FORCES AT THE HUMAN HIP JOINT.

Thesis submitted for the Degree of
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by

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NOTATION.

All terms are defined when they are first introduced. The following list defines the major terms; constants in equations are not included. Where symbols are used with suffices this is indicated below by the suffix o. Suffices and superfices are listed separately. The same symbol is used more than once in some cases to avoid complication in lettering. The context generally indicates the quantity intended.

a	span of ant. sup. iliac spines.
A	cross sectional area.
A/A	ankle in opposition.
b	span from L.ant. sup. il. spine to tail marker.
c	" " R. " " " " " " "
c	temperature coefficient of resistance
C_o	Fischer coefficient see Table 3. p.
C	Modulus of Rigidity.
d	differential operator.
e	exponential constant.
e	$x_B - x_H$
E	elastic stiffness or Young's Modulus.

f	$y_B - y_H$
f	step frequency or camera frequency.
F	force.
g	gravitational constant.
g	$z_B - z_H$
G_o	reading of galvanometer recorder.
h	height of camera above reference axes.
H	heightof test subject.
H_o	component of resultant leg/trunk force.
HS	heel strike.
i	electrical current.
I_o	second moment of cross-sectional area.
IF	inertia force.
J_o	component of resultant joint force.
k_o	radius of gyration.
k_o	length/I for force plate member.
K	viscous damping factor.
K_o	force plate calibration factor.
L_o	length of body segment.
LF	left foot.
m	natural frequency of damped vibrations.
m	fractional change in electrical resistance.

M	mass
M_o	turning moment.
n	natural frequency of undamped vibrations.
n	indefinite number.
p_o, P_o	tension in muscle group.
P_o	force in dynamometer strip.
Q	" " " "
r	coefficient of correlation.
r_o	moment arm of muscle group about reference axis.
R	electrical resistance.
R	force in dynamometer strip.
R_o	camera distance from reference axis.
RF	right foot.
S	general position co-ordinate.
S	strain gauge calibration constant.
t	arbitary time.
t	"Student" t factor.
t_o, T_o	direction cosine.
T	temperature.
T	cycle time for one leg.
TO	toe off

U	strain energy.
v, V	velocity.
V	direct load.
V	electrical voltage.
W	subjects body weight.
W_o	subject's body segment weight.
W_o	ground to foot force component.
$x_o\ X_o$, $y_o\ Y_o$, $z_o\ Z_o$	space co-ordinates. See Fig. 47, p. 152.
α, β	phase angle
γ, δ	projected inclination of femoral axis.
δ	indefinitely small fraction.
∂	partial differential operator.
Δ	damping coefficient.
η	error.
ϕ	change of slope of structure.
ϕ	projected inclination of femoral axis.
σ	longitudinal stress.
θ_o	projected inclination of body segment.

ω	angular velocity.
ω, Ω	frequency of forcing force.

SUFFICES.

A	ankle.
B	left ant. sup. iliac spine
C	tail marker.
D	right ant. sup. iliac spine.
F	centre of gravity of foot.
H	hip joint centre.
K	knee
L	leg
O,o	initial position.
P	'foot' i.e. 5th metatarsal - phalangeal joint centre.
S	centre of gravity of shank.
T	centre of gravity of thigh.
W	walkway i.e. pertaining to ground/foot interface.

SUPERFICES.

$\dot{}$	first derivative with respect to time
$\ddot{}$	second " " " " "
$'$	apparent displacement.

ABSTRACT.

The subject matter considered in this thesis is the analysis required to obtain the force transmitted between the loaded surfaces of the human hip joint. This information is desirable for the design of implants to repair fractured bones or to replace diseased joints or, more generally, toallow the discusssion of the functional behaviour of the joint in the normal person. No published work has been found giving values of hip joint force patterns during walking for normal subjects. Procedures have been described in the literature for the determination of the resultant force actions between body segments but this thesis presents the first application of dynamic measuring techniques to the functional anatomy of the body in order to determine the internal force actions.

The relevant anatomy of the hip region is described in a preliminary chapter. Thereafter, a chronological review of the studies of human body dynamics is presented covering the time period up to 1890. Further publications after that date are presented in chronological order under the separate headings of

(1) the analysis of human gait and the corresponding forces in muscles.

(2) determination of body mass properties.

(3) physiology and dynamics of muscle action.

The work undertaken by the author comprised the application and modification of known engineering techniques for dynamic and kinematic measurement to the analysis of the motion of the human body. The resultant force actions between the ground and the foot of a test subject were measured by a force platform and the positions of his leg in space were recorded by cine cameras. The resultant force and moment actions transmitted across a section of the leg through the hip joint could thus be calculated. From a consideration of the time pattern of action of the hip muscles as demonstrated electromyographically, and their spatial configuration, muscle groups were defined and consideration of the resultant forces in the groups allowed the calculation of the hip joint force. The calculation procedure required the assessment of the mass properties of limb segments, and their spatial accelerations, and was performed for every fiftieth of a second interval for a complete walking cycle for each test. The procedure was therefore arranged for calculation using a digital computer and this programme, devised by the author, is outlined in the text.

Results are presented of the analyses performed on 18 tests on three female and ten male subjects. Graphs are presented of the variation with time of the resultant joint force and its components relative to the hip and to the femur. The patterns and values obtained in these tests are substantiated by results published by Dr. N.W.Rydell of the University of Gothenburg. Rydell performed tests on two patients having a strain gauged implant replacing the head of one femur. Rydell's results are included with the present author's in a statistical analysis which shows a positive correlation between (the product of body weight and stride length)and resultant joint force. The average values of joint forces on female and male subjects were 3.27 and 5.55 times body weight respectively. These values are lower bounds corresponding to the action of the muscle groups exerting the greatest moments about the joint axes. Upper bound values are also presented.

In view of the variability of the experimental quantities measured and the complex calculation procedure, the effect of these variations is subjected to critical assessment in the discussion.

REVIEW OF PUBLISHED WORK.

Introduction.

The published work on the mechanics of the human body is small in amount and diverse in topics in the time period up to the end of the nineteenth century. Thereafter there has been a rapid expansion in the volume of published work and in its detail. After a preliminary chapter on the anatomy of the Hip Region a historical survey of relevant Biomechanics is presented covering the period up to 1890. Subsequent studies of gait and corresponding forces developed in muscles and joints are reviewed separately and followed by sections on the determination of the mass properties of body segments and the physiology and dynamics of muscle action.

Applied anatomy of the hip region.

The hip joint approximates to a spherical joint and is the connection between the femur or thigh bone and the lower part of the trunk. The combination of the two hip bones, the sacrum and the coccyx are generally referred to as the pelvis. The principal landmarks of the pelvis are shown in Fig. 1. The hip bone is formed by the fusion of three separate regions:-

1. the ilium or wing of the pelvis, comprises the multiply curved region
2. the pubis, the projecting bony part in the groin, extending

FRONT VIEW SIDE VIEW REAR VIEW

A - ANTERIOR SUPERIOR ILIAC SPINE
B - ACETABULUM
C - COCCYX
D - ILIAC CREST
E - OBTURATOR FORAMEN
F - PUBIC CREST
G - ISCHIAL TUBEROSITY
H - SACRUM
S - POSTERIOR SUPERIOR ILIAC SPINE
L - 4[TH] LUMBAR VERTEBRA.

FIG. I
THE MALE PELVIS

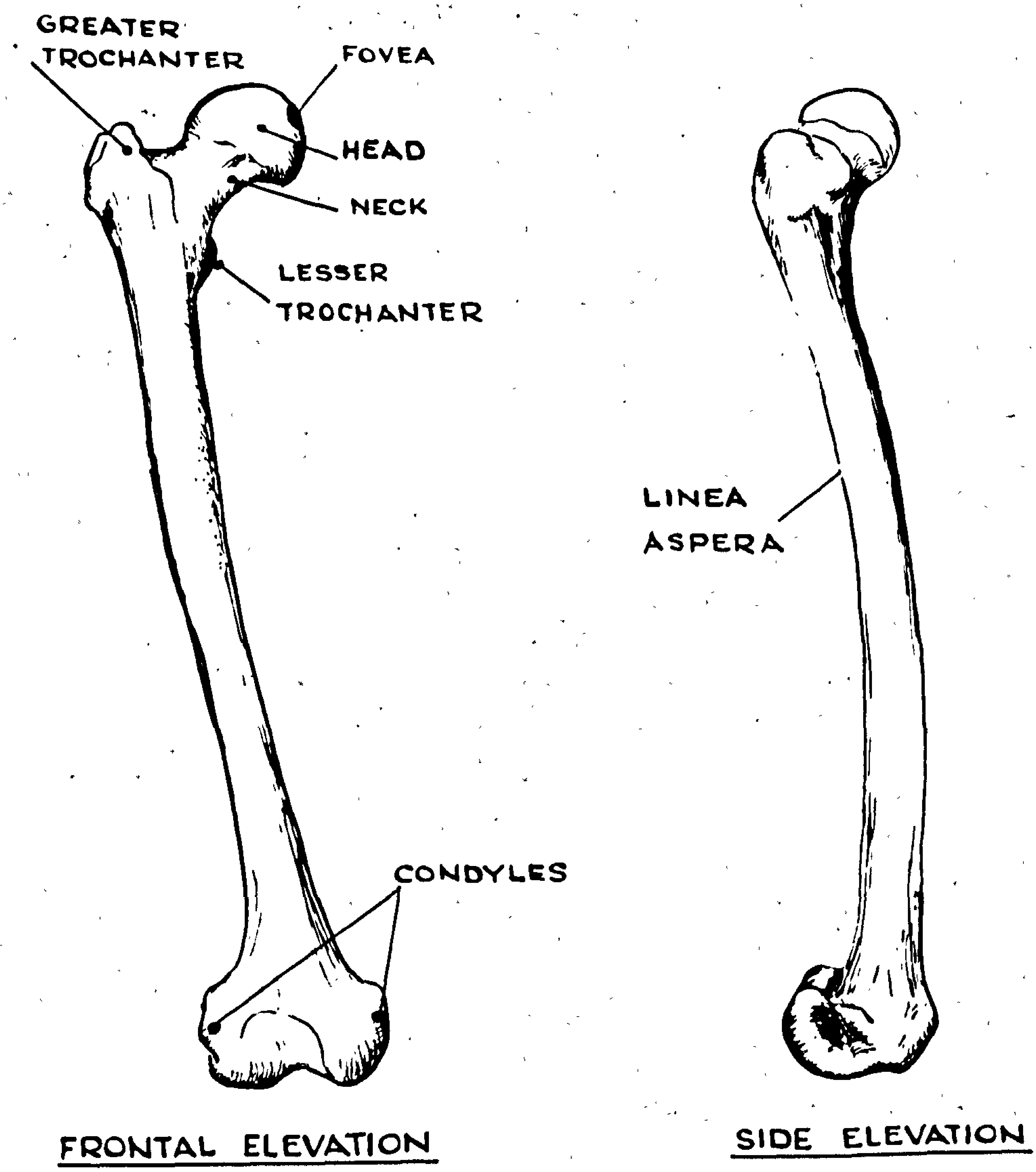
in two rami to join the acetabulum and the ischium.

