pylon.

INTRODUCTION

In a normal individual the joints between the various segments of the leg are stabilized by muscle and ligamentous action. When an artificial leg is fitted to an amputee, the joints in the prosthesis are not under direct muscular control and substitutes must be found.

In an artificial leg for a patient with an above knee amputation, knee stability can be achieved by the amputee actively extending his stump, by a mechanism built in to the artificial knee itself and by alignment of the leg.

No information is available on the inter-relation of stump capability, alignment and joint mechanism stability. Furthermore, the force actions developed in amputee activity are not known.

The work described here provides some indication of these factors and outlines a basis for prescription of prostheses for above-knee amputees.

ACKNOWLEDGMENTS

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The outline of the project was formulated by Professor C.W. Radcliffe (University of California, Berkeley) and the project itself was supervised by Professor R.M. Kenedi and Dr. J.P. Paul. I am grateful to these people for their assistance and encouragement.

Special thanks are due to Mr. W. Barclay for his active co-operation throughout the investigations. And to my wife Joanne for typing this thesis, and her tolerance and forbearance.

Philip J. Lowe

NOMENCLATURE

As confusion may arise when one symbol is used to denote more than one quantity, the following system is adopted.

* Denotes that the symbol is used only in the Statistical Analysis in section 4.4 page 116.

Symbols used only in the Appendices, are accompanied in their definition by a reference to the particular Appendix.

All symbols are accompanied by the units used in the text if applicable except for those used in the statistical analysis. (unit) is used to denote 0.1 mm. deflection on the U.V. recorder trace.

a	denotes a constant
a	distance from upper bending moment gauge to reference point (in.) (II)
A	ankle
a_j	distance from upper bending moment gauge in j plane to knee (in.)
a_{ij}	reading on load channel i ($0 < i < 6$) due to load action j ($0 < j < 6$). (unit / lbf. or unit / lbf. in.) See section 3.4.
a_{ij}	*value of effect in j ($0 < j < n_i$) stride, i ($0 < i < m$) knee device
A/K	above knee
A/P	anterior posterior
b	shank length i.e. AK (in.)
b	distance between upper and lower bending moment gauges (in.) (II)
b_j	distance between upper and lower bending moment gauges in j plane (in.)
B/K	below knee
B.S.K.	Blatchford stabilized knee with pneumatic swing phase control

c — goniometer calibration coefficient. (radians / 636 cm.)

c — constant in regression analysis (lbf.) (II)

C — centre of pressure (on foot)

c_i — *a "contrast". (An array of numbers defined on page 121)

c_i — distance between lower bending moment gauge and ankle in j plane (in.)

d — distance from knee to 'hip-centre of pressure' line (i.e. KK') (in.)

d — differential operator

D — difference between trial values of x (in.) (II)

e — normal reading error (unit) (V)

e' — zero reading error (unit) (V)

f — frequency (H_Z)

F — *tabulated values for assessing statistical significance

f_k — force in k direction

$f(x)$ — function of x

$f'(x)$, $f''(x)$, $f'''(x)$ — first, second and third derivatives of f(x) (III)

$f_o(h)$ — undeformed shape function (in.) (III)

$f_l(h)$ — stiffness function (in./lbf.) (III)

g — x coordinate of centre of pressure, C. (in.)

g_o — undeformed value of g at a particular value of h (in.) (III)

G.R. — Otto Bock Greissinger knee

h — y coordinate of C (in.)

H — hip joint

H' — projection of H on Z plane

h_o — initial trial value of h (in.) (III)

i	suffix
it	distance medially of hip joint from Z plane i.e. HH' (in.)
j	suffix
k	suffix
k	calibration coefficient (unit/lbf. or unit/lbf. in.) (II)
k	elastic modulus or stiffness (in/lbf.) (III)
k	lever arm for calibration (in.) (V)
K	knee joint (N.B. in polycentric mechanisms, K is taken to be the instantaneous centre of the knee when the thigh is at 90° to the shank)
K'	projection of K on 'hip-centre of pressure' line, CH'
l	thigh length i.e. KH (in.)
L	position of lower bending moment gauges
L	*difference between two means
L.A.	Lammers knee
m	*number of knee devices
m	slope of regression line (lbf./in.)
m_j	*mean value of an effect for a particular knee device
me	muscular effort (lbf. in. sec.)
me_{av}	average muscular effort (lbf. in.)
M_i	general term for bending moment (lbf. in.)
M_{ij}	bending moment at i in j plane (lbf. in.)
M_{HRZ}	A/P hip moment required to stabilize a single-axis, non-stabilized knee (lbf. in.)
M/L	mediolateral
MSW	*mean sum of squares within knee devices

n	number of time intervals investigated in one test
n	number of readings (II)
n	degree of polynomial (III)
N	*total number of strides for all knee devices
n_i	*number of strides for i knee device
O.B.	Otto Bock safety knee
P	component of resultant load in x' direction (lbf.)
P	load applied (lbf.) (II)
r	gauge reading (unit)
r_{ij}	reading on bending moment gauges at i in j plane (unit)
rf_x	axial load gauge reading (unit)
rT	torque gauge reading (unit)
$R\,\Delta\,F(Z)$	random error in force action F due to error in variable Z (lbf. or lbf. in.) (V)
S	*term defined by Scheffé (page 121)
$S\,\Delta\,F(Z)$	systemic error in force action F due to error in variable Z (lbf. or lbf. in.) (V)
S.A.	Single-axis, non-stabilized knee with pneumatic swing phase control
T	torque about x axis of pylon (lbf. in.)
t_i	*total of all readings for i knee device
T_n	Chebychev polynomial of degree n (III)
U	position of upper bending moment gauges
unit	0.1 mm. deflection on U.V. recorder trace
U.C.B.	University of California polycentric knee with pneumatic swing phase control
var	gauge reading (unit) (V)
var_o	zero gauge reading (unit) (V)

x, x'	directions defined in Fig.4.1 page 110 and Fig.4.2 page 111
X, X'	planes " " " " " " " " "
x	distance from reference point (in.) (II)
x	denotes an independent variable
$\bar{x}$	mean value of x (in.) (II)
$\underline{x}$	required value of x (III)
x_i	(i + l) trial value of x (III)
y	denotes a dependent variable
y, y'	directions defined in Fig.4.1 page 110
Y, Y'	planes " " " " "
z, z'	directions defined in Fig.4.2 page 111
Z, Z'	planes " " " " "
Z	rationalized reading (lbf. or lbf.in.) (V)

Greek:-

α	*level of significance
β	shank angle (radians and degrees)
∂	partial differential operator
Δ	an error term
Δt	time interval (sec.)
$\Delta F(Z)$	error in force action F due to error in variable Z (lbf. or lbf.in.)
ϕ	angle between x and x' directions (radians and degrees)
$\left.\begin{matrix}\nu_1\\ \nu_2\end{matrix}\right\}$	*degrees of freedom
σ_a	standard deviation of a

Σ	summation sign
θ	knee angle (radians and degrees)

Suffices:-

i	indicates general parameter e.g. M_i - general value of M
i	position:- A ankle L lower bending moment gauges U upper " " " K knee joint H hip joint H' projection of H on Z plane
j	planes:- X, Y, Y', Z (defined in Fig. 4.1 page 110 and Fig. 4.2 page 111)
k	directions:- x', x, y, (defined in Fig. 4.1 page 110 and Fig. 4.2 page 111)

Other symbols:-

!	factorial
$\hat{a}$	estimate of a

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CHAPTER 1

A REVIEW OF THE DEVELOPMENT OF ARTIFICIAL LEGS AND RELEVANT TECHNIQUES OF LOAD ANALYSIS

CHAPTER I

1.1 Introduction

The necessity of amputation in certain cases was realised by Hippocrates. "When gangrene sets in a fracture, the soft parts separate quickly; as for the bones, they become detached at the limit of their exposure; but much more slowly. It is necessary to remove whatever dies first below the lesion from the healthy parts, avoiding pain as far as possible, for patients die from fat embolism." (Littré (1844)). The replacement of the lost part by a prosthesis followed naturally from this realisation.

Although the majority of the development of artificial legs has occurred in this century, throughout history bursts of progress followed the major wars, particularly the Napoleonic Wars and the American Civil War.

1.2 Artificial Leg Development Prior to 1900

The actural date of the first prosthesis is not known but mention is made in writing for the first time by the Greek historian Herodotus at the beginning of the fifth century B.C. He tells of a native seer of Ellis named Hegesistratus who was thrown into prison and condemned to death by the Spartans. He was chained by his foot and in order to escape, he cut his foot off and made his way thirty miles to Tegea, where he provided himself with a wooden leg after the wound had healed. It is certain that he was present at the battle of Plataea in B.C. 479. Eventually he was recaptured by the Spartans and put to death at Zacynthus. (Herodotus IX.37).

In "The Birds" by Aristophanes, produced in 414 B.C., Pithetaenis refers to Kinesias, a player on the cithara, as "The famous lime tree man", (G. Murray (1950)). This is almost certainly a reference to a wooden leg worn by Kinesias.

Popp (1939) reports: "A picture of an Italian urn showed a prosthesis which consisted of a simple wooden capsule. The same type of prosthesis is depicted on an old Peruvian vessel. An artificial leg of remarkable workmanship was found in a grave dating back to 300 B.C. It was essentially a prosthesis with side bars made of thin bronze which were fastened with bronze nails to a wooden base."

Another leg of the same period is referred to by Haddan (1940) "It was unearthed in an ancient tomb in what was the battle-torn city of Capua in Northern Italy. It dates back to the Sammite Wars (300 B.C.) when the highway was being built and the foundation laid for the mighty Roman Empire". The leg was kept in the Royal College of Surgeons, London, but it was destroyed in the second World War.

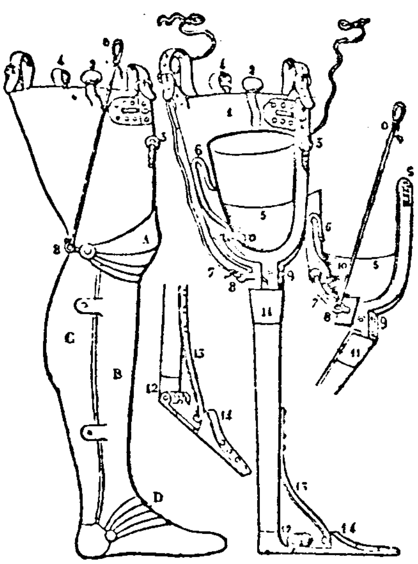
The only type of appliance used up to the 16th century was the peg leg. Various examples are clearly shown in the frescoes in the cemetry at Pisa by Oreagna and at Penni in the Vatican. References are also made to pylons and artificial limbs in ancient Telmud (Putti(1930)). The 'Acta Sanctorium' and other medieval chronicles mention wooden legs and artificial supports for those who have lost an extremity (Ficarra (1948)).

Generally these legs consisted of a pylon supported below a pad, corset or socket, which in some way was attached to the stump or suspended from the shoulders. This device provided a column which, together with the sound leg allowed the amputee to stand erect. The person could also walk although, as there was no knee joint, the gait was far from normal. In the swing phase, it was necessary for the amputee to lift the hip of the amputated side to allow the peg leg to swing through without hitting the ground, and in the stance phase he had to vault over the pylon. Although the peg leg is simple, strong, inexpensive and serviceable, the gait is asymmetric and unnatural, the amputee's performance in walking is inefficient, and his physical appearance is far from cosmetic. In the seated position, the leg is positively ugly.

The only real thought towards cosmesis in artificial legs before the so-called "conventional" leg, was attempted by armourers to cover up a missing leg of a mounted knight. It is very doubtful if these knights were able to walk on these legs as they were very heavy and clumsy. They were probably supplied with two legs, one the cosmetic leg designed purely to hide the mutilation and the other a peg leg for walking. There is an example of a purely cosmetic lower limb prosthesis designed to cover up such a mutilation for a knight on horseback in the Stibbert Museum. The knee of this leg is fixed for riding.

Paré (1510-1590) was not only a great surgeon, but developed numerous prosthetic devices including some very ingenious artificial legs. He is thought to be the first man to design a truly jointed leg prosthesis (Fig.1.1). Although the leg incorporates a simi-automatic knee lock and an articulated foot, it appears to be far too heavy to be of any practical use. Bick (1948) states that it was only used for a short time. A full description of the leg is given by Little (1922) and referring

AN ARTIFICIAL LEG OF PARÉ



(Little (1922))

Fig. 1.1

to Fig. I.I. some of the more interesting items are:-

"1) The thigh socket, with the screws and the holes of the said screws, to enlarge or tighten on the thigh (stump) which will be inside.

2) The pummel on which to place and rest the hand and to turn oneself.

3) The little ring which is in front of the thigh, to straighten and direct the limb where one wishes.

4) The two front buckles and the one behind to hold and attach the limb to the body of the doublet.

5) The hollow below, within which the thigh (stump) is placed as far as two fingers' breadth from the end, serving also to produce the beauty and shape of the leg.

6) The spring which moves the catch which closes the leg (locks the knee).

9) The hinge to allow the leg to move (at the knee), placed in front of the knee."

A, B, C and D are cosmetic coverings for the leg probably made from thin metal such as was still used for armour in 1564. The leg was constructed for Paré by a French locksmith named Lorraine.

Gibson (1955) is of the opinion that Pare's "Wooden leg made for poor Men" was probably the most useful artificial leg of that period.

An artificial leg for above-knee amputations was constructed by Gavin Wilson near the end of the eighteenth century which had a flexible knee joint and was suspended by a strap over the opposite shoulder. This may have been the first "conventional" artificial leg for above-knee amputees since no precise date can be found for its introduction.

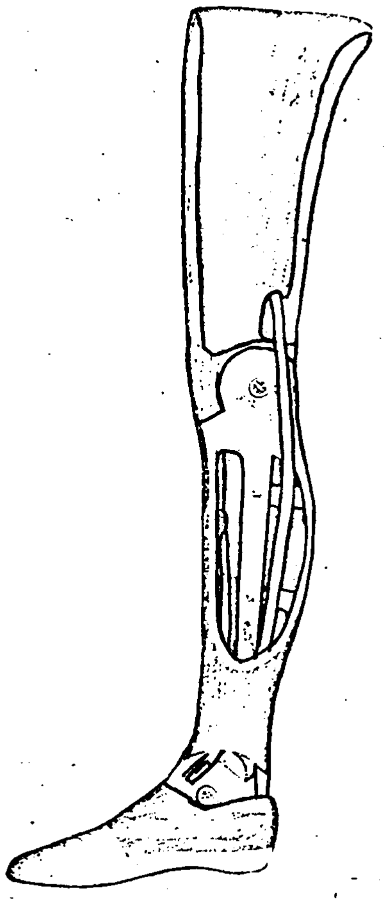
In the conventional leg, some cosmesis was aimed at in that the leg was made to look like the normal leg. The knee was hinged and could be flexed, although in the early varieties, a knee lock was provided to ensure stability in standing. A foot was attached to the shank with either a rigid or jointed ankle.

Although there are obvious advantages in this type of leg over the simple peg leg, other problems are encountered. The ability to flex the knee enables the amputee to sit, kneel and perform, in a more natural manner, other activities requiring a hinged knee joint. If a knee lock is not provided, a better gait is achieved than with a peg leg, since the leg can be brought through in the swing phase without excessive hip elevation. However, instability at the knee and ankle joints is inherent, also problems such as excessive heel rise, and shock at full knee extension just before heel strike, are introduced. These problems are still being investigated to the present day.

In 1800 a leg was patented by James Potts of London. It was fitted to the Marquis of Anglesey at the Battle of Waterloo and from then onwards it was known as the Anglesey leg. Basically it consisted of two hollow wooden segments with a steel knee joint and a wooden ankle joint. Cords from the knee controlled the ankle joint (Fig. 1.2).

Little (1922) refers to the use of the "clapper leg"; "so-called because locomotion was apt to be accompanied by a clapping sound, due to the wooden stops in front of and behind the ankle". This predated the

THE ANGLESEY LEG



(Little (1922))

Fig. 1.2

Anglesey leg but Little does not state when it was first used.

A ball joint in the ankle was a prominent feature of a complicated leg prosthesis devised by Johann George von Heine in 1810. The leg was held on by a corset around the trunk for high amputations.

According to Herrmann (1865), the first true knee mechanism was invented by Pierre Ballif zu Berlin between 1813 and 1815. As can be seen from Fig. 1.3 not only did it incorporate a swing phase control in the form of six vertical spiral springs fastened between the shank and thigh, but also an automatic knee lock. The action of this lock can be understood from the following description.

A cord is attached to a projection from the artificial toes so that dorsiflection of the toes tightens the cord. It then passes over a system of pulleys to the mechanism itself. This consists of a hinged steel bar, A, which, when fully extended, forms a rigid link between the shank and thigh. Extension is assisted by a curved spring, B. The bar is flexed by means of the cord previously mentioned. Pretensioning of the cord is achieved by passing it over a rachet, C, on the bar, and back into the shank to the uppermost pulley, D, and finally outside through the thigh front, where it can be pulled. To sit down the amputee merely pulls the same cord a little harder and the bar is flexed, allowing the knee to bend. When he stands again, he simply readjusts the tension.

The Anglesey leg was modified by William Selpho, one of Potts' workmen, in 1839 by inserting a rubber plate into the ankle joint to reduce jarring, and adding a rubber sole to prevent slipping and provide more elasticity for the foot part.

THE LEG OF BALLIF.

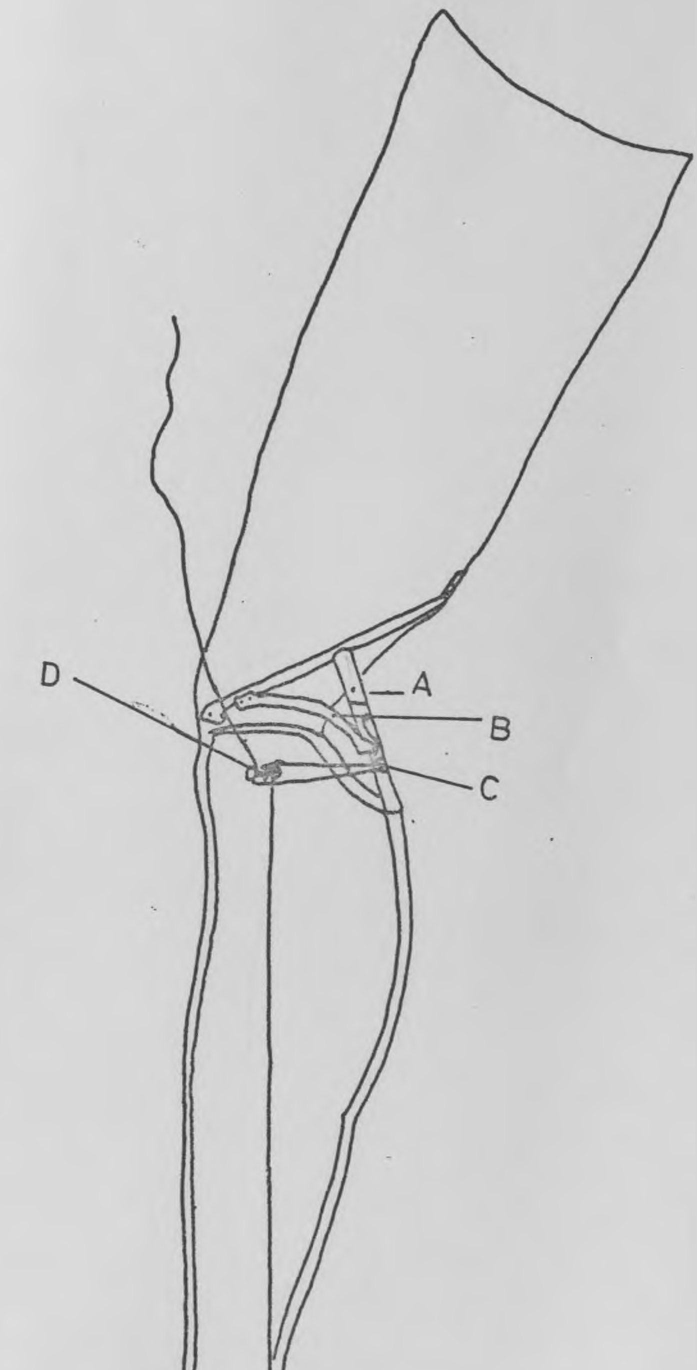


fig. 1.3

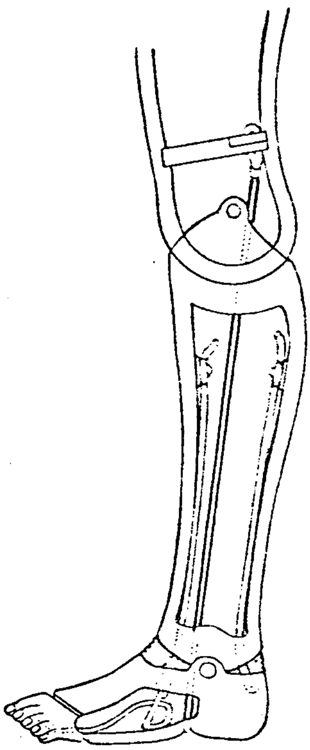
A leg designed by Martin and Charrière had the knee centre placed posteriorly so that stability was achieved during the stance phase for the first time by alignment.

Benjamin F. Palmer patented an improved version of Selpho's leg is 1846 (Fig. 1.4). This was the first patent on artificial legs issued by the United States Government. As can be seen from the diagram, Palmer's leg still used catgut cords and an anterior spring to regulate the ankle joint.

A further modification of this same device was made by Palmer in 1850, and was put into a leg made by Frees. As can be seen from Fig. 1.5 the tendons or cords were arranged in a different manner to the Plamer leg. Two cords passed behind the knee joint and one in front. All three cords were secured to a wooden cross piece. The posterior cord was used to limit knee extension. Ankle movement was achieved by the combined effect of the other two cords and a rubber bumper to the rear of the ankle. The centre section of the anterior cord was indiarubber, and the combined action of the mechanism can be understood from the diagram. One disadvantage was that the knee tended to extend when the amputee was sitting.

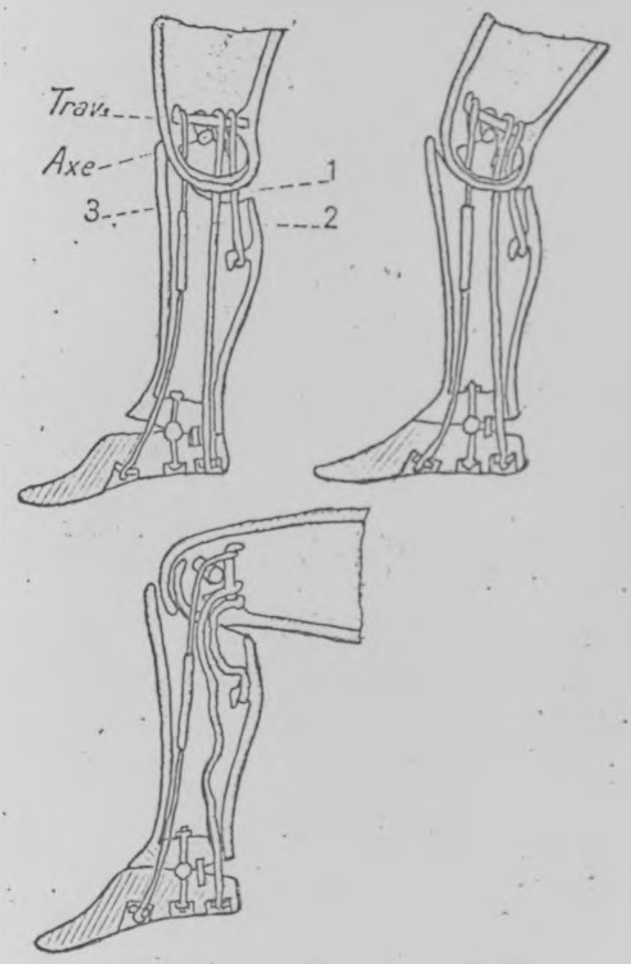
One important invention of this period was Plaster of Paris by Anthony Mathijson in 1852. He thought that if it was accurately moulded, it could take load. Plaster of Paris was used by Mathijson in the construction of his peg legs.

In 1858, Douglas Bly, M.D. invented a leg with a ball and socket ankle. Although this leg fell into disuse, it shows advanced thinking on the part of its inventor. It was intended to allow movements similar to those of the natural foot and ankle when used on uneven ground.



(Murphy (1960))

Fig. 1.4



(Broca and Duroquet (1918))

Fig. 1.5

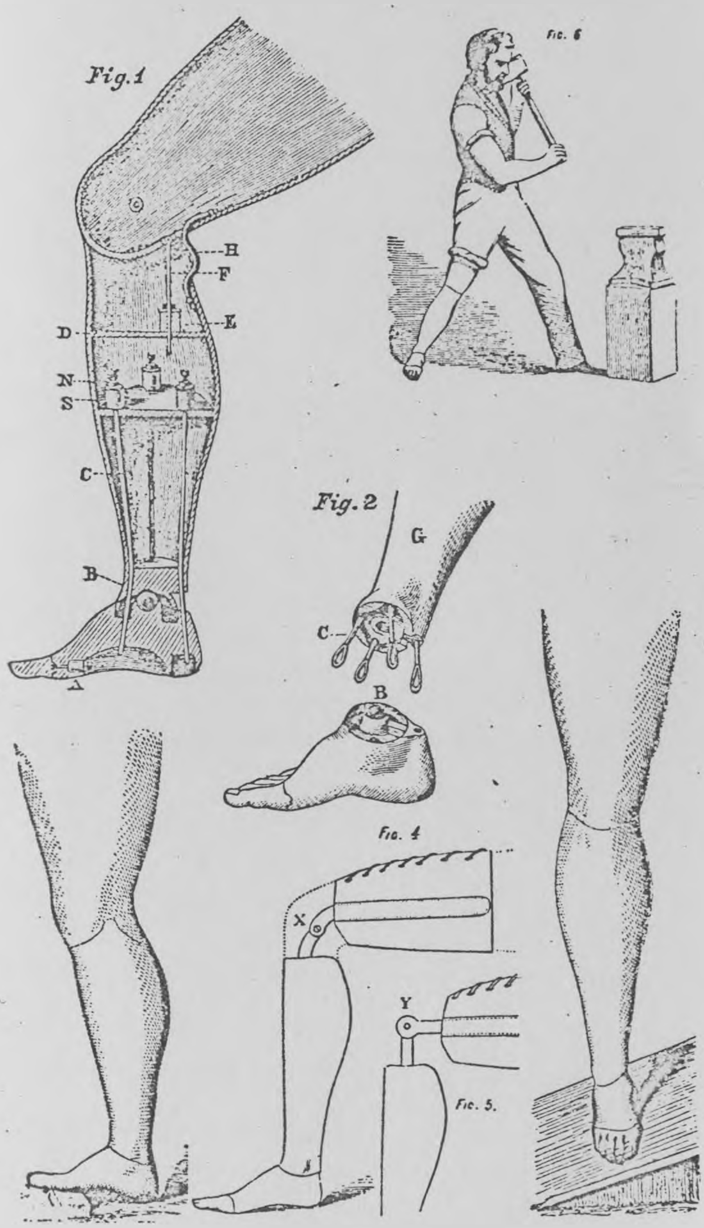
The ivory or glass ball, B, rested on a vulcanised rubber bed (Fig. I.6). Five cords or tendons, C, could be adjusted by turning the nuts, N, which were compressed when the tendons were tense. At the knee joint there was an extension stop in the form of a strap, H, but even more interesting is the swing phase control consisting of parts E and F. It is apparent that the rod F slid in the bush E during the swing phase. This bush was made of rubber with an adjusting nut on the top, and the friction could be altered by compressing the rubber as the nut was turned.

The advantage of this type of ankle joint is that the amputee can accommodate the sole of the artificial foot to various inclines of the ground. This is especially important when walking along the side of a hill, when the normal foot would be inverted or everted. If no adaptation is made, undue strain is thrown on the stump. This advantage was in practice found to be outbalanced by the disadvantage of insecurity.

The Marks leg, 1860, had a fixed foot set "in equinus at an ankle of 25 degrees to 30 degrees so that the heel is two or three centimeters from the ground (the usual height of a boot)". (Broca and Duroquet(1918)). The foot consisted of a wooden keel stretching from the bottom of the shank to a point corresponding to the middle of the metatarsus and half the thickness of the foot. The rest of the foot was formed of indiarubber stuck on to the instep piece. The whole was enclosed in a sheath of leather.

The action of the knee joint in the Marks leg is described by Broca and Duroquet, and is reproduced here as Fig. I.7. It contains an extension stop and an extending mechanism to ensure that the leg is fully extended at heel strike.

THE CONSTRUCTION OF BLY'S LEG



(Bigg (1885))

Fig. 1.6

The Marks knee.

FIGS. 52 and 53.—O, knee bolt. T, cross piece of wood, situated

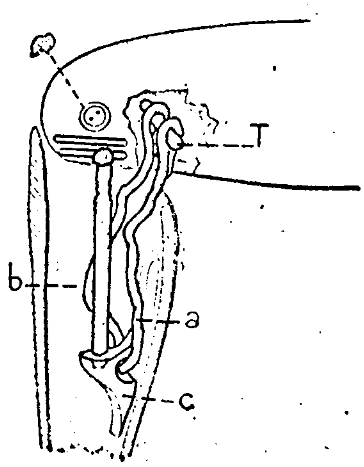


FIG. 52.

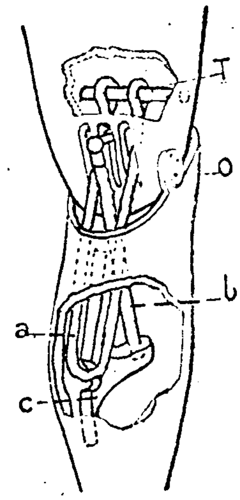
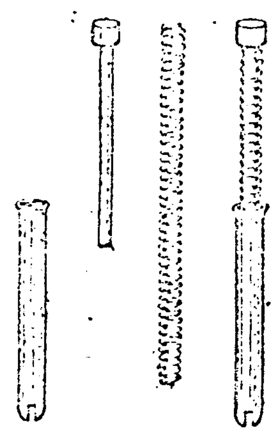


FIG. 53.

in the extended position above the knee bolt, in the flexed position behind it. C, bracket fixed halfway up the interior of the calf.

A U-shaped cord *a* passes through a hole in the bracket C and is attached at each end to the cross piece T; it limits extension. The two ends of the word enter the thigh piece by two apertures in the posterior surface, between which is fixed a metal ball which projects 2 cms. The extending spring is the rod *b* which is fixed to this ball and to a socket in the upper surface of the bracket. Figs. 54 to 57 show the parts of this spring: a tube, a spiral spring, and a rod with cup-shaped head. When the spring is in the tube and the rod in the spring (Fig. 57), it will be seen that pressure upon the head of the rod increases the tension of the spring.



FIGS. 54 to 57.

(Broca and Duroquet (1918))

Fig. 1.7

Although rubber bumpers had been used before 1861, it was in this year that Hangar developed the "cordless" ankle. The present day articulated ankle is virtually the same as the one designed by Hangar. He also introduced the wood socket and popularised the idea.

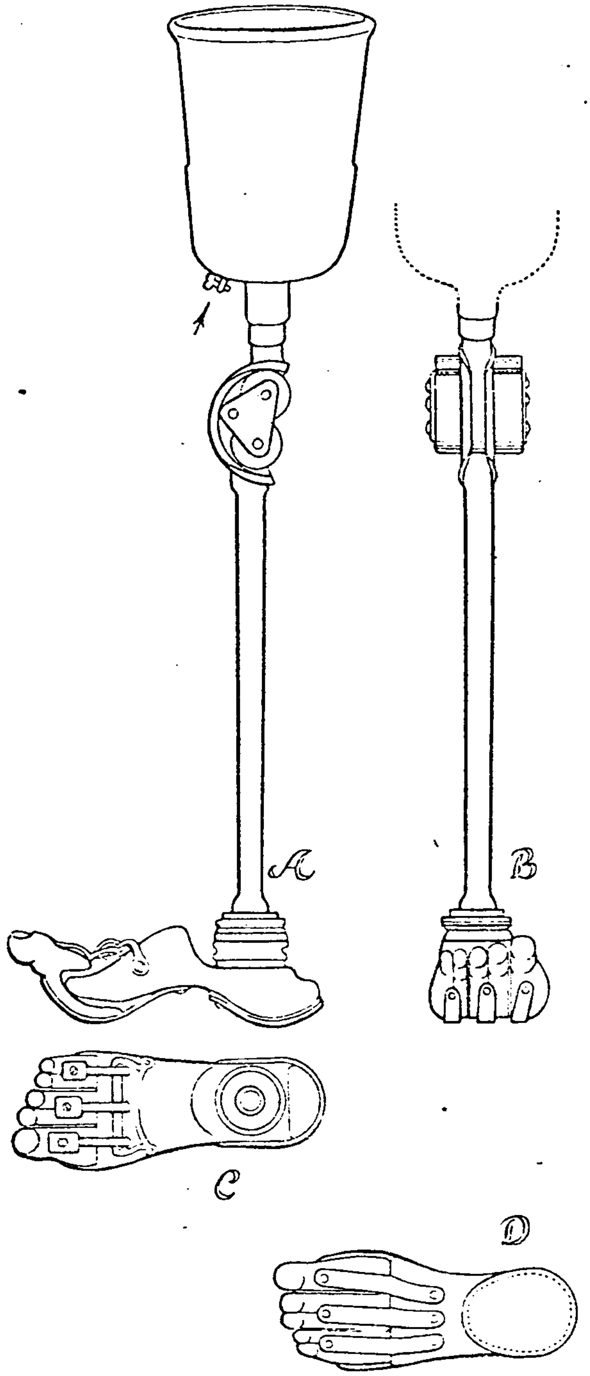
The next major invention was that of the suction socket by Dubois D. Parmelee in 1863. This type of socket had a long and variable development until it reached its present form. In his leg (Fig. 1.8) Parmelee incorporated an unusual knee unit consisting of two rollers and a concave piece of steel fitting over them. The rollers were fixed to the thigh and shank respectively and any rotation of the knee joint resulted in a rolling rather than a slipping of the two surfaces involved. It was claimed that friction was decreased.

Although Herrmann (1865) was the first to use aluminium in the place of steel he preferred to use wood for the rest of the prosthesis, since it was light and durable.

Some kind of pelvic band and hip joint for the above-knee limbs was first described by Hoffa (1891).

The last development of the nineteenth century was due to an Italian, Vanghetti. In 1898 he experimented on fowls with a technique called cineplasty. The technique is described by Lewis (1942) as the "muscles in the stump made use of, independent of the general movements of the stump itself". The first operation to be carried out on a human patient using this method was by Ceci in 1900.

THE PARMELEE LEG



(Murphy (1960))

Fig. 1.8

1.3 Above Knee Artificial Leg Development Since 1900

Probably the best way to follow the development of A/K artificial limbs in this century is to consider various aspects of prosthetic design and follow their progress through to current practice. With this in mind, it is intended to consider two components of the artificial leg relevant to the research described in this work. These are the socket and the mechanisms used to assist knee stability. Although little consideration is given to the effect of new materials, in particular plastics, reference will be made to their influence on the design of these two parts.

1.3.1 Socket development The socket forms the connection between the amputee and his prosthesis and, therefore, its design is of utmost importance in any assessment of amputee performance.

One early attempt to prevent excessive relative movement between femur and socket by suitable socket design is due to Hanausek (1917). His plaster socket had three approximately equal circumferential zones. The proximal and distal zones are bound tightly with plaster of Paris bandage, and form the main medium for force transmission. A plaster cast is also made of the end of the stump and attached to the distal zone. "The middle zone has the aim only of keeping together the proximal and distal plaster rings."

The major development in socket design has probably been that of the suction socket in its various forms.

Mechanical suspensions have been improved considerably, from the original shoulder harnesses to the modern pelvic band. Hoffa's original design was adapted by F. G. Ernst in 1915 and it is this version which is used today. The suction socket eliminates the need for this type of suspension.

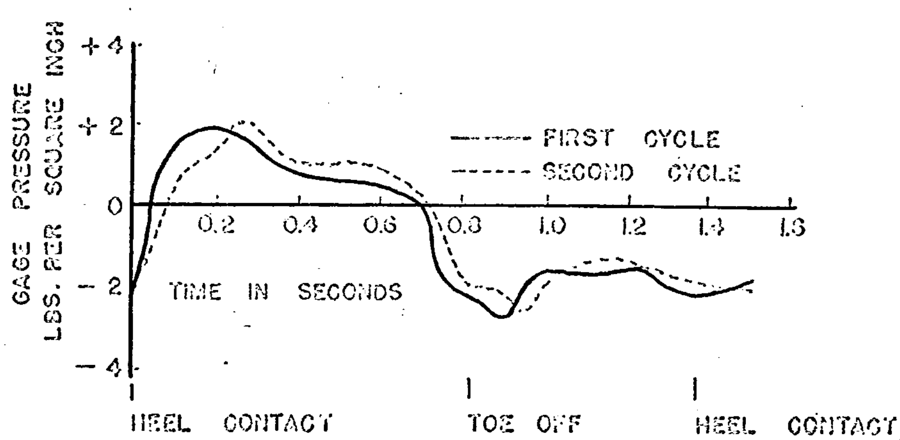
Basically the idea is that a slight negative pressure is created between the stump and socket, which is sufficient to hold the prosthesis on to the stump during the swing phase of walking (Fig. 1.9).

The remaining musculature controls all movements of the prosthesis and no sock is worn on the stump. The socket is made to fit snugly in order to maintain an air seal, but should not restrict the muscles or circulation of the stump. The fit is assisted by contraction of the stump muscles in the swing phase. When the socket is properly fitted and has no leaks, there is no relative motion between the skin and the side wall of the socket. The femur does, however, move slightly within the soft tissues of the stump.

After Parmelee's patent in 1863, a patent was issued to George C. Beacock and Terence Sparham of Brockville, Ontario in 1885 for a similar type of socket. There was no fundamental difference between the two. However, a development due to Justin Kay Toles of Stockton, California in 1911, was the addition of a rubber tube and bag lining. This could be inflated to assist in holding the socket on. In 1926 Ernest Walter Underwood of Birmingham, England varied Parmelee's design by making smooth helical grooves in the socket. These he claimed ventilated the stump as well as holding the socket in place.

Some thirty cases were fitted with suction sockets at Roehampton, England after World War I using a metal socket with a helical groove. The groove was a little over one turn round the circumference. Use of the socket was discontinued for a number of years but was revived round the end of the 1940's.

PRESSURE VARIATION IN A SUCTION SOCKET



(Daniel (1950))

Fig. 1.9

Pfau of Berlin says that the socket was known in Germany in the 1920's but was popularised by Österle of Ulm in the early thirties. Felix, a surgeon of Düsseldorf, reported on above knee suction sockets in 1941. The construction of a satisfactory valve in 1932 boosted its use in Germany, although Felix says that suction sockets were used from World War I onwards.

After a visit in 1946 of a Commission on amputations and prostheses from the U.S.A. to Europe to observe practices there, a programme was set up to determine the possibilities and limitations of the socket. This eventually lead to wide spread use of the suction socket in the U.S.A. The breadth of its use can be appreciated when it is realised that in 1952, 90% of the U.S.A.F. above knee amputees were fitted with a suction socket (Canty (1952)).

Putting the socket on is quite simple if a stump sock is used to ease the stump in. To ensure that the stump is correctly seated, the weight is put on the leg before fitting the valve.

The most popular suction socket at the moment is probably the "quadrilateral" socket, designed at the University of California on biomechanical principles. This socket follows the general requirements outlined above and a detailed description is given by Radcliffe (1955).

A recent development in Britain has been the "total surface bearing" socket. This is simple to make but takes no account of the forces acting in normal activity. However, it must be pointed out that initial trials indicate that patients seem to be quite comfortable in the socket and the gait achieved is good.

Apart from the obvious advantage of the suction socket from a comfort point of view, in that there is no belt or harness, numerous advantages are claimed. Details of advantages, disadvantages and contra-indications are given by Daniel (1950), Gillis (1957), Compere & Thompson (1957), Bancroft & Marble (1957) and Kerr & Brunnstrom (1956).

In a report by Thorndike and Eberhardt (1950) on suction sockets quoted by Furman (1962), statistics of 606 patients initially fitted with suction sockets are presented. Seventythree percent still wore suction sockets, fourteen percent wore both suction and normal sockets and thirteen percent were failures.

Until quite recently sockets were made from measurements of the stump taken by the limb surgeon. But as Broca & Duroquet (1918) pointed out "good results cannot be obtained, if, as certain people have tried, linear measurements are sent to a workshop whence an aparatus is forthwith despatched to a patient whom the maker has never seen". They also indicated that a plaster cast could be used to get an exact replica of the stump. Unfortunately these remarks were not heeded and the archaic system remained in Britain until the 1960's.

Generally in Britain the whole of the leg, including the socket, is made of aluminium. In the U.S.A., the quadrilateral socket was originally made of wood but now can be made of either laminated polyester or epoxy resin. Recently these plastic sockets have been used experimentally in Britain at one or two research centres. The technique of lamination is described fully by Foort (1963).

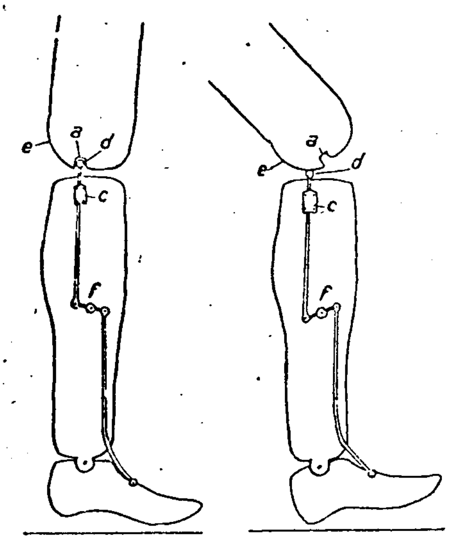
1.3.2 Mechanisms to aid the stability of the artificial knee The basic problem of instability is present with any leg which has a knee joint. With geriatrics the solution is often to have a semi-automatic knee lock. This results in a locked knee for walking and normal activity with a manual release to allow sitting down. The only concern in this thesis is automatic methods of achieving stability at the knee joint. These can be grouped into three main categories. Those which depend on positive locking of the knee, those which have a brake at the knee actuated by some means and those which are voluntarily controlled in some way.

Devices which depend on positive locking of the knee as soon as full extension is reached just before or at heel strike are generally unlocked just before toe off by a certain degree of dorsiflexion of the foot. Ballif's system for obtaining such an action has already been described in the first section of this chapter. Other legs using this same principle have been designed by Lange, Fleming and Hildebrand (1918).

Hildebrand thought it much more important to have complete fixation of the knee joint in full extension rather than a break operating at all angles, (Fig.1.10). As the leg swings into full extension, the rod is pushed into a slot on the thigh at the knee joint and the shank and thigh are rigidly locked together. Dorsiflexion of the foot pulls the rod out of the slot and the knee is free to flex ready for the swing phase.

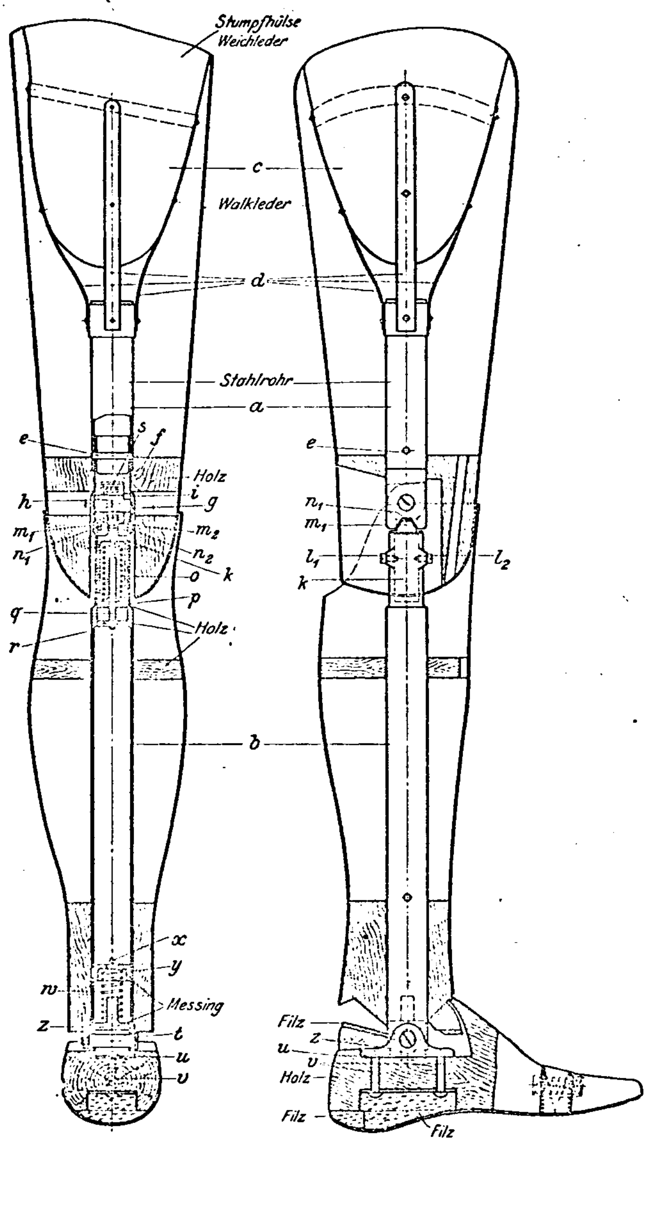
Attempts have also been made to lock the knee by body weight. Schaefer (1917) designed the load actuated lock shown in Fig.1.11. When a load is applied to the shank in the extended position, the teeth m_1 and m_2 engage

THE HILDEBRAND KNEE LOCK



(Hildebrand (1918))

Fig. 1.10



(Schlessinger (1919))

Fig. 1.11

the racks n_1 and n_2 thus locking the knee. When the load is removed the central spring o presses the two sets of teeth apart and the knee is allowed to flex.

Schlessinger (1919) describes a mechanism based on this principle but using heel movement to actuate the lock. It was designed by Baumgartel and is shown in Fig.1.12. This device is very cheap and, since the rod is hidden by the shoe is cosmetically acceptable.

Other mechanisms using the principle of body weight to actuate the knee lock have been designed by Huber, Bingler and Erlacher.

Mosberg (1918) describes a knee in which flexion is limited by a stop to a reasonable level for walking, which can be released for sitting down.

Variations on these various devices have been attempted since 1920 but no really satisfactory locking mechanism has been devised yet.

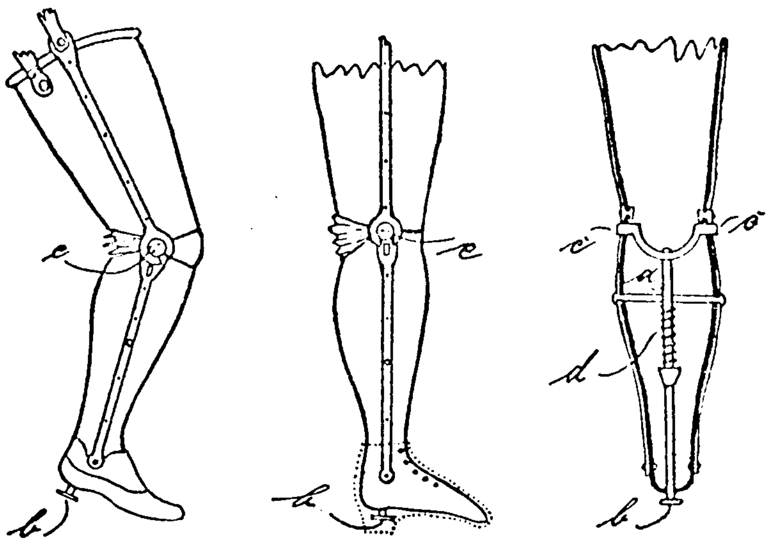
A knee brake can be actuated by body weight or by movement of the foot. The principle of using band or block brakes in artificial knees is attributed to Engels and his modified designs of 1916 are shown in Fig.1.13 and 1.14. Both these mechanism utilise body weight to force the braking surfaces together.

The Fischer leg is typical of body weight actuated devices, and is described by Schlessinger (1919) in his excellent review.

Robentisch (1919) invented a knee brake which uses the principle of a wedge being pressed into a V channel by load on the leg in much the same way as the modern Otto Bock safety knee.

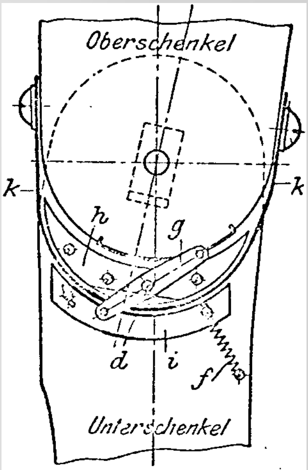
Movement of the foot is used to actuate the Rosset and Soukup knees, both of which are featured in Schlessinger (1919).

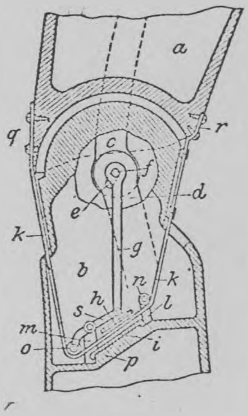
Fischer (1918) also described a knee which brakes with dorsiflexion of the foot, and this device is shown in Fig.1.15.



(Blencke (1917))

Fig. 1.12

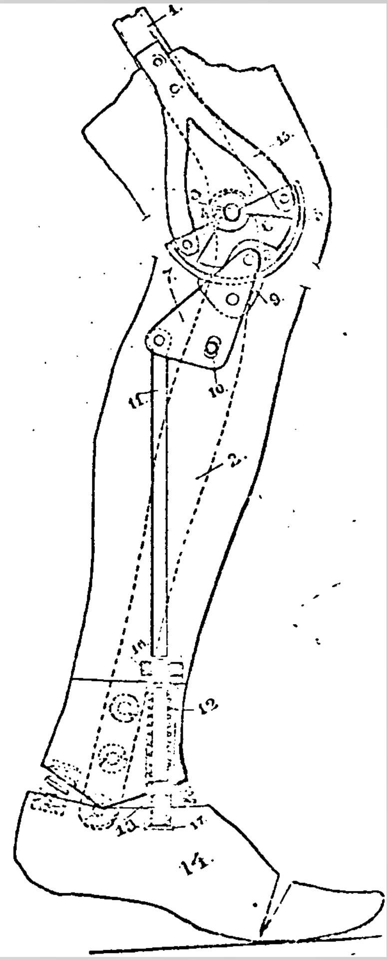




(Schlessinger (1919))

Fig. 1. 13 and Fig. 1. 14

THE FISCHER LEG



Fischer (1918)

Fig. 1.15

Schede (1917) incorporates a brake at the knee again actuated by dorsiflexion of the foot, in a steel leg.

The major development in this type of device since the early German work has been the use of hydraulics in artificial legs, although the concept was originated by Windler in 1918.

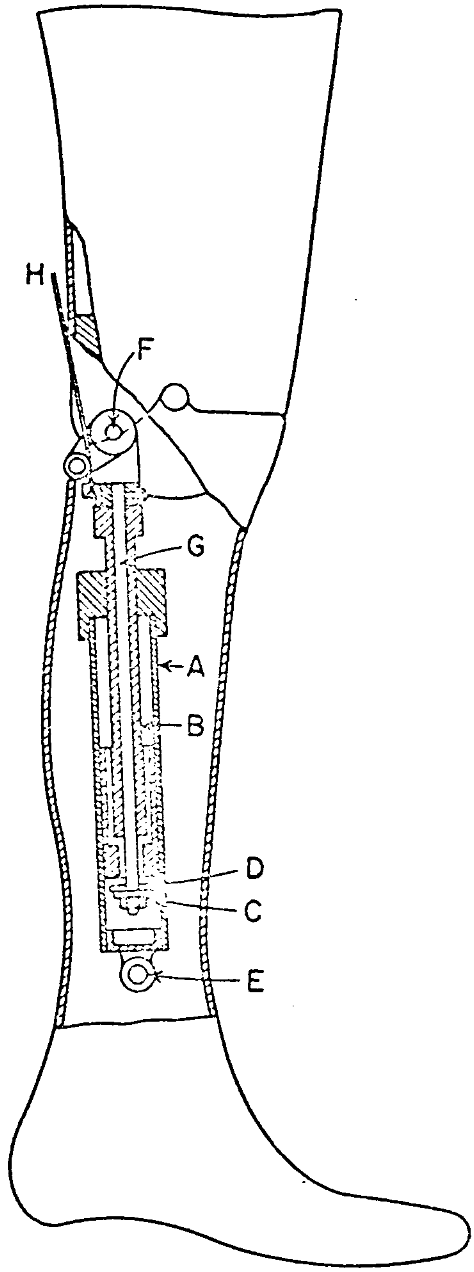
One of the early hydraulic devices was due to Filippi in 1942 and is shown in Fig. 1.16, together with an explanation of its action. At the present time numerous devices are available using a hydraulic stance phase control. Typical of these are the Stewarts-Vickers leg, the Regnell leg, and the Hensche-Mauch leg.

The Otto Bock safety knee and the Blatchford stabilised knee are two of the mechanisms which depend on body weight operating a break at the knee joint. Both these devices are investigated in this research and are described in detail, together with their characteristics under various load conditions by MacGregor (1968).

The third category of devices to consider has been losely termed "voluntarily controlled" in that although they do assist knee stability, they are dependant on some active movement of the amputee to initiate a response. True voluntary control, as well as systems giving mechanical advantage to the amputee, will be discussed in this category.

Linkage systems have been investigated by the early German workers, in fact Schede (1918) mentions that both Hermann and Sauerbruch have given some considerable thought to the idea in the nineteenth century. This type of device depends on the fact that in normal walking hip extension, knee extension and dorsiflexion of the foot occur together. The first linkage device encountered in the literature is due to Jaks (1917).

THE FILIPPI HYDRAULIC LEG



Hydraulic knee lock of Filippi. A, Cylinder; B, piston rod bearing on integral piston; C, valve disc loosely supported by the valve stem; D, valve seat on the bottom of the piston below bypass ports; E, pivot in the shank; F, pivot in the thigh; G, valve stem telescopically movable within the piston rod B; H, cord to waist belt. Tension on the cord (as in stumbling forward, or in flexing the hip at heel contact) lifts the valve stem, thus locking leg.

(Murphy (1960))

Fig. 1. 16

This leg is depicted diagramatically in Fig.1.17 and the relative movements possible are clearly shown. Jaks claims that an amputee can stand on this leg with a slightly bent knee as long as "the forward knee joint pin does not overtake the forward foot joint pin". One amputee was able to carry a load of one hundredweight on his amputated side.

The most advance design of this type is that designed by Schede (1918) and shown in Fig.1.18. At heel strike, H_2 contacts the stop A and the four bar linkage system H H_2 K_1 K allows control of the knee joint K by means of a pelvic socket (this is an idea of Schede to distribute the load over the pelvis). As the body moves over the straight leg H_2 is moved away from the stop and rotates about H_1 and can do so until the link H_2 K_1 contacts the stop A if excessive knee flexion occurs. The four bar chain is therefore released and the shank is free to swing through.

Biesalski (1919) features legs by Lange and Zuelzer which depend on the movement of the rump to tighten a cable which passes in front of the knee joint. With the Lange leg, body extension activates the knee while in the Zuelzer leg it is body flexion which is used to extend the knee. To facilitate sitting down part of the cable is elastic.

Hanausek (1917) describes a similar system where the strap passes over the shoulder of the amputated side and is secured to the rear of the socket.

An artificial leg which depended on the movement of the normal leg for its action was designed by Karsch, Schlessinger (1919).

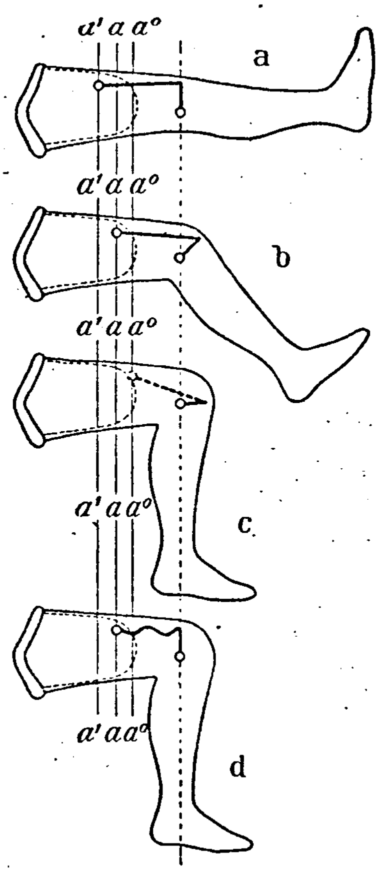
One method of achieving true voluntary control of the leg is by harnessing the power in the remaining stump muscles, for example the quadriceps. The work along these lines emanated from Vanghetti's original research on chickens. Although he considered its active use in prosthetics, the real development was due to Sauerbruch (1915). Initially Sauerbruch worked with Ruge, Felix and Stadler on a voluntarily controlled hand, but extension of his work to arms and legs is described in Sauerbruch (1916). He claimed that the lifting power source available is 13 to 15 Kg. over a distance of 6 to 10 cm. Work on a power source for legs using the same technique has been attempted by Perthes, Steinthal, Wullstein and more successfully by Blencke (1916).

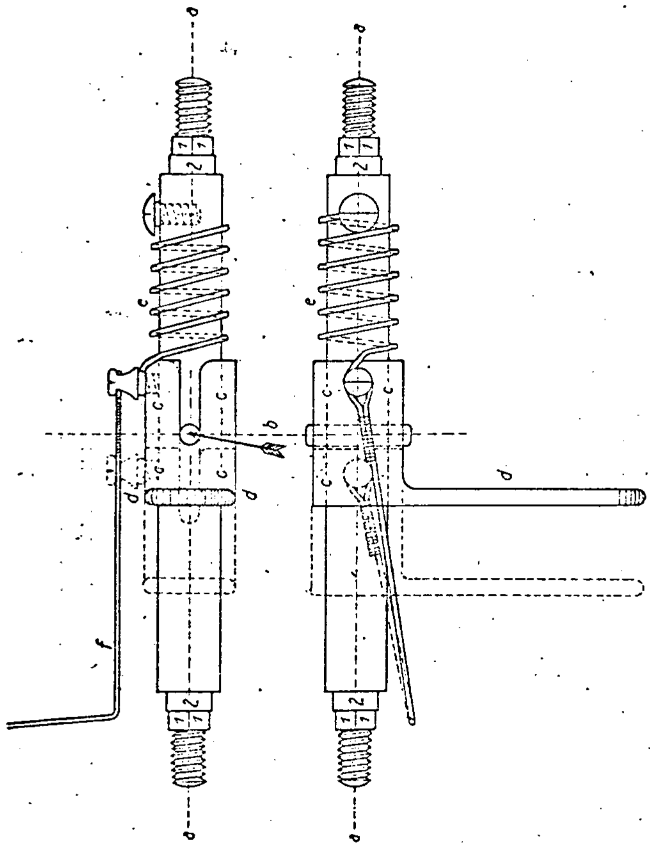
Referring to Fig. I.19, a' a is the muscle path length required for walking, while for sitting the length is increased to a^{o}. The knee release allows the muscle to maintain a resting length at "a" while sitting. Fig. I.20 shows the construction of this release. The tension from the muscle tunnel is applied to the lever d to rotate the shank about the knee axis aa. By pulling the wire f the collar c is slid along the shaft and the drive by the pin b is uncoupled so that the amputee can sit. The spring e allows automatic recoupling when the amputee stands up. In normal walking the path length is about 4 cm. and a load of 25 to 30 lb. is usual. To lift the shank up horizontally when the thigh is horizontal requires a path length of 7 cm. and a pull of about 20 lb.

Other systems of release have been designed by Schede and Luer (see Biesalski (1919)).

Work by Böhm (1918) on the use of muscles in the upper arm stump for voluntary movement of an artificial arm led Blumenthal (1919) to adopt

THE CINEPLASTIC KNEE OF BLENCKE





(Blencke (1916))

Fig. 1.19 and Fig. 1.20

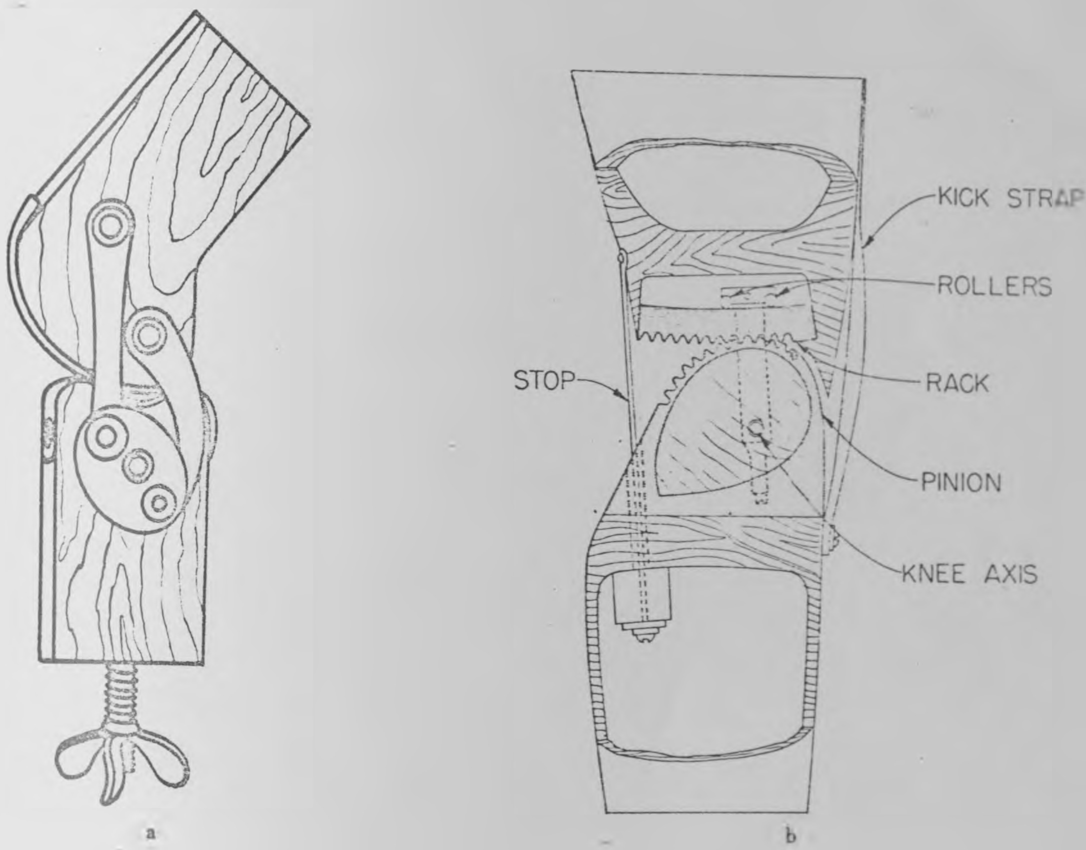
a similar system for an A/K prosthesis. With this technique a bulge is formed under the skin with the remaining stump muscle. A strap is fastened around the bulge and the power available from movement of this strap is used to release a brake at the knee joint. Henchke and Mauch (1948) describe a mechanical-hydraulic device for obtaining knee stability which is voluntarily controlled by the abdominal muscles.

Recent research in the U.S.A. by Mauch (1967), into the possibility of using voluntary control of the resistance at the knee indicates that muscle hardness may be the best source of signal to operate such a device. EMG and ultrasonics are also possible methods and are being investigated.

A further development this century is the polycentric type of knee in which the instantaneous centre of the knee axis is changed by a linkage system or condylar surface. The idea originated when Schede and Haberman attempted to copy the normal knee, Fig. 1.21. Three of the devices investigated in the research presented here are of this type. They are the Otto Bock Greissinger knee, the Lammers knee and the U.C.B. polycentric knee.

1.3.3 A brief critical review In this century there have been two major areas of prosthetic development. One a concentrated effort between 1915 and 1920 in Germany, and the other, a steady progress from about 1955 up to the present day mainly in the U.S.A. but also to a lesser extent in Europe.

Between these two, very little has contributed at all to the well-being of the amputee. The sad thing is that a number of lessons learnt and experience gained by the early German workers has been completely neglected.



(Murphy (1960))

Fig. 1.21

The principles laid down by such men as Schede and Gocht are as valid today as they were when first formulated.

With regard to the knee devices reviewed none of these is able to provide characteristics which are exactly the same as the normal knee.

It would seem that some voluntary control over the ability of the knee to sustain a moment is the best solution. The difficulty is to obtain information from the body which can be transformed into a power output and amplified to operate the knee device. Until sufficient research has been carried out in this field to warrant consideration of such a device for patient use, other mechanisms have to be supplied.

The main disadvantage of the locking type of knee is that it usually only operates in the fully extended knee position. Attempts have been made to provide some ratchet arrangement but this complicates an otherwise very simple system and produces too much noise in operation. Hydraulic devices may provide the solution but they are expensive, complicated and tend to leak in use.

A knee lock is advantageous to an amputee who has to stand for long periods at his work or who is likely to be walking over uneven ground, e.g. a ploughed field. If the knee is locked automatically at full knee extension, the amputee can choose the place and time of heel strike but the release of the lock by dorsiflexion of the foot occurs too early if the amputee goes up a steep incline.

A lock which is load actuated does not allow choice of time and place for heel strike and further, since only the component of load acting along the shank activates the device, the effect of the shear component can cause the knee to buckle before the lock is operated. One advantage of a load actuated mechanism is that its action is independent of the type of surface over which the amputee is walking.

The brake type of mechanism allows some flexion of the knee without the knee buckling. With load actuated brakes, the more knee flexion, the less effective is the brake. If dorsiflexion of the foot operates the brake, the braking effect increases with knee flexion, but dorsiflexion of the foot itself is prevented to a large extent by the device. Hence the adjustment of such a device is impossible for all conditions. At the time of heel strike, the brake will not operate since the foot is plantarflexing.

A better system would be if the brake was actuated by the direction of rotation at the ankle, i.e. as soon as dorsiflexion was initiated, the knee was braked but when plantarflexion began it became free again.

If the amputee falls with either a braked or locked knee, the results of the fall are much more serious than if there was no device in the knee.

The early attempts at voluntary control by cineplastic tunnels, although very encouraging, were discontinued due to failure of the tunnel itself (through infection) and the inability to develop sufficient power to control the knee advantageously.

The polycentric type of device appears to offer a good compromise between simplicity and effective action. Difficulty has been experienced, however, in achieving adequate cosmesis due to inevitable bulkiness of the knee.

The requirements for an ideal knee mechanism are discussed in more detail in Chapter 6.

1.4 Relevant Load Analysis Techniques

To calculate the moment exerted by a person at the hip joint, it is necessary to know the magnitude of the resultant load and the perpendicular distance from the hip to its line of action.

The system adopted at Strathclyde by Paul (1967) to calculate the hip joint force uses a strain-gauged force plate set in a raised walkpath. As the subject walks over the dynamometer, all components of the floor to foot force actions are displayed on a U.V. recorder. By filming markers at specific points on the leg of the subject as he walks over the force plate, the position of the various limb segments and pelvis can be found. This enables all the external forces acting on the hip joint to be calculated, both inertia and ground to foot forces, and the moment about hip joint can be computed.

If the gait of an amputee is investigated it is possible to incorporate the transducer into the prosthesis. This technique was first described in U.C.B. (1947) and later by Cunningham & Brown (1952). Although this method has the advantage that the amputee is not restricted to walking over a specific part of a walk path, it does introduce an encumbrance of wires to the dynamometer on the amputee.

Using a pylon technique, Cunningham & Brown reported values of force actions in the artificial leg in a A/K amputee as follows:-

Axial thrust of 0 - 250 lbf. (body weight was 190 lbf.)

A/P knee moment of -200 - +300 lbf. in.

A/P ankle moment of -150 - +1200 lbf. in.

Torque of -80 - +100 lbf. in.

No information is available in the literature regarding the hip moment exerted by an amputee in normal activity.

CHAPTER 2

A REVIEW OF THE HIP JOINT ANATOMY RELATED TO PROSTHETICS AND AMPUTATIONS

CHAPTER 2

2.1 Introduction

The power for controlling an artificial limb for an above-knee amputee is derived from the hip joint of the amputation stump. Muscles at the joint produce the majority of the force required to generate the hip moments about the two major axes, that is the mediolateral and the anterior posterior axis. A contribution to this effort is sometimes provided by the transmission of forces directly from the trunk to the socket of the artificial leg. The technique of amputation and the socket design influence the power availability and transmission.

2.2 The Skeletal System

Only two bones are important in above-knee prosthetics, the femur and the lower portion of the pelvis.

The femur (Fig. 2.1 & 2.2) is the largest, longest and strongest bone in the skeleton, and almost cylindrical in cross-section in the greater part of its length. In erect prosture, the bone is not vertical but is inclined medially at its inferior end, so that the knee joint is brought nearer to the gravity line. The main points of interest are connected with the upper extremity (consisting of the head, neck and the greater and lesser trochanter) and the shaft.

The head is spherical and is directed upwards, inwards and slightly forwards. Its surface is smooth, coated with cartilage in the healthy state except for a point just behind and below the centre, where the ligamentum teres is attached.

The neck is a flattened pyramidal process of bone, which connects the head with the shaft. In the adult it forms an angle of about 130° with the shaft, but varies inversely with the size of the pelvis and the person's height. The anterior surface of the neck is perforated by numerous vascular foramina (small holes). The posterior surface is smooth and the posterior part of the capsular ligament of the hip joint is attached to it about half an inch above the posterior intertrochanteric line. The superior border is short and thick, and terminates externally at the greater trochanter. Its surface is perforated by large foramina. The inferior border, long and narrow, curves a little backwards, to terminate at the lesser trochanter.

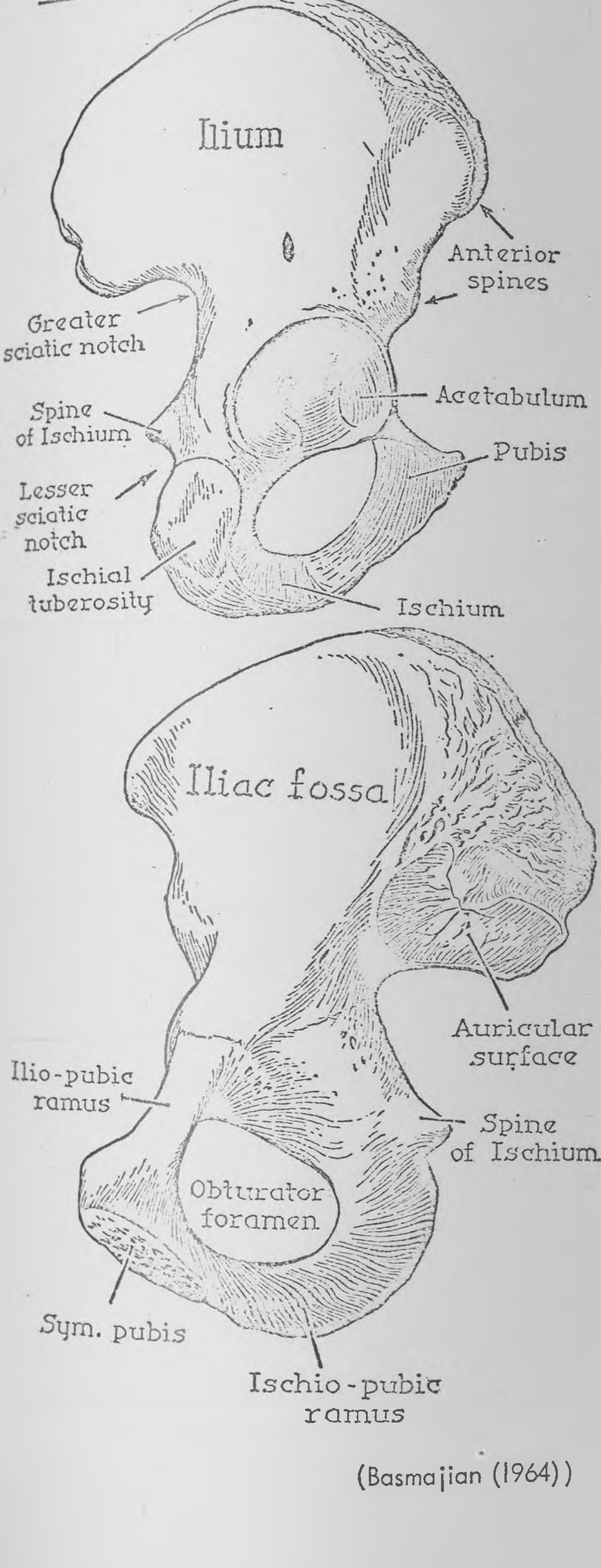
The trochanters (Greek: run or roll) are prominent processes of bone which provide leverage to the muscles which rotate the thigh on its axes.

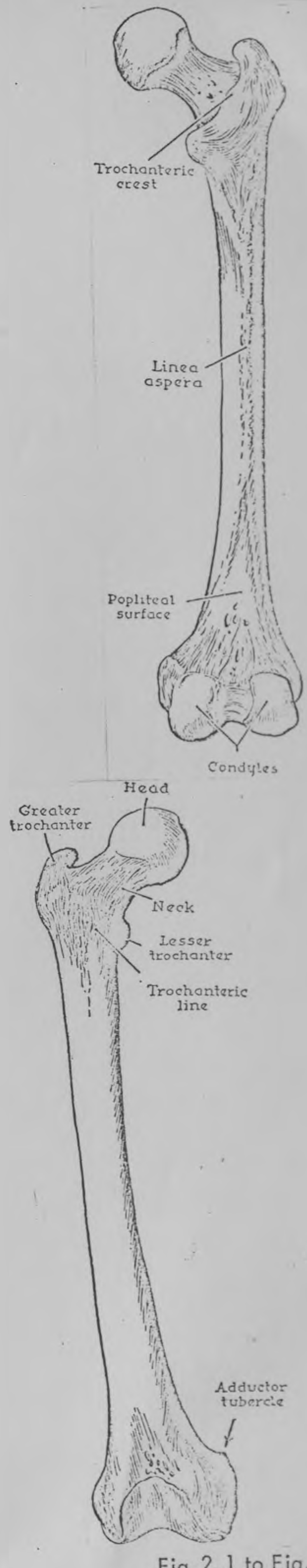
The greater trochanter is a large, irregular, quadrilateral protuberance situated at the outer side of the neck, at its junction with the upper part of the shaft. It is directed a little outwards and backwards, and in the adult, is about three quarters of an inch lower than the head.

The lesser trochanter is a conical process which projects from the lower and back part of the base of the neck. Its summit, which is directed inwards and backwards, is rough, and provides the insertion for the tendon of the Psoas major muscle.

A well-marked prominence, of variable size, which projects from the upper and lower part of the neck, at its junction with the greater trochanter, is called the tubercle of the femur. A slight ridge called the linea quadrata passes along the back part of the shaft vertically

THE PELVIS AND FEMUR





(Basmajian (1964))

Fig. 2.1 to Fig. 2.4

downwards for about 2". It is however very often not prominent and it can be seen merely as a slight thickening about the centre of the intertrochanteric line.

The shaft, almost cylindrical in form, is a little broader above than in the centre, and somewhat flattened below, in the anterior posterior direction. It is slightly arched, so as to be convex in front and concave behind, where it is strengthened by a prominent longitudinal ridge, the linea aspera. This ridge divides into three at its upper end. The most external one is very rough, and is continued almost vertically upwards to the base of the greater trochanter. At the ridge's lower end it is prolonged by two ridges which enclose between them a triangular space, the popliteal surface, upon which rests the popliteal artery. The inner ridge terminates below, at the summit of the internal condyle, in a small tubercle, the Adductor tubercle, which affords attachment to the tendon of the Adductor magnus muscle.

As far as the lower part of the pelvis is concerned, major interest is in the ischium and the os pubis Fig. 2.3 and 2.4.

The ischium (Greek: hip) is the inferior and strongest portion of the pelvis; it proceeds downwards from the acetabulum, expands into a large tuberosity, and then, curving upwards, forms, with the descending ramus of the os pubis, a large aperture, the obturator foramen. It is divisible into a thick and solid portion: the body, a large, rough process, on which the body rests in sitting; the tuberosity; and a thin, ascending part, the ramus.

The tuberosity is the most important part of the ischium and has three surfaces, external, internal and inferior.

In front of this margin the surface gives attachment to the Quadratus femoris, and anterior to this to some of the fibres of origin of the Obturator externus. The lower part of the surface gives origin to part of the Adductor magnus.

The internal surface forms part of the bony wall of the true pelvis. The inferior surface is divided into two portions: an anterior, rough, somewhat triangular part; and a posterior, smooth, quadrilateral portion. The anterior surface is subdivided by a prominent vertical ridge, passing from base to apex, into two parts. The posterior portion is subdivided into two facets by an oblique ridge, which runs anteriorly and laterally.

The ramus, or ascending ramus, is the thin, flattened part of the ischium, which ascends from the tuberosity upwards and inwards, and joins the ramus of the os pubis - their point of junction being indicated in the adult by a rough line.

The os pubis forms the anterior part of the os innominatum, and, with the bone of the opposite side, forms the front boundary of the true pelvic cavity. It is divisible into a body, an ascending and a descending ramus. Only the descending ramus is of interest in this consideration. It is thin and flattened and passes outwards and downwards, becoming narrower as it descends to join with the ramus of the ischium.

2.3 The Major Vessels and Nerves

The main artery to the thigh is the femoral artery. It originates as the external iliac from the common iliac and changes to the femoral after it leaves the abdominal cavity and passes the inguinal ligament Fig.2.5. Entering the thigh in the mid line in front, it descends vertically until, four inches above the knee, it meets the femur. The artery then passes posteriorly through an opening in the Adductor magnus and on the medial side of the femur to become the popliteal artery. Throughout its femoral course it is entirely medial to the bone and rests behind on the adductor muscles.

About two inches below its beginning, three important branches arise from the femoral artery. They supply adjacent muscles and are briefly described below.

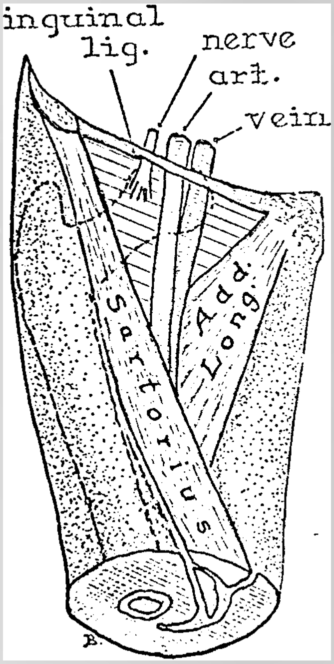
The medial femoral circumflex artery passes behind and above the adductors and joins with the inferior gluteal artery. It contributes to the supply of the hip joint and serves the upper part of the posterior of the thigh.

The lateral femoral circumflex artery is the only important branch which runs laterally. It supplies the quadriceps muscle group and provides branches which ascend or descend along the whole length of the lateral side of the thigh from hip to knee.

The profunda femoris artery is the largest branch of the femoral artery and follows the course of the main vessel but on a deeper plane. It provides blood to the back of the thigh by means of a series of three or four branches which pass close to the femoral shaft through the insertions of the adductors.

The veins correspond to the arteries described above. In the lower

THE RIGHT FEMORAL TRIANGLE



(Basmajian (1964))

Fig. 2.5

part of the thigh, the femoral vein lies behind the artery. At the root of the thigh it lies medial to the artery and changes to the external iliac vein on passing behind the inguinal ligament.

Three main nerves enter the thigh, the femoral, the sciatic and the obturator.

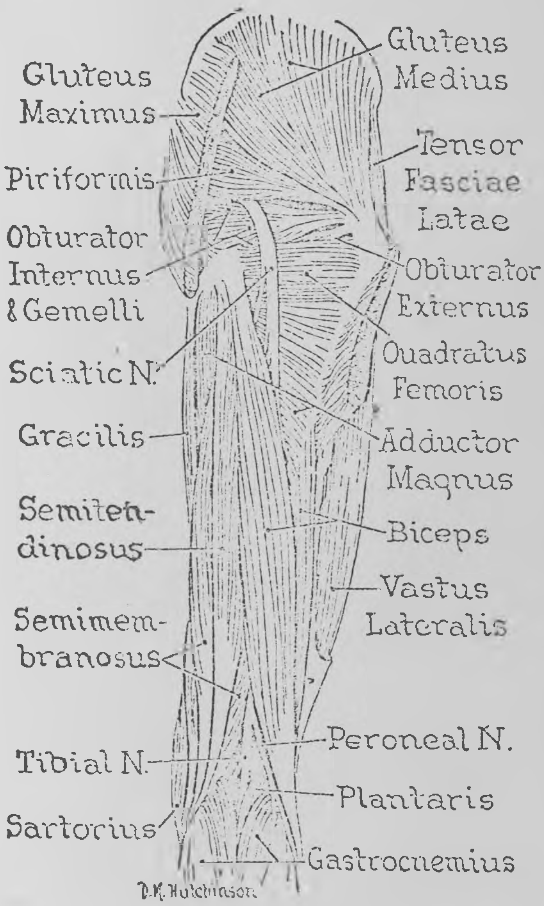
The femoral nerve passes into the thigh behind the mid point of the inguinal ligament, and immediately breaks into numerous branches. These supply the muscles and skin of the front of the thigh and the hip joint.

The sciatic nerve is the largest nerve in the body and its path is shown in Fig.2.6. It passes through the greater sciatic foramen below Piriformis, along the back of the ischium deep to the lower part of Gluteus maximus. The sciatic nerve then runs vertically downwards in the mid line of the back of the thigh deep to the long head of Biceps femoris, and it gives branches to the Semitendinosus, Semimembranosus and Adductor magnus.

The obturator nerve runs at first on the side wall of the pelvis. Reaching the thigh by passing through the highest part of the obturator foramen, it divides immediately into two branches. The anterior branch supplies Adductor longus, Adductor brevis, Gracilis (and often Pectineus), some of the skin of the medial aspect of the thigh, and the capsule of the hip joint. The posterior branch pierces and supplies Obturator externus and part of Adductor magnus.

2.4. The Effect of Amputation on Hip Musculature

Research into the effect of amputation on the muscle power in the



(Basmajian (1964))

Fig. 2.6

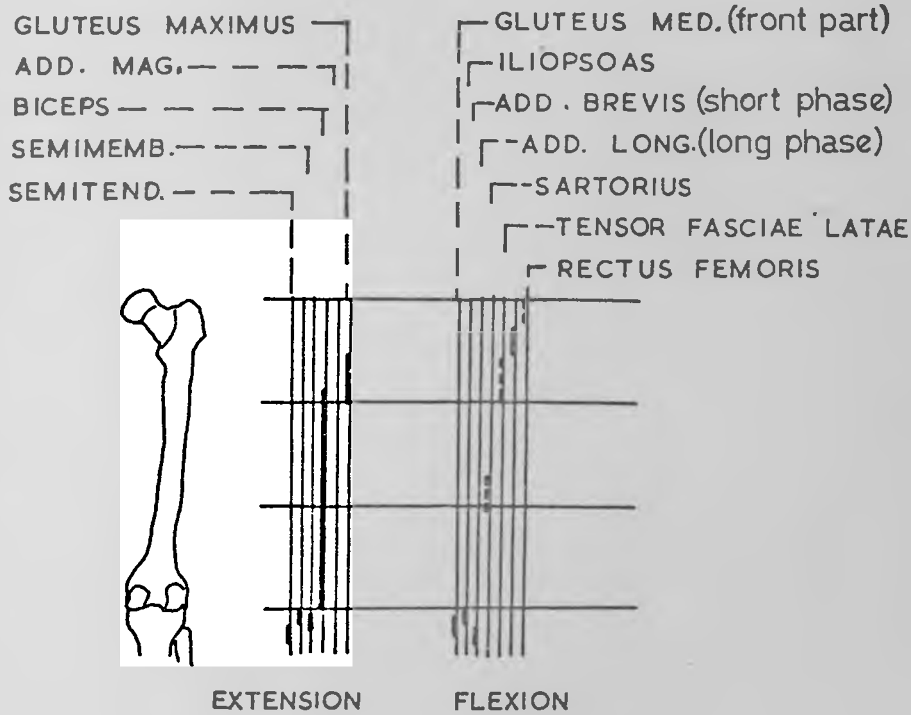
above knee stump was carried out by Mommsen (1918). He based his work and discussion on the accurate measurements made by Roith (1908). Mommsen tried to analyse the influence of the height of amputation on the muscular action at the hip. Referring to his diagrams, reproduced here as Fig.2.7, the continuous lines represent the distal insertions of muscles which act on the hip joint under favourable mechanical conditions. The dotted lines show the insertions of those muscular fibres which, according to the investigations of Roith, act under relatively unfavourable conditions.

Mommsen describes the adduction and abduction situation as being balanced if the amputation occurs below mid thigh. Above that the abductors have an increasing advantage. With regard to hip extension and flexion, the higher the amputation, the less the muscle power available for hip extension relative to that available for hip flexion. Therefore, with a short stump there is a tendency for flexion and abduction contractures.

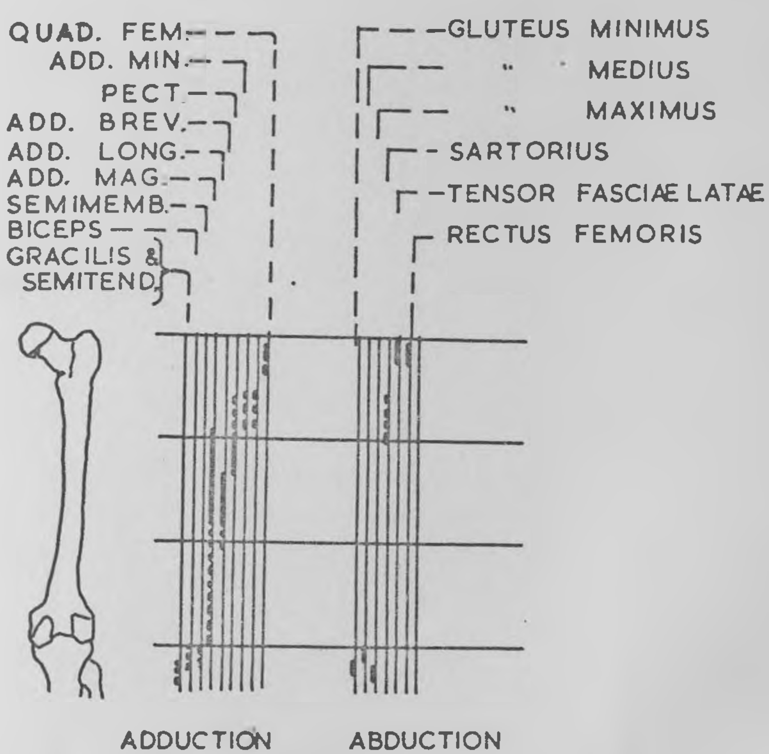
To discuss the action of various muscles in more detail, they are classified by the following anatomical groupings.

2.4.1 Muscles crossing the front of the hip joint There are two muscles in this category namely, Psoas major and Iliacus, although they are usually grouped together as the Ilio-psoas,(Fig.2.8)(a very small muscle, Psoas minor, may or may not exist in the amputee but it is of no importance in this context).

Both parts of the Ilio-psoas are concerned with flexion of the hip joint. This muscle probably contributes to the stability of the hip joint



A

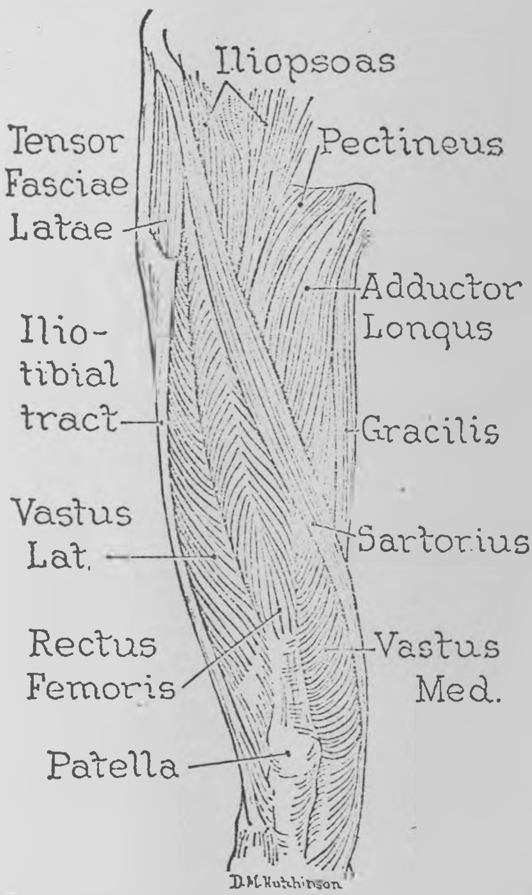


B

(Mommsen (1918))

Fig. 2.7

DISSECTION OF THE FRONT OF THE THIGH



(Basmajian (1964))

Fig. 2.8

during walking and standing upright, Basmajian (1964).

Since these muscles insert on the lesser trochanter, they are unaffected by amputation and are able to keep their full power.

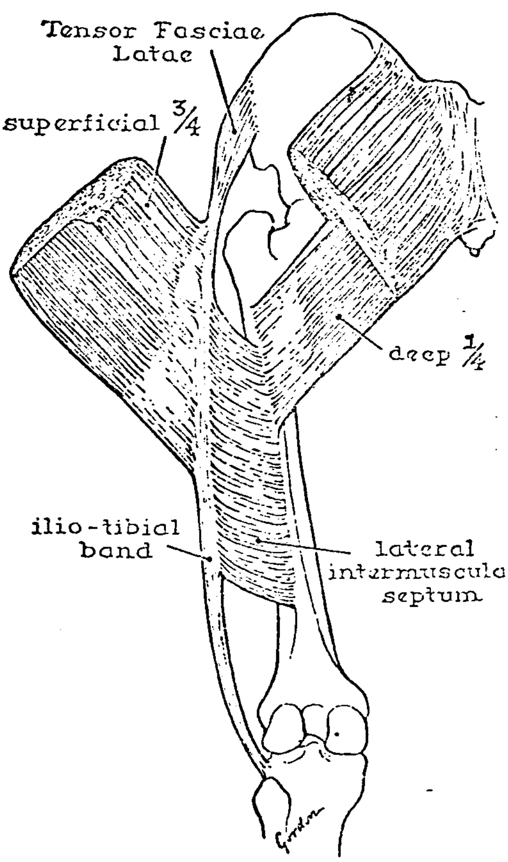
2.4.2 Gluteal muscles The three gluteal (Greek: rump) muscles are all large and powerful and the qualifying adjectives used merely indicate their relative size.

Gluteus maximus is one of the largest, thickess and coarsest-grained muscles in the body. It is placed entirely behind the hip joint and is shaped like a parallelogram whose long sides run downwards and laterally, (Fig. 2.6).

Fig. 2.9 shows the complicated insertion of Gluteus maximus but as far as the femur is concerned, it inserts along a continuous line from just below the hip joint right to the knee joint.

Gluteus maximus is an extensor of the hip joint but it is used only when the joint has to be extended with power. It is used therefore in rising from a sitting or stooped position, climbing a hill or going upstairs. Indications are that in walking on the level at normal cadence it is not used. The muscle is also a powerful lateral rotator of the extended thigh though it loses power if the thigh is flexed. It is claimed to have an abducting action but this has not been shown to be true.

From the prosthetist's point of view it should be borne in mind that Gluteus maximus can only act with full power when the femur is flexed 15° or more. It is advisable to align the stump forwards to take advantage of this powerful muscle.



Three-quarters of Gluteus Maximus run into iliotibial band. Many of these fibres then run to linea aspera via lateral intermuscular septum. (Left side, from behind.)

(Basmajian (1964))

Fig. 2.9

In amputees the major activity of Gluteus maximus occurs at heel strike at the beginning of the stance phase when it stabilizes the prosthetic knee in full extension.

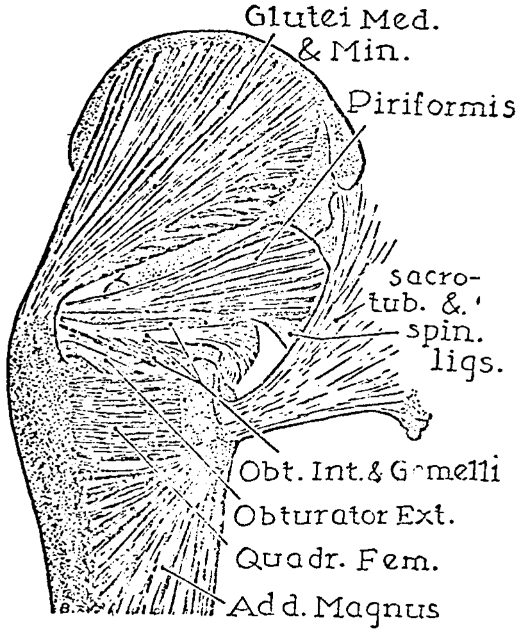
Tensor Fasciae Latae(Fig. 2. 9) runs vertically downwards and inserts into the ilio-tibial tract of the fascia lata. It lies anterio-laterally to the hip joint and can help to abduct, medially rotate and flex the hip joint. In the normal individual it assists the lateral ligament in resisting forces tending to adduct the knee in walking or when standing on one leg.

Provided the fascia lata is not allowed to retract at amputation this muscle can exert its full power on the artificial leg.

Gluteus medius(Fig. 2. 6 and Fig. 2. 10) has an extensive area of origin on the outer surface of the ilium but much further forward than the origin of Gluteus maximus. It is inserted on the greater trochanter of the femur. Since the greater trochanrer is further back than the hip joint, when Gluteus medius contracts it abducts the thigh and has a secondary rotation action.

Gluteus minimum(Fig. 2. 10) lies deep to the anterior part of Gluteus medius and is smaller and fan-shaped. It rises, therefore, well forward on the lower part of the outer surface of the iliac blade. It is inserted on the anterior border of the greater trochanter where it joins partially with Gluteus medius.

As far as these two muscles are concerned, they can be considered as one in their major actions. They aid in maintaining medio-lateral stability of the pelvis when the femur is supported in the correct amount of adduction by the lateral wall of the socket. More precisely they



(Basmajian (1964))

Fig. 2. 10

control the action of the pelvis in the stance phase by preventing excessive drop of the pelvis on the side of the limb in the swing phase. They also prevent excessive lateral shift of the pelvis which is related to the medio-lateral stability.

2.4.3 The six lateral rotators (Fig.2.10) These are Piriformis, Obturator intermus, Gemellus superior, Gemellus inferior, Obturator externus and Quadratus femoris. All these muscles lie deep in the thigh behind and below the Gluteus minimus and are hidden by the Gluteus maximus. They all pass behind the hip joint and the tendons of the first five insert into the medial surface of the greater trochanter. The sixth, Quadratus femoris, inserts in the quadrate tubercle.

Once again these muscles are unaffected by amputation and are able to function as in a non amputee.