3. the ischium the lowest and rearmost part which first extends downwards and backwards from the acetabulum and then projects forward to meet the pubic ramus.

Fusion of the separate parts is completed at or shortly after puberty.

The two hip bones join at the pubis in a cartilaginous joint. The sacrum is the wedge shaped bone formed by the fusion of the five sacral vertebrae, it articulates with the ilium at the sacro-iliac joint and supports the vertebral column. In the male very little movement is possible at the sacro-iliac joint. Gray (1958) states that in the female after puberty there is greater movement at this joint and that this movement increases in the later stages of pregnancy. The acetabulum is the female part of the hip joint and is a deep cavity facing obliquely forwards, sideways and downwards. The acetabulum is largely lined with articular cartilage and this cartilage is enlarged into a rim which extends round the border of the acetabulum retaining the femoral head.

The femur is the longest and strongest bone in the body. Three views of a typical femur are shown in Fig. 2. The head of the femur is entirely covered with cartilage except for a small area round the fovea where the ligament of the head is attached. The head is extended by a short neck to the junction with the shaft



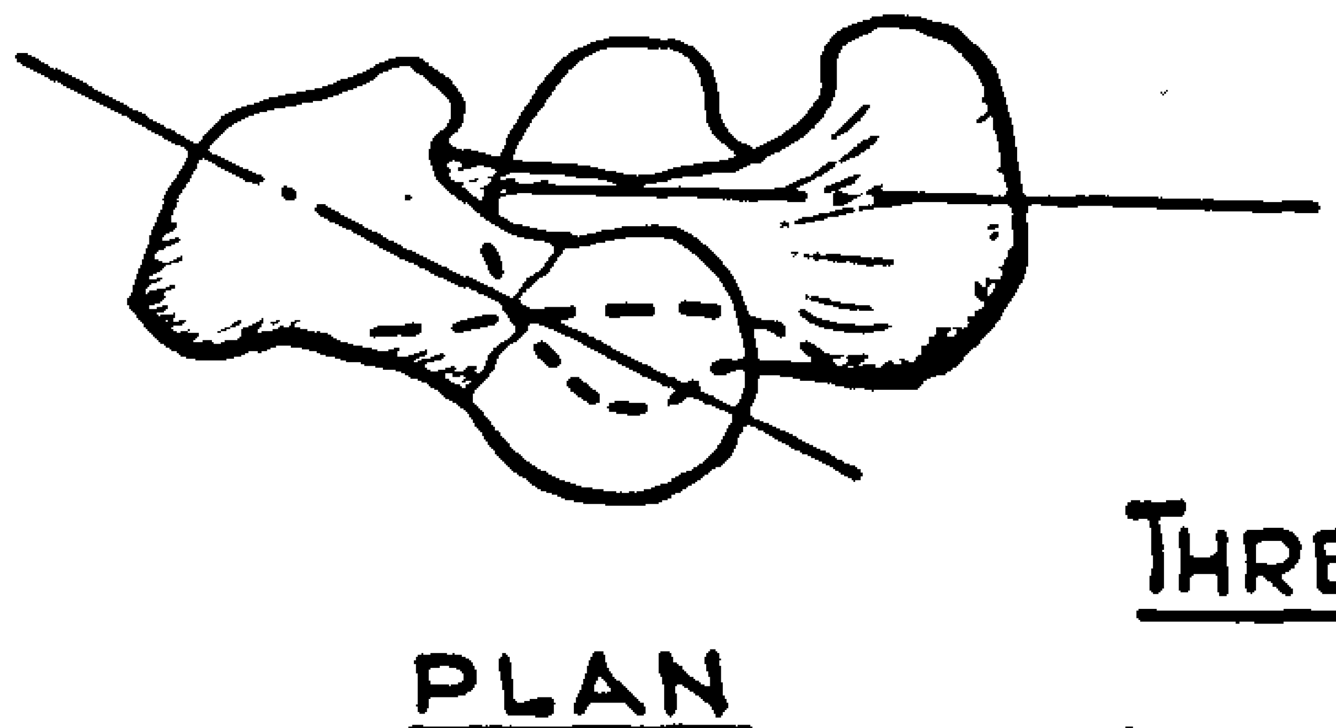
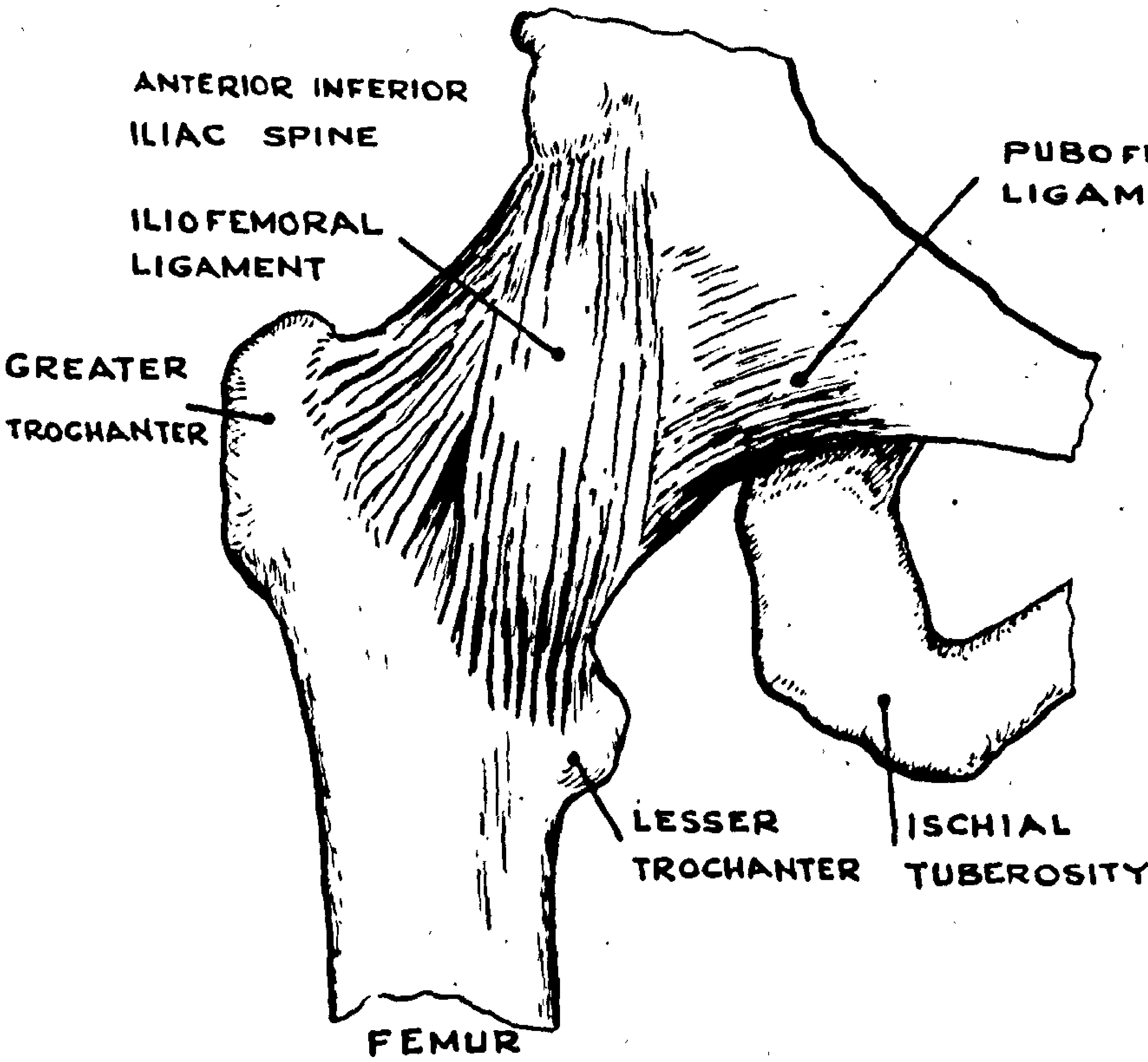


FIG. 2

THREE VIEWS OF THE RIGHT FEMUR

in the trochanteric region. The greater and lesser trochanters are bony prominences to which are attached the tendons of several muscles. The shaft of the femur is almost cylindrical over most of its length and is curved with a forward convexity. At its lower end the shaft enlarges into the femoral condyles, the underside of which are the bearing surfaces of the knee joint. In the erect posture the shafts of the femurs incline upwards and outwards. If the knee is arranged with the patella or knee cap directly forwards, the axis of the neck of the femur is inclined forwards, inwards and upwards as shown.