Piriformis may abduct and extend the thigh Anthony (1967) and all six muscles are lateral rotators of the hip.

2.4.4 Muscles of the front of the thigh (Fig.2.8) The shaft of the femur is covered by three muscles over its front portion. These are Vastus medialis, Vastus intermedius and Vastus lateralis. Since all these muscles have their origin on the shaft of the femur and are inserted into the patella at the Rectus tendon, they have no significance with respect to the hip joint. However they are important in myoplasty, since they constitute muscle bulk which can contract and hence harden. This forms a good medium for force transmission from the femur to the socket.

Two muscles of importance in this group are the Rectus femoris and the Sartorius.

Rectus femoris arises (a) by a straight tendon from the anterior inferior iliac spine and (b) by a curved tendon from the upper margin of the acetabulum. These two tendons join and the muscle runs straight down the front of the thigh where it can easily be felt. Its structure is pennate to give it great power over a short range. The insertion of Rectus femoris is into the patella at the rectus tendon.

In the normal individual, the muscle flexes the hip and/or extends the knee. This combination of movements is used to advance the limb in walking. In the above knee amputation this muscle is transected and its power is lost unless a myoplasty is performed. The same thing happens in the case of the Sartorius.

Sartorius is the longest muscle in the body and has the longest fibres. It arises from the anterior superior iliac spine and, running obliquely, makes its way as the most superficial muscle of the thigh, to the medial side of the knee where it is inserted by a thin, flat, expanded tendon, on the upper part of the shaft of the tibia.

It is not a very powerful muscle and flexes, abducts and laterally rotates the thigh whilst flexing and medially rotating the shank. It possibly plays a stabilizing role during walking.

2.4.5 <u>Muscles of the medial aspect of the thigh</u> (Fig.2.6 and 2. 8) These muscles all act to adduct the thigh and, by their bulk, fill in the gap medial to the femur.

Gracilis is a strap-like muscle descending vertically from near the mid-line symphysis pubis to the medial side of the tibia just beyond the knee. The other adductors lie between Gracilis and the femur.

Pectineus, Adductor longus, Adductor brevis and Adductor magnus arise from the front of the pubis and they run downwards and laterally to pass behind the origin of the Vastus medialis. They are triangular in outline and are arranged in layers like the pages of a book.

Pectineus is the most superior and arises from the superior ramus of the pubis as far back as the pectineal line, i.e. the front part of the pelvic brim. It lies adjacent to Psoas and is supplied by the same nerve, i.e. the femoral nerve.

Adductor longus and Adductor brevis originate from the front of the body of the pubis.

Adductor magnus arises from the ischio-pubic (inferior) ramus and extends its origin as far back as the ischial tuberosity.

Pectineus and the other three adductors (longus, brevis and magnus) are inserted along the linea aspera and medial suparcondylar line to the adductor tubercle.

Adductor longus occupies the middle six inches, Pectineus is inserted above it and Adductor brevis lies half behind Pectineus and half behind longus. Adductor magnus occupies the length of the linea aspera from a little below the quadrate tubercle to the adductor tubercle.

As a group, the adductors can produce powerful adduction yet this action is not an especially important one in walking or standing. Their true function is not known but they probably help to "stabilize the hip" (Basmajian). Because they are on a plane in front of the hip joint, most of the adductors can flex the thigh as well as adduct it. Because they pass to the back of the femur they can assist too in lateral rotation. Adductor magnus, due to some of its fibres inserting at the adductor tubercle and originating from the ischial tuberosity, can assist in extension of the hip.

Adductor longus has an important action just at the end of the stance phase of walking where it assists flexion and external rotation of the leg.

Amputation above the knee will divide Gracilis and additional adductors will be transected as the level of amputation is progressively raised.

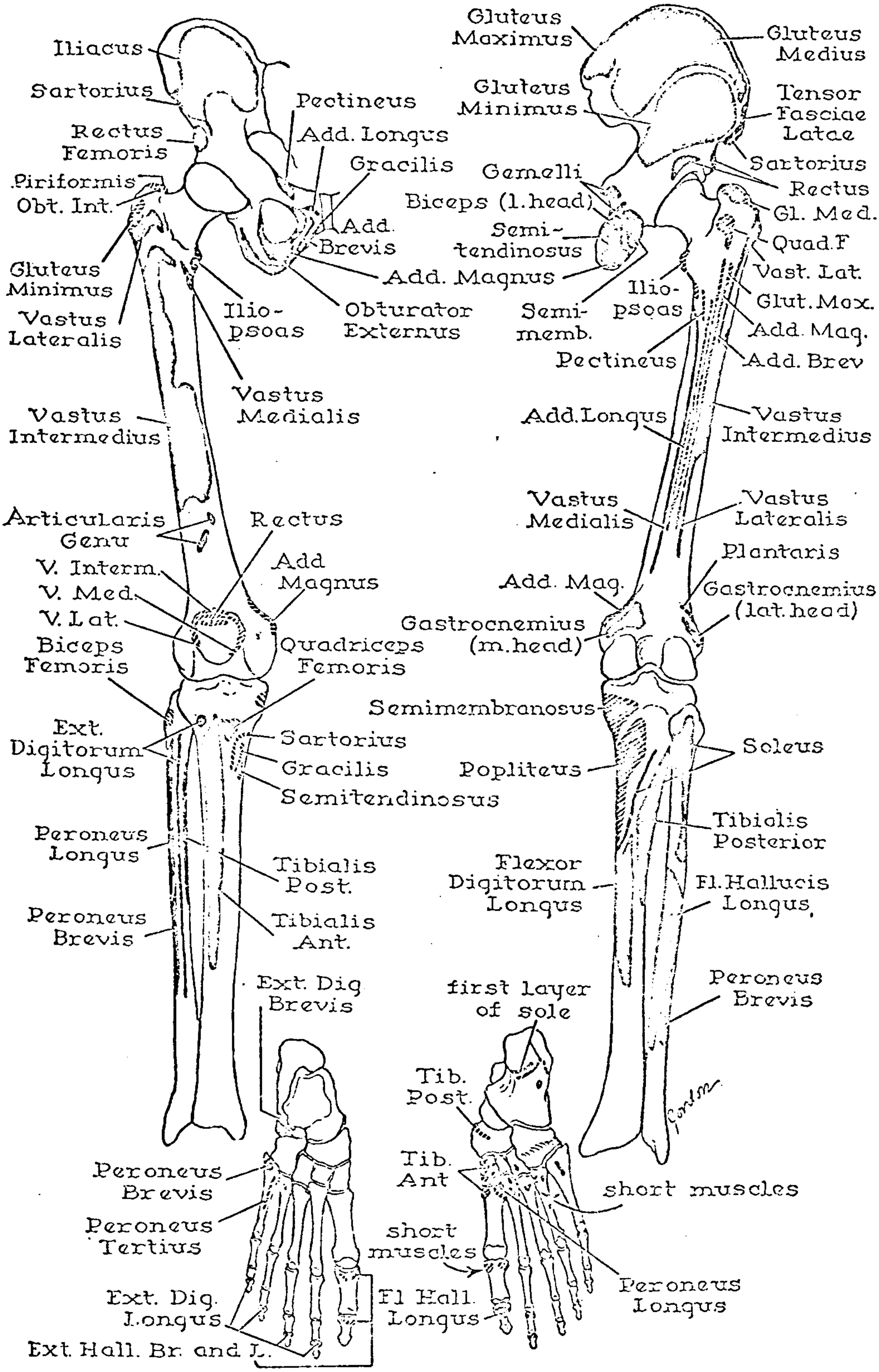
2.4.6 Muscles of the back of the thigh (Fig. 2.6) The group of muscles lying behind Adductor magnus is known as the hamstrings. These are very long and extend from a common origin on the ischial tuberosity, to beyond the back of the knee joint. They are Biceps femoris, Semimembranosus and Semitendinosus.

Biceps femoris as it name implies has two heads. The short head has very little significance in prosthetics as it arises from the middle third of the linea aspera, although in the longer above-knee stump the short head is able to contribute to muscle bulk and assist in force transmission. Both heads insert on the head of the fibula.

Semimembranosus, whose upper one third is membranous (flat sheet of fibrous tissue) is inserted on the back of the medial tibial condyle.

The Semitendinosus has a long rounded tendon that curves round the

ORIGINS AND INSERTIONS OF ALL THE LEG MUSCLES



origins (solid black) and insertions (hatched) of muscles of lower limb.

(Basmajian (1964))

Fig. 2.11

medial side of the knee to the front and is inserted into the tibial shaft in close association with Sartorius and Gracilis.

In normal individuals the hampstring group of muscles are used to extend the hip and/or flex the knee. Electromyography (Basmajian) reveals that in walking, Semitendinosus and Biceps femoris (short head) are responsible for carrying the weight of the leg from just after toe off until the hip starts to flex. Semimembranosus and Biceps femoris (long head) act when the foot is on the ground and are probably concerned with extending the hip and so shifting the weight of the body forwards.

Since the hamstring group of muscles is divided during an A/K amputation, the extensor power of the hip is reduced in an above-knee stump.

2.5 The Development of Amputation Technique

From the discussion up to this point and with regard to considerations in the fitting of a prosthesis, it can be seen that the method of amputation is of the utmost importance in any discourse on artificial legs. It is not intended to cover the technique of amputation here but rather general principles which have evolved.

The commonest type of amputation used throughout history has been the guillotine amputation. This consists of purely cutting straight through skin, muscle and bone at the same level and sewing up the skin over the bone end. As so often happens in the history of medicine, a better technique was used as long ago as A.D. 25 by Celsius but completely forgotten until 1779 when it was revived by Alanson. The method he used was to cut the skin at the lowest level, the muscle at a higher level and the bone at the highest level. When sutured the resulting stump was much better shaped and better able to be fitted by a socket.

In the late nineteenth century various attempts were made to improve the stump by surgical procedures. The most significant to these were due to Pirogoff, Gritti, Syme and Bier. Of these workers Bier probably made the greatest contributions towards a plastic amputation.

In 1893, Bier recognised the disorder associated with taking load on the cut end of the bone after amputation and called it "Knochennarbe" (bone scar). He recommended the use of a type of osteoplasty but the idea was dropped by later surgeons. Bier covered the bone end with a periosteum flap. The main fault with the technique was the length of time required for the bone plates to reform.

Hirsch, a scholar of Bier recommended a procedure; the so-called "aperiostal" amputation. To stop growth of the side bone, always a problem with early amputation techniques, he removed the periosteum one cm. above the amputation site.

Bunge also removed a ring of periosteum but he scooped out the marrow to the same level by the means of a spoon. This technique caused avascular necrosis of the stripped end of the femur and lead to it being left in the stump as a coronal sequestra. Klapp, Thomson and Marquadt had already rejected this method, however, Jenny (in his monograph) still suggests a small Bunge operation.

Some workers even stripped off the periosteum and endosteum and allowed the bone to die and be extruded as a foreign body.

None of these procedures prevented the bone from growing and they were all discarded.

Various methods have been suggested for covering the end of the bone. Wilms used the soft parts of the tendons, Rittar used the fascia and Neuber used muscles and tendons. From this point of view it appears that Neuber was the first to attempt a myoplasty. Nieny (1917) points out that to apply Neuber's technique successfully a long stump is required.

Mommsen (1918) describes a technique used by Ludloff in a Gritti amputation where he sutures the two-joint muscles on the rear side of the femur to the ligamentum patella.

Little (1922) warns that the suturing of extensors to flexors causes too much mobility but basically he is in favour of myoplastic techniques. "Muscles and tendons should be prevented from retracting by fixation to the periosteum, but not brought over the end of the bone."

After World War I, improved artificial limbs lead to amputation schemes for various levels. Both Schede (1941) and zur Verth (1937) contributed to these developments. In World War II surgeons attempted to meet the requirements of Schede and zur Verth and in fact sacrificed limb segments to do so. The guillotine operation was also adopted at the battle front to be remedied later in aseptic conditions.

The real move towards present day techniques occured after World War II when Ertl and Mondry developed plastic amputation techniques. Mondry is best known for his work in myoplasty and Ertle for osteoplasty.

Loon (1960) revived the technique of Bier but found it unsatisfactory. He then tried osteoperiosteal flaps instead of periosteal flaps and the healing time improved and satisfactory results ensued. Loon also showed, by means of venograms, the importance of the seal at the bone end. The slight pressure produced by the seal is sufficient to cause venous reflux of the blood from the rigid medullary canal. It also tends to limit oedema.

Dederich (1961) considered Mondry's modification of Ertl's method for the below knee stump satisfactory, i.e. a tube of periosteum between the fibula and tibia. He also advocated the use of myoplasty claiming that better nutrition, muscle control and less painful neuromata resulted if the muscle is closed over the bone end. However, Dederich did not fix the muscle to the bone end and problems arose due to the movements of the muscles as predicted by Little.

Previous research by Weiss (1960) into the electromyography of amputation stumps showed that no amount of exercising causes an unattached muscle to become more active and so he later modified Dederich's technique. Weiss sutured the muscle groups to the end of the bone through drilled holes. The muscles were severed at the best length to maintain the correct length - tension relationship. An osteoplasty after Loon was also performed on the femur.

The advantages of a myoplasty are enumerated by Vitalli and Harris (1964) as:-

1. Better muscular balance
2. Better femoral stability
3. Better vascularity
4. End of bone able to take some weight

Burgess and Romano (1965) described a technique based on that of Weiss after being visited by him. The technique is now referred to as myodesis and is being used by a number of surgeons in Europe and U.S.A. Together with immediate post-operative fitting, also developed by Weiss, this method allows an amputee to be walking on a permanent prosthesis three weeks after amputation. The complete I.P.O.F. system for lower extremity amputees is described in Burgess et al (1967).

2.6 The Influence of the Anatomy on Socket Design

The transmission of force between the amputee and prosthesis is discussed in Chapter 6, but it is intended to point out here certain anatomical features which necessitate some consideration whatever socket or amputation is used.

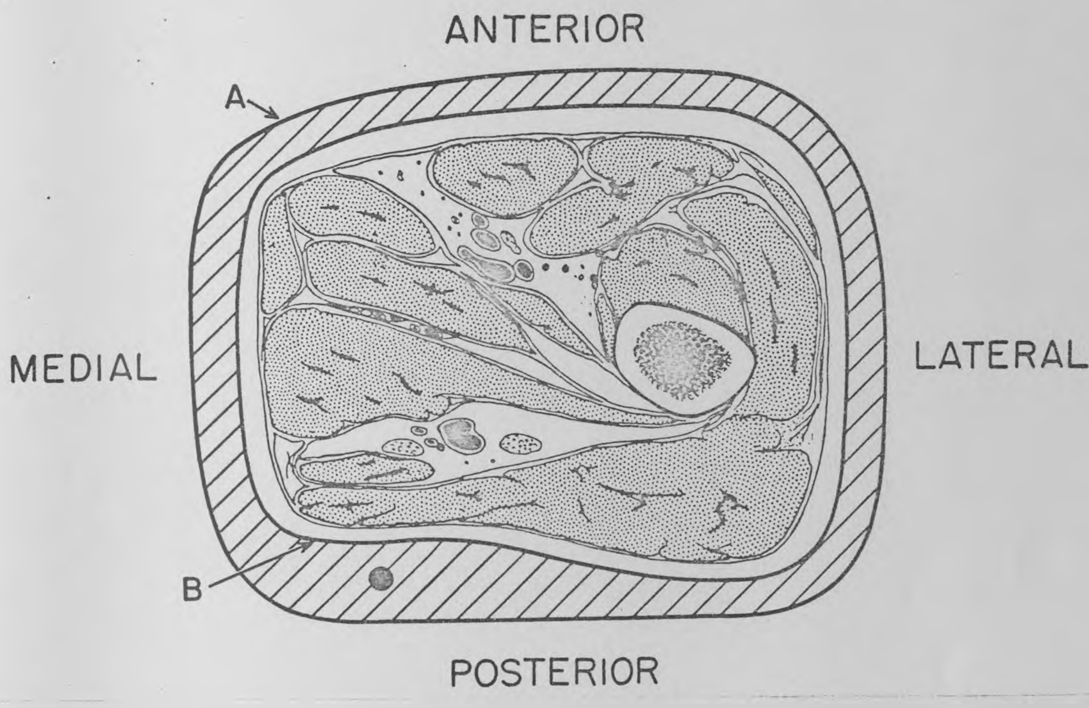
Fig. 2.12 shows a typical cross section of an above-knee stump in a Quadrilateral Socket.

The socket must be properly contoured and relieved for functioning muscles. The stump changes shape during the walking cycle and allowance must be made for this movement of the muscle groups. Where no functioning muscle exists, a stabilising force should be built in by suitable socket design.

The alignment of the socket should take into account a possible flexion or abduction contracture. The major extensors of the hip are the hamstrings, the gluteal group only acting when the hip is flexed as previously mentioned. Therefore the socket should be aligned in slight flexion. It has also been found that the abductors have more power when the socket is adducted slightly.

Pressure on the major vessels and nerves must be avoided by using large areas for force application. A total contact socket appears to be the most desirable form of socket from this point of view. It also reduces oedema, adds to stability and reduces the feeling of prosthetic weight.

THE CROSS-SECTION OF AN ABOVE-KNEE STUMP ½" BELOW THE ISCHIAL LEVEL



(Radcliffe (1955))

Fig. 2.12

Room must be left in the socket for movement of any tendons. This is especially true of the adductor tendons.

Until other methods of loading the stump are available, it appears that the majority of the weight should be taken by the ischial tuberosity.

CHAPTER 3

THE EXPERIMENTAL TECHNIQUE

CHAPTER 3

3.1 Introduction

The devices used to obtain information regarding force actions and knee angle are discussed together with the procedures used to calibrate them. The various tests performed are described and methods of accelerating the processing of the data for subsequent use by the computer are presented.

3.2 The Strain Gauged Pylon

The pylon is designed to record the following force actions as shown in Fig.3.1:-

The bending moments in the A/P or Z plane at two levels M_{UZ} and M_{LZ}.

The bending moments in the Y plane at two levels M_{UY} and M_{LY}.

The axial load f_x

The torque about the long axis T

The mechanical construction of the pylon is shown in Fig.3.2. Ideally it should be as long as possible so that inaccuracies of extrapolation are minimised (see Assessment of Accuracy, Chapter 4). On the other hand it is desirable to be able to incorporate the device in as many different types of prostheses as possible. The final design, allows it to be used in the majority of B/K amputees and gives reasonable accuracy.

Basically the pylon consists of a $1\frac{1}{2}$" internal diameter tube, 0.049 inches thick of 24 ST aluminium alloy, fitted at each end with aluminium plugs. These plugs are made to a slight interference fit, and further

FORCE ACTIONS MEASURED BY PYLON

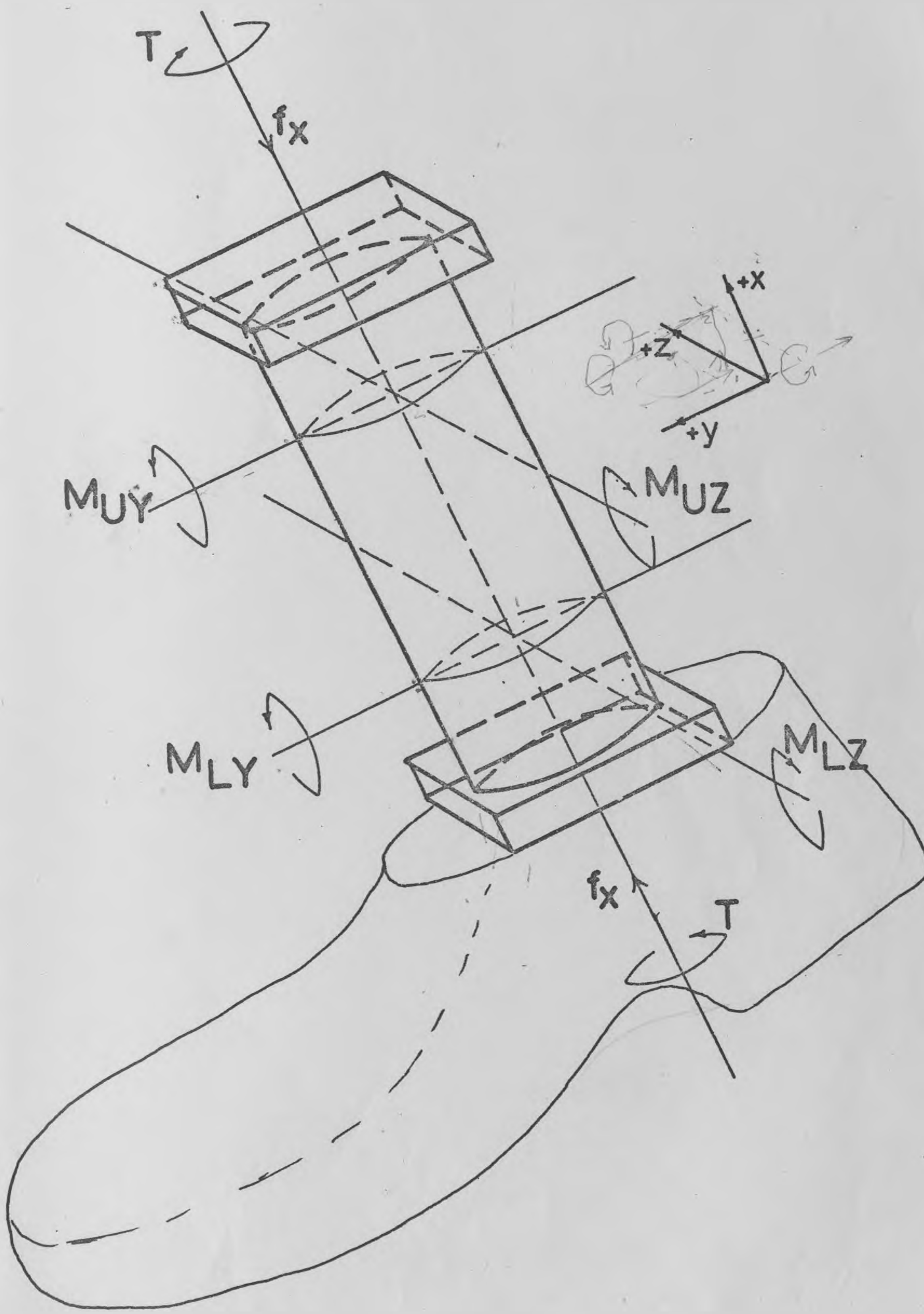
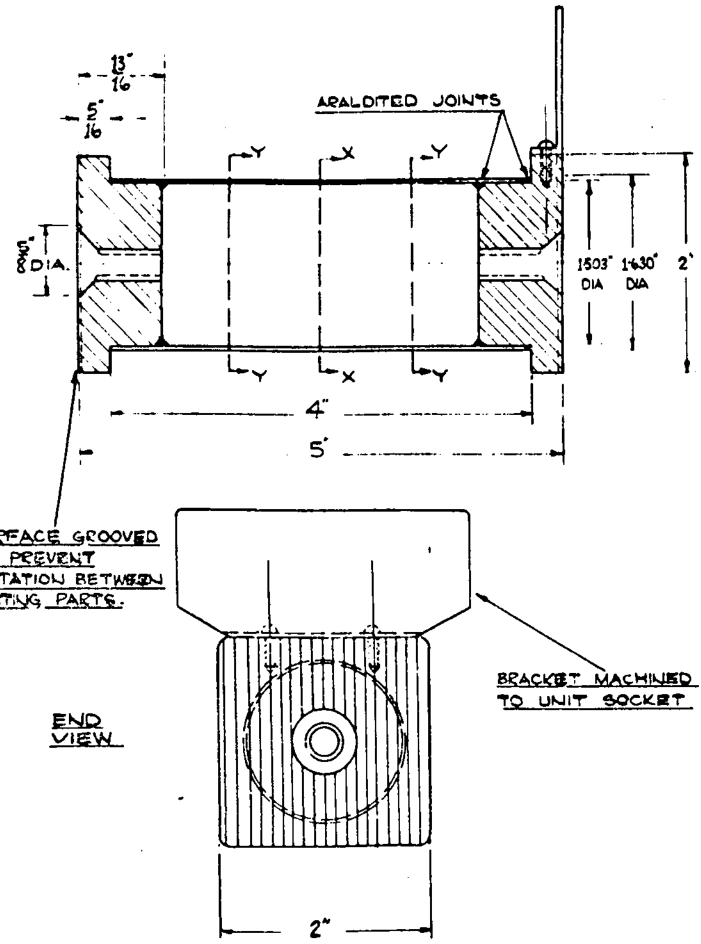


fig. 3.1

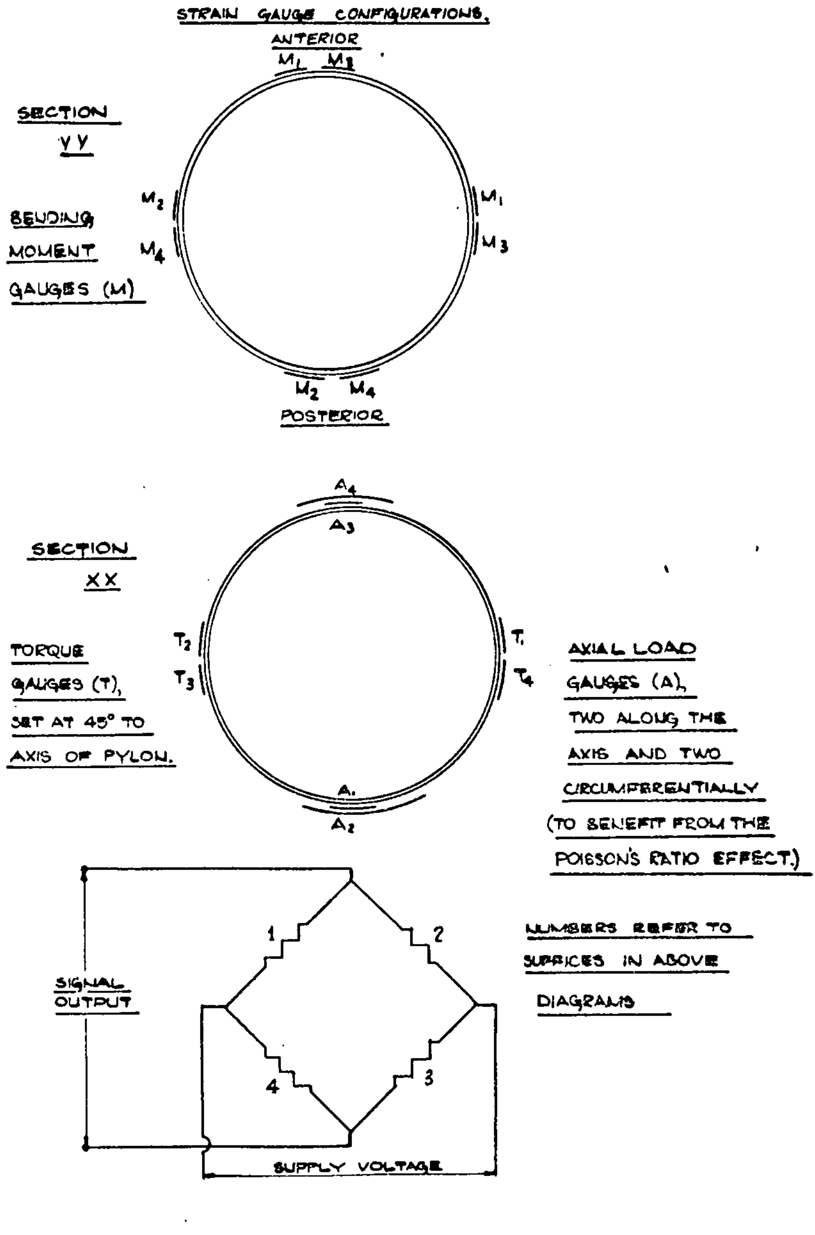
THE MECHANICAL CONSTRUCTION OF THE PYLON.



XX AND YY REFER TO STRAIN GAUGE LAYOUT.

Fig. 3.2

CONSTRUCTION OF STRAIN GAUGED PYLON



adhesion is provided by the "Araldite" keys shown in Fig.3.2. The fixation technique was tested on a dummy pylon at a bending moment of 2000 lbf. in. This was considered sufficient for use in a research centre but extensive fatique tests are required before an amputee can be allowed to walk on it continuously as part of his normal prosthesis. Grooves at each end prevent rotation between the pylon and any mating attachment.

The conical recesses provided in the ends of the pylon are for the calibration of the axial load gauges as described later.

The method of fixation to the prosthesis is also seen in Fig.3.2. If a SACH foot is used, a screw can be passed through the foot and directly into the lower end of the pylon. With a conventional foot, an adaptor has to be made.

For attaching the device to the shank, various methods can be used. If a modular prosthetic system is being used, a boss can be turned such that its O.D. is the size of the I.D. of the shank tube. This is then fastened to the top of the pylon with a screw, and the shank tube attached by a "Jubilee" clip. Alternatively, with a long B/K stump, direct attachment to the bottom of the socket is possible.

The force actions are measured by means of foil strain gauges applied to the pylon as shown in Fig.3.3. The torque gauges are Saunders Roe $\frac{1}{2}$ Torque (resistance 50 ohm.) and all the other gauges are Budd type C. 12141B (resistance 120 ohm.).

The gauges were chosen for their temperature compensation, reasonable size and relatively large surface area for heat dissipation. The strain gauge circuits are shown in Fig.3.2.

THE STRAIN-GAUGED PYLON INCORPORATED IN A PROSTHESIS

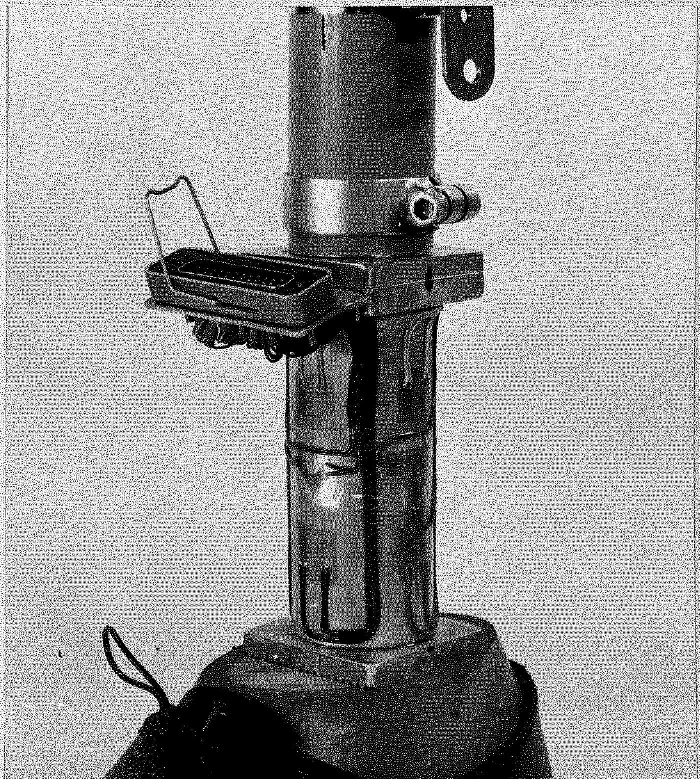


Fig. 3.3

A bridge voltage of 6 volts was used per channel and this was supplied by six Fenlow amplifiers type ZA2. The signal from the bridge was amplified by the same device and the output fed to an ultra violet recorder type SE 2100.

The galvanometers used were as follows:-

Type A3300 for M_{UY} and M_{LY}

Type A1600 for M_{UZ}, M_{LZ} and T

Type A1000 for f_x

Under typical maximum values of load, the stress and output signals are as shown in Table 3.1.

3.3 The Goniometer

For locating the hip joint in space relative to the shank axis, both the angle between the shank and thigh and the knee joint-hip joint distance have to be measured. In the work described here a simple goniometer was used to measure knee angle and its general construction is shown in Fig.3.4.

Two pieces of $\frac{3}{4}$" wide, $\frac{1}{16}$" thick brass are jointed at one end. One is nine inches long and the other fourteen inches.

The body of a Beckman type 5311 wirewound potentiometer is attached to the longer strip and the shaft to the shorter. The goniometer is energized by a stabilised power supply through a second potentiometer.

The arms are attached to the socket and shank with the potentiometer on the knee joint axis. Provision is made for relative movement between

TABLE 3.1.

Output voltages from Strain Bridge Circuits under typical maximum load conditions

For :- Bridge Voltage = 6 volt

Youngs Modulus = 10×10^6 lbf/in^2

Poisson's Ratio = 0.32

and Gauge Factor = 2.06

	Typical Max. Value of Force Action	Stress (lbf/in^2)	Circuit Output (mV)
Axial Load Gauges	250 lbf	1080	8.8
Bending Moment Gauges	1200 lbf/in	13850	171
Torque Gauges	100 lbf/in	1160	9.5

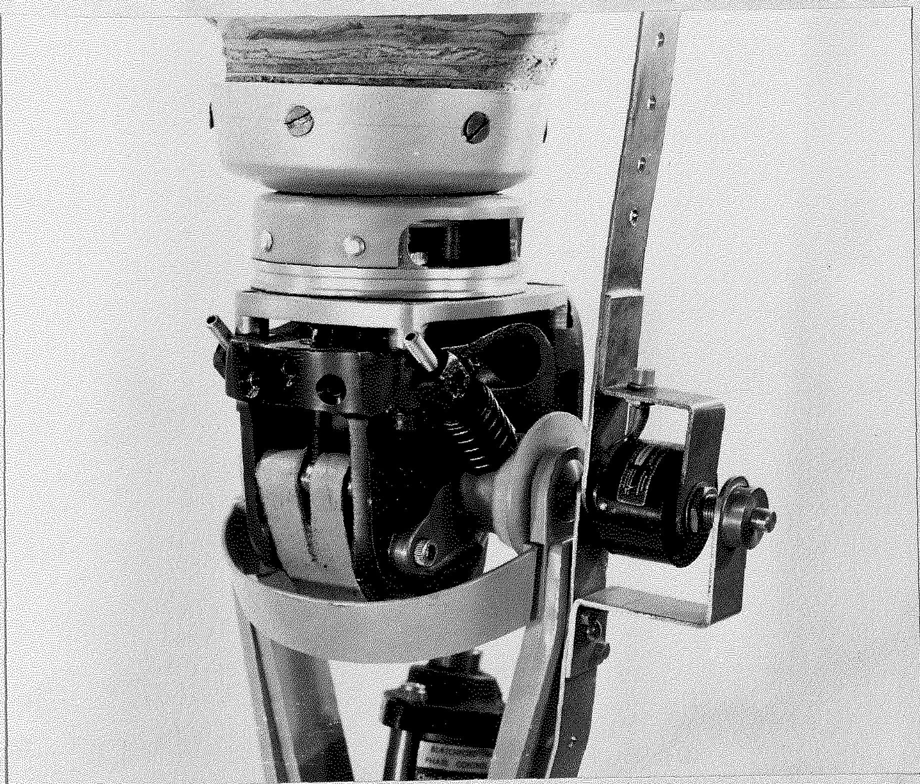


Fig. 3.4

the thigh arm and the socket by allowing the brass strip to slide between two screws at the trochanter height.

The output is recorded by means of a galvanometer type B450 in the same ultra violet recorder as used for the strain gauge readings, and is linearly proportional to the angle between the shank and the knee-trochanter line. This angle is defined as the knee angle.

Full details of the instrumentation circuitry is included for reference in Appendix I, together with the colour coding used for the wiring.

3.4 Calibration Procedure

They pylon was fastened to a stanchion as shown in Fig.3.5. A Budd Strain Indicator Model P-350 was used to examine the output from the gauges under various load conditions. A check for linearity from the bending moment and torque gauges was completed, and at the same time, the magnitude of cross effects was established.

Typical results from the six strain bridge circuits when subjected to bending in the Z plane are shown in Fig.3.6.

To examine hysteresis effects, the pylon was subjected to repeated loading with a constant weight of 85 lb. at the end of a 8 inch extension tube. A bending moment of approximately 800 lbf.in. was thus applied at the gauge positions. The readings most seriously affected were those in the Y plane and the results obtained from these are shown in Fig.3.7. The pylon was left for 2 hours with the 85 lb. load on, and the drift developed is also shown in Fig.3.7.

This hysteresis is probably due to a slight movement between the gauge

THE STRAIN-GAUGED PYLON POSITION FOR INITIAL CALIBRATION

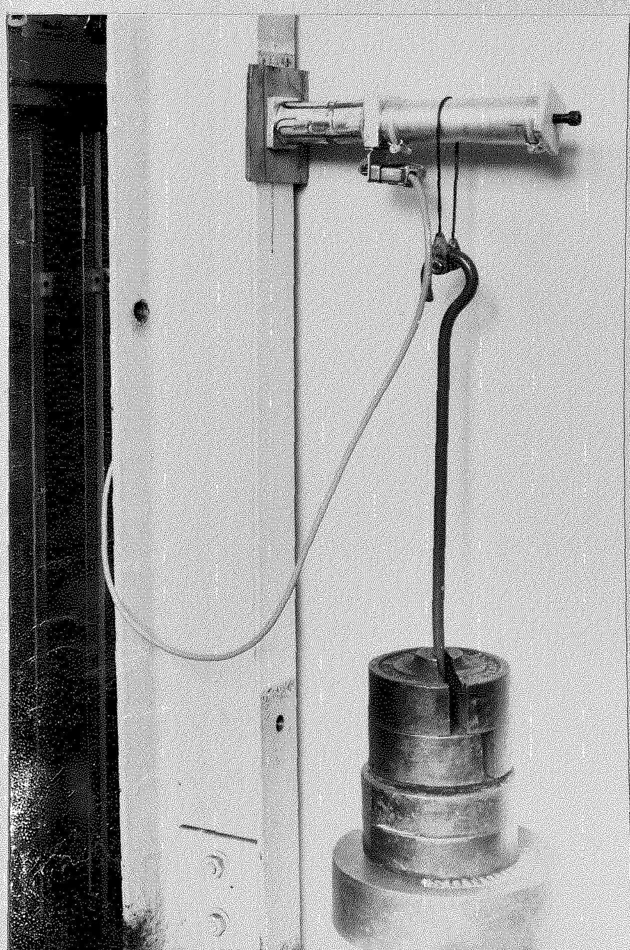


Fig.3.5

CROSS EFFECT CALIBRATION IN THE A/P PLANE
FOR BENDING

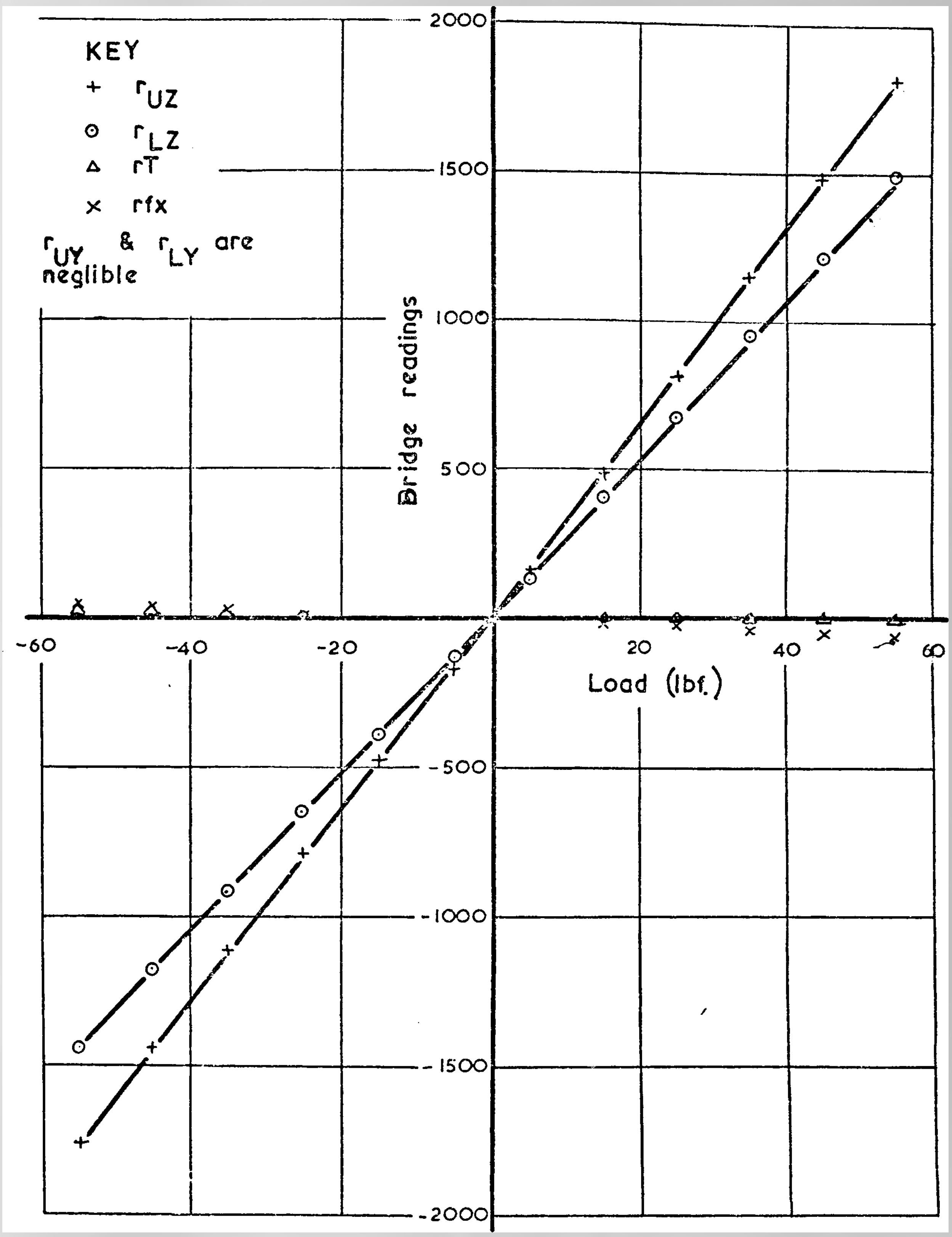
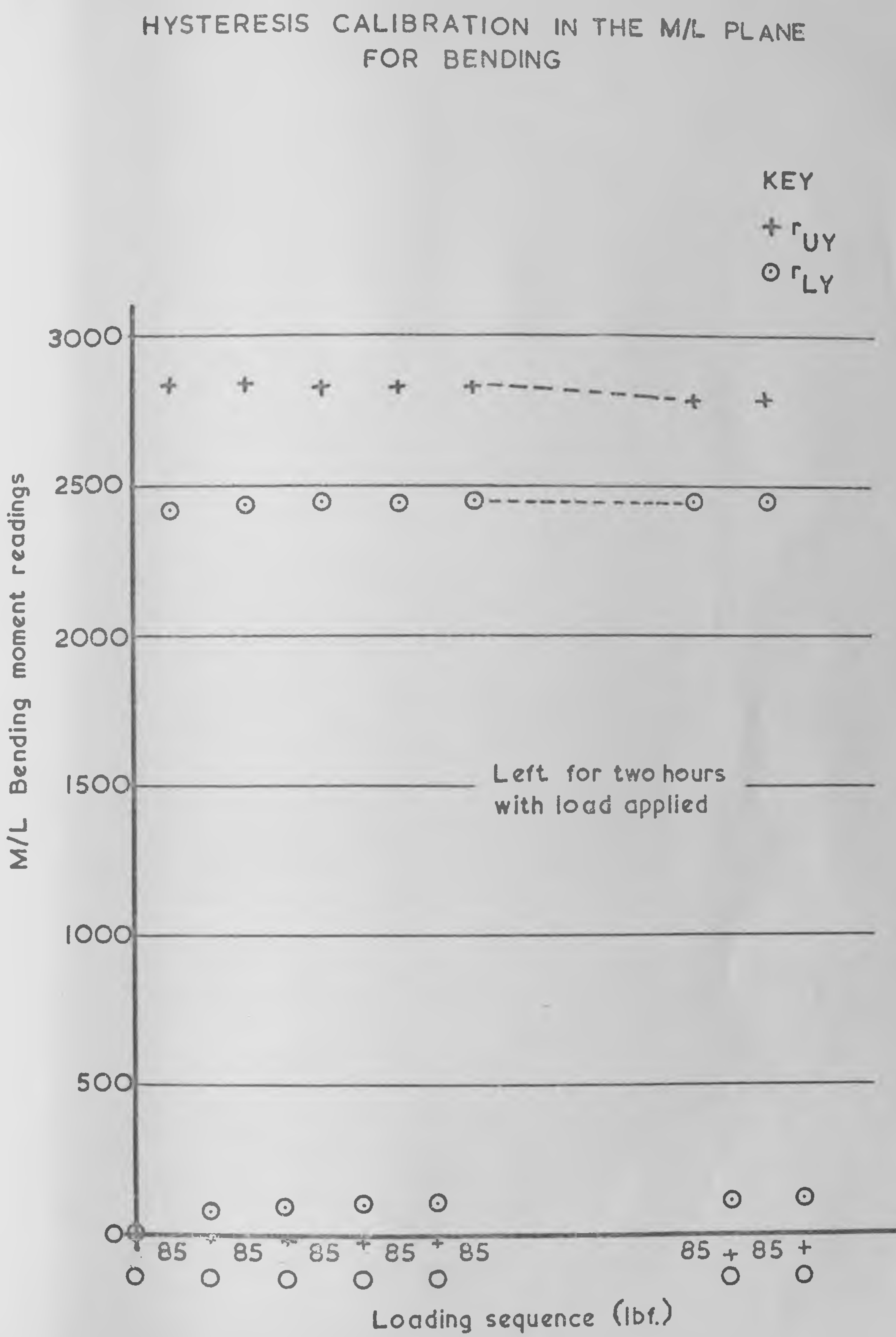


fig. 3.6



HYSTERESIS CALIBRATION IN THE M/L PLANE
FOR BENDING
KEY
+ r_{UY}
⊙ r_{LY}
3000
2500
2000
1500
1000
500
0
M/L Bending moment readings
Left for two hours
with load applied
0
85
85
85
85
85
85
85
Loading sequence (lbf.)

fig. 3.7

and the pylon surface. Since the cycle time in walking is of the order of one second, these relatively long-term effects are of no importance. For the cases when the pylon is loaded for more than this, e.g. standing on the prosthesis, the drift is assumed to be linear and by taking a zero reading before and after the test, it can be allowed for in the calculations.

Tightening of the screw attaching the pylon to the stanchion was found to affect the zero position of the lower level bending moment gauges only. It was decided, therefore, to carry out the final calibration of the bending moment gauges after the foot had been attached and firmly "bedded-in".

By subjecting the pylon to one type of loading at a time, the cross effects can be accurately determined. As discussed in Chapter 4, the equations used to calculate the true force actions are 4.1 to 4.6 and these can be rearranged to give:-

$$r_{UZ} = a_{11} M_{UZ} + a_{12} M_{UY} + a_{15} f_x + a_{16} T \quad \ldots\ldots \tag{3.1}$$

$$r_{UY} = a_{22} M_{UY} + a_{21} M_{UZ} + a_{25} f_x + a_{26} T \quad \ldots\ldots \tag{3.2}$$

$$r_{LZ} = a_{33} M_{LZ} + a_{34} M_{LY} + a_{35} f_x + a_{36} T \quad \ldots\ldots \tag{3.3}$$

$$r_{LY} = a_{44} M_{LY} + a_{43} M_{LZ} + a_{45} f_x + a_{46} T \quad \ldots\ldots \tag{3.4}$$

$$r f_x = a_{55} f_x + a_{51} M_{UZ} + a_{52} M_{UY} + a_{56} T \quad \ldots\ldots \tag{3.5}$$

$$rT = a_{66} T + a_{61} M_{UZ} + a_{62} M_{UY} + a_{65} f_x \quad \ldots\ldots \tag{3.6}$$

If a bending moment is applied in the Z plane only to the pylon, and neither axial load nor torque, the readings from the various bridge circuits, r, using equations 3.1 to 3.6 are:-

$$r_{UZ} = a_{11} M_{UZ}$$

$$r_{UY} = a_{21} M_{UZ}$$

$$r_{LZ} = a_{33} M_{LZ}$$

$$r_{LY} = a_{43} M_{LZ}$$

$$r f_x = a_{51} M_{UZ}$$

$$r T = a_{61} M_{UZ}$$

and since M_{UZ} and M_{LZ} are known, these particular interaction coefficients may be determined. A similar procedure was adopted to find the other coefficients, applying bending moment in the Y plane a pure torque and a pure axial load.