The hip joint is enclosed by the synovial membrane which is attached to the femoral head, the margin of the cartiliginous surface distally and the margin of the acetabular cartilage proximally. Outside the membrane is a strong dense fibrous capsule which forms along certain well defined lines shown and named in Fig. 3. The ilio-femoral ligament comprises two parts having tensile strengths of 200-500 lb. This ligament becomes taut when the thigh is extended. It also assists in limiting inward rotation when the thigh is extended and outward rotation when the thigh is flexed. The ischio-femoral ligament tightens in extension, in inward rotation with extension, and in abduction. The pubo-femoral ligament checks outward rotation when the hip is in



VIEW FROM THE FRONT

VIEW FROM BEHIND

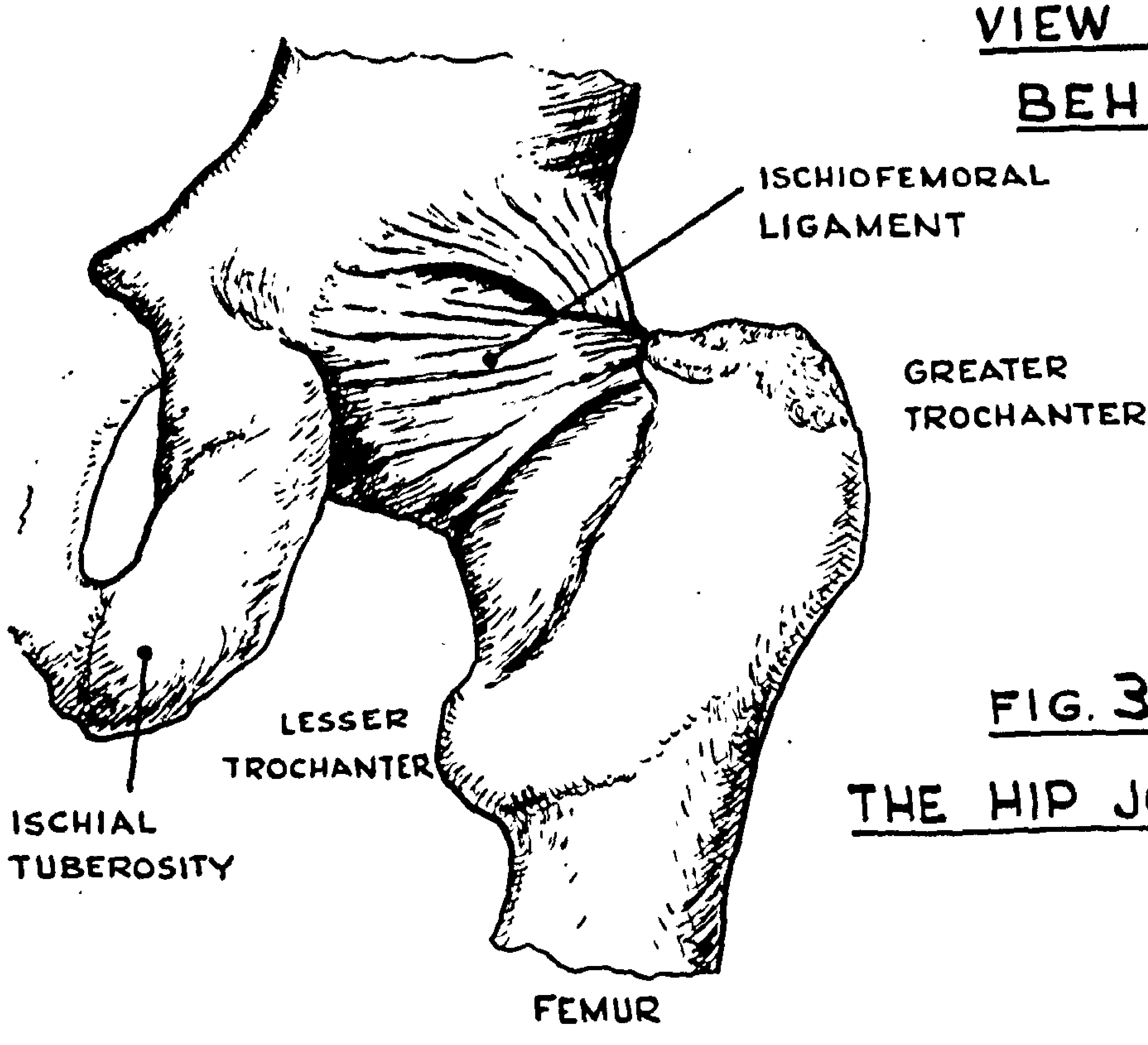


FIG. 3

THE HIP JOINT

extension, checks abduction, and when the thigh is flexed checks abduction and outward rotation. The ligament of the head (ligamentum teres) is stated to limit adduction when the thigh is partly flexed but generally to have little mechanical significance (Gray 1950) Steindler (1935). Generally the action of the ligaments is to retain the femur lightly in the capsule, to prevent extension of the femur much beyond the straight position and to limit abduction and rotation movements. The range of movement in the normal living subject is 120 degrees in flexion: 20 degrees in extension : 45 degrees each in abduction and adduction : and inward and outward rotation totalling 90°. For movements generally within this range the capsular ligaments exert no force, although in certain combined movements the range is somewhat reduced.

The muscles acting to produce movement at the hip joint number 22 in total. Six of these act principally to rotate the femur on the hip in an outward direction. Generally they are of small size and are deep seated. Their moment action and contribution to joint force will therefore be small by comparison with muscles of larger cross-section. The remaining muscles may conveniently be described according to the major rotational actions which they exert, i.e. flexion, extension, abduction or adduction. They may be further subclassified as two-joint or one joint depending on the number of skeletal joints between their origins and insertions or those of their tendons.

The flexors of the hip comprise Psoas (major), Iliacus, Rectus Femoris, Tensor Fascia Lata, Sartorius and Pectineus. The positions of the origins and insertions of these muslces are shown in Figs. 4 and 5.

Psoas (Major) is unipennate in form, has its origin on the lumbar and lowest thoracic vertebrae and their intervertebral cartilages,and ends in a tendon of insertion which passes over the front of the capsule of the hip joint before joining the femur at the lesser trochanter. Iliacus is a flat triangular muscle, which occupies and arises from the inner surface of the iliac bone. Most of its fibres converge and join the tendon of Psoas but some are inserted directly on the femur in front of and below the lesser trochanter.

Rectus femoris is a bipennate muscle, arising by a tendinous head from the anterior inferior iliac spine and by a tendinous head from a groove above the rim of the acetabulum. The two heads unite at an acute angle and spread downwards on the forward side of the muscle from which the muscular fibres arise. Rectus femoris is a two-joint muscle and the insertion is into a flat tendon joining the upper pole of the patella, and the patella

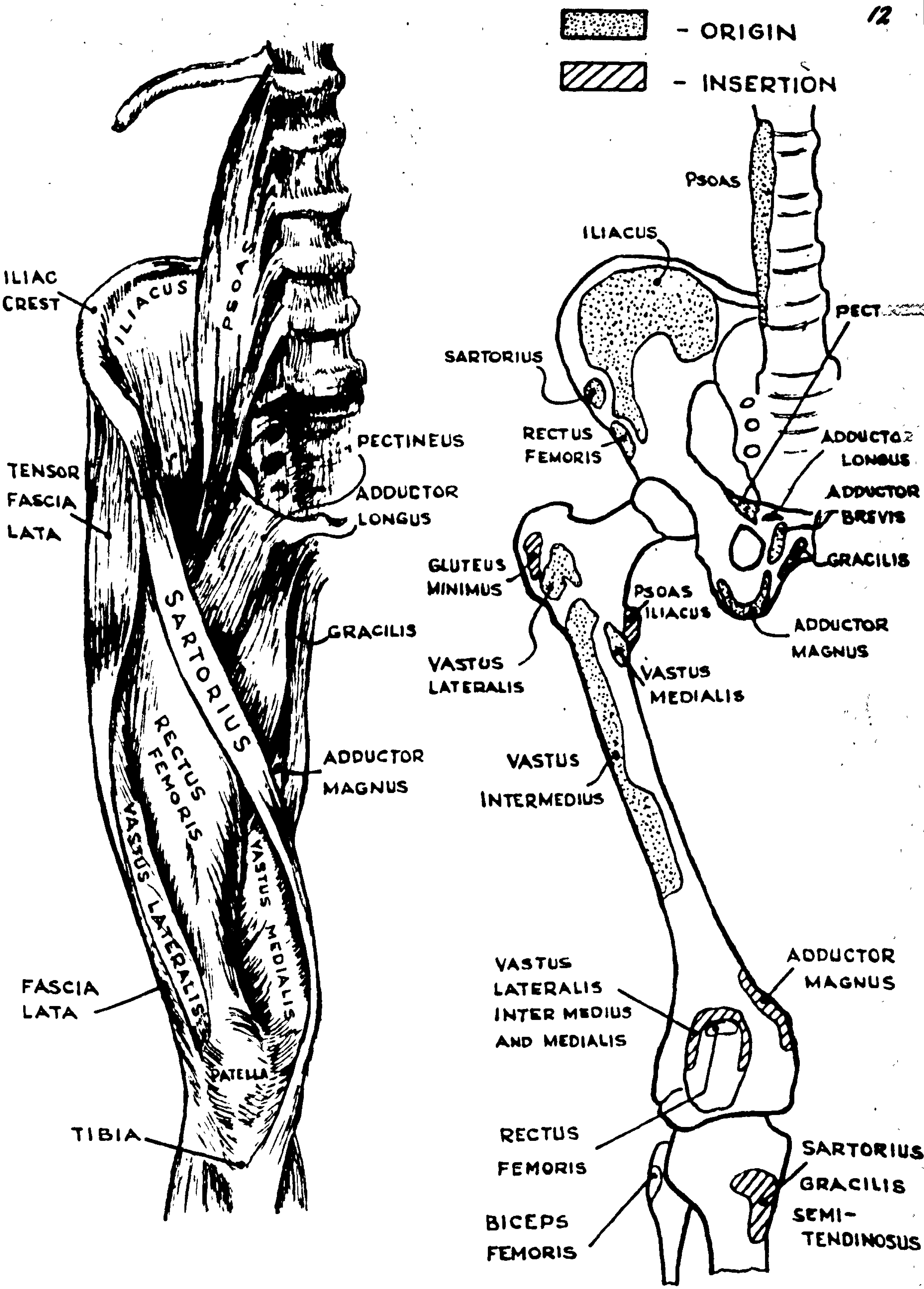


FIG. 4

HIP MUSCLES - FRONTAL VIEW

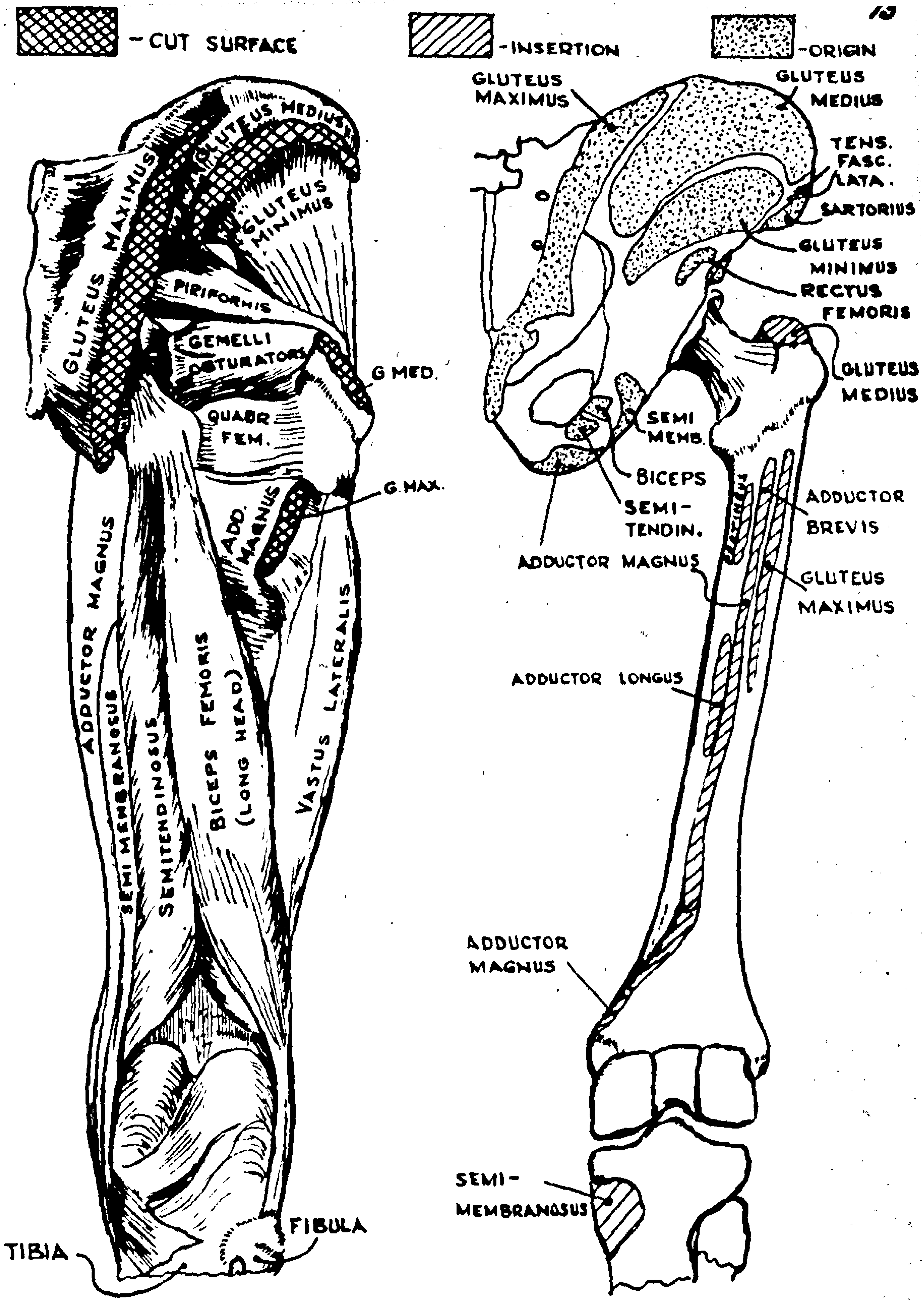


FIG. 5

HIP MUSCLES - POSTERIOR VIEW

is connected by the patellar tendon to the front of the tibia.

Tensor fascia lata arises from the outer lip of the iliac crest and the outer surface of the anterior superior iliac spine and is inserted into the fascia lata, an extensive sheath enclosing the muscles of the thigh. Over the lateral surface of the thigh the fascia lata is specially thickened and forms a strong band termed the iliotibial tract. At its lower limit the ilio-tibial tract is attached to the lateral condyle of the tibia. Tensor fascia lata acts therefore as a two-joint muscle.

Sartorius the longest muscle in the body is narrow and ribbon-like. It arises from the front surface of the anterior superior iliac spine and passes diagonally down the front of the thigh from outside to inside and becomes a thin flattened tendon running down the inside of the knee to insert on the inside surface of the tibia. Sartorius is thus also a two-joint muscle.

Pectineus is a muscle of quadrangular section having its origin in the upper ridge joining the pubis to its junction with the ilium and inserted on the femur between the lesser trochanter and the linea aspera.

Although the major action of these muscles is to flex the thigh on the hip, because of the positions of their lines of action relative to the head of the femur, they will have other rotational actions in addition, many of these depending on the relative angular position of the femur and the pelvis. Table 1 is of interest in showing the conflict amongst the authorities on the function of specific muscles.

The extensors of the hip comprise Gluteus Maximus, the long head of Biceps Femoris, Semitendinosus and Semimembranosus. These are illustrated in Fig. 4 and 5,Gluteus Maximus is a broad thick fleshy muscle forming the prominence of the buttock. Its origin extends along the hindmost quarter of the outer surface of the iliac crest, the posterior surface of the sacrum close to the ilium and the side of the coccyx. The insertion is along a rough line 4 inches long on the posterior aspect of the femur between the greater trochanter and the linea aspera.

Biceps femoris (long head), semimembranosus and semitendinosus are all two joint muscles having a common origin on the ischial tuberosity. The tendons of the three form the hamstrings. Biceps femoris has its insertion on the outside of the tibia and the fibula. The other two muscles have tendons of insertion leading to the inner surface of the medial tibial condyle.

The principal actions of Gluteus Medius and Gluteus

ACTION / MUSCLE	FLEXION	EXTENS^N	ABDUCT^N	ADDUCT^N	ROTATION INWARD	ROTATION OUTWARD
PSOAS.	BG RS				(G)	(S)
ILIACUS	BG RS				(G)	(S)
SARTORIUS	G (R) S		(B)(G)(R)(S)			(G)(R) S
RECTUS FEMORIS	BGRS		(R)			
PECTINEUS.	BGR(S)			(B)GRS		(R)(S)
TENSOR FASCIA LATA.	(B)(R)S		BG(R)S		BGR	
GLUTEUS MAXIMUS		BG RS	(R)	(R)(S)		R
BICEPS FEMORIS (LONG HEAD).		BGRS		(S)		(R)
SEMI TENDINOSUS		BGRS		(S)	(R)	
SEMI MEMBRANOSUS		BGRS.		(S)	(R)	
GLUTEUS MEDIUS	(R)(S)	(R)(S)	BGRS		B(G)(R)	(R)
GLUTEUS MINIMUS	(R)(S)	(R)(S)	BG(R)S		B(G) R	(R)
GRACILIS.	G(R)S			(B)(G)RS	G(R)	
ADDUCTOR LONGUS	(B)(G)(R)(S)	(G)		BGRS		BG(R)
ADDUCTOR BREVIS	(B)(G)(R)(S)	(G)		BGRS		BG(R)(S)
ADDUCTOR MAGNUS	(B)(G)(R)	B(G)(R)		BGRS	B (R)(S)	BG(R)
SIX ROTATORS.						BGR.