The bending moments and torque were applied to the pylon while it was attached to the stanchion as shown in Fig.3.5. To apply a pure axial force, a simple calibration device was constructed and its use is demonstrated in Fig.3.8. The location at the bottom of the pylon is by means of a steel ball which fits into the conical recess machined in the aluminium end pieces of the pylon. At the top of the pylon, a tapered brass plug is located in a hole which is concentric to the O.D. of the aluminium boss on which the extension tube is attached. This ensures that the line of action of the load is along the x axis of the pylon. The load is applied through a linear ball bearing to ensure minimum friction effects.

The interaction coefficients were determined using exactly the same instrumentation as was used during the amputee tests.

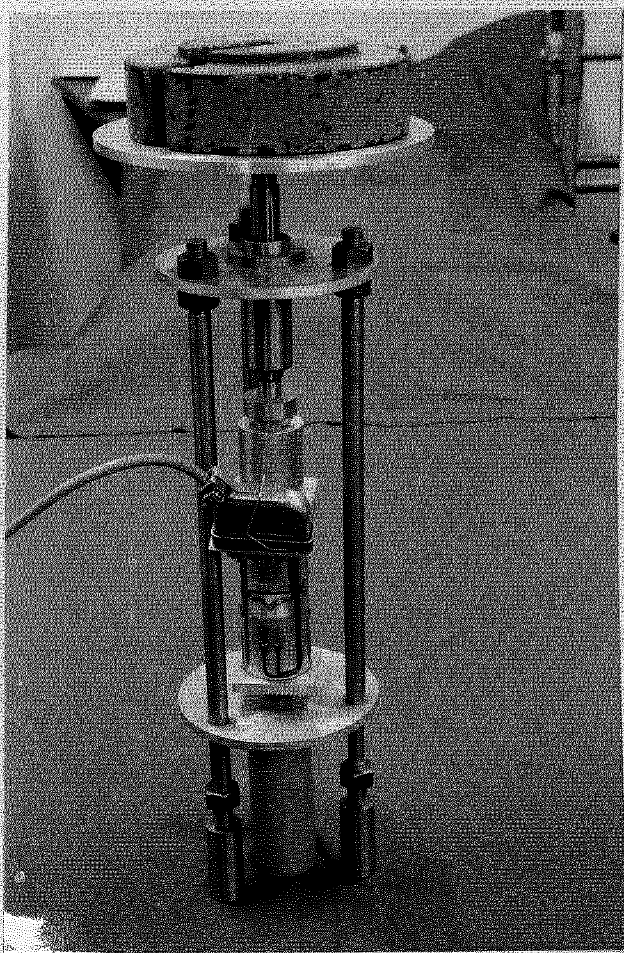


Fig. 3. 8

To carry out final calibration of the bending moment gauges, extensive use is made of the Static Stability Rig described by MacGregor (1968). This allowed various combinations of load action to be applied to the pylon when it is actually attached to the foot at one end, and a boss at the other, as shown in Fig.3.9.

The goniometer was found to be linear with angle over the range within which it was expected to operate, i.e. 0 - 90°. At the time of amputee evaluation, a reading is taken with the angle at 0° and one with 90° of flexion. The reciprocal of the difference between these two is used in the computer programme for the calibration factor.

3.5 Outline of Test Routine

The amputee was fitted with a quadrilateral suction socket. A description of this type of socket can be found in Foort (1963). Although the original sockets were made of wood the modern quadrilateral socket is constructed from laminated epoxy resin.

The strain-gauged pylon is attached to a SACH foot of the correct size for the amputee by a $\frac{3}{8}$" B.S.W. Allen head set screw. A shoe is put on and the knee unit to be investigated is secured to the upper end of the pylon with a length of tube between the two. The height can be set by a suitable tube length. "Jubilee" clips are used to attach the tube to the pylon and knee units.

The socket is joined to the knee unit generally, by means of a "Staros-Gardner" alignment device. When carrying out tests on the Blatchford modular system, the socket is fitted directly to the built-in alignment mechanism.

CALIBRATION USING THE STATIC STABILITY RIG

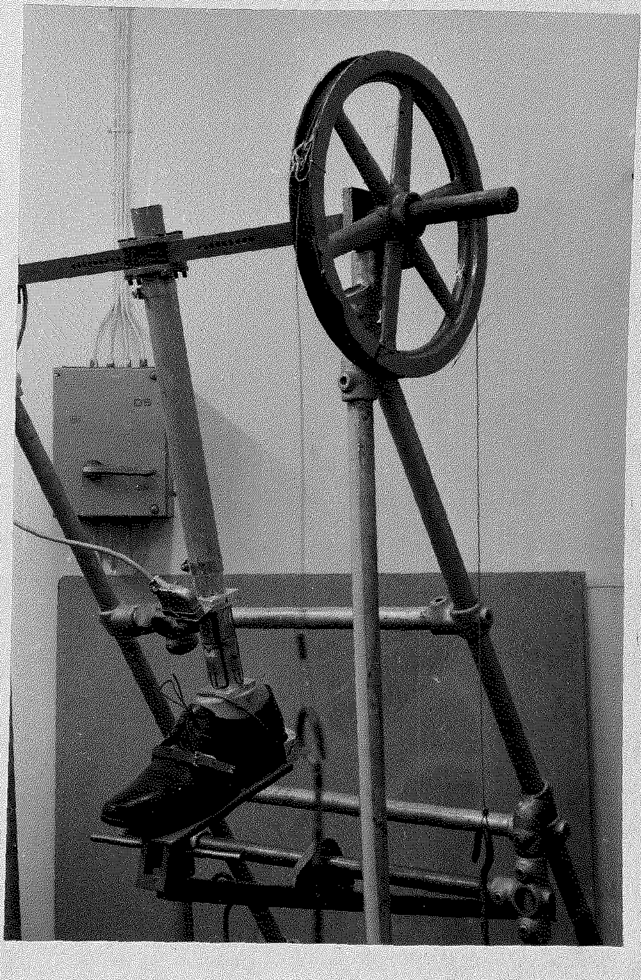


Fig. 3.9

Once the leg is assembled and the total length from the ankle to the ischial seat of the socket is adjusted correctly, it is roughly aligned before being fitted to the amputee. Final alignment and adjustment to the knee mechanism is carried out using conventional methods of amputee assessment, that is by watching the gait and asking the amputee if any adjustment is an improvement. The complete set of tests is carried out with the optimum alignment based on these considerations. The amputee is asked to walk up and down to get used to the particular leg involved.

A subjective impression of the knee mechanism is obtained by asking the amputee to speak about the device into a portable tape recorder. At the same time he is asked if the socket is comfortable, if the stump is properly in and if there is any pain.

The goniometer is attached and checked to ensure free movement. This is especially important in the polycentric types of knee, where movements of an inch are possible between the upper arm of the goniometer and the trochanter position on the socket from the hyperextended position of the knee to the position when the amputee is seated. This results in an error in the thigh length of one inch but as demonstrated in Appendix V, this is insignificant compared with that due to other factors. The method of attachment varies with the type of knee under consdideration but the general arrangement can be seen in Fig.3.10.

The amplifiers are switched on at least an hour before the tests commence. This ensures stabilisation of the strain-gauge bridge voltage. Although the pylon is connected to the amplifiers for this warming up

AN ABOVE-KNEE PROSTHESIS WITH THE GONIOMETER AND STRAIN-GAUGED PYLON INCORPORATED

Fig. 3. 10

period, it is disconnected during the alignment procedure since the lead tends to become twisted after a while due to movement of the amputee. It is reconnected just prior to the test and fastened securely to the socket with adhesive tape. The goniometer lead is likewise secured and the two cables are led up to an overhanging beam which pivots about a fixture on the wall.

The amplifiers are connected to the ultra violet galvanometer recorder. The bridge voltages are noted and the gains on each amplifier adjusted to their calibrated value.

To facilitate the taking of a zero reading in the test procedure and for balancing the bridge, a simple standard method was adopted. The amputee was asked to raise the artificial leg off the ground and let the shank hang vertically. The only difference noted on the recorder between this reading and a true zero reading, i.e. with the pylon disconnected from the leg and standing on the floor, was a slight reduction in the axial load gauges due to the weight of the foot and shoe. This resulted in a 3 mm. difference on the U.V. recorder trace. An adjustment of this amount was made to all the axial load zero readings.

Balancing is not as critical with the arrangement used as it is with some systems. The important parameter with regard to linearity is the input impedance of the device connected to the bridge. By using amplifiers with a high input impedance, the problem of connection to the low impedance galvanometers (approximately 50 ohm.) is obviated.

The galvanometers are set such that the spots of light are at specific positions along the paper which correspond to those used in the calibration.

They are adjusted to give a sharply focused spot, and the amputee is then asked to stand on the leg and a check is made to see that the pylon is working correctly.

The U.V. recorder controls are checked to make sure that the paper speed is set at 200 mm/sec and the time markers at 0.01 sec. It is worthwhile to ensure that the "trace" and "grid" intensity are high enough to get a good image by running a foot or so of paper out at the setting used.

During the actual test it is an advantage to have an assistant who can pull the overhead beam round so that the wires do not interfere with the amputee. This is not essential as one person can operate the U.V. recorder by remote control. He can then walk along with the beam and control the U.V. recorder at the same time.

A trial walk is first carried out before preparing for the actual test. Zero readings are taken before and after each different activity.

The activities investigated are:-

(1) Level walking

(2) Walking up and down a ramp

(3) Walking up and down stairs

(4) Standing up and sitting down

(5) Lifting and lowering a weight

(6) Stepping over an object

(7) Walking sideways

(8) Running

The first test to be studied is level walking. A walk of at least six strides on the artificial leg is aimed at. The U.V. recorder is started and the amputee asked to commence the walk. At the end of the test, the recorder is stopped and, after rechecking the zero reading, the amputee sits down and rests while the U.V. trace develops in the room light. The result is checked and if, for some reason, it is not satisfactory, the test is repeated.

Generally a rest of about five minutes is taken by the amputee whilst preparations are made for the next activity.

After level walking, the gait is investigated when the amputee walks up and down ramp. This particular incline has a gradient of 1 in 7.

Performance whilst ascending and descending a staircase is examined using stairs which have a rise of 7 inches and a length of 7 inches. It should be pointed out that when descending the stairs, the amputee, in these tests, placed the artificial leg on the next step and this was followed by the normal leg to the same step. Normally he would descend step over step as a normal individual, but as the pylon is not very well protected against hard knocks, it was felt safer to adopt this particular technique.

Standing up and sitting down starts with the amputee in his normal seated position. He is asked to practice once the act of standing and sitting to see how much he can use the leg, then after sitting normally again the test is repeated with the recorder switched on. The activity is carried out in one smooth movement with the amputee standing up straight only momentarily before sitting again.

The lifting of the weight is carried out with a piece of turned steel shown in Fig.3.11 and weighing 14 lbf. Once again a practice run is completed before any reading is taken, and the movement is smooth as with the last activity. The amputee is standing comfortably, he bends down, picks up the weight, stands up straight, lowers it to the floor again, releases it and stands up straight.

One of the most difficult activities for an A/K amputee is stepping over an object. The object used in these tests is a block of wood, 5" wide x 6" high. The amputee is asked to stand comfortably at a distance from the wood which suits him best, and when told to, steps over the piece of wood. The difficulty is that the amputee has to put his artificial leg over the object first and bring his normal leg over after. Normally an amputee would choose to use his normal leg first, for example in climbing stairs. However, if he did this when stepping over something, the artificial leg would almost certainly catch on the object and cause him to stumble.

Not only does the amputee stand in front of the piece of wood and step over it, but he is asked to walk up to the object from about twelve feet away and then step over it. This is to examine the way an amputee adjusts his stride so that he gets to the object with the correct leg on the ground. The most significant step is still the one after he has stepped over the piece of wood.

The next activity to be investigated is walking sideways with first the artificial leg leading and then with the normal leg leading. The difference between the two methods can be appreciated and any problems encountered in walking sideways seen.

THE WEIGHT USED FOR THE "LIFTING WEIGHT" TEST

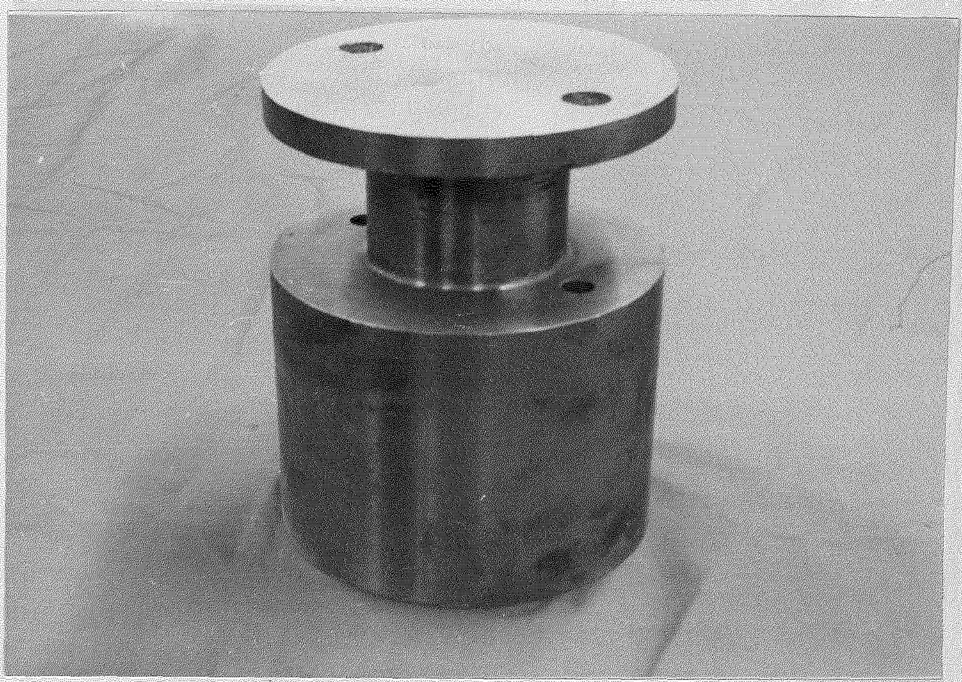


Fig. 3.11

In the case of running, an amputee has to skip on his normal leg to allow time for the artificial leg to swing through.

As part of the research into the amputee performance, it is essential to assess the actual capability of the amputee so that a comparison can be made between what hip moment is used and the maximum hip moment available. Ideally the capability should be investigated over a range of hip flexion angles. The knee is locked by means of a wooden stay and the amputee asked to stand on both legs with his artificial leg approximately at the angle of hip flexion under study. He then extends and, if possible, flexes the hip as hard as possible against the resistance of the floor. The range studied was from 20° of hip flexion to 20° of hyperextension. The angle of hip flexion was measured by a goniometer on the shank. The method is illustrated in Fig.3.12.

3.6 Reduction of Data

3.6.1 Preparation of input data for computer Once the U.V. traces have been obtained of the strain gauge and goniometer signals, this information must be put into a form suitable for analysis by the computer. The three possibilities are punched cards, magnetic tape and punched paper tape, but all three demand digital information. Punched paper tape was chosen as the most convenient to handle and check in the event of error.

Two data tapes are specified in the programme, one containing information for a series of tests, i.e. calibration factors, dimensions, scale factors etc., the other consisting of actual input readings. The first tape is prepared by means of a Flexawriter in the usual way.

The second tape consists of the readings from the U.V. recorder traces. A machine, the D-mac Pencil Follower, was used to produce this rapidly

THE ARRANGEMENT USED FOR HIP MOMENT CAPABILITY TEST

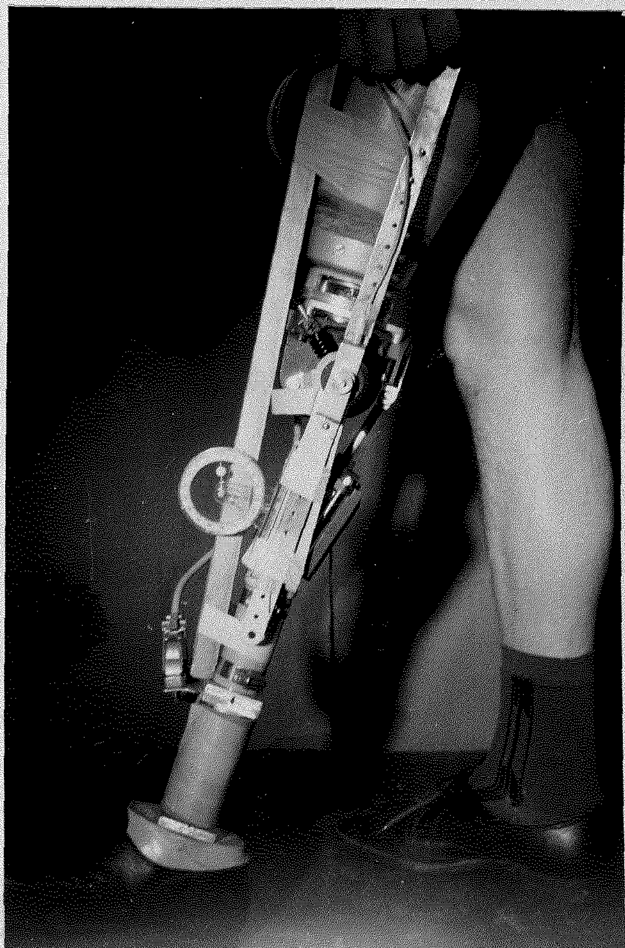


Fig. 3. 12

and without excessive reading errors.

It consists of two main units, the Reading Table and the Electronics Console. The pictorial information to be analysed is placed onto the surface of the Reading Table. Analysis is effected by following the trace with the Reading 'head'. An automatic mechanism beneath the table surface follows the 'head' accurately and, on depression of a switch, position signals are passed to the Electronics Console, where they are visually displayed and converted into suitable form for feeding the output device, a 8 channel tape punch suitable for the ICT 1900 computer. The reading surface is a sheet of plate glass covered with a hard white plastic surface on which the drawing is placed or the image is projected, of working area 100 x 45 cms (40" x 18").

The Reading Head consists of a coil which is inductively coupled to a detector head which is mounted on a trolley on a gantry beneath the reading surface.

The U.V. paper record is laid on the table so that the bottom edge of the paper is approximately parallel to the bottom edge of the table, and weights placed at the corners to hold it in position. The record must be placed centrally on the table and a check made that none of the points whose position is required lies outside the range of the machine. The final levelling of the record is achieved by aligning the extreme ends of one of the straight lines produced on the paper by the U.V. recorder, within the accuracy of the D-mac, i.e. 0.1 mm. A quick check that the centre of this particular line chosen also has the same y coordinate should be made, since it is possible for the paper to move sideways, relative to this line when it is coming out of the recorder. The record is now taped to the table securely and the levelling rechecked.

Before readings are taken, the number of intervals to be fed in, n, and the cycle time in tenths of a second are typed out on the keyboard. The counter on the console is zeroed and the "position" made of operation selected.

Initially the readings of all the traces were taken for each of the time intervals in turn beginning with the zero reading. It was found that with this technique numerous errors were made due to mistaken identity of the traces involved. The computer programme was then altered to allow it to store all the information for one stride, rather than store only that for one interval. The data could then be read in a channel at a time with the first reading being the zero. The information is then reorganised in the programme and the values for each interval made available in the correct order. Not only did this technique reduce the errors to a negligible amount but the time taken was cut from approximately one hour per stride to fifteen minutes.

It is worthwhile leaving about a foot of blank tape between the data from each set of readings. If the total shown by the console is different from n + 1 (the zero reading is included), the whole channel is simply remeasured and the tape spliced.

3.6.2 Representation of the output from the computer   The output force actions are required to be displayed in a manner such that they can easily be understood. A graph of the relevant parameter with respect to time is probably the most meaningful method of achieving this.

It was decided to make use of the graph plotting facility available on the Strathclyde computer. The scaling factors are fed in on the first data tape, they are adjusted so that the graphs are a convenient size and the headings and titles added.

Care must be taken when using the graph plotter that sufficient safe-guards are incorporated in the programme to allow the results to be plotted. In effect the value of the output is limited by the physical size of the graph paper and not, as is usual, by the computer. If the plotter does fail due to an attempt to plot an excessively high value, it will not continue and the rest of the output information will be lost.

The force actions to be plotted are stored in array "plot (j, i)" and are plotted off line to save computer time.

3.7 Socket Prescription

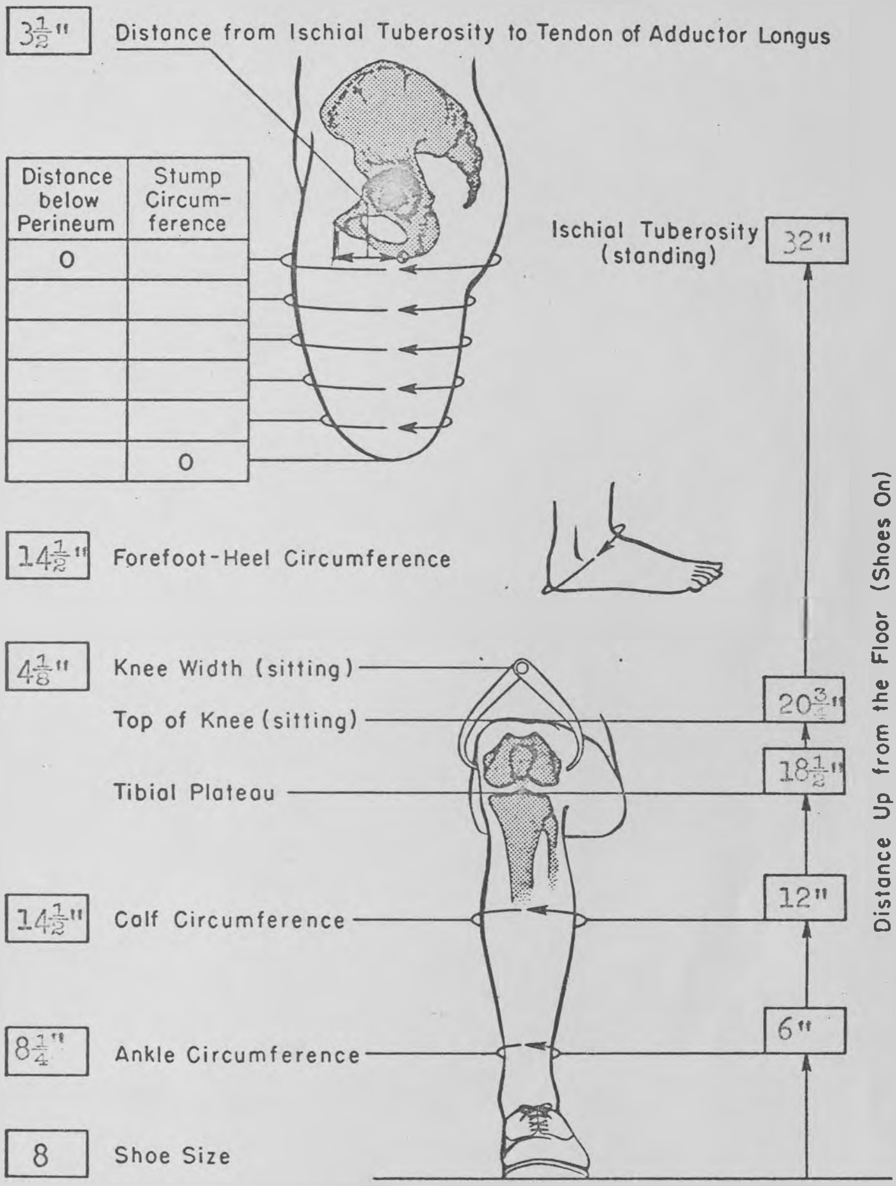
Information relevant to the fitting of the prosthesis is completed on the University of California Form 7B and inserted as Fig.3.13.

A description of the relevant stump details is given in Form 7A (Fig.3.14). Photographs of the lateral, anterior, posterior and inferior aspects of the stump are presented as Fig.315 to Fig.3.18.

PROSTHETIC INFORMATION FROM THE AMPUTEE

Amputee ________________ Date ________________

Right or Left Amputation LEFT Prosthetist ________________



UNIVERSITY OF CALIFORNIA
PROSTHETIC DEVICES RESEARCH

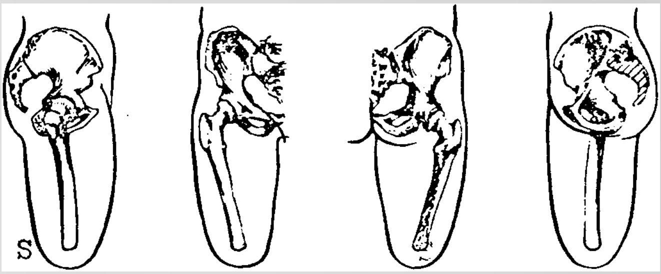
FORM 7B

Fig.3.13

STUMP DESCRIPTION OF THE AMPUTEE

Amputee __________ Date __________
Height 5' 9½" Prosthetist __________
Weight 162 lb.
Age 28
Sex Male

Show location of stump details, identifying with letter code:



Lateral Anterior Posterior Medial

- A. Abrasion
- B. Boil
- BS. Bone Spur
- D. Discoloration
- E. Edema
- I. Irritation
- M. Muscle Bunching
- P. Pressure Point
- R. Redundant Tissue
- S. Scar
- T. Trigger Point

Ischium: Toughened (X) Pressure sensitive ( )
Muscle padding (X) Prominent ( )
Previous ischial bearing? yes(X) no( )

Subcutaneous tissue: Heavy (X) Light ( )
Skin condition: Tough (X) Thin ( )

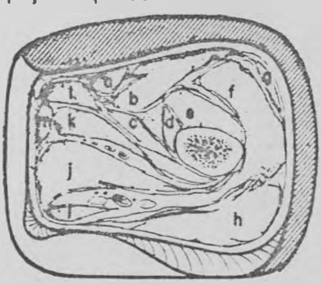
Stump lateral contour: Convex outward (X) Flat ( ) Concave inward ( )

Stump Musculature	Soft	Average	Hard	
General		X		
Hamstring Group			X	Prominent with stump extension? YES
Gluteal Group			X	Prominent with stump extension? YES
Rectus Femoris		X		Prominent with stump flexion? YES
Adductor Longus		X		

Stump flexion contracture: 5 degrees; abduction contracture: 0 degrees

Show modification of basic socket shape, if required:

Muscle code:
- a. Sartorius.
- b. Rectus femoris
- c. Pectineus
- d. Vastus medialis
- e. Vastus intermedius
- f. Vastus lateralis



- g. Tensor fasciae latae
- h. Gluteus maximus
- i. Hamstrings
- j. Adductor magnus
- k. Adductor brevis
- l. Adductor longus
- m. Gracilis

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PROSTHETIC DEVICES RESEARCH

FORM 7A

Fig. 3.14

THE AMPUTEE'S STUMP - LATERAL ASPECT

Fig.3.15

Fig.3.16

THE AMPUTEE'S STUMP - POSTERIOR ASPECT

Fig. 3.17

THE AMPUTEE'S STUMP - INFERIOR ASPECT

Fig. 3.18

CHAPTER 4

THEORETICAL ANALYSIS

CHAPTER 4

4.1. Introduction

The strain gauge and goniometer data is obtained in the form of curves on recorder paper together with time interval markers. It is now necessary to calculate the required force actions from this information. Due to the large number of readings involved, a computer was used to evaluate these parameters and produce the output in the form of graphs. A technique for comparing one device with another in terms of the effort produced at the hip joint is suggested. A statistical analysis of the more relevant force actions is then presented. Finally, an assessment of the inaccuracies involved in measurement and calculation is made.

4.2. Derivation of the Basic Equations

It is convenient to define two coordinate systems. The pylon orientated system is referred to as x, y, z, and the auxiliary system as x', y', z'. Figs. 4.1 and 4.2 show the axes of the two sets of coordinates.

The long axis of the pylon is defined as the x axis.

The line joining the centre of pressure on the foot, C, with the projection of the hip joint, H, on the Z plane of the pylon, H', is defined as the x' axis.

The upper case letters X, Y and Z refer to planes while x, y, z are reserved for directions.

Since both sets of coordinates lie in the Z plane, Z' is the same as Z, and z' the same as z.

LATERAL VIEW OF FORCE ACTIONS AND COORDINATE SYSTEMS

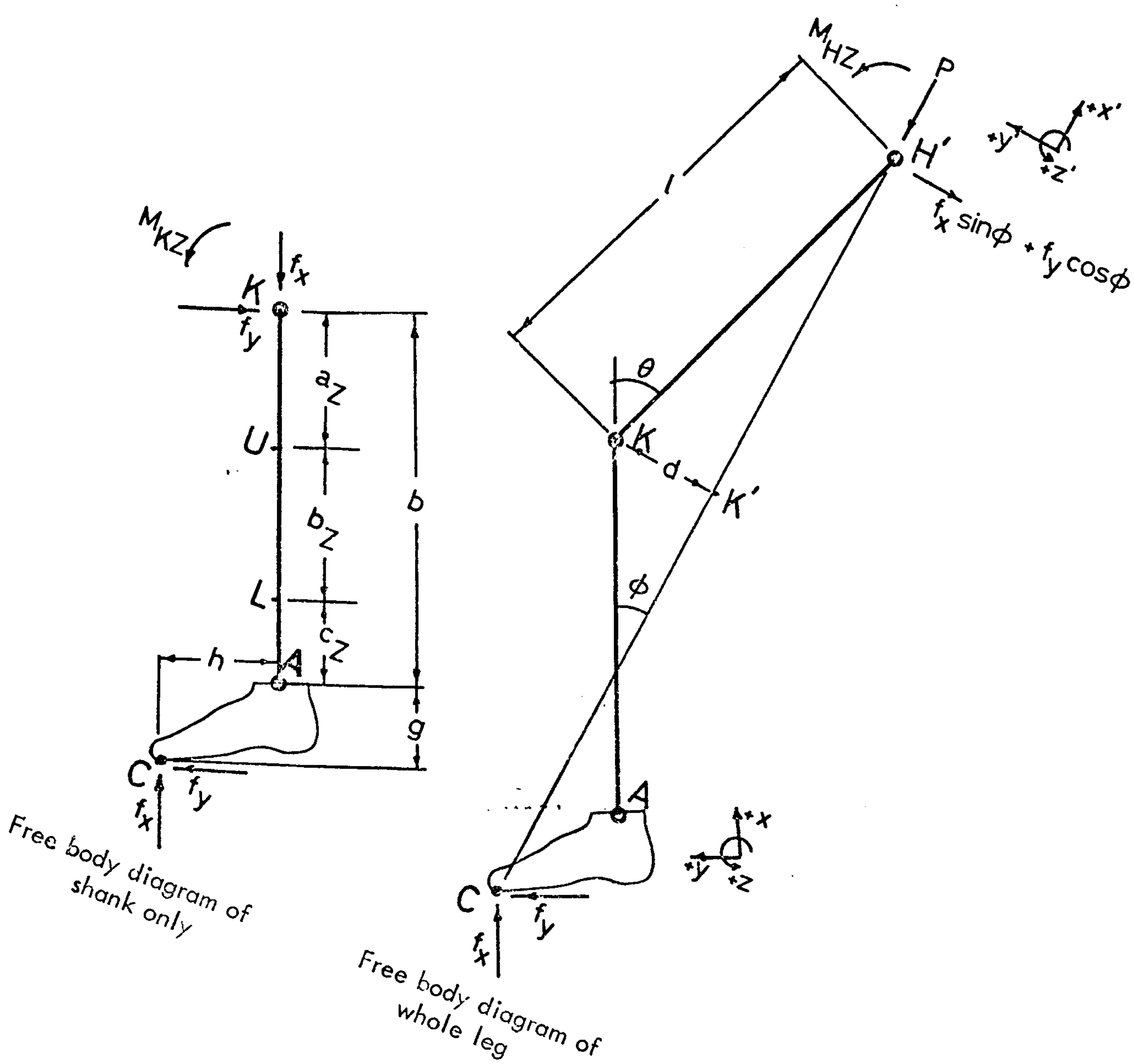
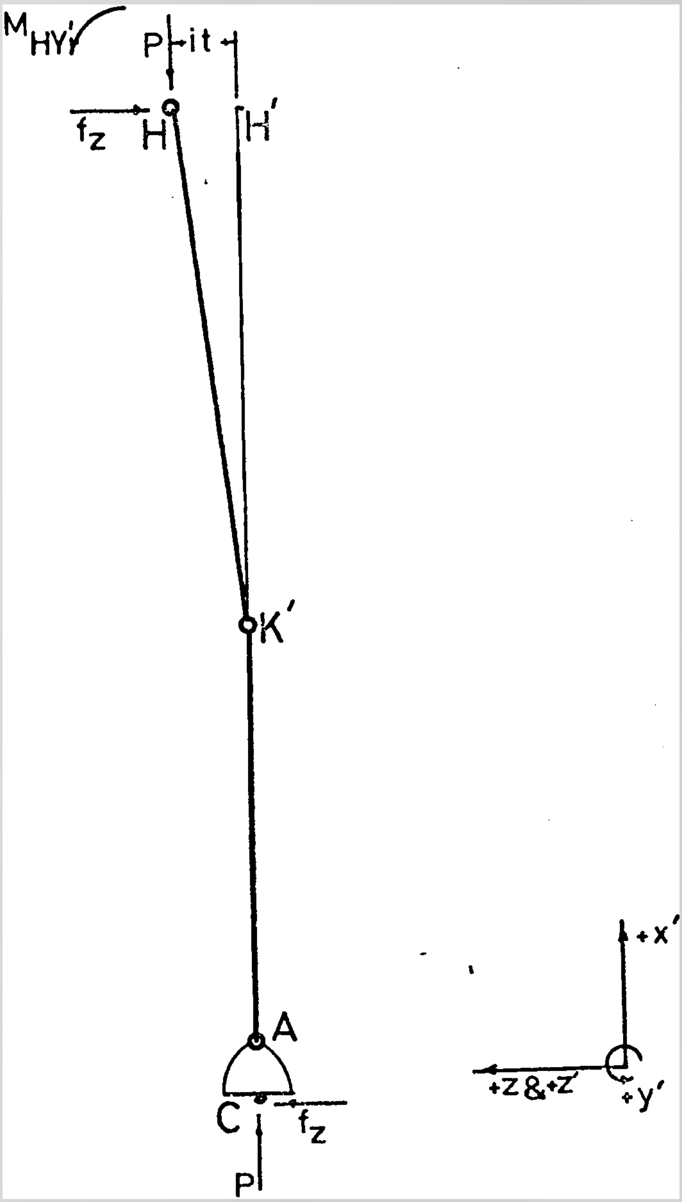


fig. 4.1



Free body diagram of leg in Y' plane

Fig. 4.2

The anterior/posterior or A/P plane is defined as the Z plane.

The mediolateral or M/L plane is defined as the Y' plane.

In the tests performed by MacGregor (1968) on the static stability rig, the coordinate system used corresponds to the auxiliary system defined above.

Referring to Figs. 4.1 and 4.2 the force actions required are:-

M_{KZ} the knee moment in the Z plane i.e. the A/P knee moment.

M_{HZ} the hip moment in the Z plane.

M_{AZ} the ankle moment in the Z plane.

R the total resultant load acting on the foot.

T the torque about the pylon axis.

$M_{HY'}$ the hip moment in the Y' plane i.e. the M/L hip moment.

M_{HRZ} the hip moment required to stabilize the knee in the Z plane assuming it is completely free and single axis.

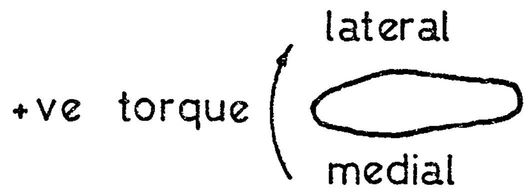
The sign convention used throughout is shown in Fig. 4.3.

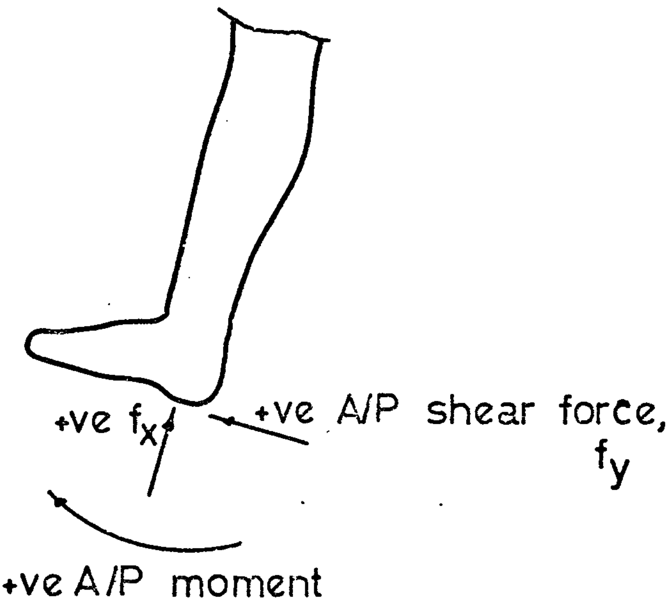
It is possible that a particular bridge circuit will respond to more than one load action due to minute and unavoidable misalignment of the gauges. Correction for such effects has been incorporated in the computer programme.

Considering as an example the upper Z plane bending moment gauge reading, r_{UZ}, the main contribution will be due to the upper Z plane bending moment, M_{UZ}. However, there will be an effect from the Y plane bending moment, one from the axial load and finally one from the torque. Therefore, these effects must be subtracted before final division by the calibration factor a_{11}, i.e:-

$$M_{UZ} = (r_{UZ} - a_{12} M_{UY} - a_{15} f_x - a_{16} T) / a_{11} \quad .... \qquad 4.1.$$

SIGN CONVENTIONS USED





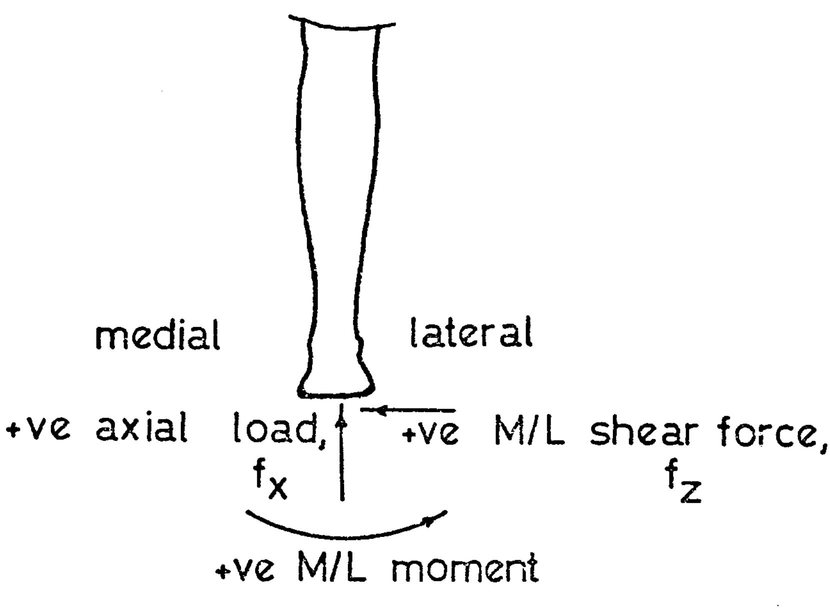


Fig. 4.3

Similar equations can be arrived at for the other parameters as follows:-

$$M_{UY} = (r_{UY} - a_{21} M_{UZ} - a_{25} f_x - a_{26} T)/a_{22} \ldots\ldots \qquad 4.2.$$

$$M_{LZ} = (r_{LZ} - a_{34} M_{LY} - a_{35} f_x - a_{36} T)/a_{33} \ldots\ldots \qquad 4.3.$$

$$M_{LY} = (r_{LY} - a_{43} M_{LZ} - a_{45} f_x - a_{46} T)/a_{44} \ldots\ldots \qquad 4.4.$$

$$f_x = (rf_x - a_{51} M_{UZ} - a_{52} M_{UY} - a_{56} T)/a_{55} \ldots\ldots \qquad 4.5.$$

$$T = (rT - a_{61} M_{UZ} - a_{62} M_{UY} - a_{65} f_x)/a_{66} \ldots\ldots \qquad 4.6.$$

(The factors 'a' are the interaction coefficients considered in Chapter 3)

These equations are solved by an iterative method. Initial values of $M_{UY} = \frac{r_{UY}}{a_{22}}$, $f_x = \frac{r f_x}{a_{55}}$ and $T = \frac{rT}{a_{66}}$ are substituted in equation 4.1. The value of M_{UZ} thus obtained is substituted together with f_x and T in equation 4.2. and a new value of M_{UY} evaluated. The process is continued until the difference in the value of each parameter on successive iterations is less than 0.05 (for M_{UY}, M_{UZ}, M_{LY} and M_{LZ}) and 0.1 (for f_x and T).

The majority of the relevant force actions occur in the Z plane. Referring to Fig. 4.1.:-

The shear force in the y direction is given by:

$$f_y = \frac{M_{UZ} - M_{LZ}}{b_Z} \quad \ldots\ldots\ldots \qquad 4.7.$$

Therefore for the shank section between L and K taking moments about K gives:

$$M_{KZ} = M_{UZ} + f_y . a_Z$$

$$= M_{UZ} + \frac{a_Z}{b_Z}(M_{UZ} - M_{LZ}) \ldots\ldots \qquad 4.8.$$

Similarly for the whole shank taking moments about A gives:

$$M_{AZ} = M_{LZ} - c_Z . f_y$$

$$= M_{LZ} - \frac{c_Z}{b_Z}(M_{UZ} - M_{LZ}) \quad .... \qquad 4.9.$$

To find M_{HZ} take moments about H considering the thigh and shank above U

$$M_{HZ} = f_x . l. \sin\theta + M_{UZ} + f_y (a_Z + l. \cos\theta)$$

$$= M_{UZ} + f_x . l. \sin\theta + \frac{(a_Z + l. \cos\theta)}{b_Z}(M_{UZ} - M_{LZ}) \quad ..... \qquad 4.10.$$

Referring to Fig. 4.2, the leg in the M/L plane:-

The shear force in the z direction is given by:

$$f_z = \frac{M_{UY} - M_{LY}}{b_Y} \quad .... \qquad 4.11.$$

The resultant force R is given by:

$$R = \sqrt{f_x^2 + f_y^2 + f_z^2} \quad .... \qquad 4.12.$$

Referring to Fig. 4.1., if the resultant force in the A/P plane is resolved in the directions x' and y' the component in the x' direction, P, corresponds to that resultant force which would occur if no hip moment were produced. Assuming that the knee joint has no built in device for assisting in its stability under load, it is necessary for the amputee to exert a hip moment, M_{HRZ}, of sufficient magnitude to cause the resultant force to pass through the instantaneous centre of the knee joint to prevent buckling. If the knee is hyperextended, the back check prevents collapse due to extension and thus can be thought of as an assistive device in this configuration.

Let g, h be the coordinates of the centre of pressure from the ankle in the -x and y directions.

If the resultant load is to pass through the knee joint's centre, then the moment about the knee must be zero, i.e.

$$0 = P.d - \frac{M_{HRZ}}{CH'} CK'$$

$$\therefore M_{HRZ} = \frac{P.d.CH'}{CK'} \quad \ldots\ldots \qquad 4.13.$$

where $P = f_x.\cos\phi - f_y.\sin\phi$ ..... 4.14.

$d = (b+g)\sin\phi - h.\cos\phi$ ..... 4.15.

$CH' = h.\sin\phi + (b+g)\cos\phi + l.\cos(\theta - \phi)$ ..... 4.16.

and $CK' = h.\sin\phi + (b+g)\cos\phi$ ..... 4.17.

Now $\tan\phi = \dfrac{h + l.\sin\theta}{b+g+l.\cos\theta}$ ..... 4.18.

considering the equation of foot, taking moments about C

$$f_x.h = M_{AZ} + g.f_y$$

$$h = \frac{M_{AZ} + gf_y}{f_x} \quad \ldots\ldots \qquad 4.19.$$

Examining equations 4.13 – 4.19, the only unknown required to calculate M_{HRZ} is g. This was assumed to be constant at 3.2 in. The effect of this approximation is discussed in Appendix V and the last section of this chapter.

Referring to Fig. 4.2, the mediolateral view, the M/L ankle moment is generally very small and is not shown in this work. The M/L knee moment is also not presented but an appreciation of its magnitude can be made by consulting the M/L hip moment graph. Since the thigh and shank are approximately equal in length, and $M_{AY'}$ is very small, $M_{KY'}$ will be approximately one half of $M_{HY'}$.

To calculate $M_{HY'}$, take moments about H in the Y' plane.

$$M_{HY'} = f_z.CH' - it.P \quad \ldots\ldots \qquad 4.20.$$

where P, f_z and CH' are given by equations 4.14, 4.11 and 4.16.

The moment calculated in each plane is the resultant of the moment due to the hip musculature and the body-socket contact forces. As yet no technique is available for ascertaining the exact location of the point where the load is taken by the body, and therefore the contribution due to the two components cannot be separated.

When attempting to compare the performance of an amputee on two different prostheses, criteria are required. It is suggested that the maximum moments exerted by the hip in extension and flexion are two important parameters, but some assessment of the overall effort produced by the amputee is desirable.

Nubar and Contini (1961) suggest that muscular effort be defined in terms of muscle tension and duration of contraction. The simplest expression suiting these requirements is:- muscular effort (me) = tension x time. This can be expressed in terms of the hip moment as me = moment x time. Since the moment varies at each time interval, Δt, the total muscular effort over n intervals $= \sum_{i=0}^{n} M_i . \Delta t$. To allow for positive and negative moments, the expression is modified to:- the absolute value of $\sum_{i=0}^{n} M_i . \Delta t$ or $\sum_{i=0}^{n} \left| M_i . \Delta t \right|$. This is a very simple quantity to to calculate as the experimental observations are taken at equal Δt intervals.

i.e. $\sum_{i=0}^{n} \left| M_i . \Delta t \right| = \Delta t . \sum_{i=0}^{n} \left| M_i \right|$.

Since strides may not be of constant length with different mechanisms, and the purpose of walking is in fact to get from A to B, presumably with the least effort, a better measure of performance may be the average muscular effort. This can be defined as:- $me_{av} = \frac{\Delta t}{n} \sum_{i=0}^{n} \left| M_i \right|$.

4.3 Layout of Computer Programme

The programme was prepared originally on punched paper tape. Once it had been corrected and test results checked for various extreme values of imput parameters, the compiled version was stored on magnetic tape. This saves compilation time from Algol to machine code each time a set of results is computed.

The layout of the programme is indicated in Fig. 4.4. and a typescript copy is given in Appendix IV.

4.4 Statistical Analysis of Results for Level Walking

Since all the observations and measurements needed to calculate the force actions required involve some degree of experimental error, conclusions based on the magnitude of these parameters must in some way take account of this inaccuracy. It is intended to compare four different criteria between the six knee devices (with the particular alignment used) for the activity of level walking. These are:-

(1) The total muscular effort in the A/P plane per stride, me.