AUTHORITIES. B - BASMAJIAN (1960) G.-GRAY. (1958)
R - RASCH & BURKE (1963) S-STEINDLER. (1955)

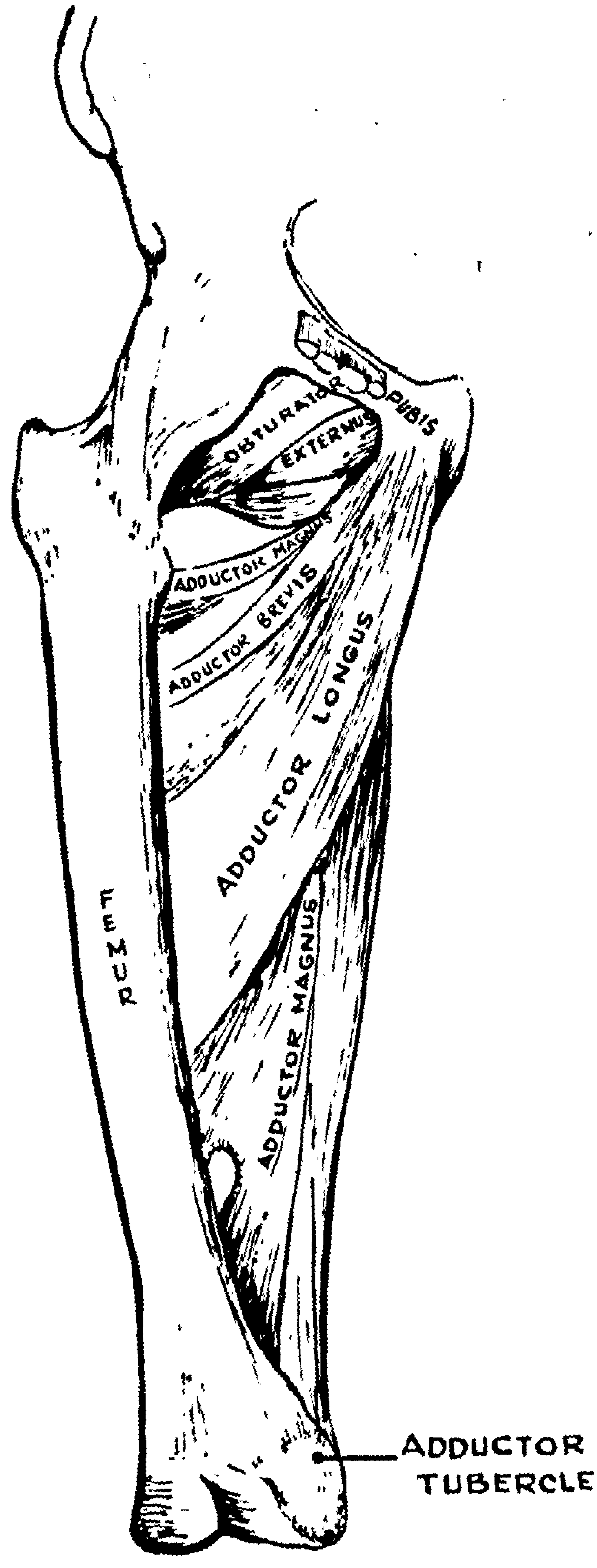
( ) DENOTES SMALL MAGNITUDE OR RESTRICT^D. RANGE OF ACTION.

TABLE. 1
ACTIONS OF THE HIP MUSCLES.

Minimus are abduction and they are the only muscles having this prime function. Their origins are on the outer surface of the ilium. Gluteus medius is the larger muscle and lies on top of gluteus minimus. Both muscles have their insertions on the greater trochanter, minimus on the front part of the top and medius on the lateral surface.

The adductors of the hip comprise Gracilis, Adductor Brevis, Adductor Longus and Adductor Magnus. Gracilis is a slender two joint muscle extending from the pubis to the medial side of the upper part of the shaft of the tibia below the condyle. The three adductors, brevis, magnus and longus are fan shaped muscles lying deep on the inside of the thigh as shown in Fig. 6. The origins are on the front of the pubis : just below the crest for longus, on the outer surface of the inferior ramus for brevis and along the line from the front of the pubis to the ischial tuberosity for magnus. The insertion of these muscles extends from the top of the linea aspera to the adductor tubercle on the femoral condyles. Adductor Magnus is one of the largest muscles in the body, yet it is surprising that reference books can ascribe little general function to the adductor group in normal activity.

Six small muscles, Piriformis, Obturator internus, Obturator



DEEP MUSCLES OF THE MEDIAL FEMORAL REGION

FIG. 6

externus, Quadratus femoris, Gemellus Superior and Gemellus Inferior have their origin on the posterior portions of the pelvis, their insertions on the greater trochanter of the femur, and act principally as outward rotators. Gray (1958) suggests that these muscles also act to assist the ligaments to retain the integrity of the hip joint on the occurrence of sudden strains when the great muscles are relaxed.

Historical Review

The first recorded studies of the mechanics of the human body stem to ancient Greece. Hippocrates (c.460 B.C.) Aristotle (c.350 B.C.) and Archimedes (c.250 B.C.) showed some insight into the application of the principle of moments to the movements of the body produced by muscular contraction. Galen (c.200 A.D.) showed that the Roman philosophers had some knowledge of nerve and muscle association and defined the terms agonist antagonist and tonus as referred to muscle action. The next recorded analysis of body mechanics is that of Leonardo da Vinci (c.1500 A.D.) whose knowledge of anatomy and mechanics enabled him to demonstrate the mechanics of standing, walking, jumping and rising from a sitting position. These studies required the concept of the centre of gravity of the human body. Galileo

(1638) , a student of medicine before the studies of dynamics for which he is renowned, indicated in his dialogues the application of classical mechanics of motion to the human body. Borelli (1675) furthered the work of Galileo and developed a theory of the chemistry of muscle contraction. His book 'De Motu Animalium' illustrates many applications of mechanics to the movement of limbs under muscle action and indicates a method for measuring the position of the centre of gravity in man. Baglivi (1700) continued the work of Borelli and differentiated between the structure of smooth and striated muscles, hypothesising on their function.

The principal event of the 18th century was the discovery of the phenomenon of the electrical stimulation of muscle as reported by Haller (c.1740) and Whytt (c.1740). In his "Commentary on the Effects of Electricity on Muscular Motion" Galvani (1791) made an explicit statement of the presence of electric potentials in nerve and muscle and this is probably the first published reference to this relationship. Duchenne (1867) reported extensive studies of the function of individual muscles as observed when the muscles were subjected to externally applied electrical stimulation. This procedure showed the response of an isolated muscle to stimulation although Duchenne realised that isolated muscle action does not occur in nature.

Weber and Weber (1836) published a treatise on the mechanics of human motion which reported anatomical studies on cadavers, measurement of the position of the centre of gravity of the human body and the mathematical analysis of certain aspects of gait. Their contention that the movement of the leg in the swing phase of walking was a pendulum action without the assistance of muscle action has been refuted by several of their successors.

Meyer (1853) determined all co-ordinates of the location of the centre of gravity in various standing postures, "normal", "military" and "comfortable" and proposed a theory of the self-locking action of the leg joints whereby standing would be maintained with a minimum of muscle action. These postures are not however clearly defined.

The first use of serial photography to investigate the characteristics of gait is that of Muybridge (1882). The gait of race horses was investigated by the use of a series of cameras viewing from the side. The successive firing of the cameras was obtained by physical contact of the horse with a series of trip wires. Further investigations were performed on human subjects walking, rising from sitting position, sitting from the erect position, dancing and performing various other movements. In some of the studies the background is marked into squares to allow estimation of the displacements.

Marey (1887) and Carlet (1872) reported investigations into the characteristics of human walking using a pneumatic system of recording displacements of parts of the body, foot pressures and periods of muscular activity. The subject walked in a circular path and variation of the experimental quantities was made to alter the air pressure within a series of closed systems. The indicating transducers were located at the centre of the walking circle and the connecting pipes were carried on a rotating radius arm which followed the test subject. Muscular activity was measured by the displacement of a cuff arranged over the appropriate muscle. Demeny (1887) also reports a balance for the determination of the centre of gravity of a subject in any configuration of his limbs. In this, the subject lay on his side on a platform supported on balances at three positions. The centre of mass is defined in terms of the balance readings and the geometry of the supports. The principal advance of these investigations was in their use of chronophotography to register the successive positions of the test subject in walking. The apparatus comprised a plate camera with a mechanical drive to the shutter allowing exposure at predetermined intervals. The subject carried markers which could be lights or reflective material and the successive positions of these markers were recorded on the film.

<u>Analysis of Gait and the Forces in Muscles</u>.

The first investigator to perform a comprehensive analysis of the three dimensional movements of the parts of the human body was Fischer (1898). Fischer performed three tests on one subject. The subject was dressed in black and carried incandescent tubes, eleven in number, strapped to segments of the body as shown in Fig. 7. The subject dressed in dark clothing walked in a darkened room in the field of view of four plate cameras whose shutters were closed and opened cyclically at a controlled speed. The exposed images of the tubes on the plate gave "stick" diagrams representing the successive positions of the body parts. The four cameras were situated near the subject: one on each side trained perpendicular to the axis of progression and two viewing the subject in oblique frontal aspect along lines at 30° to the axis of progression as shown in Fig. 8. The corrections to the readings required to eliminate errors due to perspective were therefore complicated. The information

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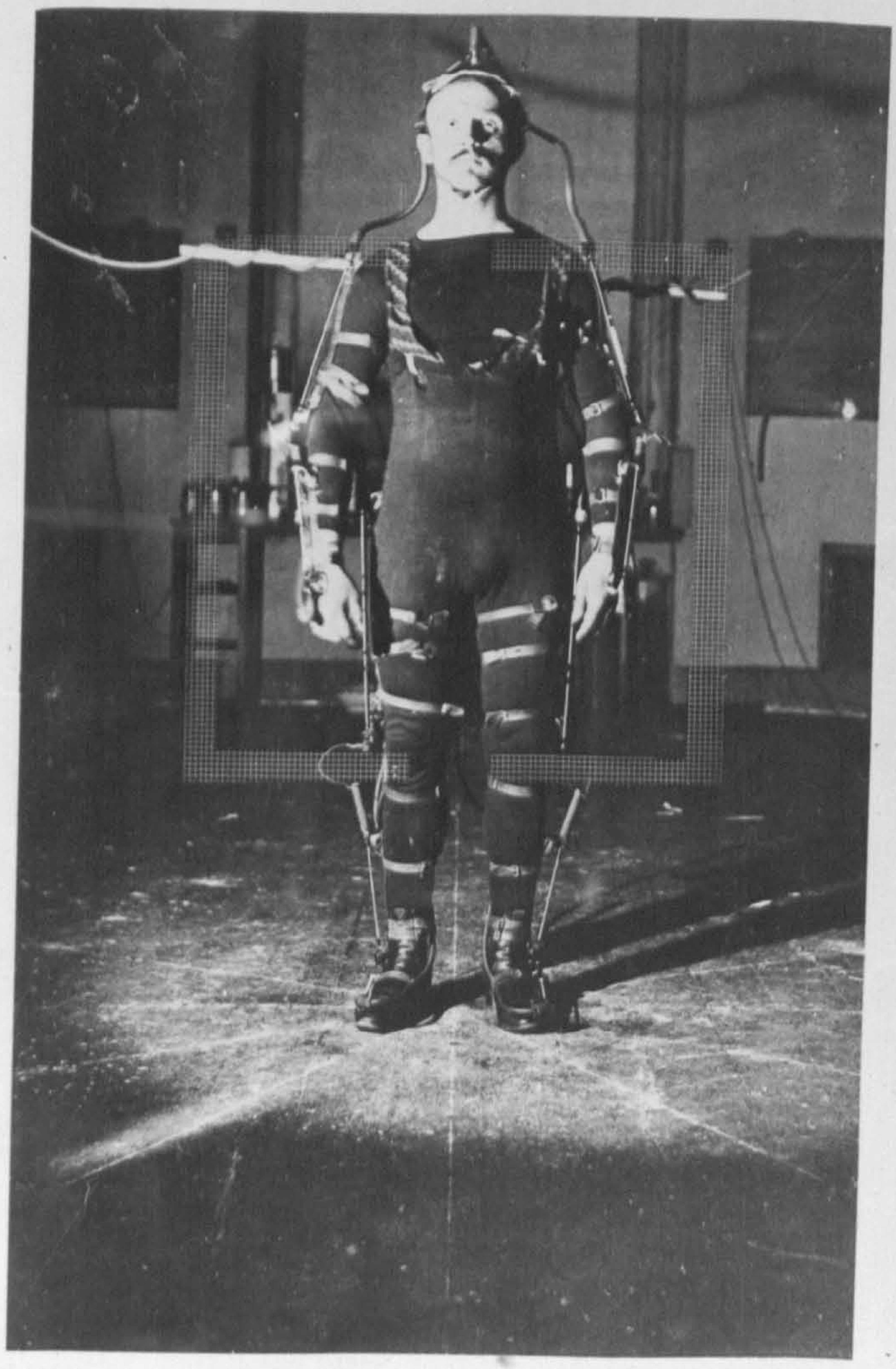
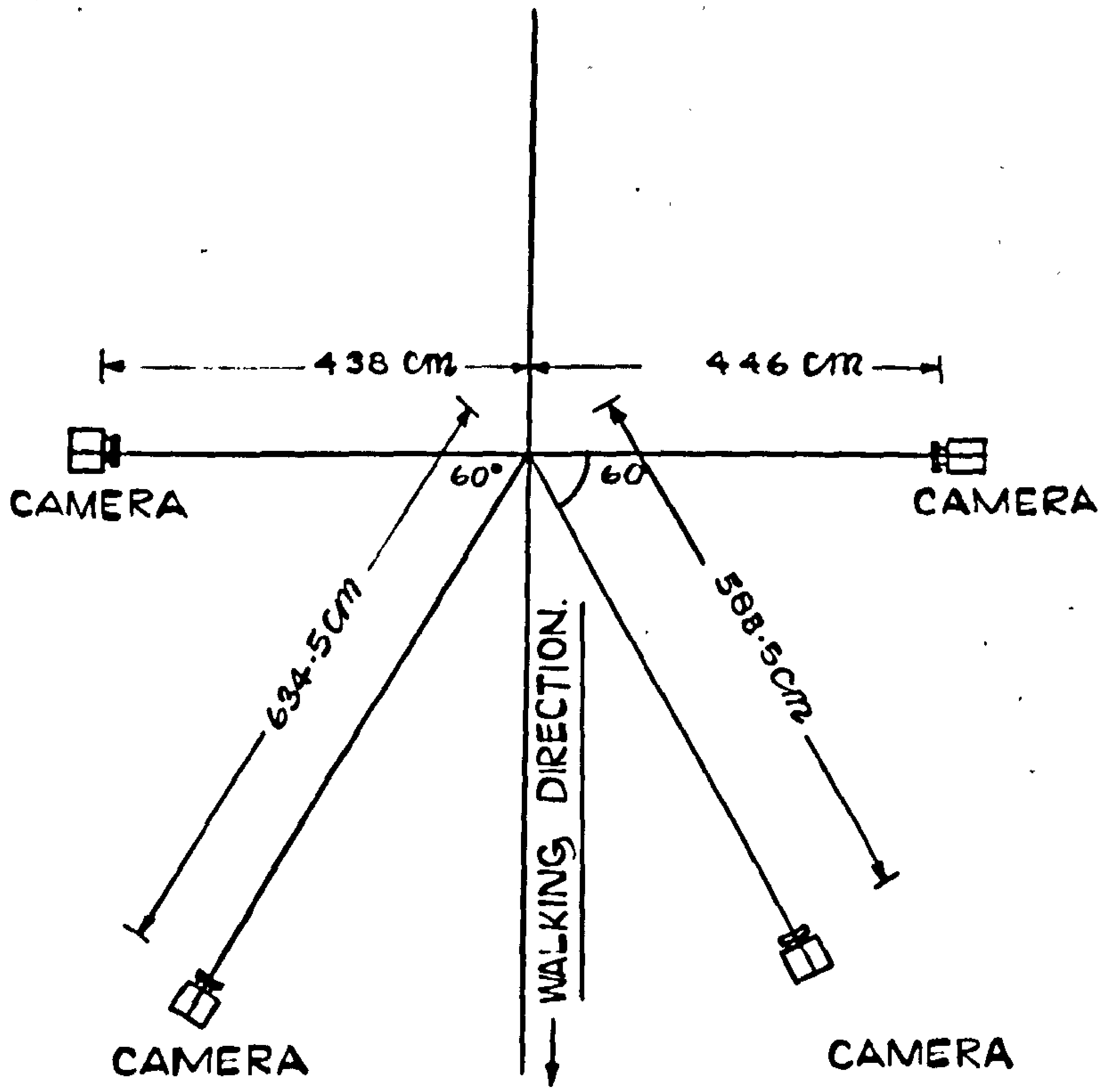


Das Versuchsindividuum in voller Ausrüstung.

Test Subject from Fischer (1898 - 1904).

Fig. 7.



PLAN OF EXPERIMENTAL LAYOUT
BRAUNE & FISCHER (1895)

FIG. 8

was analysed to obtain curves of the variation with time of the displacement, velocity and acceleration in three reference directions of landmarks and centres of gravity of the body segments. In each test one double step was analysed and the cycle divided into 31 phases. From the values of acceleration and displacement for each segment of the body Fischer calculated the resultant forces, combined these vectorially and presented tables quoting the magnitude of the ground to foot force at each phase. Fischer's data on the kinematics of gait has been referred to by all subsequent investigators and his findings largely conditioned the mechanical design of external prostheses for the first half of the present century. In addition to the analysis of the external force actions Fischer published a further volume, Fischer (1908), in which the internal mechanics of the body in respect of ligament and muscle forces are presented. Tables and graphs are shown describing the variation of the moment action of certain muscles as the angulation of the body segments changes.

Fick (1904) contends however that Fischer's results in the walking tests are inaccurate since the centres of gravity of Fischer's subjects were determined from measurements taken on cadavers which were in the recumbent position.

The degree of spinal curvature for an erect subject requires that the centre of gravity be considerably further forward than Fischer assumes. Fick's publications describe also the complete anatomical range of movement of the joints of the human body, together with the corresponding lengths and moment arms of appropriate muscles. Fick's tables and diagrams are based on a knowledge of the positions of the muscular attachments to the body structure and the values quoted are calculated trigonometrically. Strasser (1908 - 1917) uses a joint model "Globusmuskelphantom" to obtain corresponding information on the mechanics of muscles and joints.

The analysis of the force systems present during movement was significantly advanced by Amar (1916, - 1917) with the invention of a "force plate" or instrument to measure ground to foot forces. The measuring system was mechanical, spring deflection under load operating an indicator. The instrument measured vertical, sideways and backward force components only and gave no measure of forward force components or of moment actions about any reference axis. Amar's force plate was used in investigations of the gait of subjects wearing external prostheses and the findings were used to suggest improvements.