(2) The average muscular effort in the A/P plane per stride, me_{av}.

(3) The maximum value of the A/P hip extension moment per stride.

(4) The maximum value of the A/P hip flexion moment per stride.

To see if there is a significant difference between the knee mechanisms in respect of each of these criteria, the usual analysis of variance technique can be used. Briefly this involves comparing the variation between knee devices with the variation within each knee device for the number of strides considered. This ratio is checked against tabulated values called the

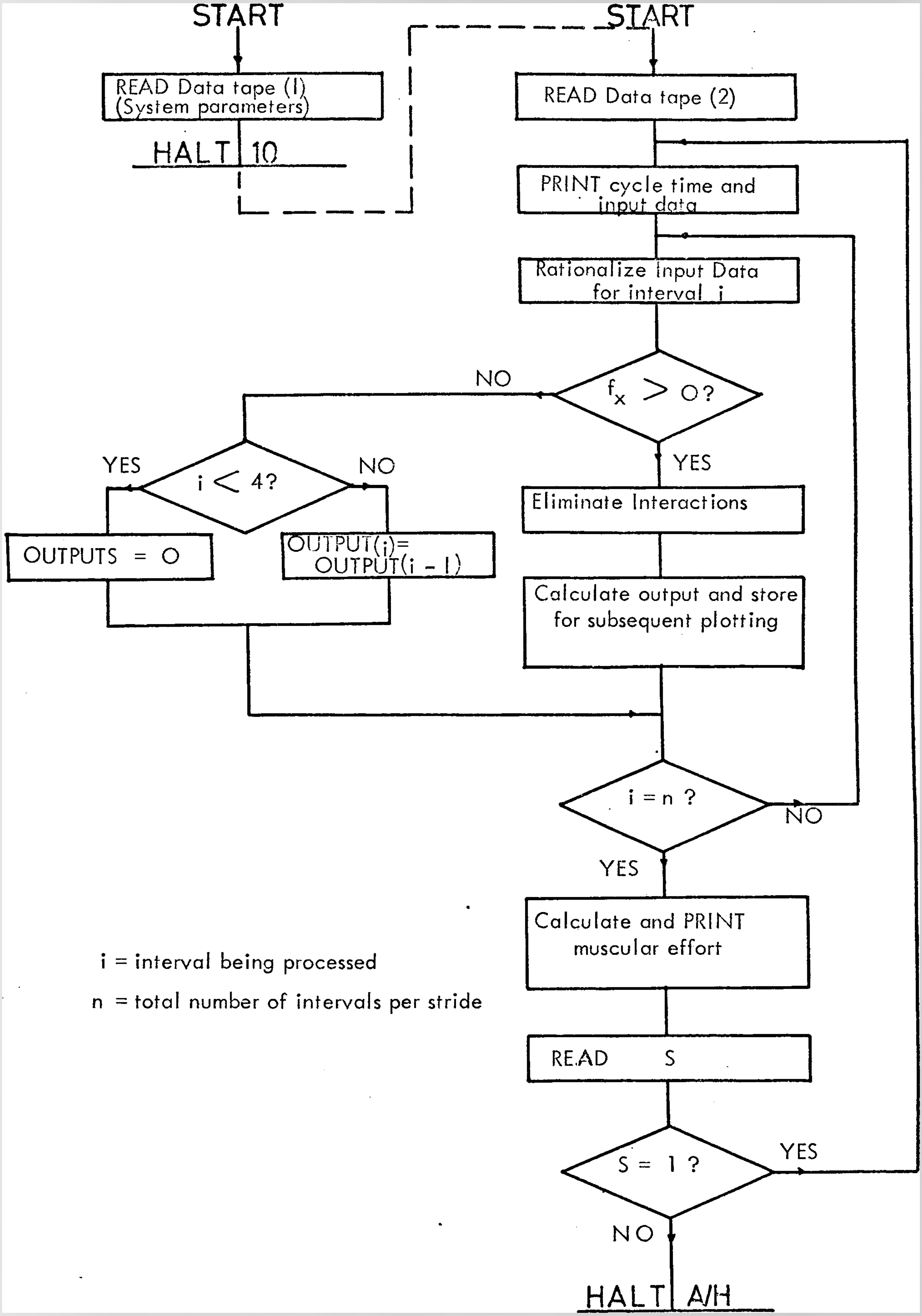

START
READ Data tape (1)
(System parameters)
HALT 10
START
READ Data tape (2)
PRINT cycle time and input data
Rationalize Input Data for interval i
fx > 0?
NO
YES
i < 4?
YES
NO
OUTPUTS = 0
OUTPUT(i)= OUTPUT(i - 1)
Eliminate Interactions
Calculate output and store for subsequent plotting
i = n ?
NO
YES
Calculate and PRINT muscular effort
i = interval being processed
n = total number of intervals per stride
READ S
S = 1 ?
YES
NO
HALT A/H

Fig. 4.4

F distribution. A difference is considered significant if there is only a 5% chance that they are the same. The techniques are explained by Johnson and Leone (1964).

In this particular case, it would be much more useful to be able to compare each mechanism with every other and arrive at a table showing all the possible combinations and whether or not there is any significant difference. Techniques are available for doing this but extreme care must be taken especially if the comparisons to be made are based on the results obtained and not decided before the experiment is conducted. If there are m different groups, m-1 comparisons can be made in a particular way using "orthogonal contrasts" (see Johnson and Leone) without too much difficulty, but strictly speaking the contrasts used should be chosen before the experiment is carried out.

The method used here is that described by Scheffé (1953), although a technique of Tukey (1953) can be used if the replications within groups are equal (i.e. if the number of strides for each knee device was the same). Full details of the methods and limitations are given by Scheffé (1959) and Guenther (1964).

It is necessary to estimate the mean value, m_i of the effect considered for each knee mechanism, by merely averaging the readings of the strides and arrange these in descending order. If the difference between any two means m_1 and m_2 is L then an estimate of L, $\hat{L}$, is given by $\hat{m}_1 - \hat{m}_2$. The spread or confidence limits of L have to be found now. Three levels of significance will be taken, the 5%, 1%, 0.1% levels, i.e. 95%, 99%

and 99.9% chance that the limits given actually contain the true value of L. The level of significance is designated by α and the other parameters are defined below:

m is the number of groups (knee devices).

n_i is the number of readings (strides) in the ith group.

N is the total number of readings taken i.e. $\sum_{i=1}^{m} n_i$.

$F_{\alpha;\, \nu_1,\, \nu_2}$ is the F value (from tables) at the % level for degrees of freedom ν_1 and ν_2.

a_{ij} is the jth reading in the ith group.

t_i is the total of all readings in the ith group i.e. $\sum_{j=1}^{n_i} a_{ij}$.

MSW is the mean sum of squares within groups.

$$MSW = \frac{1}{N-m}\left(\sum_{i=1}^{m}\sum_{j=1}^{n_i} a_{ij}^2 - \sum_{i=1}^{m}\frac{t_i^2}{n_i}\right) \quad \ldots\ldots \quad 4.21.$$

Scheffe has proven that the probability is $1-\alpha$ that all imaginable contrasts will be captured by the set of intervals given by:-

$$\hat{L} - S\,\hat{\sigma}_{\hat{L}} \leqslant L \leqslant \hat{L} + S\hat{\sigma}_{\hat{L}}$$

where $S^2 = (m-1)\ F_{\alpha;\ m-1,\ N-m}$

and $\hat{\sigma}_{\hat{L}}^2 = MSW \sum_{i=1}^{m} \frac{c_i^2}{n_i}$

The array c_i is called a contrast and is used to pick out which of the various combinations of comparisons available one wishes to make. The only constraint on c_i is that $\sum_{i=1}^{m} c_i = 0$. For simply comparing one mean with another, c_i will take the form of 1 -1 0 0 0... or 0 0 1 -1... etc.

To illustrate the technique, assume that knee device 1 is to be compared with knee device 2.

$\hat{m}_1 = \frac{t_1}{n_1}$, $\hat{m}_2 = \frac{t_2}{n_2}$

Therefore $\hat{L} = \frac{t_1}{n_1} - \frac{t_2}{n_2}$

MSW must be calculated according to equation 4.21 and is, of course, constant for all the comparisons throughout any particular effect.

S^2 is also constant, for any particular level of significance, for any criterion.

$\hat{\sigma}^2_{\hat{L}} = MSW \left(\frac{1}{n_1} + \frac{1}{n_2}\right)$ for this case, and the range of L is given by:-

$$\left\{\left(\frac{t_1}{n_1} - \frac{t_2}{n_2}\right) - S\sqrt{MSW.\left(\frac{1}{n_1} + \frac{1}{n_2}\right)}\right\} \leqslant L \leqslant \left\{\left(\frac{t_1}{n_1} - \frac{t_2}{n_2}\right) + S\sqrt{MSW.\left(\frac{1}{n_1} + \frac{1}{n_2}\right)}\right\}$$

From the rationale on which the technique is based, if this interval includes 0 then the two means are not significantly different at this level. Another way of saying the same thing is that if

$$\left|\frac{t_1}{n_1} - \frac{t_2}{n_2}\right| \quad > + \; S\sqrt{MSW\left(\frac{1}{n_1} + \frac{1}{n_2}\right)}$$ then there is a significant difference between means.

Scheffé has shown that although from a theoretical point of view the technique requires random samples from a normal distribution, the results are valid for widely different conditions and hence the levels given are very "safe".

A small computer programme was written to evaluate all possible combinations and list the significance, if any, of the differences. This programme is included in Appendix IV.

4.5 Assessment of Accuracy

Errors introduced into the calculation and hence final results of the calculation can be divided into two main categories, and it is important that the difference between these categories is appreciated. They will be referred to here as systemic errors and random errors, and are defined below.

Systemic errors are those errors which are inherent in the measurement of system parameters.

Random errors are those errors due purely to the inaccuracy of the readings taken from the output data from the pylon and goniometer.

In general, random errors can be tolerated provided the sampling frequency of the data is high enough. Systemic errors result in an overall inaccuracy of any graph drawn as a result of the calculations and any inference made from these graphs. They should, therefore, be minimised to an acceptable level and this level should be determined. Ideally, random errors should also be as small as possible but the demand for their reduction is relatively unimportant.

Before an analysis is made of the particular calculations involved, some general points about sampling frequency should be made. This is a compromise between three factors. Briefly they can be enumerated as follows:

The paper speed, and with it the time of collecting the data.
The store space and time requirements for the computer per stride.
The frequency content of the activity considered.

In any system of sampling frequency, f Hz., the highest frequency that can be derived from the digitised data is f/2 Hz. (Bracewell 1965)). The dominant frequency in walking is about 1 Hz., but U.V. recorder traces obtained indicate that they can be as high as 100 Hz. at heel strike.

When deciding on a sampling rate, the magnitude of the random error must be borne in mind since it forms a frequency component commensurate with the highest component that can be extracted at this sampling rate. This means in practical terms, that the highest frequency that can be relied upon decreases as the magnitude of the random error increases for any specified sampling rate.

A sampling frequency of 100 Hz. was used throughout the work. This was directly available on the U.V. recorder and no additional instrumentation was required to generate it.

The maximum frequency that can possibly be relied on from the data at this sampling rate is 50 Hz., and the author would consider 30 Hz. a more realistic figure. It would have been better to have introduced a high frequency filter into the circuit before recording the data if a purely digital output were taken, to exclude all components higher than say 40 Hz. However, the information was initially recorded in analogue form and is a true record of all frequency components. This allows an examination of the frequency content with respect to the time interval traces. The data read into the computer could be passed through a digital filter in the programme but since the output was in the form of a graph, the trend of the particular force actions is easily recognised. Further, the random errors are not masked and their effect can be appreciated.

To ensure that all the frequency components are transferred to digital form, a sampling frequency of at least 300 Hz. is desirable. It was thought that the additional information made available in this way did not warrant the extra effort required in the data preparation. Moreover the computer time used would be increased accordingly. In fact the time taken to prepare the data using the D-mac is approximately 15 minutes per stride and the computer time is approximately 6 minutes per stride. The graph plotting is carried out "off-line" and is not included in this time.

Ideally the paper speed of the U.V. recorder should be as high as possible so that the intersection of the curves with the time interval traces is clearly defined. Practically a limit is imposed due to the amount of paper used. A compromise of 200 mm/sec. was decided on. If a lower paper speed is used, the time interval traces are too close together to avoid errors of misreading. The minimum distance apart that can be tolerated is about $\frac{1}{10}$ inch.

A check was carried out on the D-mac to establish the maximum reading accuracy that can be obtained. Twenty readings were taken of a single point approaching it from various directions. It was found that the machine's accuracy could be specified as a standard deviation of 0.6 on the y coordinates and 1.5 on the x coordinate. For the use made of the machine here, only the y coordinate is of interest. Further, since the resolution of the D-mac is 1, the random variation is not significant. In practical terms it may be stated that the machine will measure to within 1 unit or 0.1 mm, in the y direction. However this does not mean that the reading error is of this magnitude, since it is influenced by the speed of reading and by the ability to define exactly where the recorder traces intersect the time interval markers. It is assumed that the reading error for zero readings is $\pm$ 0.1 mm. and for other readings is $\pm$ 0.2 mm.

It was decided at the onset of the work to avoid the use of filming, since it introduces more labour in the preparation of data and difficulties of synchronizing pylon and film parameters. This means that unless some system such as described in Appendix III is used, information regarding the configuration of the limb segments with respect to the ground is not available. The effect of using this simplified approach will be briefly discussed.

No account in the analysis is taken of inertia force actions. In the normal individual these represent, according to Frankel and Bresler, a maximum of 10% of the ground-to-foot forces in the stance phase of walking and due to a decreased weight of the leg in an amputee, this effect will be reduced accordingly, since the accelerations are approximately of the same magnitude (U.C.B. report (1947)).

Since the hip angle is not known, no account of the weight of the prosthesis on the A/P hip moment can be taken. The affect of this additional force action is usually to cause the hip moment, calculated from the input data, to be greater than the actual hip moment produced. At heel strike and toe off this difference is a maximum, due to the size of the hip angle, and this can be estimated by using the body weight coefficients given by Frankel and Bresler. The weight of an artificial leg for an above knee amputee is approximately 8lbf. and the foot and shoe constitute about 2lbf. of this. By far the most important contribution is the weight of the stump but assuming that it is about $\frac{3}{4}$ of the volume of a normal thigh its weight would be about 14 lbf. The total weight above the pylon is, therefore, 20 lbf. If the hip angle at heel strike and toe off is 35° and the centre of gravity of the suprapylon weight is 8 inches from the hip joint, the moment due to this weight is given by:-

$$M = 20 \times 8 \times \sin 25^{\circ} = 67 \text{ lbf in.}$$

The technique used to analyse the other inaccuracies introduced by reading errors etc. is detailed in Appendix V. A summary of the various effects and the contributions to them of the different measurements is given in Tables 4.1 to 4.3.

Table 4.1 outlines the random errors, Table 4.2 indicates the overall systemic errors, and Table 4.3 shows the errors involved when two strides are compared, assuming that the length and weight of the leg are the same and that the calibration errors have not changed.

TABLE 4.1

The Random Errors

Errors Assigned to Input Parameters	Random Errors Produced in Output Force Actions						
	M_{AZ} (lbf.in.)	M_{KZ} (lbf.in.)	M_{HZ} (lbf.in.)	$M_{HY'}$ (lbf.in.)	R (lbf.)	T (lbf.in.)	M_{HRZ} (lbf.in.)
M_{UZ}	1.5	10.8	23				
M_{LZ}	2.8	9.4	22				
M_{UY}				12.8			
M_{LY}				12.8			
f_x					0.2		0.8
T						0.4	
θ			20	0.02			
TOTAL	3.2	14	38	18	0.2	0.4	0.8

TABLE 4.2

The Maximum Systemic Errors Involved when Comparing Two Strides

Errors Assigned to Input Parameters	Maximum Erros Produced in Output Force Action						
	M_{AZ} (lbf. in.)	M_{KZ} (lbf. in.)	M_{HZ} (lbf. in.)	$M_{HY'}$ (lbf. in.)	R (lbf.)	T (lbf. in.)	M_{HRZ} (lbf. in.)
M_{YZ}	0.7	5.4	11.5				
M_{LZ}	1.4	4.7	11				
M_{UY}				6.4			
M_{LY}				6.4			
f_x					0.1		0.4
T						0.2	
θ			20	0.02			
TOTAL	1.6	7.1	26	9	0.1	0.2	0.4

TABLE 4.3

The Overall Systemic Errors

Errors Assigned to Input Parameters	Errors Produced in Output Force Actions						
	M_{AZ} (lbf. in.)	M_{KZ} (lbf. in.)	M_{HZ} (lbf. in.)	$M_{UY'}$ (lbf. in.)	R (lbf.)	T (lbf. in.)	M_{HRZ} (lbf. in.)
$M_{UZ} < 250$ lbf.in.	0.7	5.4	11.5				
$M_{UZ} = 1000$ lbf.in.	1.7	12	25.5				
$M_{LZ} < 250$ lbf.in.	1.4	4.7	11				
$M_{LZ} = 1000$ lbf.in.	3. 1	11	25				
$M_{UY} < 100$ lbf.in.				6.4			
$M_{UY} = 500$ lbf.in.				17.2			
$M_{LY} < 100$ lbf.in.				6. 4			
$M_{LY} = 400$ lbf.in.				14.3			
Gauge positions in Z plane	0-1.3	0-8.5	0-19.5				
Gauge positions in Y plane				0-19.4			
Thigh length l			0-6.5	0 - 2			
Shank length b				0 - 2			
f_x			0-1.6		0. 1		0.4
$T < 50$ lbf. in.						0.2	
$T = 90$ lbf. in.						0.3	
Knee Angle θ			0-20	0-0.02			
Assumption that g = constant				0-24			0-6%
TOTAL: MINIMUM	1.6	7.1	16	9	0.1	0.2	0.4
MAXIMUM	3.7	18.3	50	39	0. 1	0.3	6%

CHAPTER 5

ANALYSIS OF RESULTS AND DISCUSSION

CHAPTER 5

5.1 Introduction

The force actions are presented in graphical form in Appendix VI. To illustrate particular aspects of the discussion, certain relevant graphs are also reproduced here.

The graphs of A/P hip moment are presented in order of knee mechanism as follows:-

S.A.	Single-axis, Non-stabilized knee with pneumatic swing phase control (Fig.VI.1 to Fig.VI.16)
B.S.K.	Blatchford Stabilized knee with pneumatic swing phase control (Fig.VI.17 to Fig.VI.31)
O.B.	Otto Bock Safety knee (Fig.VI.32 to Fig.VI.46)
G.R.	Otto Bock Greissinger knee (Fig.VI.47 to Fig.VI.60)
U.C.B.	University of California Polycentric knee with pneumatic swing phase control (Fig.VI.61 to Fig.VI.75)
L.A.	Lammers knee (Fig.VI.76 to Fig.VI.90)

In the case of the activities; standing, sitting down and lifting and lowering a weight, the resultant load on the prosthesis is reproduced adjacent to the A/P hip moment to assist in comparing the effectiveness of the devices.

The remaining force actions are illustrated for the single-axis, non-stabilized knee only, to indicate the typical variations between activities.

A/P knee moment	(Fig.VI.91 to Fig.VI.100)
A/P ankle moment	(Fig. VI.101 to Fig.VI.110)
M/L hip moment	(Fig.VI.111 to Fig.VI.120)

Torque	(Fig. VI. 121 to Fig. VI. 130)
Resultant load	(Fig. VI. 131 to Fig. VI. 140)

In general the discussion will aim at a critical examination of the results of the tests performed together with definitive clincial aspects associated with their interpretation.

5.2 The Definition of Relevant Parameters

For all the activities involving cyclic variations with time, the graphs begin at heel strike and end at toe off. These points are defined as the first and last time interval traces at which the axial load gauges register a positive value of load. The cycle time for any stride is the time between the preceding toe off and toe off of the stride considered.

The beginning and end of all the other activities are defined thus:-

Stepping over an object:	Heel strike and toe off
Standing up:	Initiation and completion of knee extension
Sitting down:	Initiation and completion of knee flexion
Lifting and lowering weight:	Approximate time when weight is first lifted from the floor and finally released

An explanation of the use of the M_{HRZ} results is presented below.

If a single-axis, non-stabilized knee is incorporated in an artificial leg and a force applied between the centre of pressure on the foot and the hip joint, then the knee will collapse in flexion or hyperextension, unless this force passes through the knee centre. This can be prevented by the application of a hip moment by the amputee which causes the line of action of the resultant load to pass through the knee joint. It is this moment that is referred to in the results as M_{HRZ}. Hyperextension of the knee is prevented by a check strap or stop so collapse can only occur in flexion. Therefore, if the value M_{HRZ} is positive, i.e. an extensor moment is required to maintain stability of the knee, the knee will collapse if no hip extensor moment is exerted. Further, when knee flexion is initiated prior to toe off, the hip moment required to cause this is exactly M_{HRZ} and the two curves reproduced on the A/P hip moment graphs should intersect at this point.

5.3 Detailed Presentation of the Results of one Test

The particular test to be presented is level walking with S.A. Variations of the reading from the strain gauged pylon and goniometer with time are depicted in Fig.5.1. The time interval marks are not reproduced but they occur at 2 mm. intervals along the trace. The various parameters are indicated, together with datum lines representing a zero reading of that particular strain gauge circuit.

Table 5.1 shows the digital data from the d-mac pencil follower, for the strain gauge reading r_{LZ}. The y coordinate has a + sign associated with it, and is the only coordinate used in the following computation. Ideally the x coordinate should increase by 20 units for each time interval, and provides a useful check if any data is missing. The discrepancies from this figure are due to two factors. One is the variability of the motor speed in the U.V. recorder causing the time interval traces to occur at irregular distances from each other. The other is an inherent inaccuracy in the d-mac follower represented by a standard deviation of 1.5 units.

The calibration factors and lengths of leg segments are detailed in Table 5.2.

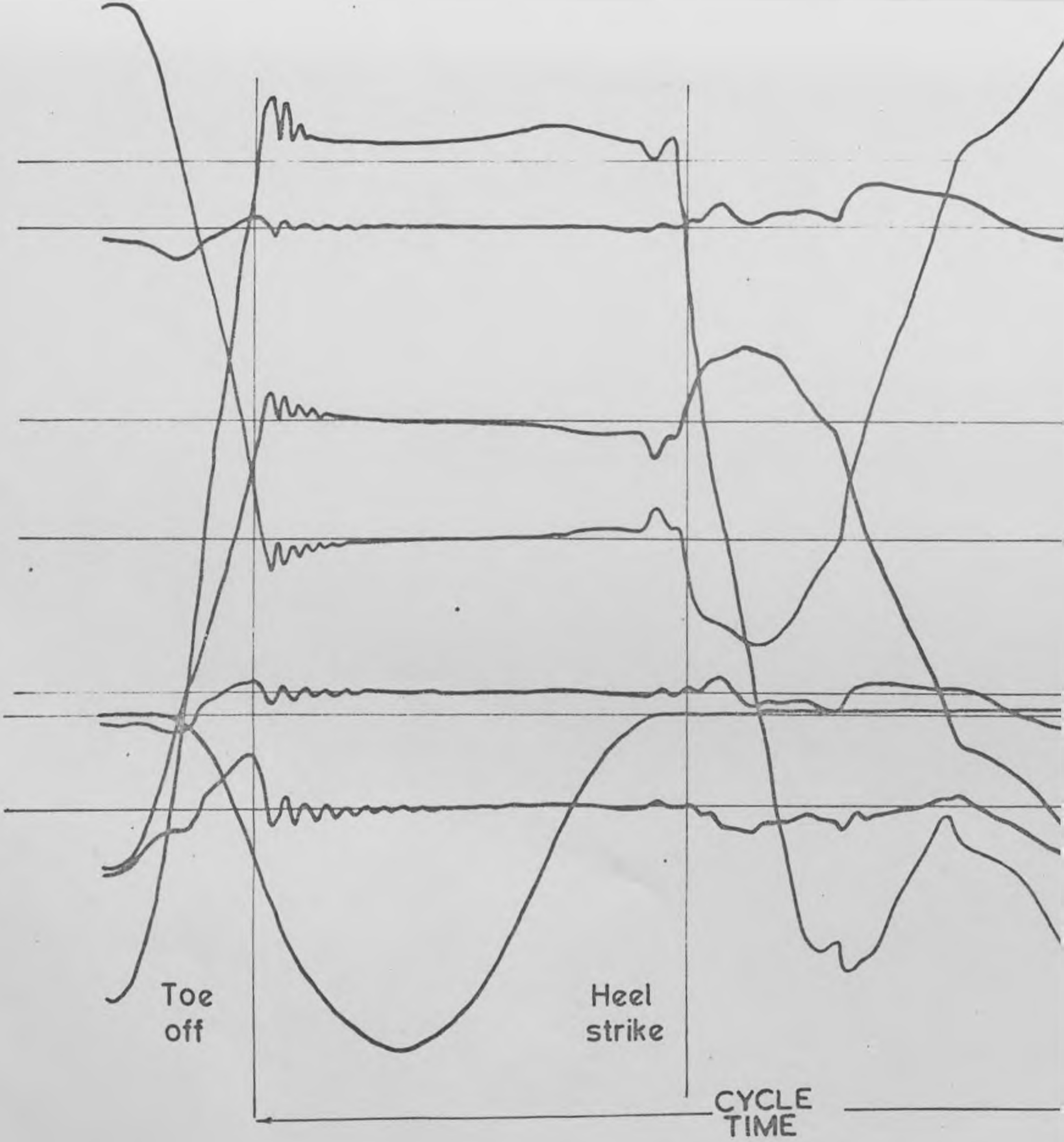
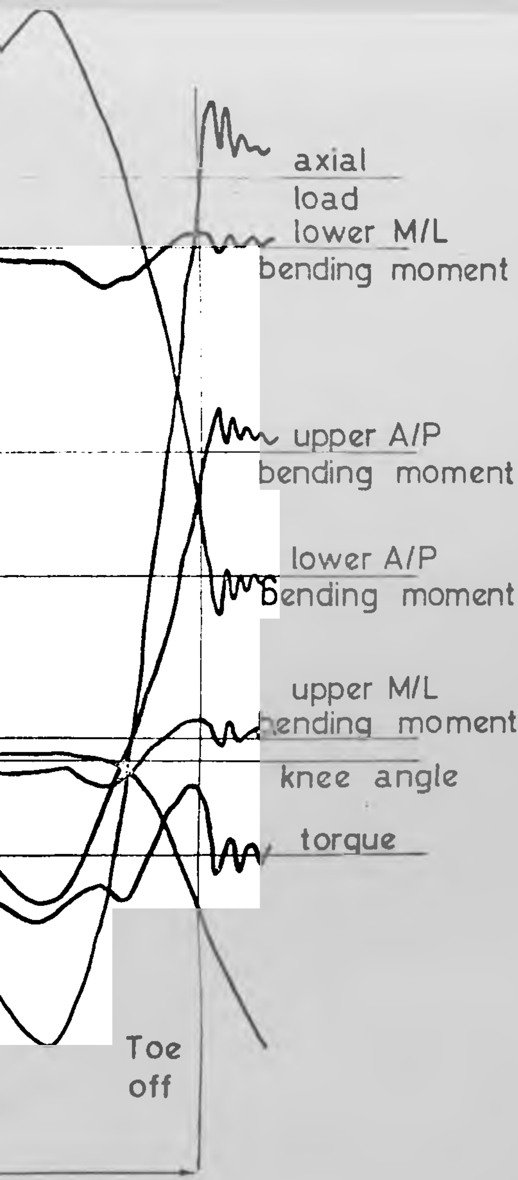


Fig.5.1



TYPICAL U.V. RECORDER TRACE

TABLE 5.1

DIGITAL OUTPUT FROM D-MAC

9513	+1667	3609	+1611	3586	+1525	3566	+1492	3550	+1484
3529	+1477	3503	+1457	3483	+1438	3465	+1429	3446	+1429
3427	+1437	3408	+1451	3389	+1469	3367	+1500	3343	+1541
3326	+1571	3305	+1603	3283	+1632	3264	+1692	3245	+1771
3224	+1857	3205	+1925	3183	+1982	3162	+2041	3143	+2096
3125	+2147	3104	+2195	3082	+2252	3063	+2307	3042	+2371
3023	+2436	3003	+2497	2983	+2517	2965	+2531	2941	+2545
2919	+2564	2900	+2581	2882	+2597	2863	+2618	2844	+2642
2823	+2672	2801	+2709	2783	+2732	2762	+2761	2739	+2797
2724	+2818	2703	+2836	2682	+2850	2662	+2847	2639	+2831
2620	+2797	2601	+2756	2582	+2707	2564	+2651	2542	+2596
2523	+2543	2500	+2471	2479	+2381	2459	+2293	2437	+2208
2413	+2120	2396	+2015	2375	+1912	2354	+1815		

TABLE 5.2

Constants used in the Calculations for Level Walking with a Non-Stabilised, Single-axis Knee

Thigh length, l	=	15.1 in.
Shank length, b	=	16.2 in.
Goniometer calibration factor, c	=	-0.001059 radians per 636 cm. deflection.

Pylon calibration factors, a_{ij} :-

a_{ij}	j = 1	2	3	4	5	6
i = 1	-1.518	0.031	0	0	0	0
2	0	2.242	0	0	-0.093	0.079
3	0	0	1.680	0	0	0
4	0	0	0	3.160	-0.036	0.065
5	-0.156	0.345	0	0	-10.96	0.651
6	0.248	0.056	0	0	0	4.68

The units of a_{ij} are either mm. deflection per 10 lbf.in. or mm. deflection per 10 lbf.

To illustrate the method of calculation, the value of M_{HZ} will be calculated at the points of maximum extensor hip moment and maximum flexor hip moment.

Equation 4.10 gives:

$$M_{HZ} = M_{UZ} + f_x . l . \sin\theta + (a_Z + l . \cos\theta)(M_{UZ} - M_{LZ}) / b_Z$$

For maximum hip extensor moment

Equation 4.1 gives:

$$M_{UZ} = (r_{UZ} - a_{12} M_{UY} - a_{15} f_x - a_{16} T) / a_{11}$$

Since a_{15} and $a_{16} = 0$,

$$M_{UZ} = (r_{UZ} - a_{12} M_{UY}) / a_{11}$$

$$= (r_{UZ} - 0.031 M_{UY}) / -1.518$$

To a first approximation, $M_{UY} = \dfrac{r_{UY}}{a_{22}} = \dfrac{r_{UY}}{2.242}$ (From equation 4.2)

$$\therefore M_{UZ} = (r_{UZ} - 0.014 \; r_{UY}) / -1.518$$

$$= \left[(1955 - 1926) - 0.014 (1297 - 1325)\right] / -1.518$$

$$= (29 - 28 \times 0.014) / -1.518$$

$$= (29 - 0.39) / -1.518$$

The second term represents only 1% of the total and is well below the reading accuracy of $\pm$ 2 units.

$$\therefore M_{UZ} = -\frac{29}{1.518} = -19.1 \text{ lbf. in.}$$

Similarly $M_{LZ} = -\dfrac{96}{1.68} = -57.2$ lbf. in.

and $f_x = \dfrac{1753}{10.96} = 160$ lbf.

$$\theta = -3 \times 1.57 \times -0.001059$$

$$= 0.005 \text{ radians}$$

$\therefore \sin\theta = 0.005$

and $\cos\theta = 1$

$$\therefore\ M_{HZ} = -19.1 + 160 \times 15.1 \times 0.005 + (12.6 + 15.1 \times 1)(-19.1 + 57.2) / 1.69 \text{ lbf.in.}$$

$$= -19.1 + 12.1 + \frac{27.7 \times 38.1}{1.69} \text{ lbf.in.}$$

$$= -19.1 + 12.1 + 625 \text{ lbf.in.}$$

$$\therefore\ M_{HZ} = 618 \text{ lbf.in.}$$

Similarly for maximum hip flexor moment

$$M_{UZ} = \frac{-771}{-1.518} = 508$$

$$M_{LZ} = \frac{984}{1.68} = 586$$

$$f_x = \frac{1566}{10.96} = 144$$

$$\theta = -16 \times 1.57 \times -0.001059$$

$$= 0.0266$$

$$\therefore\ \sin\theta = 0.0266$$

$$\cos\theta = 1$$

$$\therefore\ M_{HZ} = 508 + 144 \times 15.1 \times 0.0266 + 27.7\,(508 - 586)/1.69 \text{ lbf.in.}$$

$$= 508 + 58 - 1280 \text{ lbf.in.}$$

$$= 714 \text{ lbf.in.}$$

These points together with the rest of the values of A/P hip moment calculated are plotted by the computer and presented here in Fig. 5.2.

In the calculation of the maximum hip flexor moment, errors are introduced due to the subtraction of one large value from another. An estimation of the magnitude of these effects is fully discussed in Appendix V and summarised in Tables 4.1 and 4.3.

The values calculated for M_{HRZ} are also reproduced on the same graphs as the A/P hip moment variation.

TYPICAL COMPUTER OUTPUT OF HIP MOMENT

(LEVEL WALKING WITH S.A.)

cycle time. 1·10 seconds

me. 231·5 lbf. in. sec.

me_{av}. 352 lbf. in.

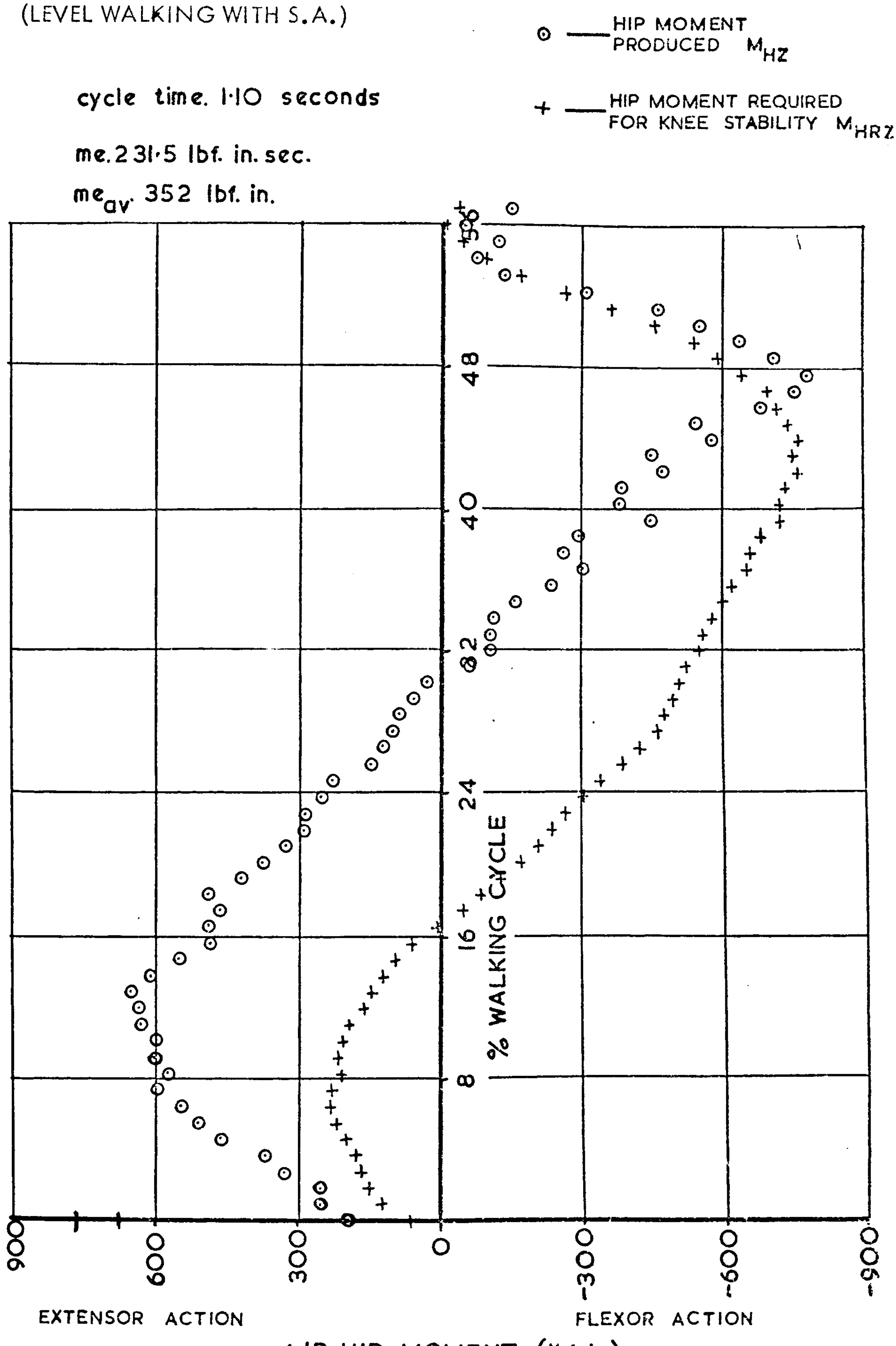


fig. 5.2

5.4 A/P Hip Moment Variations for the Various Knee Devices

These are the most important results obtained from the tests carried out with respect to the evaluation of amputee activity and more attention will be given to them than to other force actions.

The various knee devices investigated are briefly described below.

The Blatchford stabilized knee and Otto Bock safety knee are load sensitive friction devices, the Greissinger knee is a condylar type of mechanism and the Lammers and U.C.B. knees are of a four-bar linkage construction.

The Greissinger, Lammers and U.C.B. assist the amputee to maintain stability of the artificial knee by raising the instantaneous centre of the knee as the knee extends. They are therefore termed polycentric devices. A detailed description of all these mechanisms together with a theoretical analysis of their performance characteristics and practical response to various static load conditions can be found in MacGregor (1968).

5.4.1 Level walking The variation of the A/P hip moment for level walking with S.A. is reproduced as Fig.5.3 to show the general pattern. The main points of interest are discussed here.

Approximately 55% of the cycle is occupied by the stance phase. The hip moment increases to a maximum of +750 lbf.in. at 12% of the walking cycle and falls away, almost linearly, to a minimum of -600 lbf.in. at 47% of the cycle. The point of zero hip moment occurs at 32% of the walking cycle.

On the other hand, the hip moment required to stabilize the knee reaches

A/P HIP MOMENT IN LEVEL WALKING WITH S.A.

——— HIP MOMENT PRODUCED M_{HZ}

— — — HIP MOMENT REQUIRED FOR KNEE STABILITY M_{HRZ}

cycle time 1.10 seconds

me 225.9 lbf. in. sec.

me_{av} 371 lbf. in.

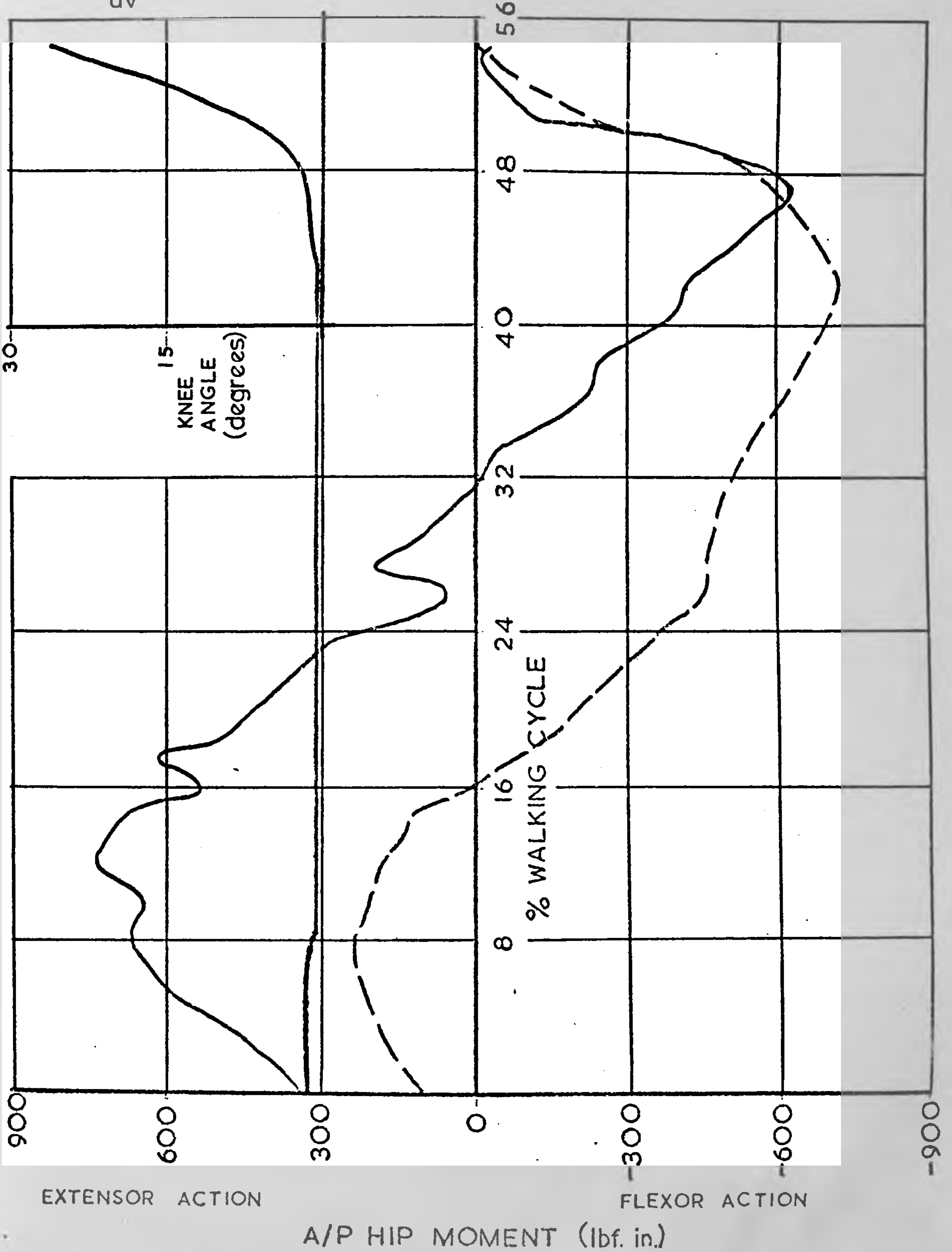


fig.5.3

a maximum of +280 lbf. in. at 8% of the cycle, passes through zero at 16% of the cycle and reaches a minimum of -700 lbf. in. at 41% of the cycle. The two curves intersect at 47% of the cycle.

The hip moment required to stabilize the knee at heel strike is easily achieved by the amputee and he is able to increase his extensor action as the requirement increases. The danger of the knee buckling is present up to 16% of the cycle at which stage, the hip-centre of pressure line passes through the knee joint. Beyond this point, stability is assured without any effort on the part of the amputee.

To exert a hip flexing moment on the normal leg at the initiation of the swing phase, a hip extensor action is required on the prosthesis. This is demonstrated by the results in the persistence of hip extensor action until 32% of the cycle.

The intersection point of the two graphs at 47% of the cycle is the time at which flexion at the knee is initiated. Ideally, the two curves should coincide from this point up to toe off. Not only the static but also the dynamic characteristics of the knee device can, however, influence the hip moment during the latter part of the stance phase, as the knee is flexing.

As far as amputee effort is concerned, the maximum values of hip extensor and flexor moment in association with some indication of their duration are the important parameters. A measure of the latter criterion is given by the muscular effort ($\Delta t . \Sigma M_i$) as discussed in section 4.2. The results of the statistical analysis Table 5.3 to Table 5.6 provide the best means of comparing the various devices for level walking with respect to all these variables. The method adopted of displaying the information is explained on the table concerned but an example will be used to clarify the procedure.

TABLE 5.3

Comparison Between Knee Devices for Muscular Effort in Level Walking

(a) The values of muscular effort obtained for the various knee devices (in lbf.in.sec.)

O.B.	265.3	284.0	251.4	254.5	239.5	236.0	
U.C.B.	241.1	234.8	249.7	242.4			
S.A.	223.1	231.5	225.9	226.4	214.1	213.0	217.7
G.R.	203.4	175.7	192.7	177.6	175.9		
B.S.K.	168.4	185.8	200.0	166.0	177.7	172.5	
L.A.	172.1	151.4	157.7	145.7	143.4	155.1	157.7

(b) The results of the statistical analysis

	O.B.	U.C.B.	S.A.	G.R.	B.S.K.
L.A.	xxx	xxx	xxx	xx	x
B.S.K.	xxx	xxx	xxx	-	
G.R.	xxx	xxx	xxx		
S.A.	xx	-			
U.C.B.	-				

The horizontally listed devices have a larger value of muscular effort than those listed vertically. The significance of this difference is indicated thus:-

xxx	0.1%
xx	1%
x	5%
-	Not significant

TABLE 5.4

Comparison Between Knee Devices for Average Muscular Effort in Level Walking

(a) The values of average muscular effort obtained for the various knee devices (lbf. in.)

O.B.	393	430	399	392	368	347	
U.C.B.	344	351	367	362			
S.A.	338	368	371	349	334	318	325
G.R.	299	266	292	269	259		
B.S.K.	224	266	286	244	262	246	
L.A.	236	226	228	218	205	225	219

(b) The results of the statistical analysis

	O.B.	U.C.B.	S.A.	G.R.	B.S.K.
L.A.	xxx	xxx	xxx	xx	-
B.S.K.	xxx	xxx	xxx	-	
G.R.	xxx	xxx	xxx		
S.A.	x	-			
U.C.B.	-				

The horizontally listed devices have a larger value of average muscular effort then those listed vertically. The significance of this difference is indicated thus:-

xxx	0.1%
xx	1%
-	Not significant

TABLE 5.5

Comparison Between Knee Devices for Maximum Hip Extensor Moment in Level Walking

(a) The values of maximum hip extensor moment obtained for the various knee devices (in lbf. in.)

O.B.	780	840	780	765	700	705	
G.R.	627	534	645	550	570		
S.A.	576	626	730	498	477	471	447
U.C.B.	351	375	390	380			
L.A.	375	363	345	396	273	321	333
B.S.K.	447	471	285	273	280	300	

(b) The results of the statistical analysis

	O.B.	G.R.	S.A.	U.C.B.	L.A.
B.S.K.	xxx	xxx	xxx	-	-
L.A.	xxx	xxx	xxx	-	
U.C.B.	xxx	xx	x		
S.A.	xxx	-			
G.R.	x				

The horizontally listed devices have a larger value of hip extensor moment than those listed vertically. The significance of this difference is indicated thus:-

xxx	0.1%
xx	1%
x	5%
-	Not significant

TABLE 5.6

Comparison Between Knee Devices for Maximum Hip Flexor Moment in Level Walking

(a) The values of maximum hip flexor moment obtained for the various knee devices (in lbf. in.)

U.C.B.	840	804	797	870			
S.A.	666	768	633	756	762	670	747
B.S.K.	543	600	780	750	790	675	
O.B.	265	450	450	444	480	440	
G.R.	291	450	414	480	411		
L.A.	351	450	339	405	387	426	399

(b) The results of the statistical analysis

	U.C.B.	S.A.	B.S.K.	O.B.	G.R.
L.A.	xxx	xxx	xxx	-	-
G.R.	xxx	xxx	xx	-	
O.B.	xxx	xxx	xx		
B.S.K.	-	-			
S.A.	-				

The horizontally listed devices have a larger value of hip flexor moment than those listed vertically. The significance of this difference is indicated thus:-

xxx	0.1%
xx	1%
x	5%
-	Not significant

If it is required to compare the B.S.K. with the S.A. for muscular effort, the single point concerned with the two devices in the array of Table 5.3b is located. In this case it is the third column, second row. For this example, S.A. is listed horizontally and B.S.K. vertically. Hence S.A. has the higher value of muscular effort. Whether this difference is significant or not is assessed using the technique described in section 4.4 (page 118) and the result displayed as explained in the table.

Although multiple comparisons have been tested for each of the four parameters, the discussion will be limited to the comparison between each knee device and the single-axis, non-stabilized knee, S.A. Apart from O.B., the results obtained for the muscular effort and average muscular effort are identical, therefore only muscular effort will be discussed in these cases.

B.S.K. had a significantly lower value of muscular effort and maximum hip extensor moment than S.A. Although there was an increase in the maximum hip flexor moment produced, this was not significant in the experiments carried out. The amputee felt very secure on the leg and the addition of the swing phase control resulted in a smooth transition from swing to stance phase.

A very significant increase in muscular effort and a significant increase in average muscular effort were obtained with the O.B. This is mainly due to the very high maximum hip extensor moment produced by the amputee although the maximum hip flexor moment was significantly lower. It is thought that the results from the O.B. were influenced to a large extent by an extension spring mounted in the knee. This is designed to assist in knee extension during the swing phase of walking. The spring action was, however, too strong for the amputee. As the knee extended before heel strike, the inertia of the shank caused the hip to flex and the amputee to overstep his usual stride length. A greater push off was required from the normal leg and, due to the longer stride length, a larger hip extensor moment was needed on the prosthesis.

Apart from this fault of the design, which can be overcome by fitting a tension adjustment to the spring, the amputee was happy with the leg. It gave him a definite stabilizing effect in the stance phase.

The amputee was pleased with the G.R. but commented on the excessive heel rise that occurred with this particular mechanism. The muscular effort was significantly lower than S.A. but no significant decrease in maximum hip extensor moment was recorded. On the other hand, a highly significant decrease in hip flexor moment was obtained.

A significantly larger value of muscular effort is recorded with the U.C.B. due mainly to a very high hip flexor moment. At the same time the maximum hip extensor moment is significantly lower than that obtained with S.A.

With this device, heel rise, just after toe off, was reduced by the swing phase control and a very smooth gait was achieved. The amputee could "feel" his artificial leg going into full extension after heel strike. This is approaching the gait achieved by a normal individual in this phase of the walking sequence. One failing of polycentric mechanisms was exposed during this particular investigation. With a polycentric device it is difficult, if not impossible, to extend the hip without extending the knee. A typical situation where this can occur is the first stride of a walking sequence, where ideally the knee should be in slight flexion to give a smooth push off. As the amputee extends the hip to balance the body weight, the knee extends and is raised giving a jolt as the first stride is taken.

The Lammers knee, L.A., gave the best set of results for muscular effort and maximum hip flexor moment, and although the tests indicate that the B.S.K. required less hip extensor moment, this was not significant in the statistical analysis.

The only comment against the L.A. is the lack of a swing phase control, and hence the cadence of the amputee is limited to that imposed by the mass properties of the shank-foot assembly.

5.4.2 Walking up a ramp When walking up ramp (Fig.5.4) the knee is stable from 6% of the cycle onwards, but hip extensor action is required for a higher percentage of the cycle (34%) than in level walking (20%).

This increased effort is needed to lift the body of the amputee up the gradient of the ramp. The maximum hip moment of +700 lbf.in. occurs much later than the maximum of that required for knee stability (10% against 3%). The stance phase occupies 54% of the cycle, very similar to that of level walking. However, knee flexion is initiated later at 48%.

The only knee to produce any advantage in ascending a ramp is the Lammers (Fig.5.5). Both the hip extensor and flexor moments are decreased, but as with S.A. the peak extensor moment occurs later in the cycle than the peak M_{HRZ}. A much lower hip flexor moment than M_{HRZ} is required to initiate knee flexion showing the effect of the polycentric device on the hip moment.

A/P HIP MOMENT IN WALKING UP RAMP WITH S.A.

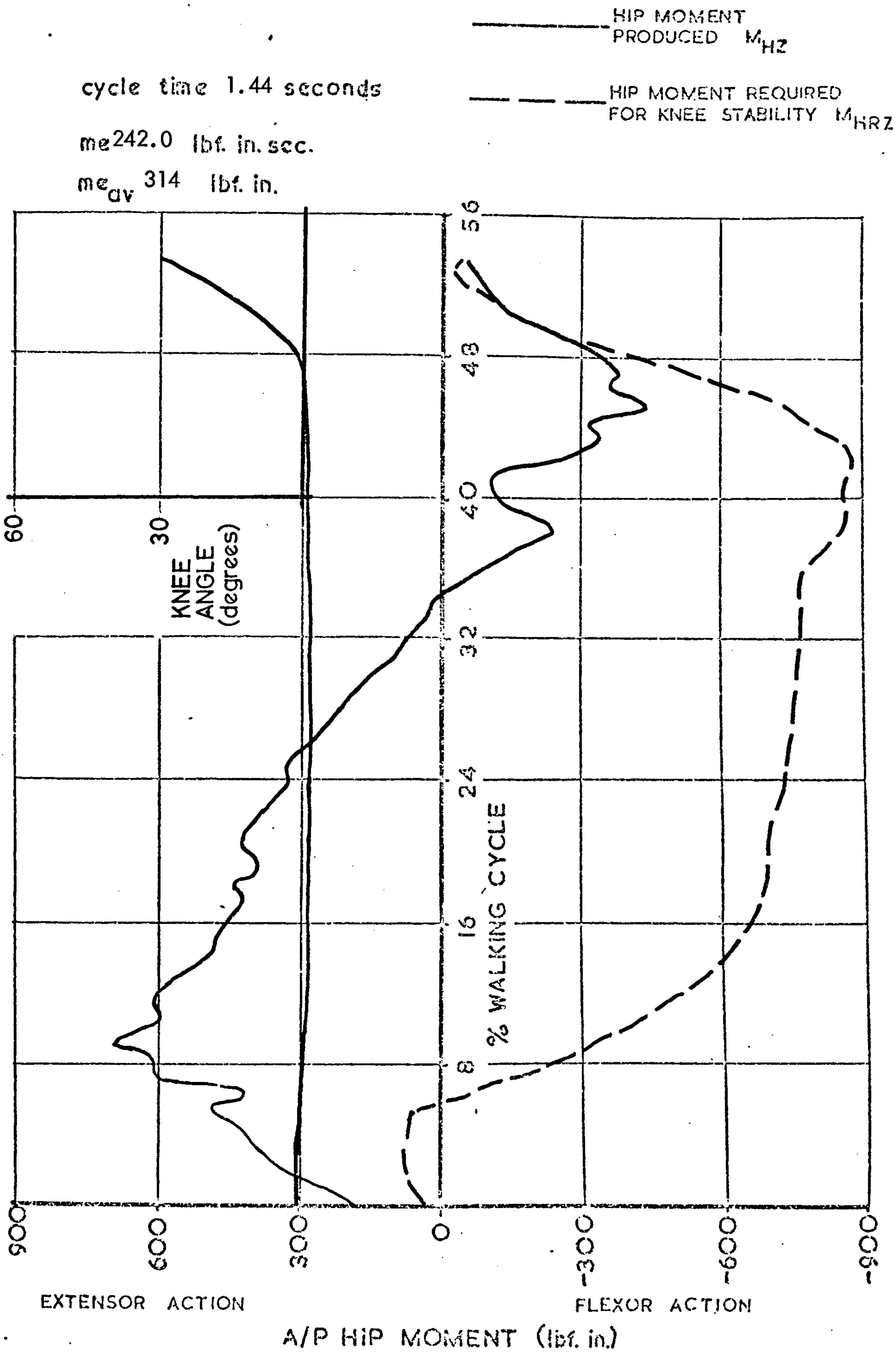


fig. 5.4

A/P HIP MOMENT IN WALKING UP RAMP WITH L.A.

cycle time 1.54 seconds

me 157.9 lbf. in. sec.

me_{av} 194 lbf. in.

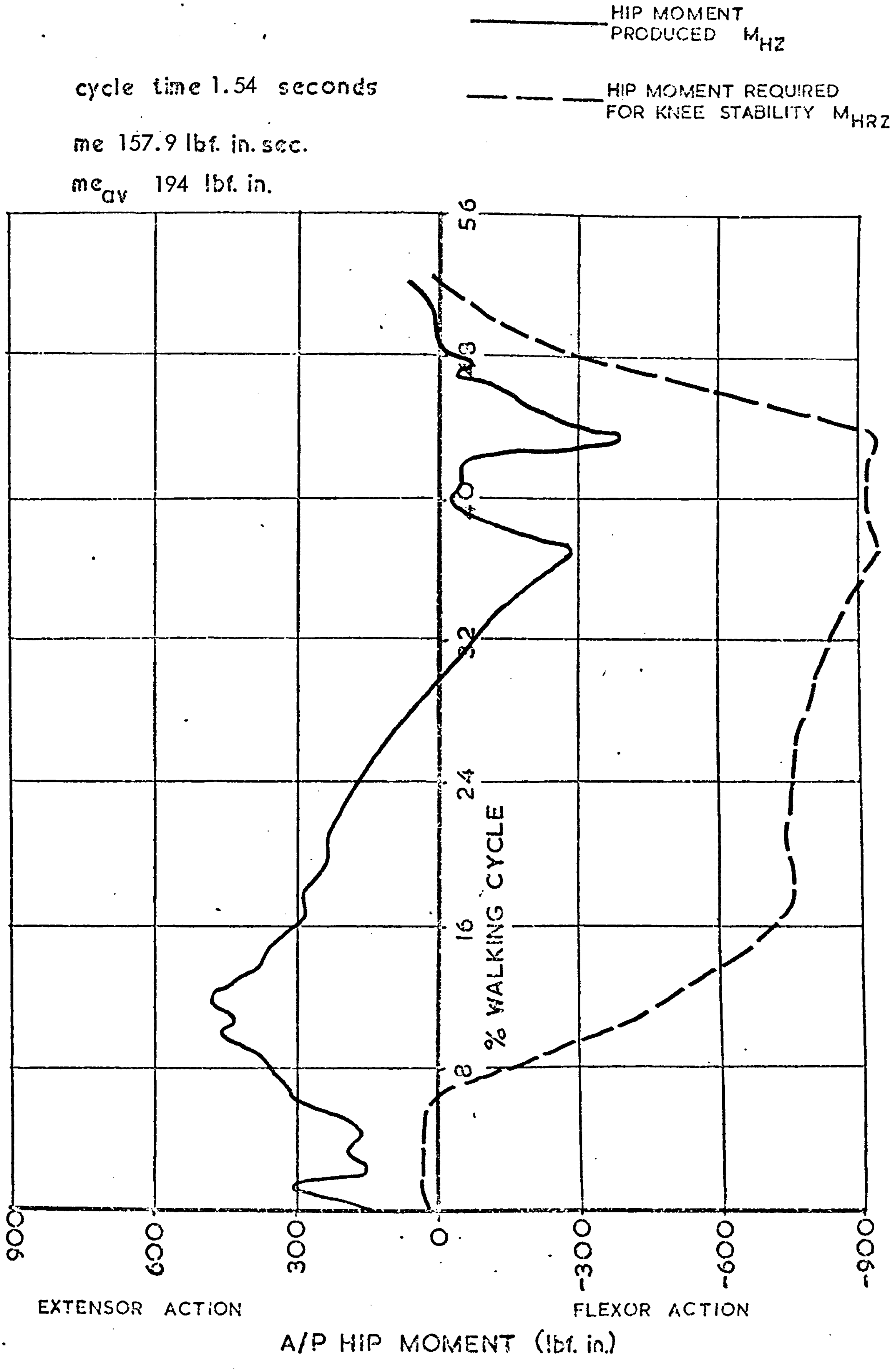


fig. 5.5

5.4.3 Walking down a ramp Walking down ramp (Fig. 5.6) demonstrates a marked difference from level walking. The danger of buckling of the knee is present up to 24% of the cycle and M_{HRZ} reaches a maximum of 330 lbf. in. at 9%. This appears to be the main reason for hip extensor action. After the initial peak at heel strike, the hip moment drops to a level where it just maintains knee stability. It follows the M_{HRZ} curve much more closely than in level walking and changes to a flexor action earlier (31%). Knee flexion is also initiated earlier than in level walking. In walking down ramp, the stance phase occupies 53% of the total cycle time.

With the B.S.K. (Fig. 5.7), the maximum hip extensor moment is not reduced in magnitude but soon decreases. The curve of M_{HRZ} shows a marked difference from that with the S.A. which may be due to the shorter length of the stride. Before starting to walk down the incline with the O.B. (Fig. 5.8), the amputee could stand comfortably at the top of the ramp with weight on his slightly flexed artificial leg. The large peak (900 lbf. in.) at 1% of the cycle is caused by the effect of the extension spring, as explained above when discussing level walking.

With the U.C.B. (Fig. 5.9), the swing phase control allows a more controlled heel strike and the polycentric knee permits an easy rollover requiring very little hip moment from the amputee.

The L.A. (Fig. 5.10) produces a similar curve but does give a sharp peak at heel strike unlike the U.C.B. However, the hip flexor moment is less than that with U.C.B.

5.4.4 Walking up and down stairs In both walking up and walking down stairs, the amputee walks "foot to foot". That is, the normal and prosthetic foot

A/P HIP MOMENT IN WALKING DOWN RAMP WITH S.A.

——— HIP MOMENT PRODUCED M_{HZ}

— — — HIP MOMENT REQUIRED FOR KNEE STABILITY M_{HRZ}

cycle time 1.14 seconds

me 197.2 lbf. in. sec.

me_{av} 318 lbf. in.

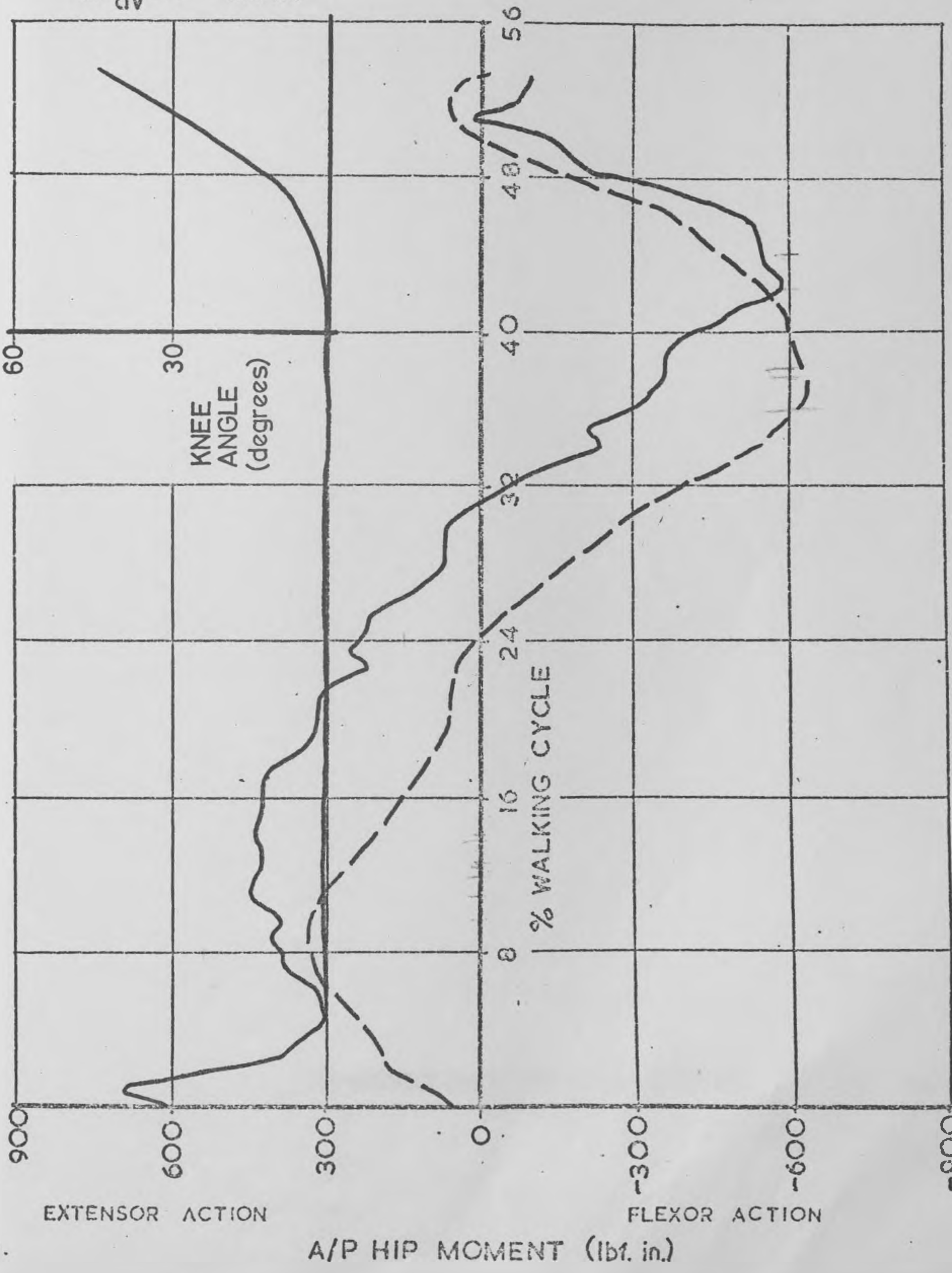


fig. 5.6

A/P HIP MOMENT IN WALKING DOWN RAMP WITH B.S.K.

cycle time 1.12 seconds

me 164.9 lbf. in. sec.

me_{av} 290 lbf. in.

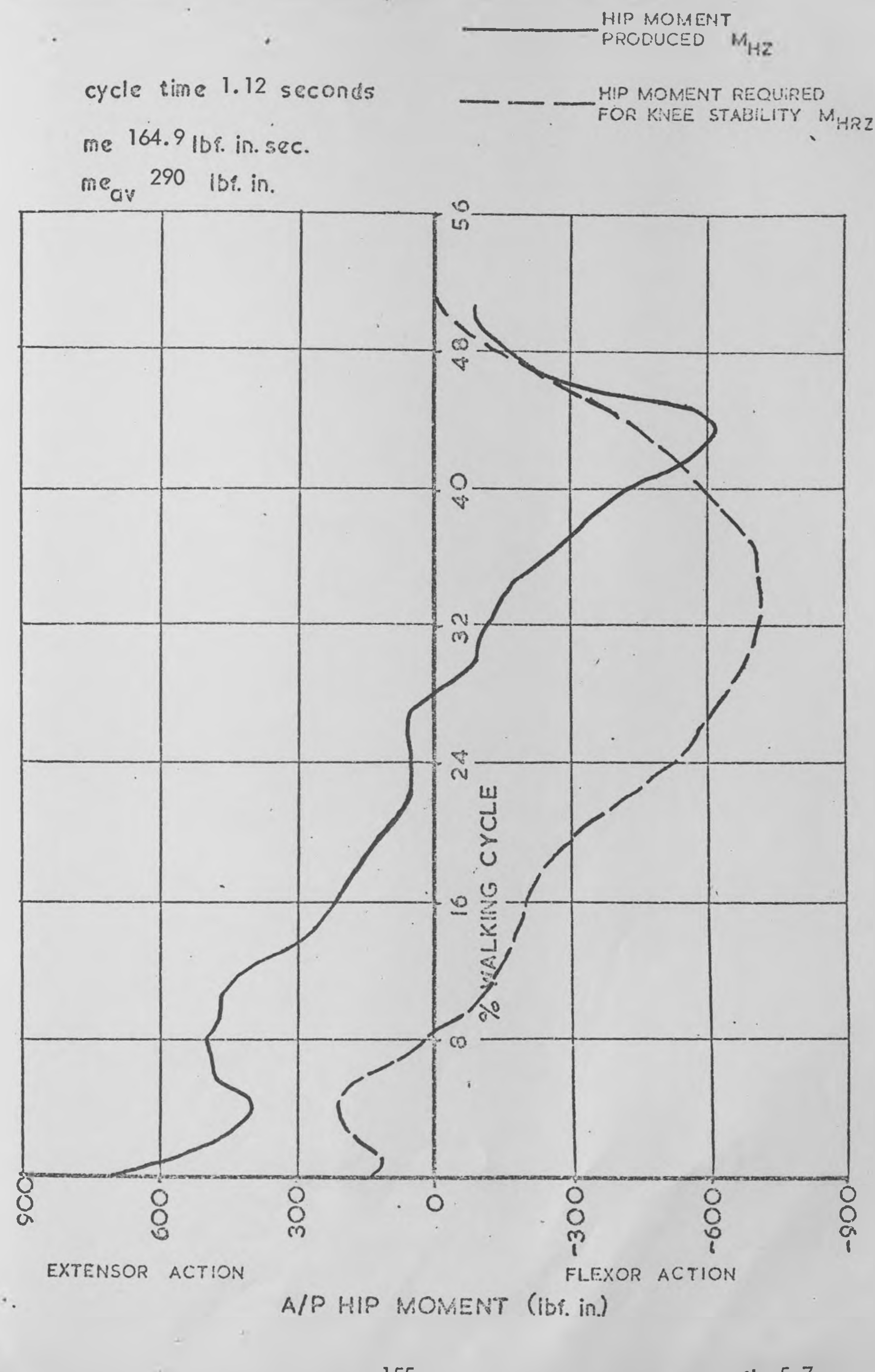


fig. 5.7

A/P HIP MOMENT IN WALKING DOWN RAMP WITH O.B.

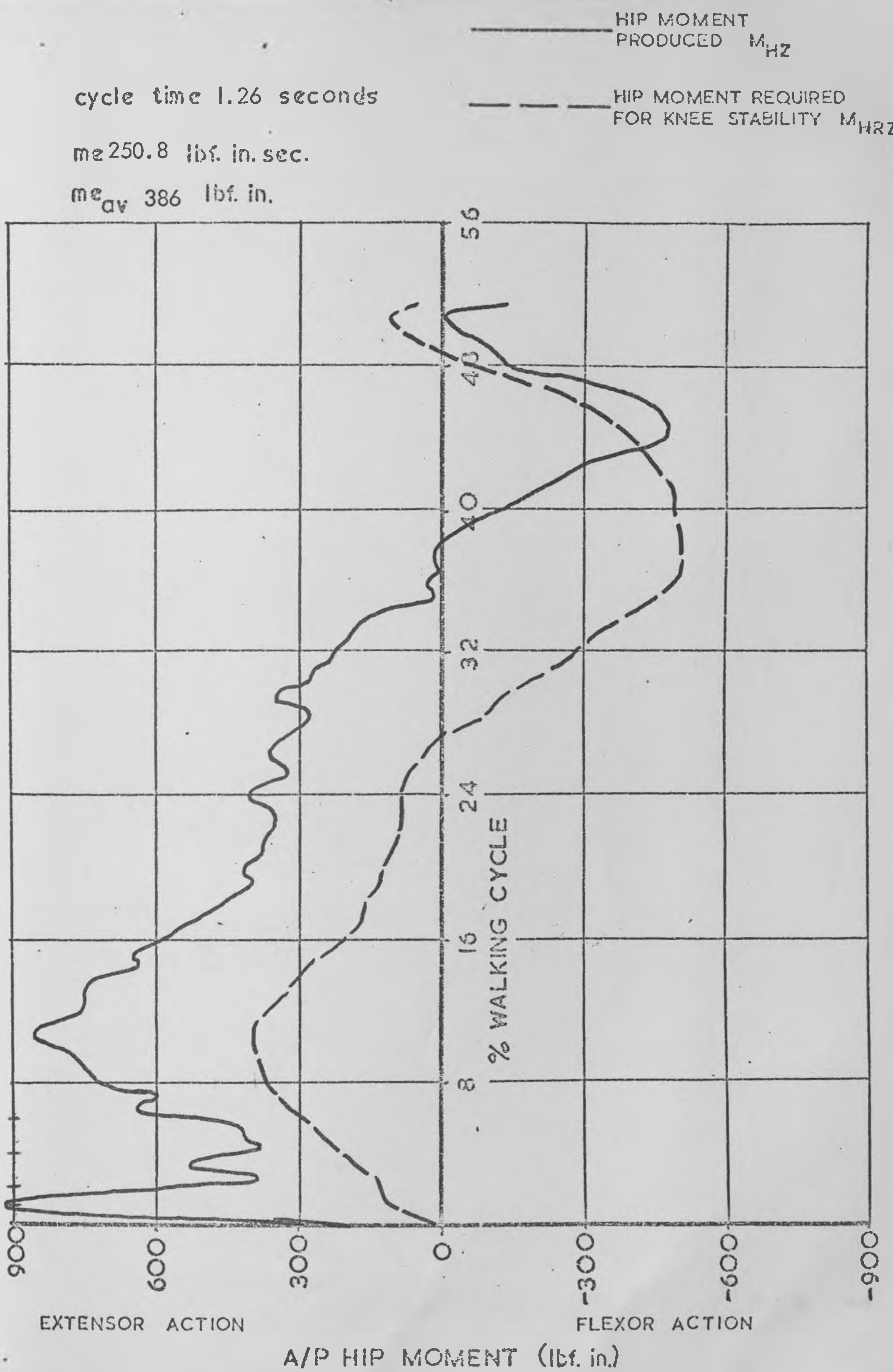


fig. 5.8

A/P HIP MOMENT IN WALKING DOWN RAMP WITH U.C.B.

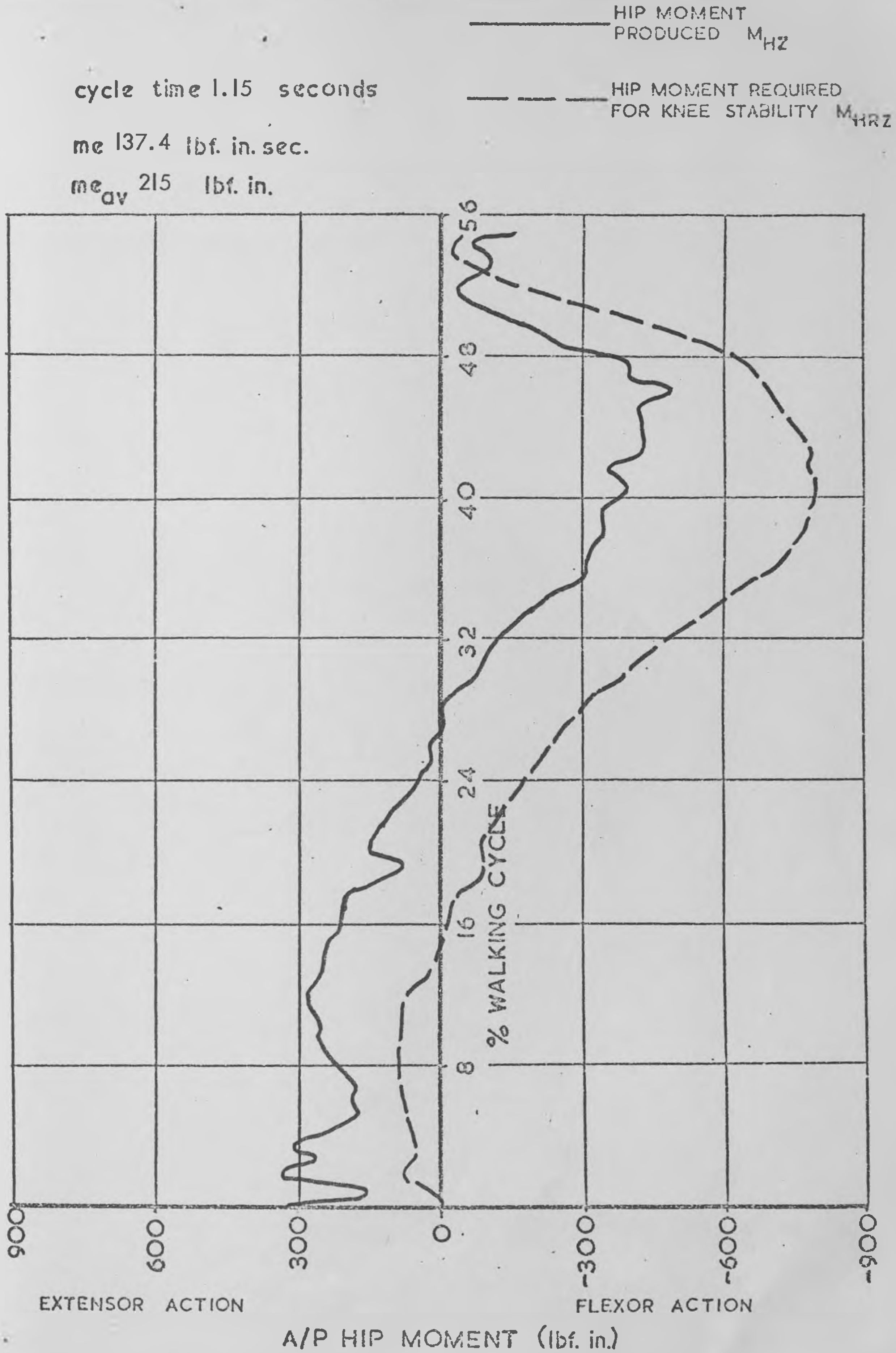


fig. 5.9

A/P HIP MOMENT IN WALKING DOWN RAMP WITH L.A.

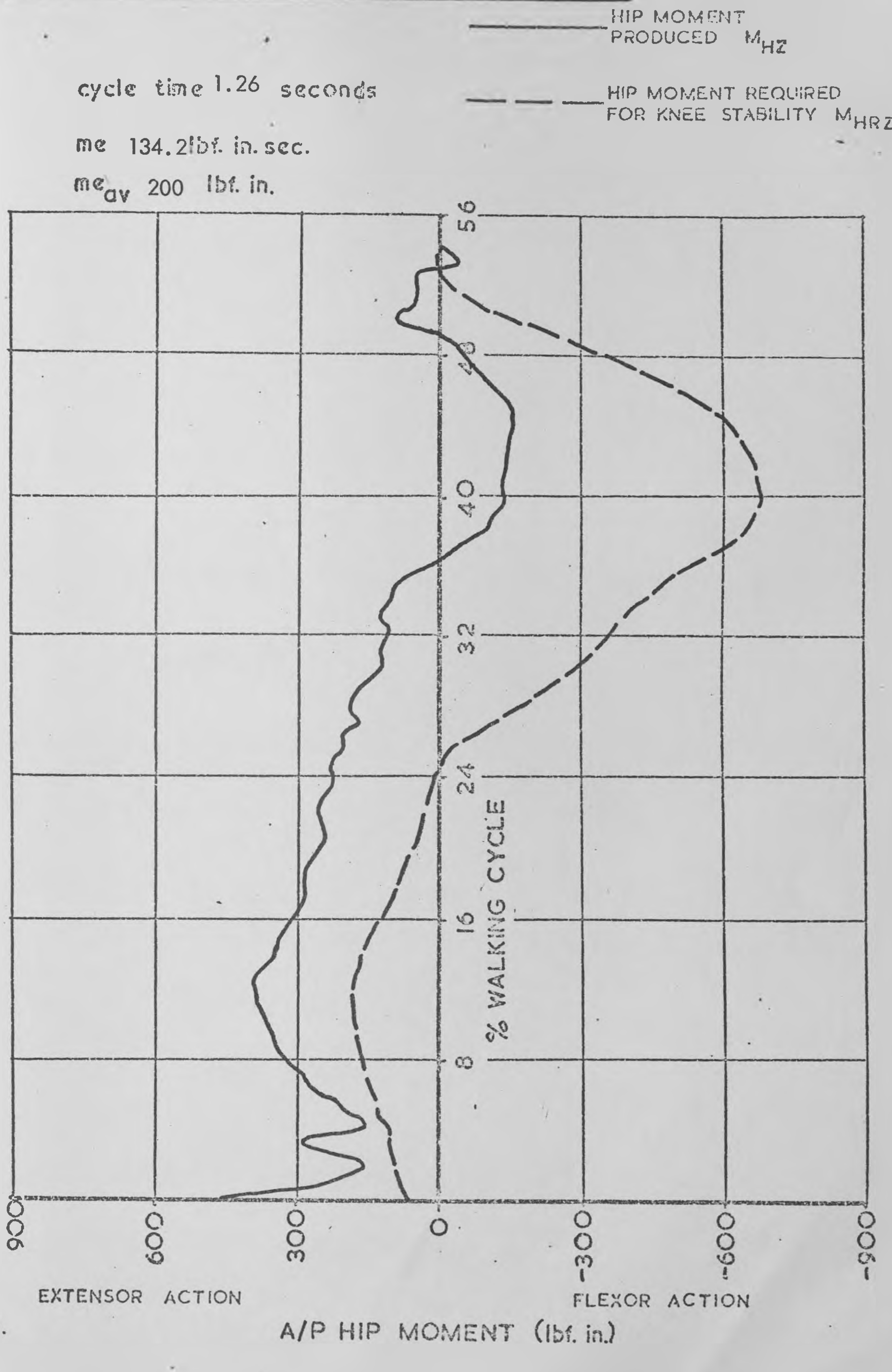


fig. 5.10

are together on the stair after each stride has been taken. When walking up stairs, the normal leg leads but in walking down stairs, the prosthesis leads.

The variation in hip moment for the S.A. in walking up stairs is shown in Fig.5.11. The knee is stable throughout the loaded phase and is never flexed. The extent of the stability can be appreciated by examining the graph of M_{HRZ}. The stance phase occupies 55% of the cycle and the hip moment is low and varying between extensor and flexor action.

When walking down stairs with the S.A. (Fig.5.12), the possibility of buckling at the knee depends on whether the whole of the artificial foot is placed on the step or only the heel. In this particular stride, hip moment has to be provided throughout the stance phase to stabilize the knee. When using the G.R. (Fig.5.13), only the knee device prevents the knee from buckling in the region between 1% and 12% of the cycle.

5.4.5 Stepping over an object If an amputee meets an object in his path i.e. a kerbstone, he has to step over it like anyone else. However, it introduces serious problems to the amputee since he must put his prosthesis over first. Any attempt to step over with his normal leg first involves the prosthesis hitting the object due to the inability of the amputee to flex the artificial knee sufficiently to clear it. As shown in Fig.5.14 very high values of hip moment are recorded at the time of heel contact although to stabilize the knee, much less than this is required. These are probably due to the high ground to foot inertia forces produced when the leg is decelerated by the ground, and are unlikely to be a true record of the actual hip moment produced by the amputee.

A/P HIP MOMENT IN WALKING UP STAIRS WITH S.A.

cycle time 1.39 seconds

me 58.0 lbf. in. sec.

me_{av} 76 lbf. in.

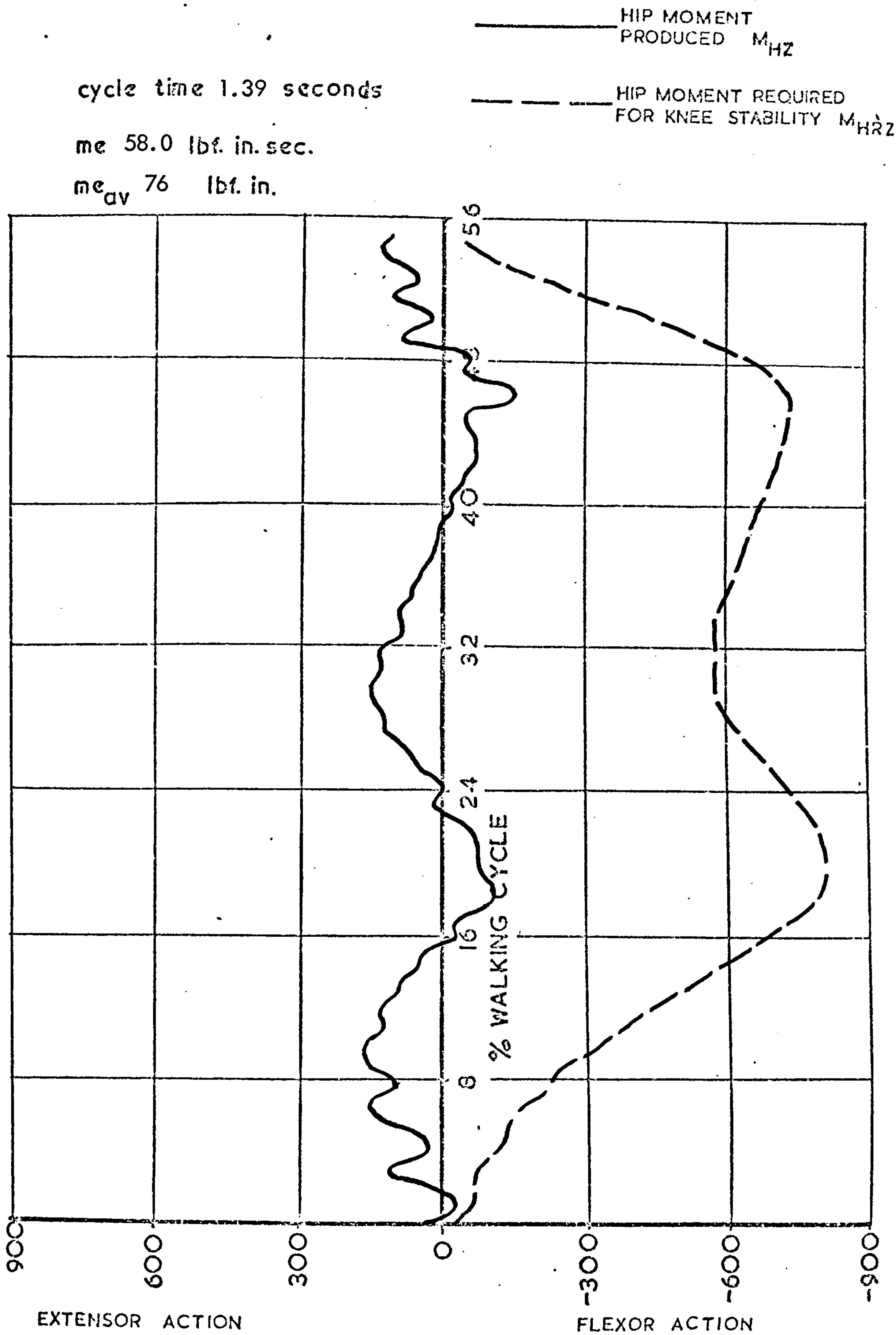


fig. 5.11

A/P HIP MOMENT IN WALKING DOWN STAIRS WITH S.A.

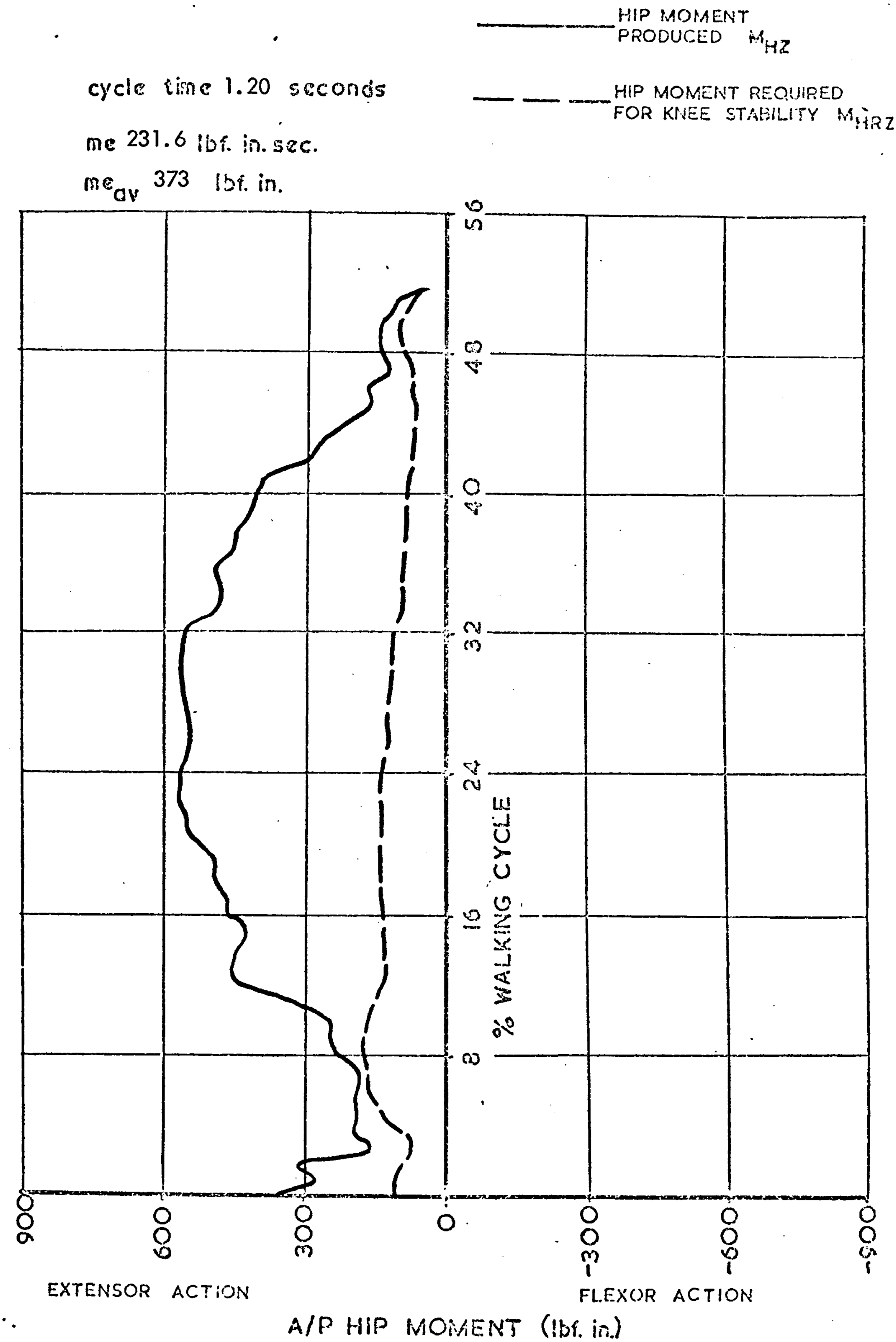


fig. 5.12

A/P HIP MOMENT IN WALKING DOWN STAIRS WITH G.R.

HIP MOMENT PRODUCED M_{HZ} (solid line)

HIP MOMENT REQUIRED FOR KNEE STABILITY M_{HRZ} (dashed line)

cycle time 1.35 seconds

me 118.9 lbf. in. sec.

me_{av} 192 lbf. in.

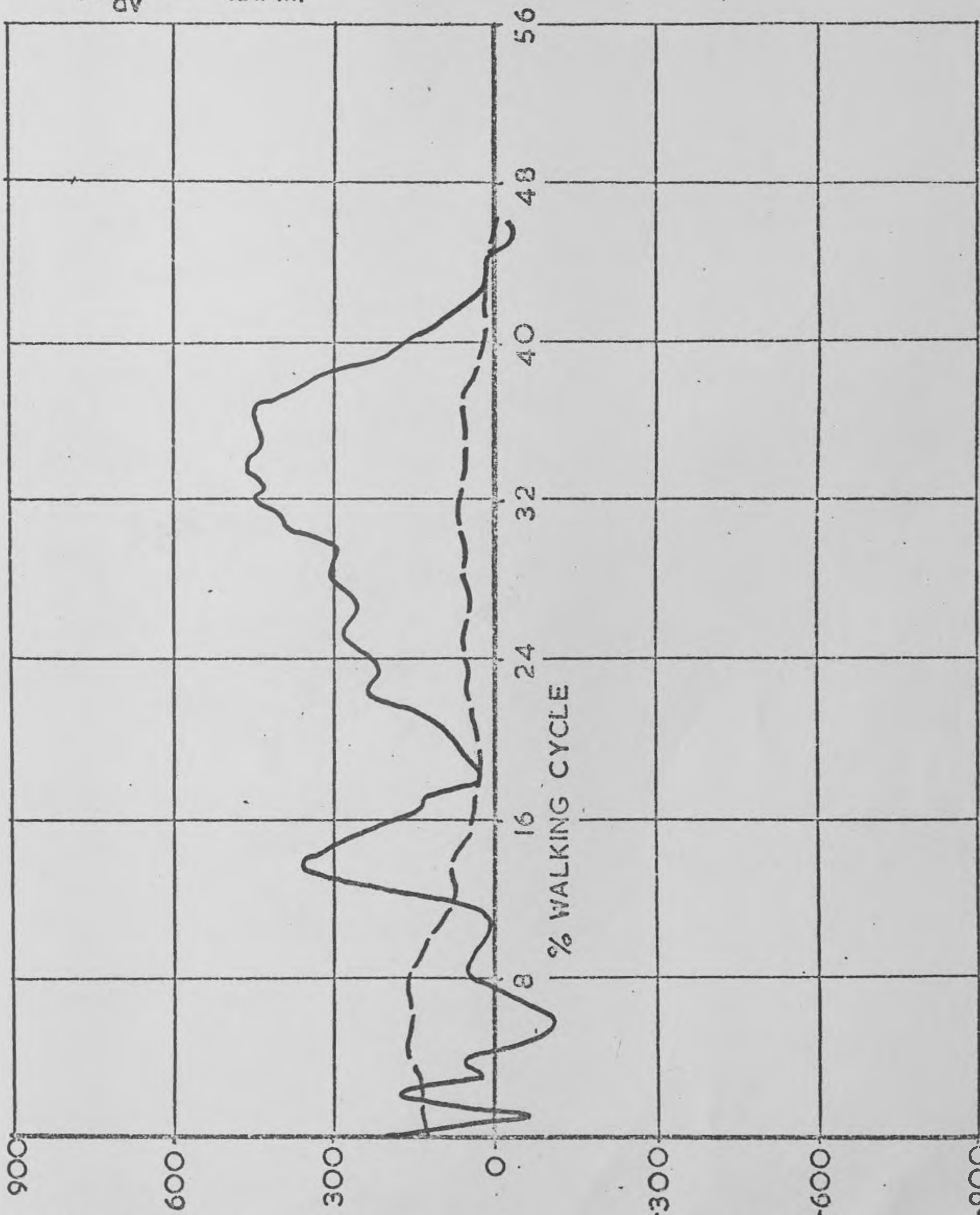


fig. 5.13

A/P HIP MOMENT IN STEPPING OVER AN OBJECT (from a standing position) WITH S.A.