Fenn is principally known for his studies on electromyography which are reviewed in another chapter but his paper Fenn (1930) presents an analysis of the variation of "kinetic energy of the parts of the body relative to the trunk" during activity. The relative kinetic energy R.K.E. is defined as

$$R.K.E. = \sum \frac{W_1}{2g} (V_1 - V_o)^2 + \sum \frac{W_1}{2g} k_1^2 \omega_1^2 \quad \text{----} (1)$$

where W_1 is the mass of a part of the body, k_1 its radius of gyration, V_1 the linear velocity of the C.G. and ω_1 the angular velocity. V_o is the velocity of the whole body. The statement is made that the total kinetic energy K.E. is given by

$$K.E. = \frac{W_o}{2g} V_o^2 + R.K.E. \quad \text{-------} (2)$$

where W_o is the weight of the whole body. The total kinetic energy is in fact given by

$$\text{True } K.E. = \sum \frac{W_1}{2g} V_1^2 + \sum \frac{W_1}{2g} k_1^2 \omega_1^2 \quad \text{------} (3)$$

(2 and 3) may be written as

$$\text{Fenn's total } K.E. = 2 \times \frac{W_o}{2g} V_o^2 - 2 \sum \frac{W_1}{2g} V_1 V_0 + \text{True } K.E. \quad (4)$$

It appears therefore that the concept of relative kinetic energy is incorrect and unacceptable.

Work performed at the Institute of Experimental Medicine in Moscow over the period 1930 - 1935 is reported by Bernstein (1935). The work reported is an extension and improvement of the original work of Fischer. The experimental technique is improved by increasing the frequency of the exposure of the film giving more closely defined "stick" diagrams for the limb configuration. Tests are performed on more subjects than Fischer and more data is analysed. Table 2 is reproduced from this volume. Data is also presented on the effects of weights carried on the shoulders on the gait characteristics and also on variations with the onset of fatigue. A later volume, Bernstein (1940), reports the development of gait characteristics in children starting from the first unsupported step. This volume also includes the analysis of the gait in athletic performance in sprinting, running and the long-jump. Chkhaidze (1957, 1958) continues the investigation, and reports the changes in gait on the level, up and down inclines, for trained and untrained subjects performing at high altitudes. These works are grouped for discussion since the experimental techniques, method of analysis and presentation of results are the same.

Single plate exposures of successive views of the subjects were taken from two sides on cameras whose shutters operated at a frequency of 80 per second. Markers were placed at 20 positions on the body from the head to the feet. There was no instrumentation to record ground to foot forces. The only movements recorded are those occurring in planes parallel to the plane of progression. Bernstein states that sideways

RESEARCHER.	BRAVNE & FISHER.	BERNSTEIN.
FILM PLATES.	12 (3 x 4)	> 800
Nº. OF TESTS.	1	> 65
ANALYSIS OF EXPERIMENTS.	3	
SINGLE PACES EXAMINED.	6	7,500
MEASUREMENTS	5,000	400,000
PHASES OF MOVEMENT.	93	C. 24,000.

TABLE. 2

COMPARISON OF RANGE OF TESTS OF BRAVNE & FISHER & BERNSTEIN.

movements are of no significance to studies of gait.

It is submitted that, in fact, sideways movements are of considerable importance in the gait of the human, since deviations from normal due to pain, malfunction or the use of a prosthesis are most readily detected in the view from the front. Furthermore a large amount of the energy used in walking is required in the musculature which controls the sideways movement. Velocities and accelerations were determined by numerical differentiation of the displacement values as follows.

If S_n is the typical position co-ordinate at exposure n

Velocity at this exposure $= (S_{n+2} - S_{n-2}) \times \frac{f}{4}$

where f is the frequency of exposures.

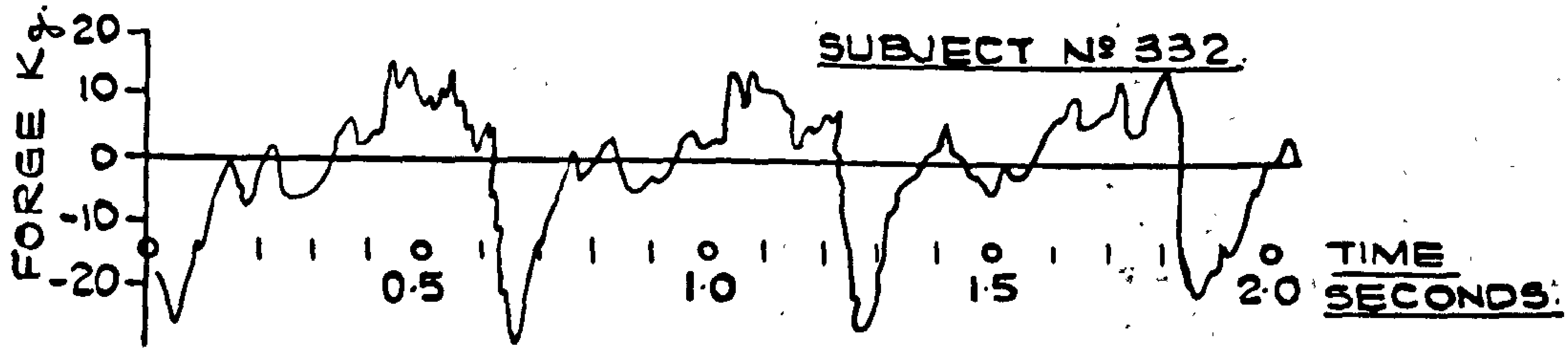
Similarly acceleration $= (S_{n+4} - 2\,S_n + S_{n-4})\ f^2/16$

Bernstein is insistent that the curves of displacement should not be smoothed prior to differentiation since essential information about the gait characteristics is lost thereby.

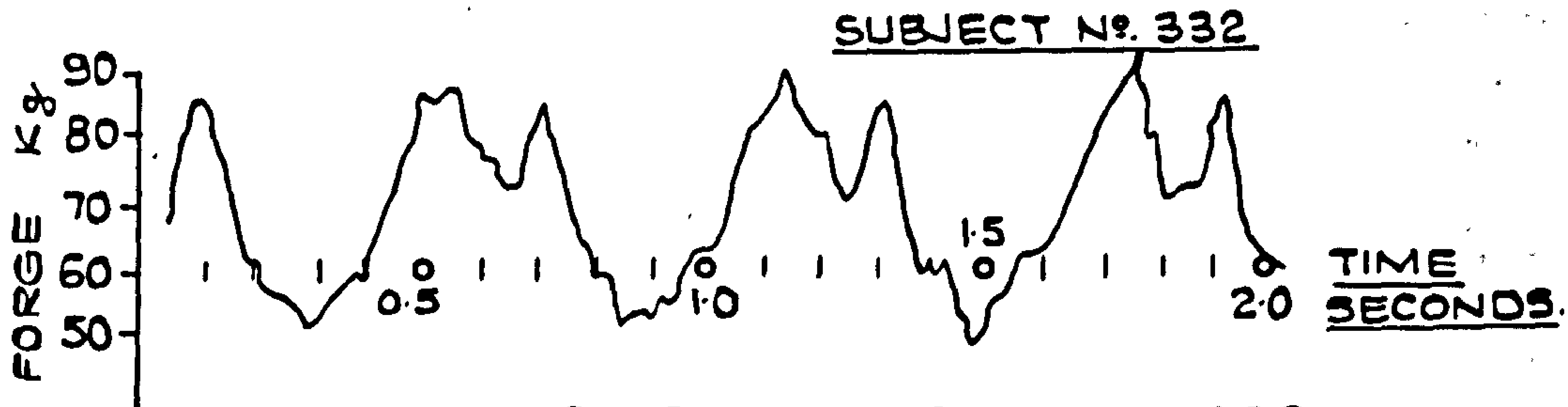
This contention requires to be treated with reserve, however, whenever the sensitivity of measurement is considered. Bernstein quotes no figures for the probable measurement error of his photographic records. It appears that the displacements of the skin markers were obtained to the nearest millimetre. With a shutter speed of 80 per second and a mean walking speed of 5 mile/hour the change in horizontal position of a point on the body between frames would be 28 mm. If the measurements are taken to an accuracy of $\pm$ 0.5 mm the possible fractional error in velocity is $\pm$ 1/56 and the corresponding error in acceleration might be $\pm$ 1/28. Where the velocity of a part of the body is considerably less than the mean, the fractional errors increase and the errors in acceleration to and from rest may be quite high. The amount of the errors could be reduced either by smoothing the curves to eliminate the effects of measurement errors or by taking the velocities and accelerations from curves fitted to the experimental points using a least square of deviation technique.

Fischer's coefficients for the mass of body parts as a proportion of total mass are used to obtain the inertia forces at the centres of gravity of the parts of the body. These forces correspond to linear accelerations only, angular accelerations being ignored. The inertia forces are combined to give resultant force at C.G. of leg, at C.G. of left and right sides of the body and at the C.G. of the whole body. In these manipulations no account is taken of the moments about vertical axes or about horizontal axes parallel to the line of progression due to the offset of the vertical planes in which the C.G.'s of the parts lie.