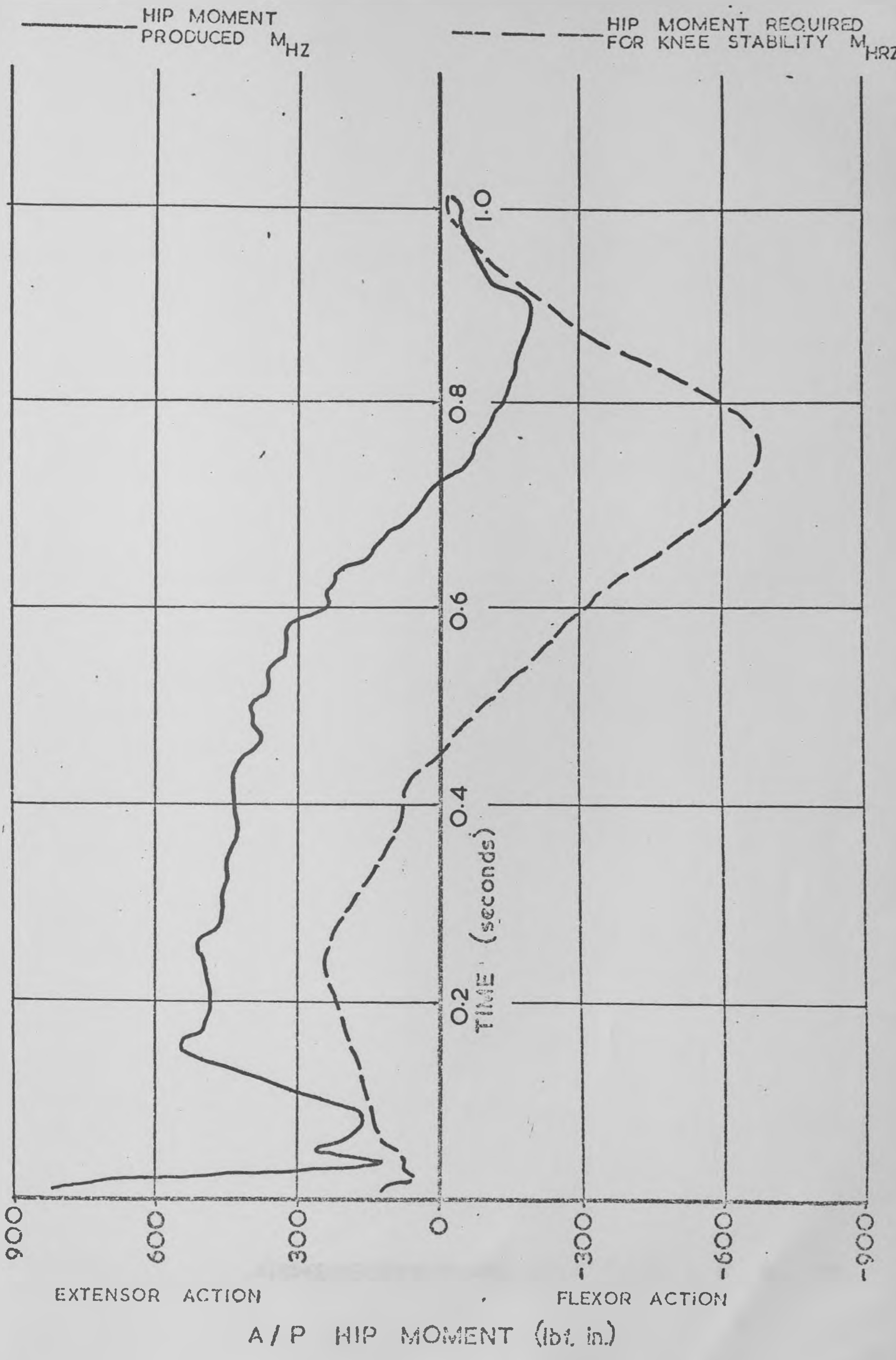


fig.5.14

The results from stepping over the object after walking up to it are similar to Fig. 5. 14 and are not reproduced here but can be found in Appendix VI. Both graphs show a marked insecurity of the amputee just after heel strike which is quickly counteracted by a sharp increase in hip moment. Hip extensor action is required for the first 0.45 seconds to stabilize the knee in each case. When the amputee walks up to the object before stepping over, less time is spent on the prosthesis and extensor action ceases earlier (0.6 sec. against 0.7 sec.).

With the B.S.K. (Fig. 5. 15) the knee would buckle at 0.04 sec. and 0.08 sec. if the device did not provide sufficient resistance to bending. The paradoxical continuation of hip extensor action throughout this test is due to the amputee stopping after stepping over the object. Knee flexion does not occur, hence a hip flexor moment is not required.

The extension spring helps the amputee to step over the object (Fig. 5. 16) with the O.B. as it assists knee extension even when the hip is flexed. Hence the time taken is reduced from 1.0 sec. to 0.85 sec. Knee instability is prevented by the knee device at 0.03 sec. Hip extensor action is, however, higher than with the S.A.

With the U.C.B. (Fig. 5. 17) considerable difficulty was experienced in stepping over an object especially when standing in front of it beforehand. The knee began to flex due to the weight of the shank and foot before the leg contacted the floor on the other side of the object.

The lack of swing phase control (Fig. 5. 18) in the L.A. meant that coupled with the disadvantage of the U.C.B., an unusually high peak of hip moment is recorded. This is followed by a large hip extensor moment

A/P HIP MOMENT IN STEPPING OVER AN OBJECT (from a standing position) WITH B.S.K.

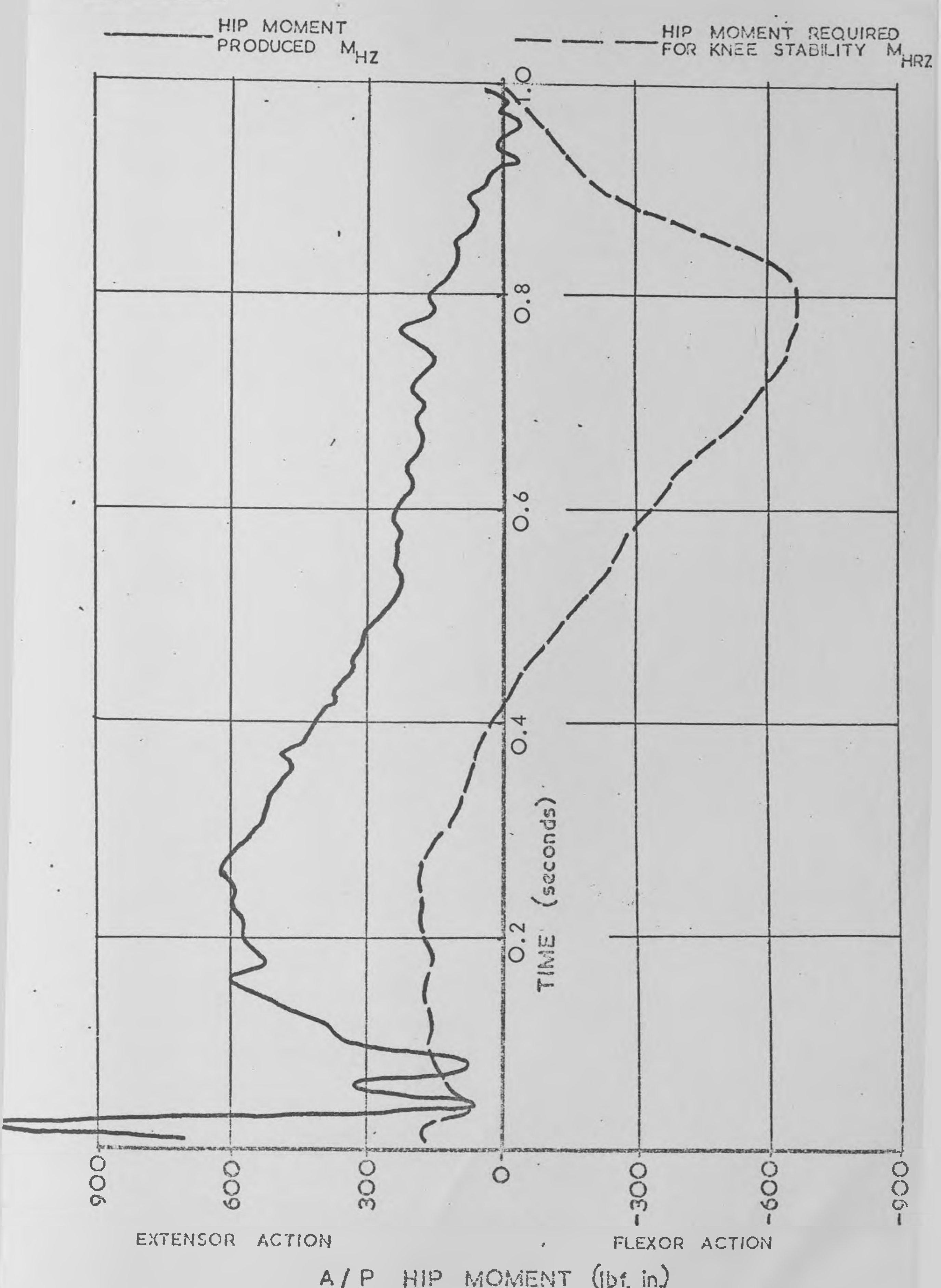


fig.5.15

A/P HIP MOMENT IN STEPPING OVER AN OBJECT (from a standing position) WITH O.B.

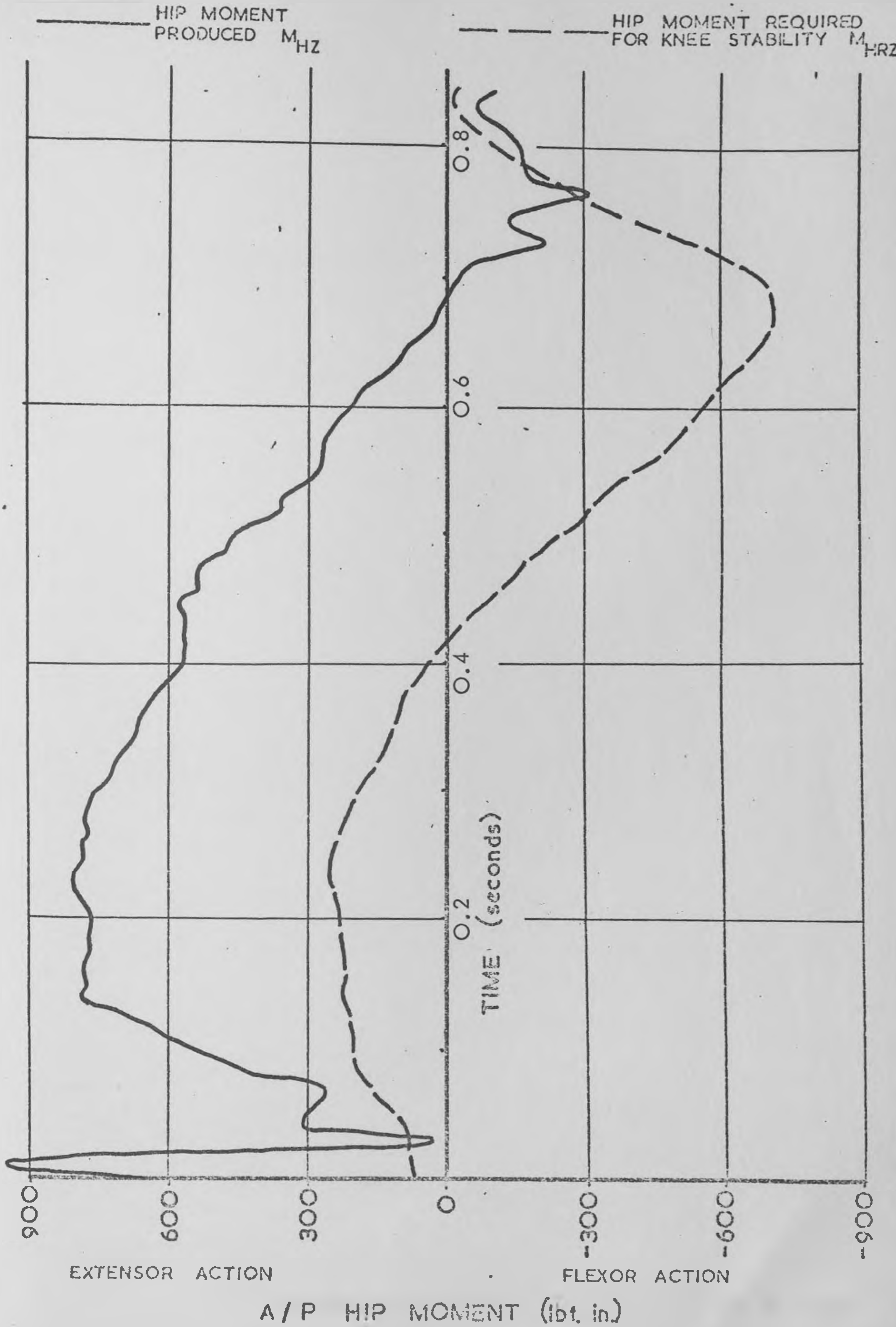


fig. 5.16

A/P HIP MOMENT IN STEPPING OVER AN OBJECT (from a standing position) WITH U.C.B.

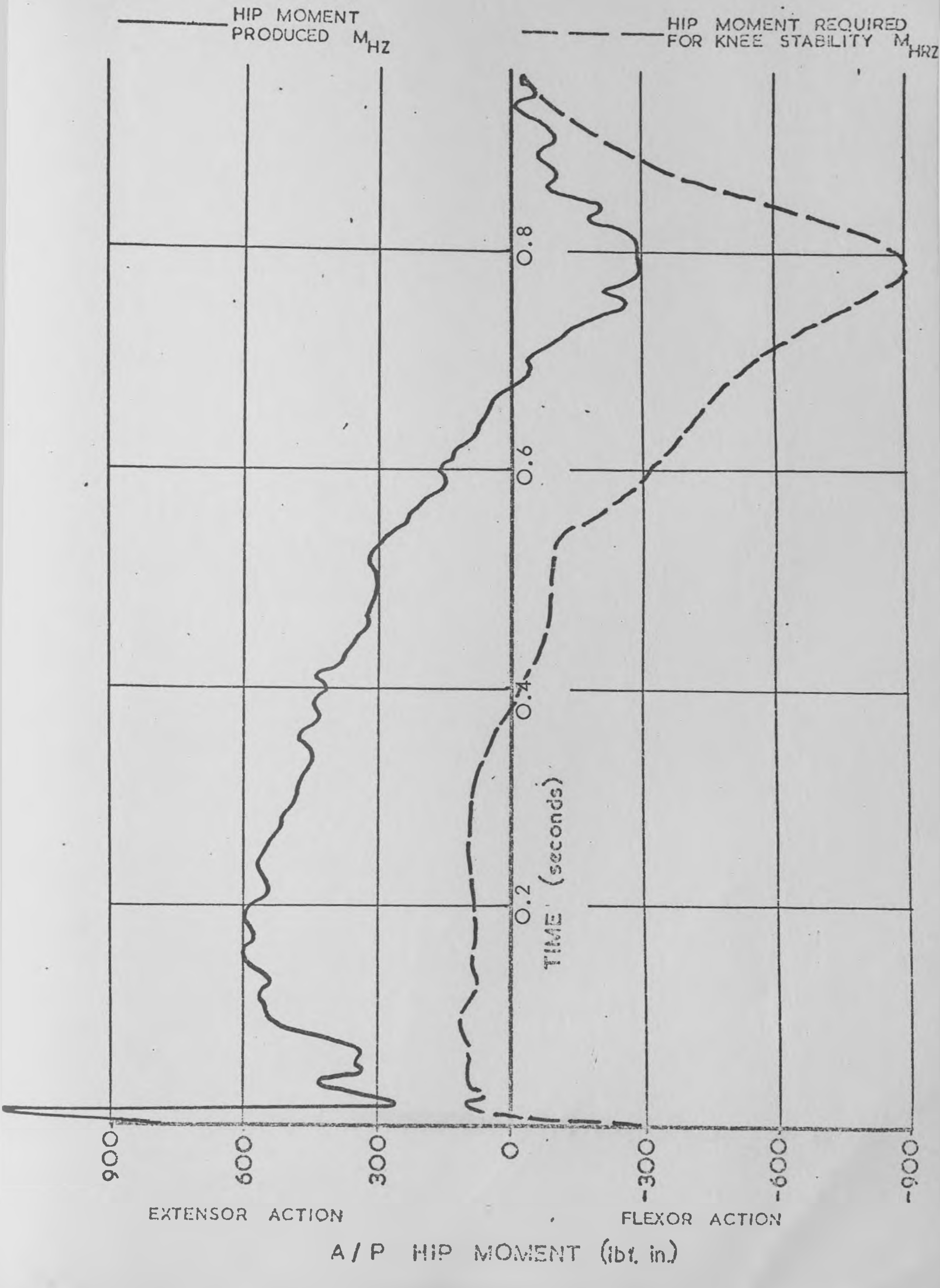


fig.5.17

A/P HIP MOMENT IN STEPPING OVER AN OBJECT (from a standing position) WITH L.A.

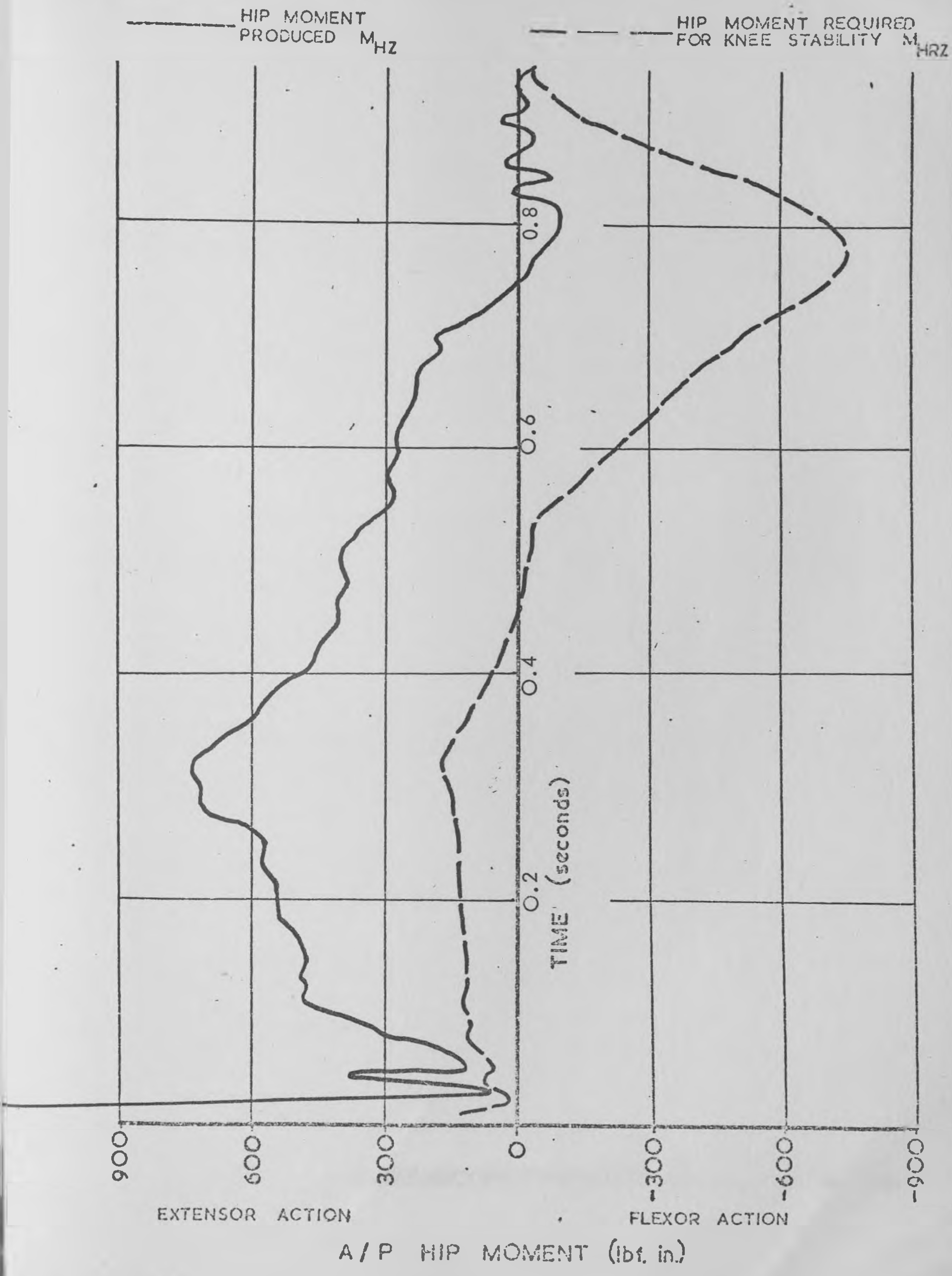


fig. 5.18

later in the cycle due to the natural rhythm being upset by stepping over the object.

5.4.6 Running This is such an erratic activity that no comparison between mechanisms can sensibly be made, hence only the results with the S.A. are presented (Fig. 5.19).

An A/K amputee runs by skipping on his normal leg while waiting for the artificial leg to swing through. This means that he is only using his prosthesis for a short part of the total cycle time, in this case 42%. As with stepping over an object ground to foot inertia forces cause a peak in hip extensor moment just after heel strike. This is followed at 4% by some insecurity which is overcompensated to a maximum extensor action of 600 lbf. in. at 15%. The danger of knee instability is present up to 12% of the cycle but hip extensor action persists up to 26%. Knee flexion is initiated at 34% of the cycle.

5.4.7 Other activities In standing up, sitting down and lifting a weight, the prosthesis is not used until it is almost fully extended when it supports more than half the total load.

No outstanding difference between mechanisms was found.

5.5 Variations in the Other Force Actions

For each force action considered, its variation will be discussed for the S.A. only, in level walking and the activity producing the largest value of that force action.

5.5.1 A/P knee moment The A/P knee moment must be positive if no knee device is present to prevent flexion. The initiation of knee flexion occurs when

A/P HIP MOMENT IN RUNNING WITH S.A.

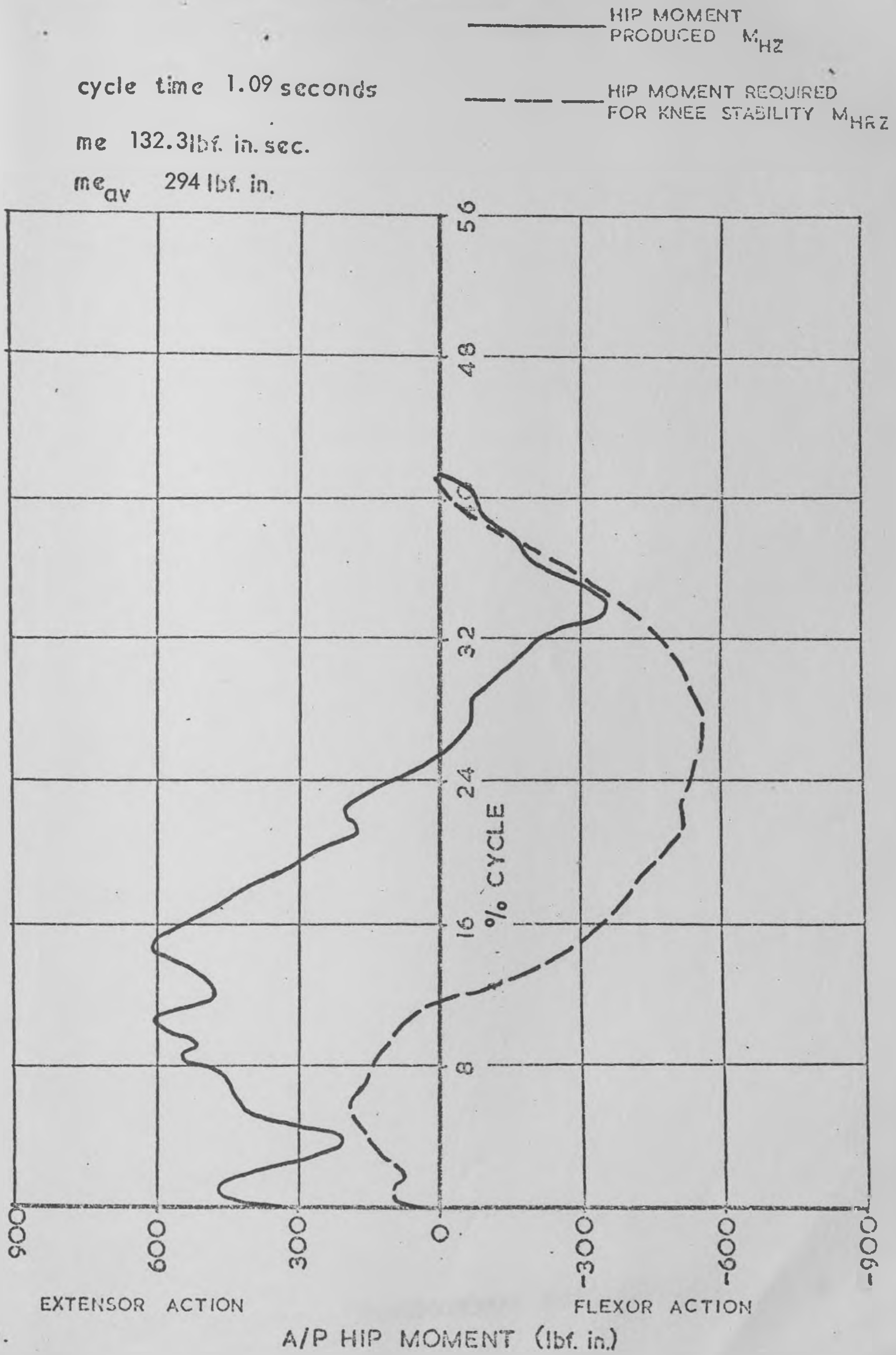


fig. 5.19

the A/P knee moment decreases to zero.

In level walking (Fig.5.20) the value rises to a peak of 350 lbf.in. at 16% of the cycle, is maintained at that value until 28% of the cycle when it begins to fall away to zero at 47% of the cycle. The maximum value of 600 lbf.in. occurs in walking up ramp (Fig.5.21) from 12% to 21% of the cycle.

5.5.2 A/P ankle moment The ankle is the most stressed part of the prosthesis due to the high bending moments imposed on it. In level walking (Fig.5.22) the moment is negative or plantarflexing to 16% of the cycle reaching a maximum of -150 lbf.in. The maximum positive moment of 750 lbf.in. is reached at 42% of the cycle after which the moment decreases linearly to 150 lbf.in. at toe off.

Walking up ramp (Fig.5.23) produces the largest positive value of ankle moment, 850 lbf.in. at 42% of the cycle.

5.5.3 M/L hip moment Very little adductor action is recorded in most of the activities investigated and if it is present it acts for short periods only, just after heel strike and just before toe off.

In level walking (Fig.5.24) a maximum hip adductor moment of equal magnitude, 120 lbf.in., is exerted at just after heel strike and just before toe off. Abductor action reaches a maximum of 500 lbf.in. at 16% of the cycle but is maintained at about 400 lbf.in. for 33% of the cycle. This is required to balance the moment exerted about the hip joint due to the body weight.

The most important activity concerned with the M/L hip moment is walking sideways. The two methods are shown in Fig.5.25 (prosthesis leading) and

A/P KNEE MOMENT IN LEVEL WALKING WITH S.A.

cycle time 1.10 seconds

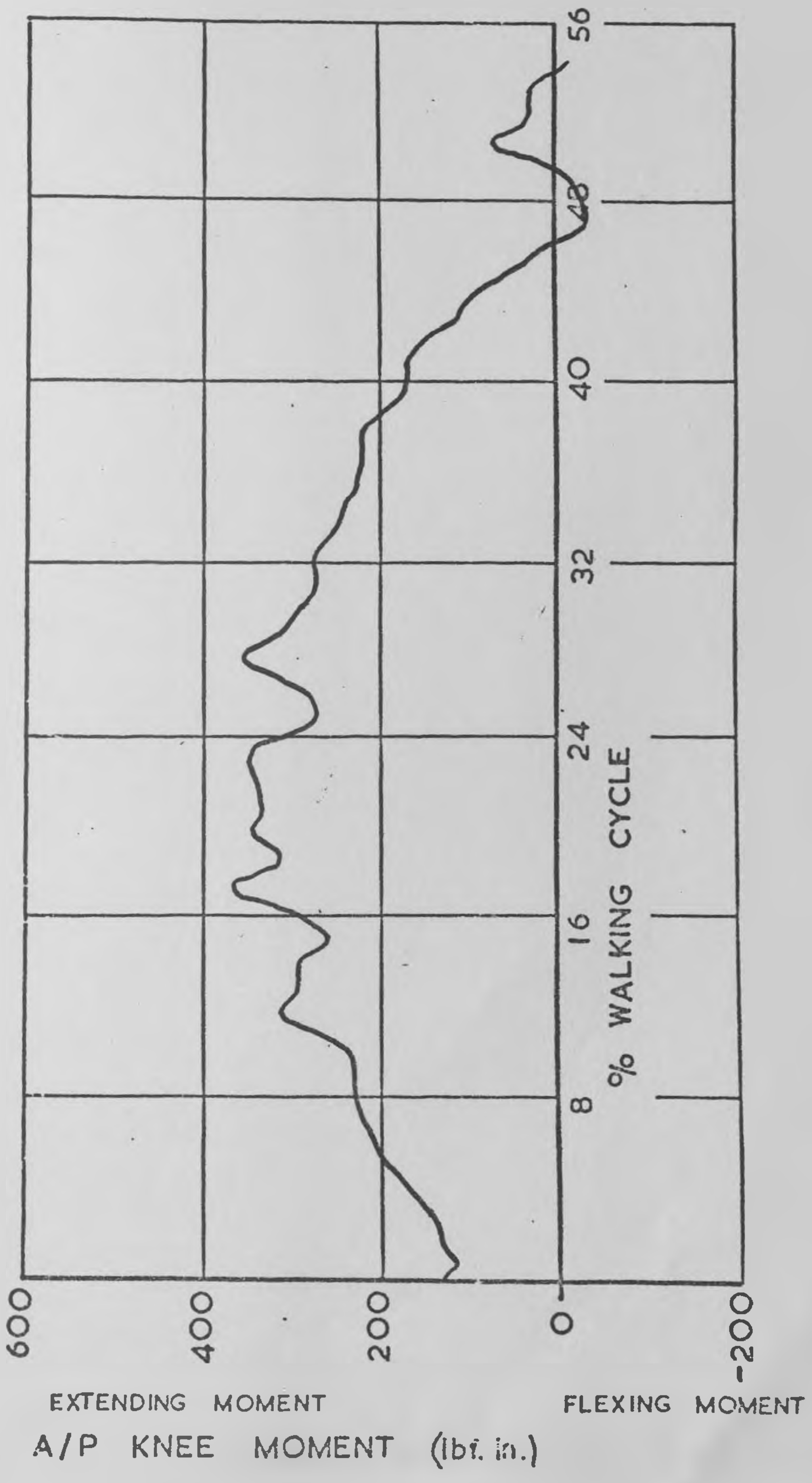


fig.5.20

A/P KNEE MOMENT IN WALKING UP RAMP WITH S.A.

cycle time 1.44 seconds

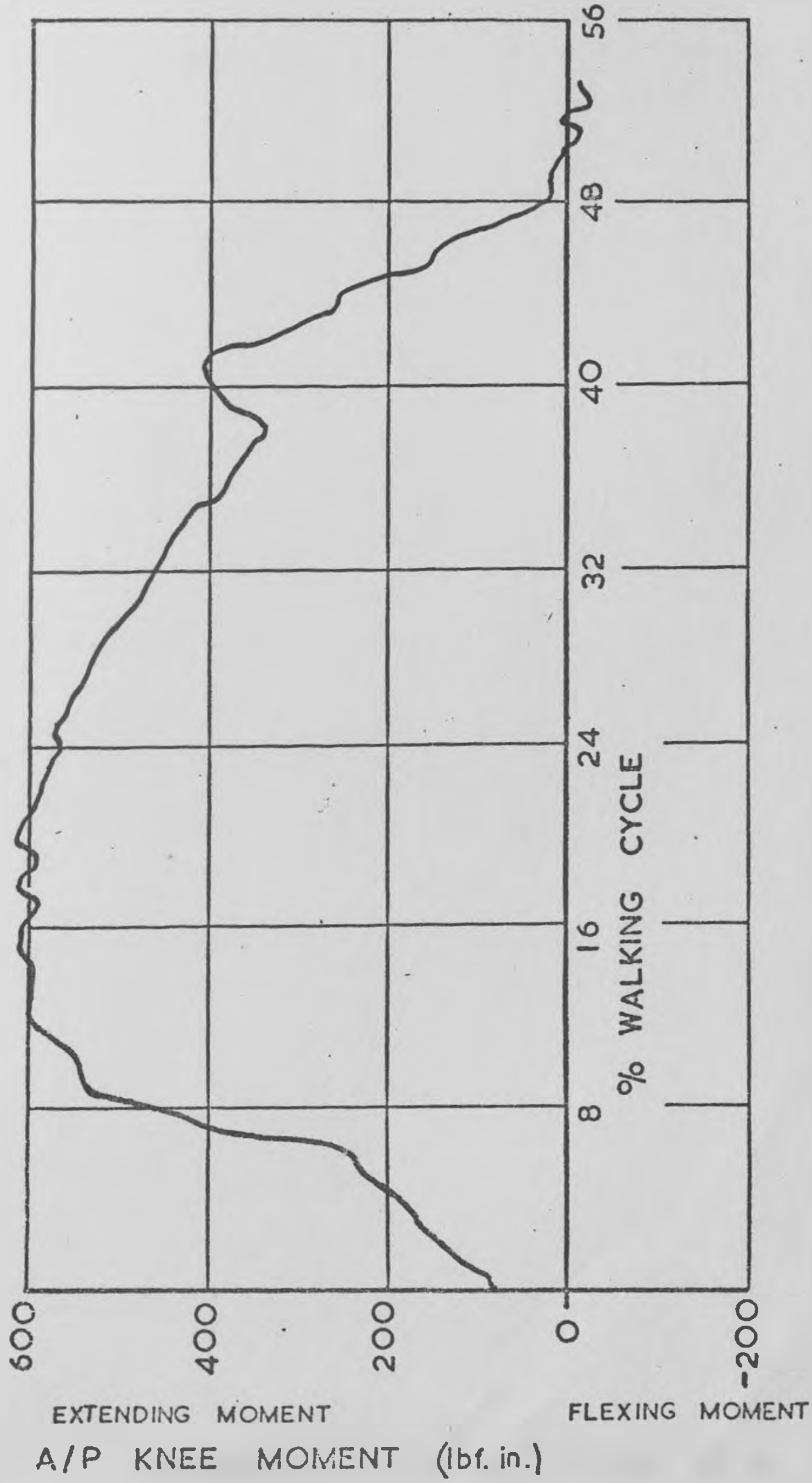
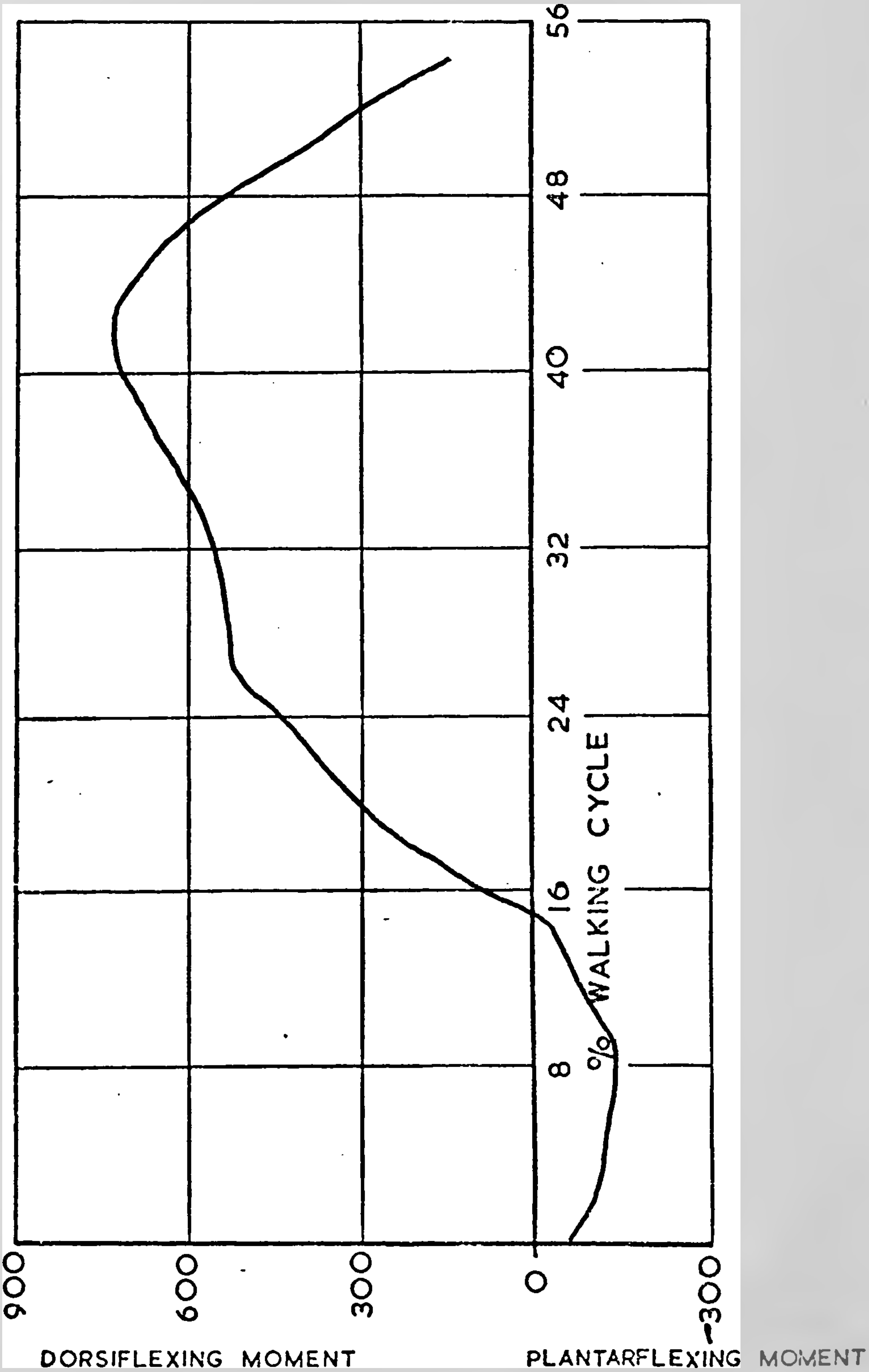


fig. 5.21

A/P ANKLE MOMENT IN LEVEL WALKING WITH S.A.

cycle time 1.10 seconds



A/P ANKLE MOMENT (lbf. in.)

fig. 5.22

A/P ANKLE MOMENT IN WALKING UP RAMP WITH S.A.

cycle time 1.44 seconds

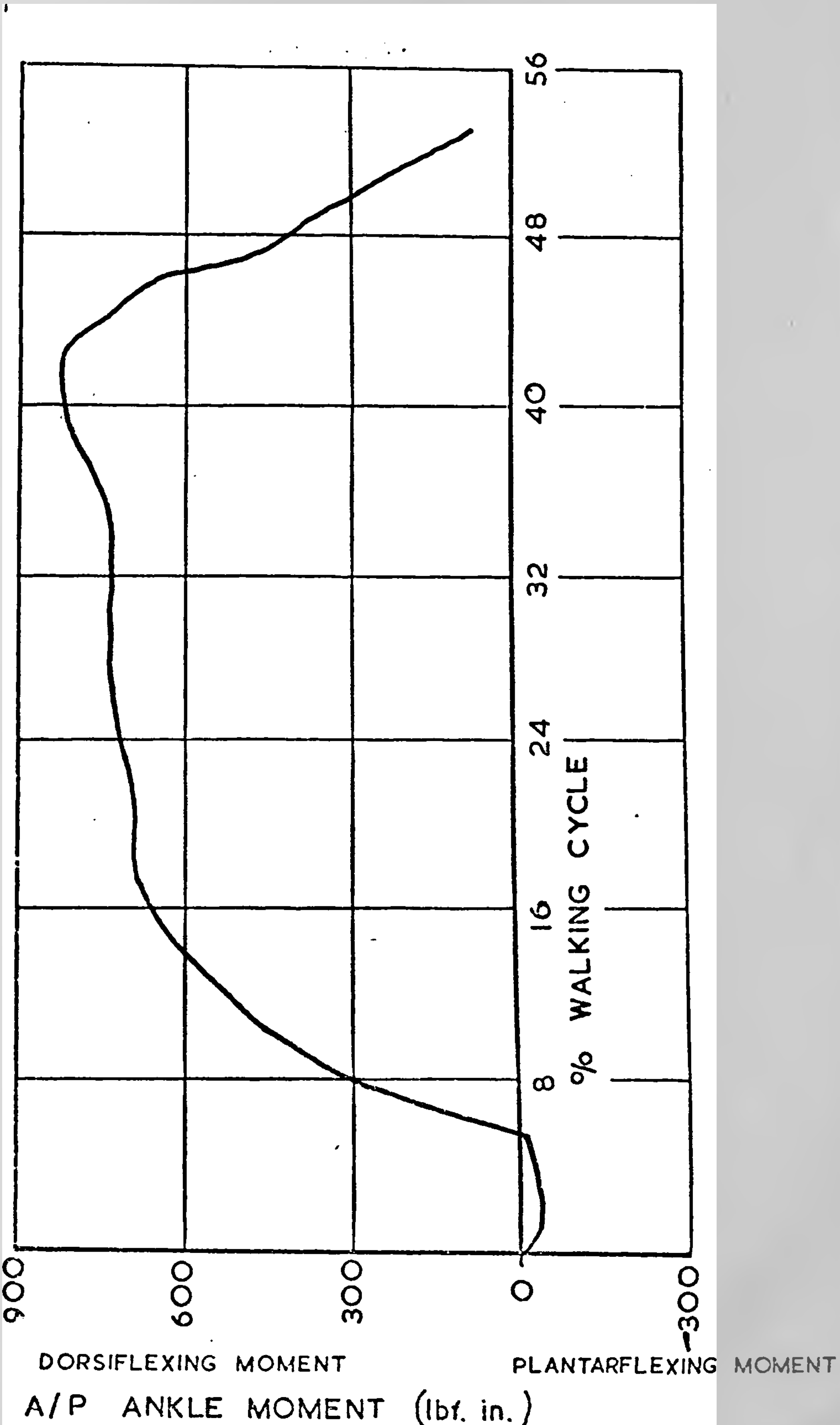


fig. 5.23

M/L HIP MOMENT IN LEVEL WALKING WITH S.A.

cycle time 1.10 seconds

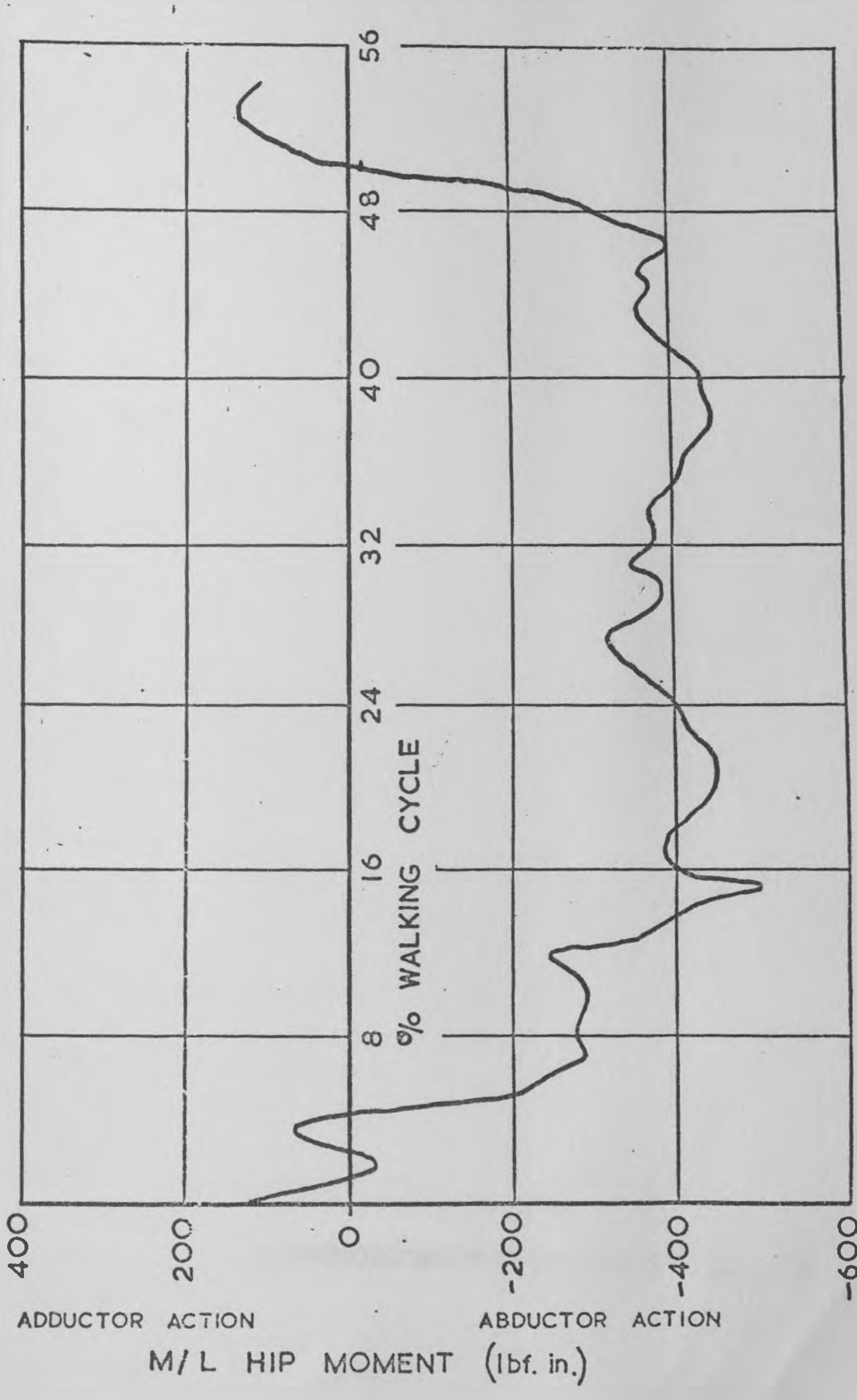


fig. 5.24

M/L HIP MOMENT IN WALKING SIDEWAYS (Prosthesis Leading) WITH S.A.

cycle time 1.00 seconds

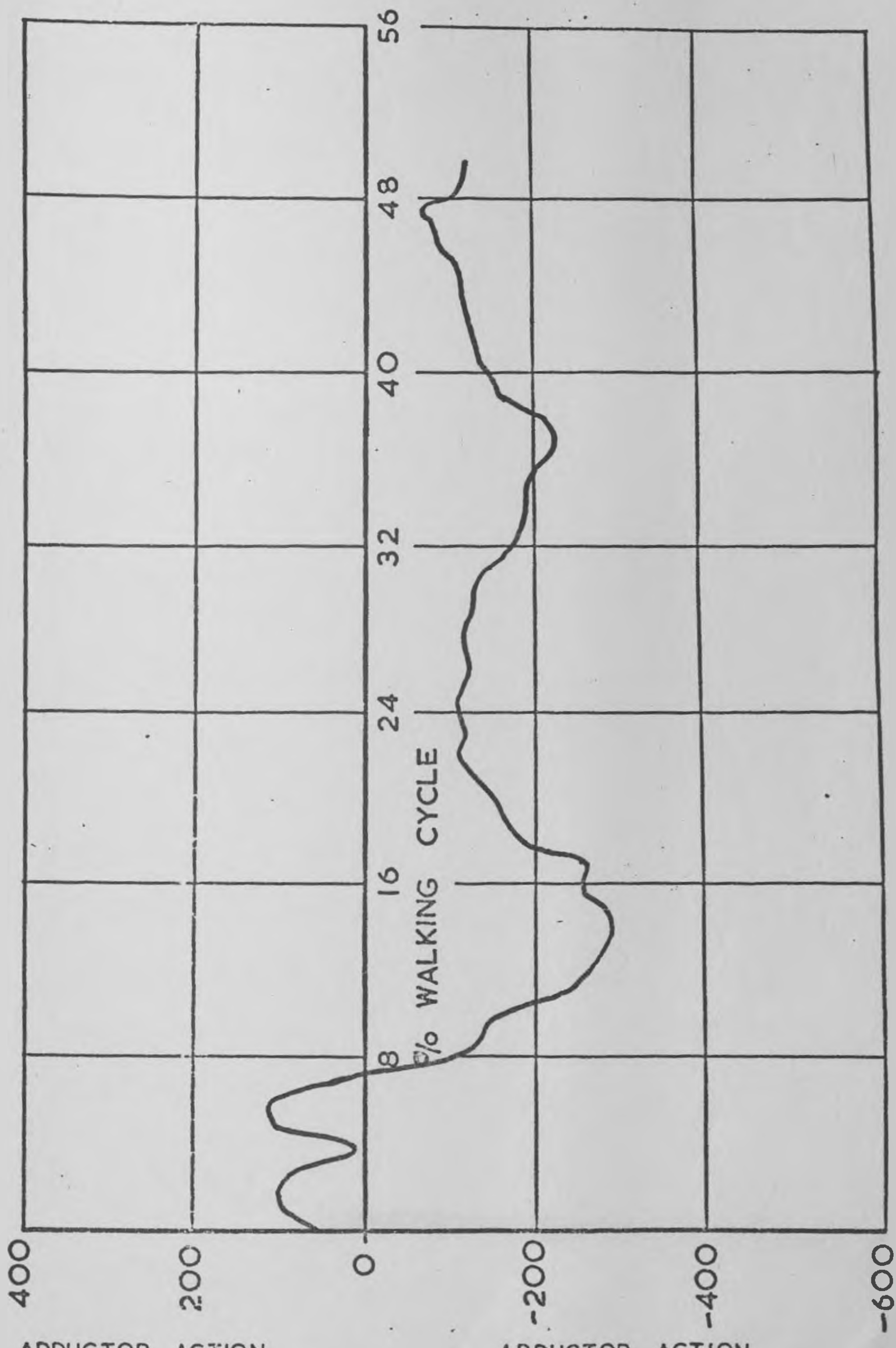


fig. 5.25

Fig. 5.26 (normal leg leading). With the prosthesis leading, very little M/L hip moment is required but when the normal leg is first, the largest values of 500 lbf. in. at 3% of the cycle and -840 lbf. in. at 35% of the cycle.

5.5.4 Torque about the long axis of the pylon In level walking (Fig. 5.27) the torque builds up to a maximum of -60 lbf. in. at 42% of the cycle. This is caused by rotation of the pelvis relative to the prosthesis as the walking cycle progresses. As the load is reduced on the prosthesis, the artificial leg can rotate back and release the stress built up by this action. However the inertia of the prosthesis causes the torque to increase to +30 lbf. in. before toe off. A damped oscillation then occurs at the start of the swing phase.

The maximum torque -80 lbf. in. at 0.7 sec. occurs in stepping over an object (Fig. 5.28).

5.5.5 Resultant load The typical double peaking of the resultant load variation reported by other workers in the field (U.C.B. (1947)) is observed in the majority of the activities. The body weight of the amputee is 160 lbf. and is shown on the graphs. The initial rise in level walking (Fig. 5.29) is 175 lbf. in. at 17% of the cycle, whereas the second is 165 lbf. in. at 42% of the cycle.

The highest resultant load on the prosthesis occurs in running (Fig. 5.30) and is 210 lbf. at 13% of the cycle.

5.5.6 Comparison of results with published work The main reference for A/K amputee force actions is the U.C.B. Report (1947). No results of moments

M/L HIP MOMENT IN WALKING SIDEWAYS (Normal Leg Leading) WITH S.A.

cycle time 0.82 seconds

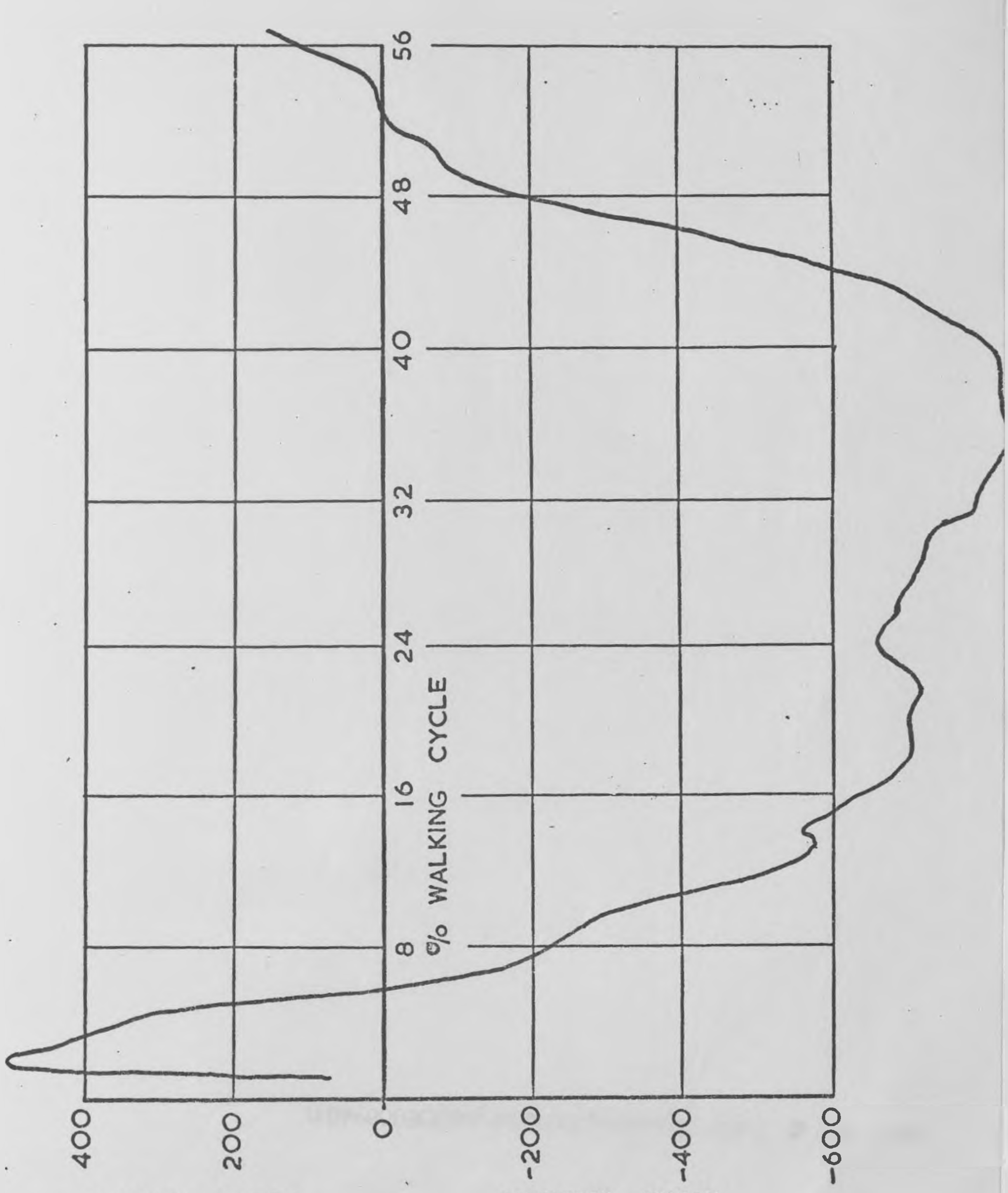


fig. 5.26

TORQUE IN LEVEL WALKING WITH S.A.

cycle time 1.10 seconds

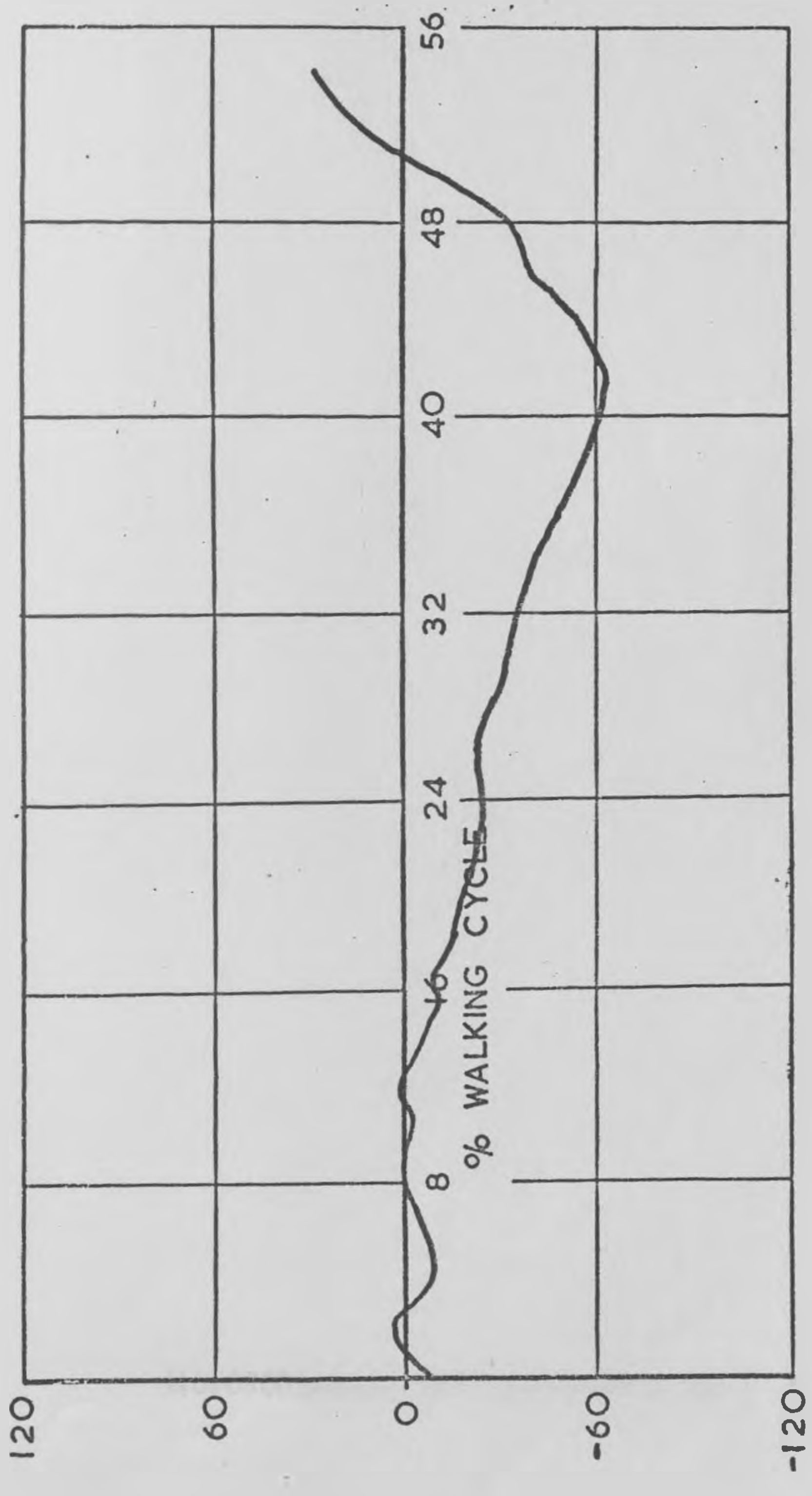
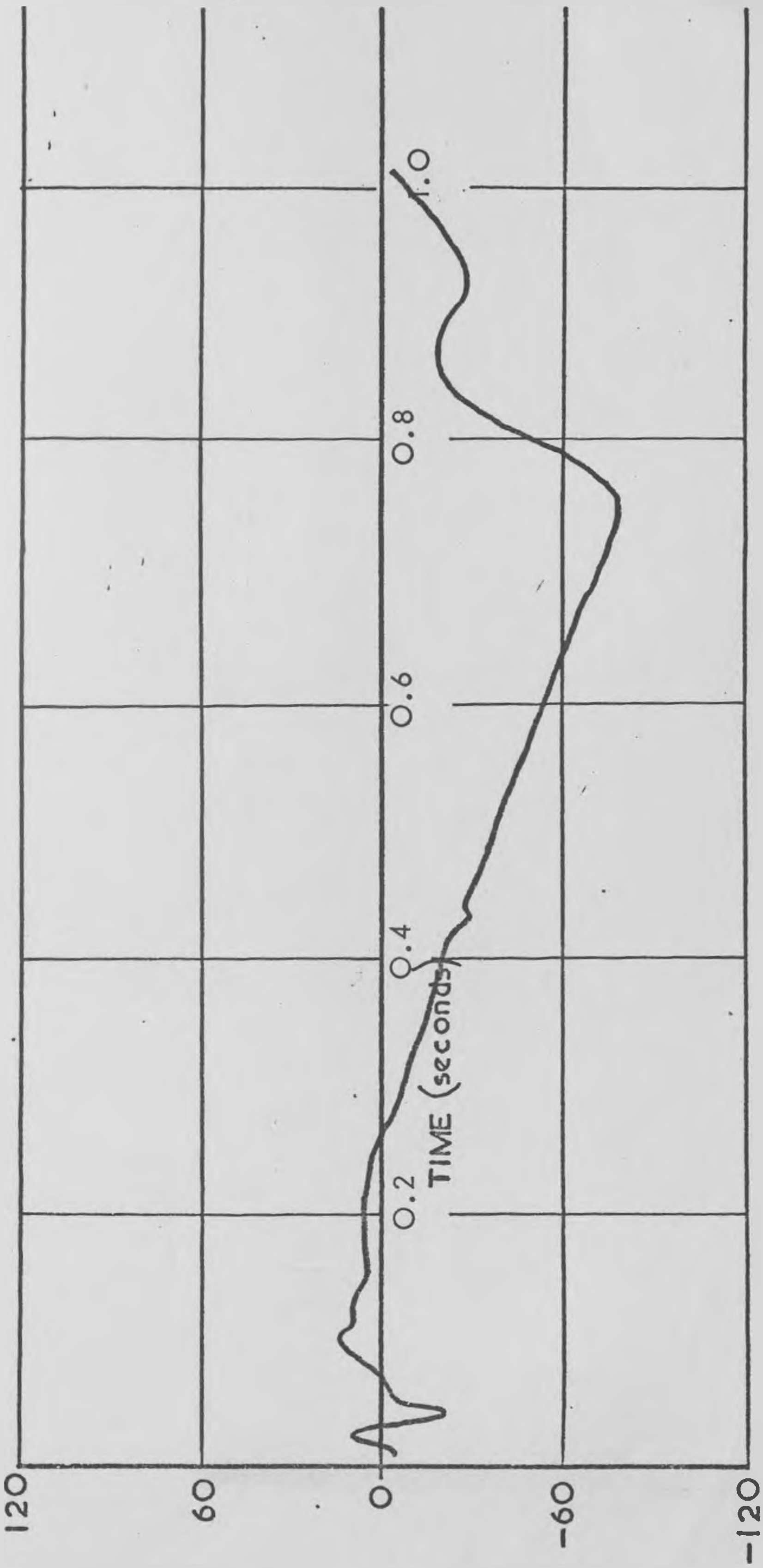


fig. 5. 27

TORQUE IN STEPPING OVER AN OBJECT (from a standing position) WITH S.A.



TORQUE (lbf. in.)

fig.5.28

RESULTANT LOAD IN LEVEL WALKING WITH S.A.

cycle time 1.10 seconds

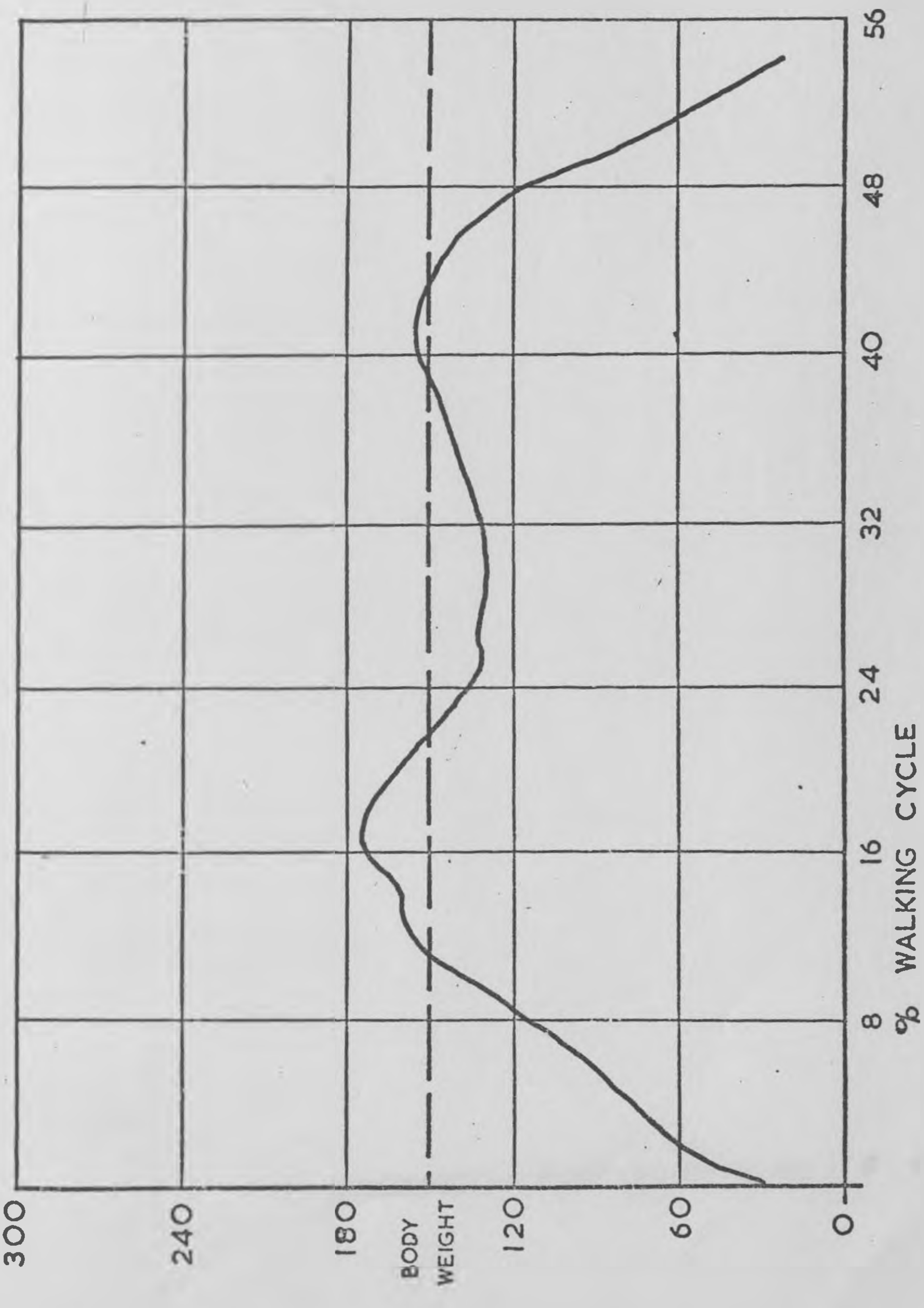


fig. 5.29

RESULTANT LOAD IN RUNNING WITH S.A.

cycle time 1.09 seconds

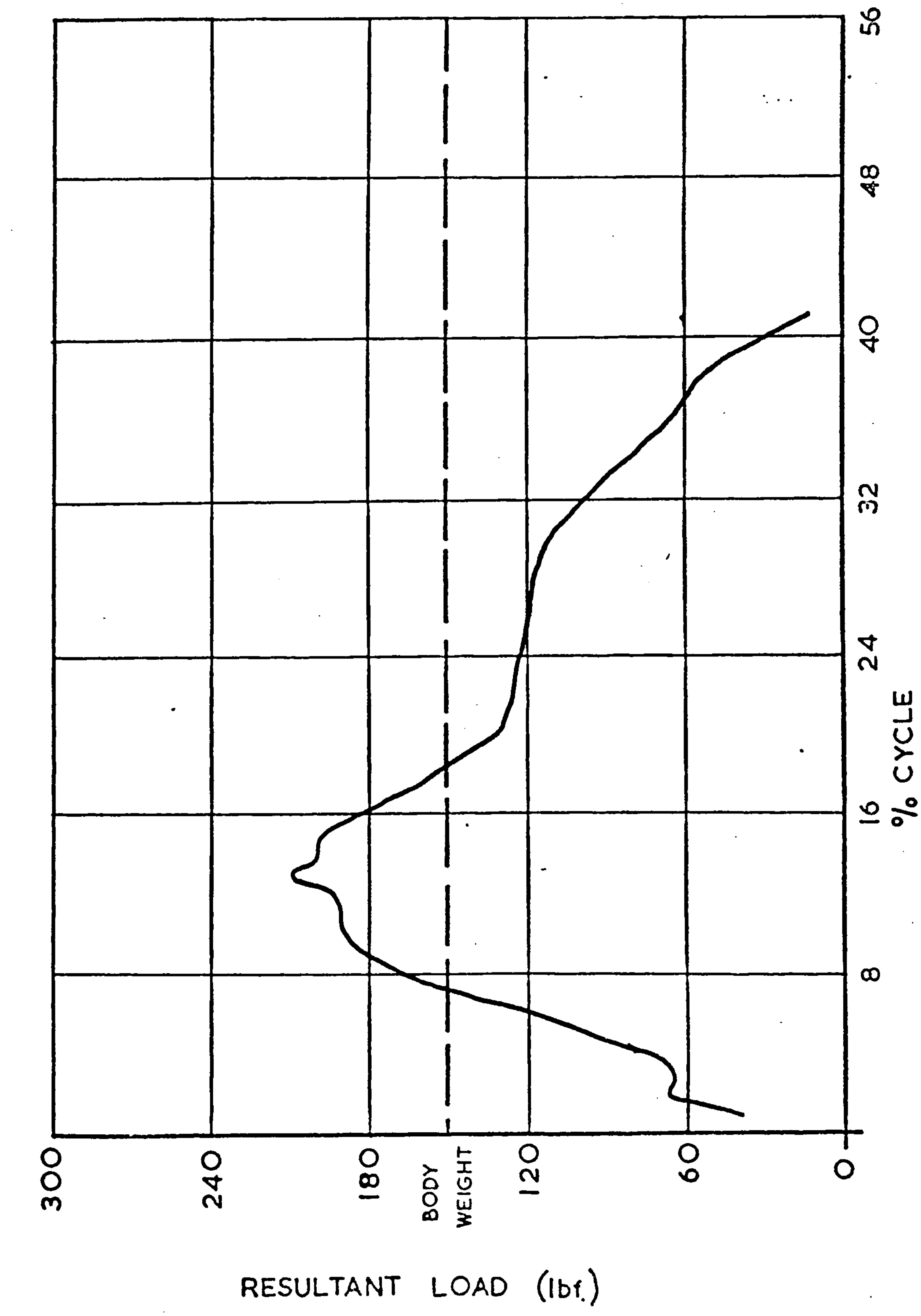


fig. 5.30

about the hip joint are available in this report but other force actions were measured. Fig. 5.31 is a selection of graphs presented in this report for force actions in level walking.

The amputee used weighed 145 lbf. and his cadence (45 strides per min.) is slightly slower than in the tests reported in this thesis. A suction socket was used and the results were obtained from a strain-gauged pylon.

The resultant load is lower in the U.C.B. Report but the general shape of the curve is the same. The fall in load is not as marked since the cadence is less.

A different sign convention has been adopted for the torque, but U.C.B. show a slightly reduced value of the maximum (55 lbf.in. against 60 lbf.in.).

The ankle moment again has a lower maximum value (580 lbf.in. against 750 lbf.in.) but the -ve maxima are identical. The lower value of resultant load will cause the ankle moment to be less.

The graphs of the variations in knee moment from the two sets of results are virtually the same.

5.6 Ideal Mechanism Characteristics

Schede (1919) defined the objectives of a knee control as follows:-
"It is, therefore, necessary that the bent and loaded knee be controlled without movement and above all that it can be actively extended".

At the present state of knowledge, this ideal cannot be realised. A knee device is required which allows the amputee to carry out the normal activities of his daily life with the minimal amount of inconvenience due to his deficiency.

It is essential that the knee is stable during the "critical phases" of amputee activity. A critical phase can be defined as:- That part of the activity in which

TYPICAL FORCE ACTION VARIATIONS FROM THE U.C.B. REPORT (1947)

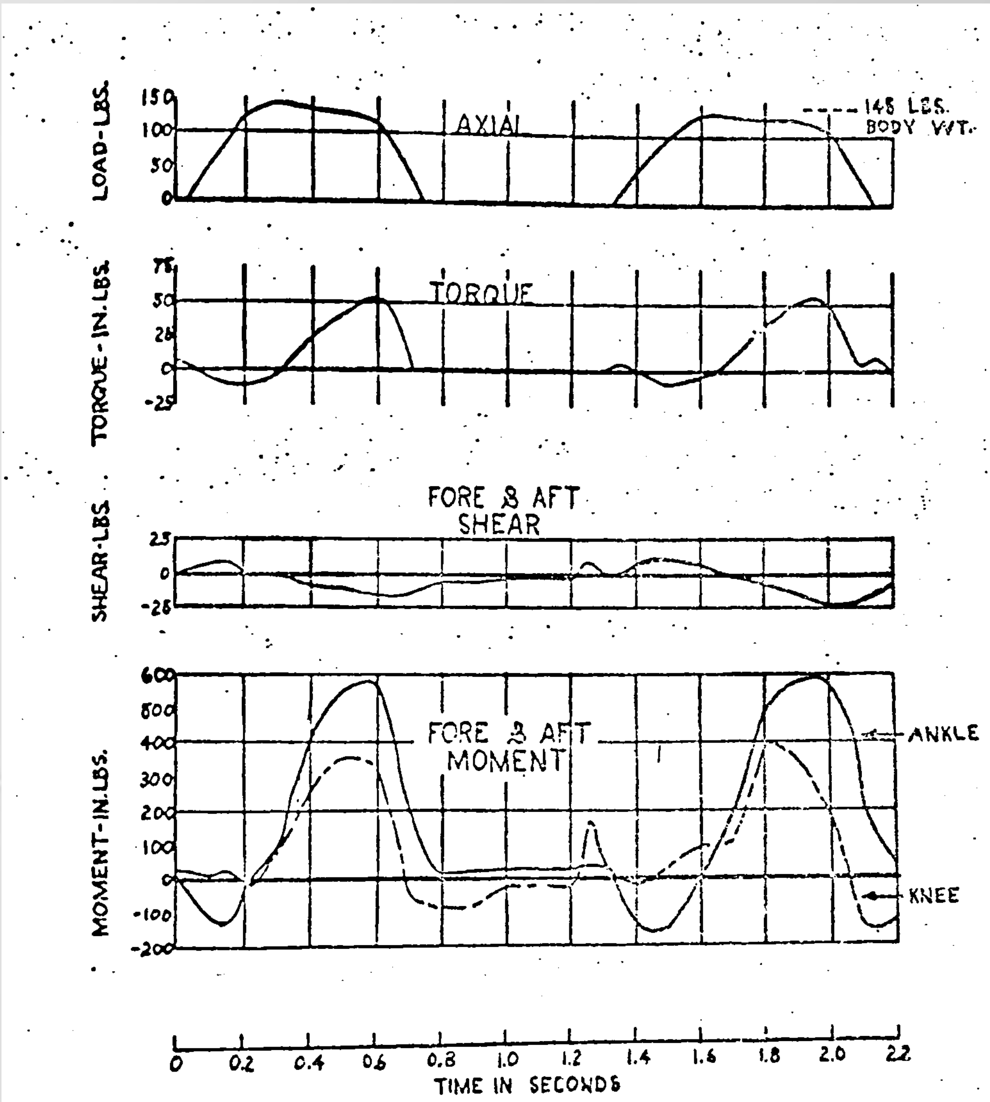


Fig. 5.31

the knee would flex, due to the load imposed on the prosthesis, if no measures were taken to prevent it. In terms of the results presented here, this can be interpreted as; any part of an activity when the hip moment rquired for stability is positive.

A further requirement of the knee mechanism is that the amputee should be capable of initiating knee flexion when it is necessary, without undue difficulty. The first point just before toe off at which the hip moment produced is equal to that required for knee stability in the case of the single-axis, non-stabilized knee, is the time when the amputee begins to flex the knee.

The ability of any device to meet these two conditions can be investigated by the use of a static stability rig such as that described in MacGregor (1968).

MacGregor measured the moment resisting flexion at the knee of various devices when the leg was loaded between the simulated hip joint (on the static rig) and the rear of the foot. Similar tests were carried out when the load was applied to the front of the foot. These results have limited application since the point of maximum instability does not occur at heel strike but at the point where M_{HRZ} is a maximum. Further, when knee flexion is initiated, the load is applied not to the front of the foot but posterior to this. These points of loading are indicated in Fig.5.32.

Static tests in general have limited application since it is apparent from the investigations described in this report that the hip moment applied by the amputee has functions other than stabilization of the knee. The static rig can however give important information on the characteristics of a knee device if the load at the important phase in the cycle and its point of application are determined in the manner described below.

COMPARISON OF LOADING POINTS ON STATIC RIG

A Points used by MacGregor

B Approximate point when maximum knee instability is present i.e. M_{HRZ} is maximum.

C Approximate point when initiation of knee flexion occurs i.e. M_{HRZ} and M_H graphs intersect.

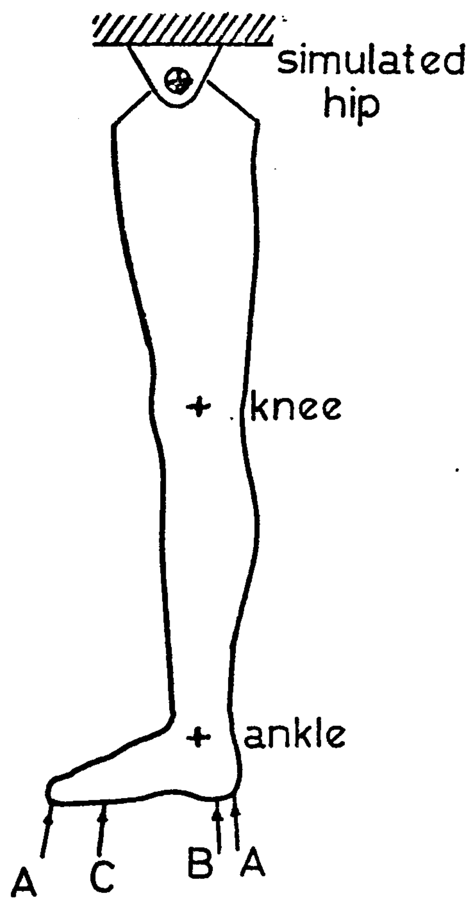


Fig. 5.32

From the variation of M_{HRZ}, the percentage cycle time of the critical phase of an activity can be found, in particular the maximum. By consulting the A/P ankle moment results, the corresponding bending moment at the ankle can be obtained. In a similar manner, the resultant load at this time is also estimated from the particular graph of resultant load. Since the angle between the pylon axis and the resultant load line is small, when the centre of pressure on the foot is near to the ankle, the distance of the centre of pressure from the ankle can be estimated by dividing the ankle moment by the resultant load. As the centre of pressure moves along the foot towards the toe, bending of the foot causes the x coordinate to decrease. A similar calculation can therefore be carried out to determine the load conditions and point of load application at the initiation of knee flexion.

The leg can be aligned in the static rig with the load found by this technique applied at the calculated distance from the ankle joint, and the hip moment required to flex the knee or to maintain extension can be established.

As an example consider level walking with the S.A. for a man of 160 lbf. The M_{HRZ} graph (Fig. 5.3) has a maximum extensor action value at 8% of the cycle and the point of intersection with the hip moment produced is at 47% of the cycle. Consulting the A/P ankle moment graph (Fig. 5.22) and the resultant load graph (Fig. 5.29) the results obtained for the two points of importance are:-

For maximum extensor action	R = 115 lbf. and M_{HZ} = 150 lbf. in.
For knee flexion	R = 130 lbf. and M_{HZ} = 600 lbf. in.

∴ To investigate the performance of a stance phase knee mechanism in level walking statically, the two tests to be carried out are:-

(1) The hip moment required to maintain stability of the leg when a load of 115 lbf. in. is applied at a point on the foot $\frac{150}{115}$ in. (i.e. 1.3 in.) posterior to the ankle joint.

(2) The hip moment required to flex the knee when a load of 130 lbf. is applied at a point on the foot, $\frac{600}{130}$ in. (i.e. 4.6 in.) anterior to the ankle joint.

If the amputee has a weight different from 160 lbf., the values indicated above for load must be multiplied by the ratio of the amputee's weight to 160 lbf.

5.7 Mechanics of Force and Moment Transmission at the Stump/Prosthesis Interface

For an A/K amputee to transmit the force to the trunk from his artificial leg as in a normal individual, it would be necessary to connect the prosthesis directly to the femur. At this time this is not possible and it may in fact turn out that it is not desirable. The main problems are rejection of the implanted part of the prosthesis and infection at the point where the artificial leg passes through the skin. However, research into the possibility of direct femoral attachment is being carred out Esslinger (1964-66). The results obtained with dogs are very encouraging and it appears that an excellent end bearing stump can be achieved even if direct femoral attachment is impractical. The technique used is to insert a mushroom shaped piece of silicone rubber into the end of the femur after amputation and suture the skin over this. The next stage would be to attempt to have this insert actually protruding through the skin and for complete healing to occur so that the internal environment is effectively sealed from possible infection. An artificial leg could then be attached to the plastic.

Although most authors agree that, excluding direct femoral attachment, end bearing is the best method of supporting the body of an artificial leg, it is not often possible due to discomfort to the amputee. The early German writers carried out a tremendous amount of therapy on their patients to achieve a degree of end bearing and claims of 60% and 70% A/K end bearing stumps have been made, Blencke (1917). The method used at the present time is to take the load mainly on the ischial tuberosity ; although the possibility of a total surface bearing socket is being examined at Roehampton.

Considering the mechanics of the ischial bearing socket, it is obvious that since the ischial tuberosity is medial to the hip joint, the amputee gains some mechanical advantage in abduction. In respect of the A/P plane the ischial tuberosity is posterior to the hip joint and hence the amputee has an advantage when flexing the hip joint but is at a disadvantage when extending it.

As Schede (1919) observes, when load is taken on the ischial tuberosity, the pelvis must fall forward. The trunk attempts to maintain itself upright, and lordosis of the lumber vertebrae is caused by the pelvic rotation. The limiting point of pelvic rotation is reached when further lordosis is impossible. However the stump moves with the pelvis until the end of it presses against the posterior socket wall and its anterior surface presses against the anterior upper socket brim. This results in a forced position of the pelvis by the trunk on one side and by the stump on the other. This means that in the latter part of the stance phase, hyperextension is often difficult and a danger of premature knee flexion exists.

Once solution proposed by Schede was a different design of socket.

The result is seen in Fig. 1.18. A steel strip goes obliquely down under the ischial tuberosity, round the adductors, climbs tightly over the inguinal ligament to the iliac crest. Then it bends round and is connected to its other end at the hip joint. The A/K socket is then joined by a steel link to a point on the pelvic socket corresponding to the hip joint.

Anderson et al (1960) report that 10° of anterior pelvic rotation can be telerated without damage to the spine. An amputee can hyperextend his hip 5°, so the total angular movement possible is 5° + 10° i.e. 15°.

Brunnstrom (1947) also stresses the importance of realising the limitations of ischial bearing, and reports that an A/K amputee should be able to extend the thigh slightly more than the normal individual. By setting the stump in slight flexion in the socket, some assistance is given in obtaining this.

Another problem of ischial bearing is the atrophy of the tissues of the stump. Nieny (1917) reports that Kienbock, Sudeck and Sick have shown that not only muscle but bone also atrophies when the load is taken on the ischial seat. Other workers since have reiterated this finding. Recently Loon (1960) found that osteoporosis set in in the pelvis as well as the femur. He also found degeneration in the hip joint. An end-bearing prosthesis would assist circulation in the bone and cartilage.

To transmit the moment to the socket from the femur, considerable lost motion occurs due to the relative softness of the tissues between these two solid members. Research into this problem was first carried out by Blumenthal (1917) who used X-rays to examine the movement of the femur in the socket.

5.8 Selection of Mechanism and Alignment for Individual Cases

Knee mechanisms can be investigated under various load conditions in the static stability rig discussed in section 5.6. It is suggested that the critical phases and the points of knee flexion of normal activity be established by the technique also described in section 5.6. The range of these activities should be extended to cover sporting activities etc. By subjecting the device to the load indicated by this method, applied at the position on the foot calculated and finding the hip moment required to stabilize the knee or flex it, the limitations and advantages of the particular mechanism can be established.

When an amputee requires fitting with a leg, he is asked to complete a questionnaire enquiring about his hobbies, sports, do-it-yourself and any other activities which he would want to continue. An interview should then be arranged to discuss the answers to the questionnaire and any other particular difficulties or problems that may arise.

The hip moment capability of the amputee should be objectively measured through the complete range of hip movement used by him in normal activity. The graph presented as Fig. 5.33 shows the hip moment capability of the amputee used in these tests (as measured by the technique described in section 3.5), through the range $+20^\circ$ to -20° of hip flexion. The hip extensor capability decreases linearly with angle from 750 lbf.in. as the effect of Gluteus maximum diminishes to zero at -15°. Hip flexor capability reaches a maximum at -13° of 700 lbf.in. If these moments are compared with those obtained in the tests, it is seen that much higher values have been recorded. This is not important since only comparisons between the hip moments will

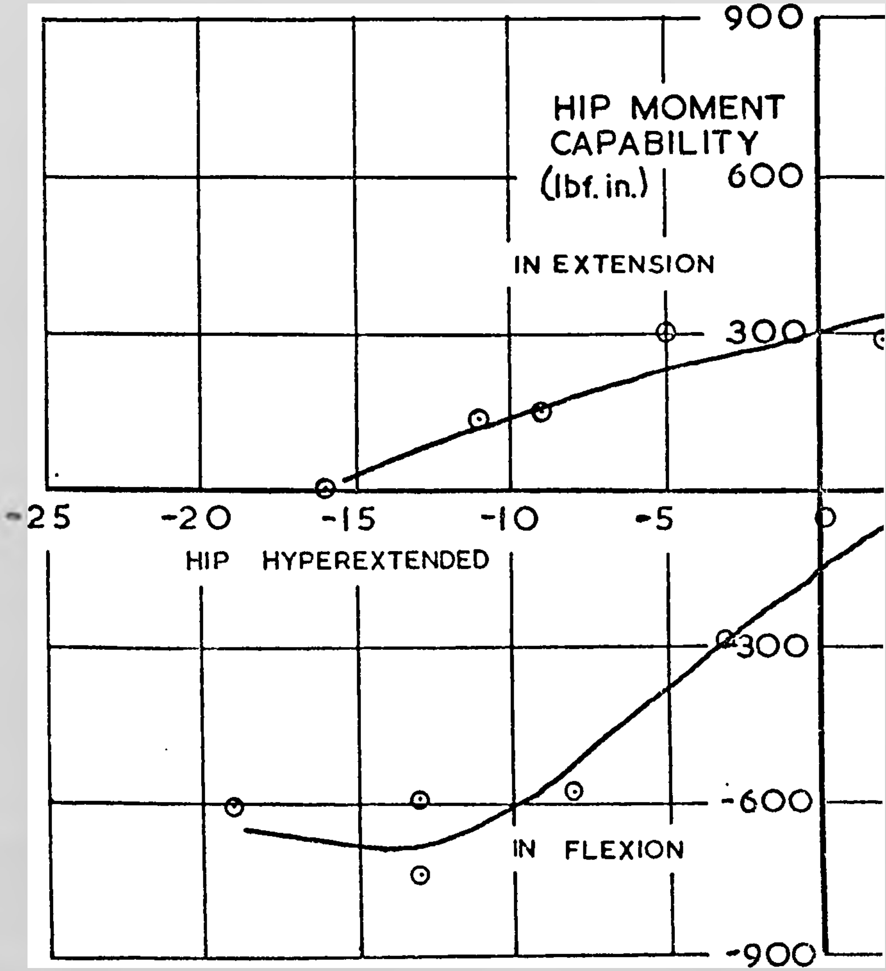
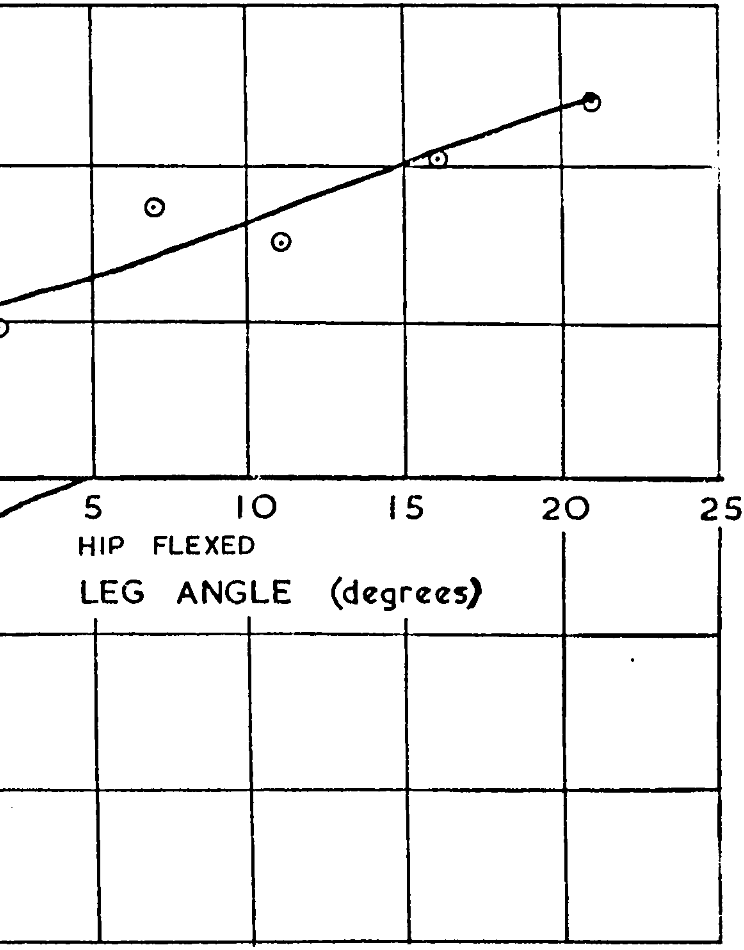

900
HIP MOMENT
CAPABILITY
(lbf. in.)
600
IN EXTENSION
300
-25
-20
-15
-10
-5
0
HIP HYPEREXTENDED
-300
-600
IN FLEXION
-900

Fig.5.33



VARIATION IN HIP MOMENT CAPABILITY WITH HIP ANGLE be made, and Grieve (1969) has pointed out the possibility of exceeding in a dynamic situation the isometric strength of muscules.

It is important, however, to measure hip moment capability in as near as possible the conditions which are expected in normal activity. It is suggested that the socket to be used in the final prosthesis, be fitted to the amputee with a tube of the correct length, connected between it and the foot. The strain-gauged pylon is incorporated in the tube and the test carried out in the manner described previously. It must be remembered that the hip moment capability is not purely a muscular action but also depends to a large extent on the contact between socket and amputee.

By consulting the results from static tests on the requirements which the amputee expects from the leg the knee device can be chosen. Some consideration must be given even at this stage to the hip moment capability as it may in some cases influence the choice of device used.

The leg can now be assembled, adjusted and aligned in the normal way, but with the strain-gauged pylon incorporated in the shank. Assuming that a system exists as is described in Chapter 6, it is suggested that the following approach be made to final adjustment and alignment.

The amputee is asked to walk up and down to get used to the leg and the variation in hip moment examined on an oscilloscope. The muscular effort (me) is displayed on a digital voltmeter after each stride and the leg is adjusted and aligned so that me is a minimum but with the following provisos:

(1) The hip extensor moment and hip flexor moment should be biased according to the amputee's capability, i.e. if he has good extensors but weak flexors, align the leg so that he requires less flexor action and more extensor action.

(2) It is better to hyperextend slightly more than necessary since:

(a) with ischial bearing, very little "muscular" effort is required to flex the knee. He simply sits on the ischial seat.

(b) extensor action is required for a longer period of time than flexor action.

By adopting these criteria, a better alignment for the devices investigated here would have been:-

G.R. and O.B. move knee centre back.

U.C.B. move knee centre forwards.

Finally the ability of the amputee to negotiate steps, ramps and uneven surfaces should be investigated.

CHAPTER 6

SUGGESTIONS FOR FURTHER WORK

CHAPTER 6

6.1 Experimental Technique

The development of pylon orientated techniques of force analysis in amputee activity should take two different paths. The one being a clinical tool and the other a research instrument for fundamental studies.

6.1.1 Clinical pylon

The requirements of any device to be made available for general use under clinical conditions are as follows:

(I) It should be inexpensive

(2) It must be simple to use, as untrained personnel may be required to operate it.

(3) The information must be made available as soon as possible. Ideally any calculations should be carried out "on-line". However, the possible use of a hybrid computer should not be overlooked providing the cost is not prohibitive.

(4) It should be as robust as possible. Any delicate mechanism on the device should be fully protected against accidental damage and mis-use by the operator.

Bearing these principles in mind the clinical pylon can be developed.

In its final form it is envisaged that the calculations at present carried out on a large digital computer be performed by a small on-line computer on the amputee himself and the output be transmitted by a telemetry link to the display console. However, the effect of the inevitable simplifications involved must be fully evaluated and understood in all the possible activities that it is intended to examine. It is suggested that this work be carried out in a research centre equipped with diverse facilities that is: digital computer, photographic facilities etc., using a system similar to that described below.

As an interim stage the analogue computation can be carried out away from the amputee where the limitations of size and weight do not restrict modifications to circuitry and do not preclude the use of sophisticated matching components.

6.1.2 The use of the pylon for fundamental research The technique of data processing used should be improved by the use of analogue to digital converters, with a suitable interface, replacing the manual method described here. A sampling frequency of approximately 1 m.sec. per channel is suggested. The complete set of input information for one sample would then take 7 m.sec., and the frequency response would be of the order of 70 H_Z (see section 4.5). To examine in more detail the force actions at the time of heel strike, the sampling frequency would have to be doubled. This is the lowest sampling frequency at which the 100 H_Z content of the signal at heel strike would faithfully be reproduced in digital form.

6.2 Further Investigations

The results from the pylon should be compared with those from the force plate by filming an amputee walking over the force plate whilst wearing a strain-gauged pylon.

Techniques for obtaining information on the spatial configuration of the limb segments without resorting to filming should be developed. It is hoped that Appendix III may prove to be of use in this respect. The dynamic effects at heel strike and the contribution due to the weight of the artificial leg could be calculated.

In further studies on other amputees, it is suggested that the most important of those activities investigated here with regard to knee stability are level walking, walking down stuirs, walking down ramp and stepping over an object. However the range of tests should be extended to cover walking over uneven, soft and polished surfaces.

Similar investigations on hip disarticulation amputees and through knee amputees would reveal any specific differences in their use of knee devices.

The relationship between muscular effort, as defined in Chapter 4, and metabolic cost should be investigated, and an ideal specification of amputee gait arrived at, if possible. This entails consideration of energy cost to the amputee as well as possible degenerative effects and force actions in the hip joint, and the joints in the remaining leg.

The effect of learning should be examined by carrying out tests on an amputee throughout his walking training. An attempt could be made to correlate hip moment capability at each stage with the physiotherapy treatment of the amputee.

CHAPTER 7

CONCLUSIONS

CHAPTER 7

The following conclusions have been based on the results of investigations into the gait of one amputee with various knee mechanisms. The ability of the amputee to learn to use his leg in the best possible way has not been examined.

The strain-gauged pylon can be incorporated into a prosthesis and provide useful information for both clinical and research applications.

Its development into a clinical tool and research instrument is a viable proposition.

A static rig is desirable to assess knee mechanism characteristics, thereby assisting in the selection of a device to suit any amputee.

The hip moment produced by an A/K amputee in level walking is about three times that required for knee stability in the early part of the stance phase when no mechanism is incorporated at the knee joint. The hip moment was reduced in some activities but there were very few occasions when the amputee approached knee instability.

Generally an amputee can make use of a knee device but not to the extent theoretically possible.

The danger of knee instability is greatest just after heel strike, (8% of the cycle in level walking).

The most unstable activity investigated was walking down a ramp.

The Lammers knee gave the best stance phase control of all devices but the lack of a swing phase control restricted the amputee's cadence.

The most stable knee device tested was the Blatchford stabilized knee. In fact the amputee was able to bear his whole weight on the leg when the knee was flexed to an angle of 25°.

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