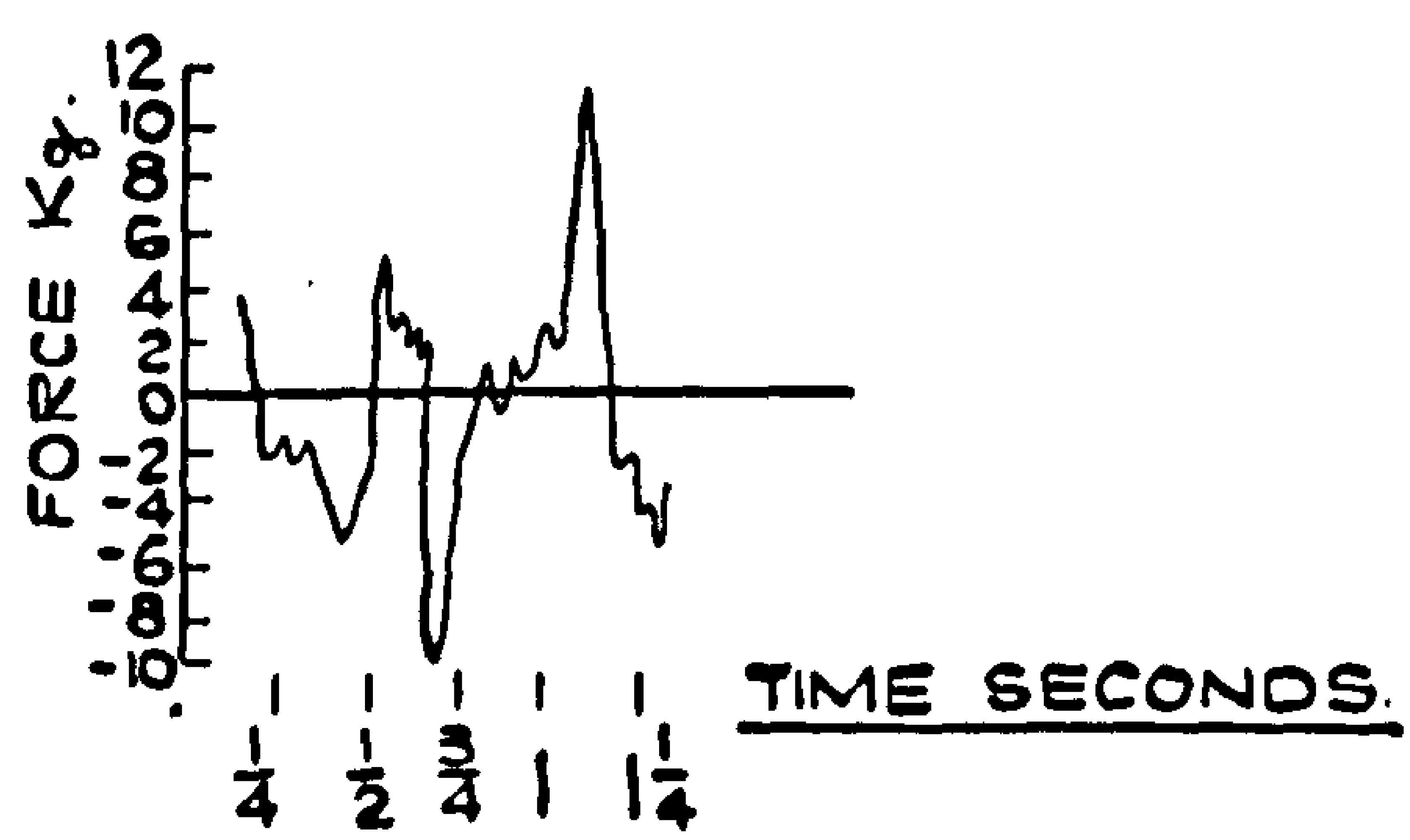
Typical curves of variation of inertia force at C.G. of whole body as quoted by Bernstein and at the C.G. of the thigh as presented by Chkhaidze are shown in Fig. 9. The characteristic acceleration peaks are stated to be physiological in origin corresponding to the pattern of nervous response in the higher and lower control centres. These peaks are taken to be characteristic of the individual and are shown to be reproducible in subsequent tests and to be affected by physiological changes such as fatigue and change of altitude. It must also be considered that these characteristic peaks may equally correspond to movements of the skin of the subject relative to the skeleton. This could give characteristic patterns for each subject and vary with physiological conditions. Curves of the variation in moment about certain joints are presented but these are the moments about the horizontal axes of the joints perpendicular to the line of progression and take account only of linear acceleration forces. They are of no significance therefore to leg joints during the phase of double support since ground forces will then be of overiding importance: and even in the swing phase their value will be reduced due to the omission of angular acceleration terms.



COMPONENT OF INERTIA FORCE IN DIRECTION OF MOVEMENT FOR THE WHOLE BODY (BERNSTEIN)



VERTICAL COMPONENT FORCE AT C.G. OF THE WHOLE BODY (BERNSTEIN.)



COMPONENT OF INERTIA FORCE IN DIRECTION OF MOVEMENT FOR THE WHOLE BODY (CHKAIDZE.)

FIG. 9

Pauwels (1935) performed an analysis to determine the magnitude and direction of the resultant force transmitted between the femoral head and the acetabulum during stationary weight bearing on one foot, and concluded that the value of this force was 2.92 times body weight. When the dynamic component of the vertical force actions was taken into account this value increased to 4.5 times and during walking it was suggested that values of 6 to 8 times were not implausible.

Elftman (1938, 1939) performed gait analysis in walking and running using a force plate in which components of force are measured by the deflection of elastic systems and are recorded by cine camera. No information is quoted on the amount of force plate deflection under load, its natural frequencies of vibration or on its hysteresis. Since the assembly comprised two plates, one sliding on ball-bearings on top of the other against spring resistance, it is surmised that its dynamic behaviour might not be ideal. There was no provision for the measurement of the moment exerted between ground and foot about a vertical axis through the contact area.

In the analysis of walking Elftman (1939a) reports the results of tests performed using the forceplate and measuring displacements by a cine camera viewing from the side only.

Only movements parallel to the plane of progression are considered but all gravity and inertial effects producing force actions in this plane are considered in the calculations for the resultant moments transmitted at the ankle, knee and hip joints. It is shown that angular acceleration has a very small effect on the values obtained. No attempt is made to calculate the actual values of muscle force required to transmit these moments. The analysis compares the rates of change of kinetic energy of parts of the leg with the rate of transmission of energy at the joints. This transmission rate was obtained from the product of the moment transmitted at the joint and the relative angular velocity of the parts meeting at the joint. In this way Elftman indicates that to some extent the muscles have to function as energy transmitting connections as well as energy developing and absorbing members. The following paper Elftmann (1939b) discusses the role of muscles which traverse more than one joint.

It is shown that in normal activity loading occurs at adjacent joints which can be transmitted by the energisation of a single two joint muscle. If the joint movements during this phase are such as to maintain a constant contracted length for the muscle, then the muscle need only be loaded in isometric tension, in which circumstance its efficiency of energy transformation is greatest. If the same duty were being performed by a one-joint muscle at each joint, one would be developing mechanical work while the other was absorbing it - obviously a less efficient procedure, and, in addition, the muscles would not be working isometrically so that their efficiency would be lower on this account also.

It is concluded that the human anatomical structure of the leg would be more economical of energy if a three-joint muscle system were present.

Corresponding analysis for the estimated rate of energy expenditure of the leg muscles in running is reported in a further paper Elftman (1940).

The National Research Council Committee on Artificial Limbs acting in an advisory capacity to the Veterans Administration and Office of the Surgeon General, United States Army, financed research by the College of Engineering and the Medical School of the University of California from 1945 onwards. Their report (University of California (1947))covers a large range of work and is considered in detail.

Previous analysis of gait had been deficient in description of the components of the rotation of body segments about the long axes of the segments. Tests were performed on 29 subjects in which pins were driven into the following bony prominences.

1. Iliac crest of the pelvis.
2. Adductor tubercle of the femur.
3. Tibial tubercle.

Wooden rods with spheres attached extended sideways from the pins far enough to be clear of the outermost projection of the shoulder. The subjects walked along a straight path viewed by three cine cameras situated above, in front and to the right of the subject. The cameras ran at 48 frames/sec and a clock mechanism, viewed simultaneously by the three cameras allowed the synchronisation of the records. The records were analysed at first taking account of perspective and parallax errors. It was found subsequently that during the middle sixty per cent of the stance phase the values measured from the orthogonal projection were correct within 2° although the value at toe-off could be in error by 6°. The projected angles were therefore used subsequently since the amount of this error was less than the variations between individuals. The averages of the ranges of angles measured for the pelvis, femur and tibia of 12 subjects were respectively 8, 15 and 19 degrees. The worst errors would be for the tibia which has the greatest range of angular movement in the sagittal plane but even so it would appear worth-while to apply a correction where an error of 6 parts in 19 is possible. The average rotation of the femur relative to the pelvis for ten subjects was 8.24 degrees. It should be pointed out also that the pelvis marker passed through the field of swing of the arm and the subjects required to fold one arm across the chest. Since rotations of the trunk about the vertical axis depend critically on arm movements, it would appear that all these values should be treated with some reserve.

Interrupted light and cine camera techniques were used in studies of the movements of normal and prosthetic legs in planes parallel to the plane of progression. A single camera was placed 16 ft from the plane of progression and a line of dowels at 1 ft centres was laid out 13 ft from the camera. The field of view was approximately 18 feet. The plate size used in the camera was 5 x 7 inch and the cine film was 35 mm. The interrupted light frequency was 30 frames per second and the cine-camera speed was not stated. Measurements from the negatives were taken using a measuring micrometer reading to one ten-thousandth of an inch. Graphs of velocity and acceleration were obtained by numerical differentiation of the displacement data as follows:-

$$\dot{x}_{1.5} = (x_2 - x_1)/t$$
$$\ddot{x}_1 = (\dot{x}_{1.5} - \dot{x}_{0.5})/t \qquad \text{--------} (5)$$

where t is the uniform time interval.

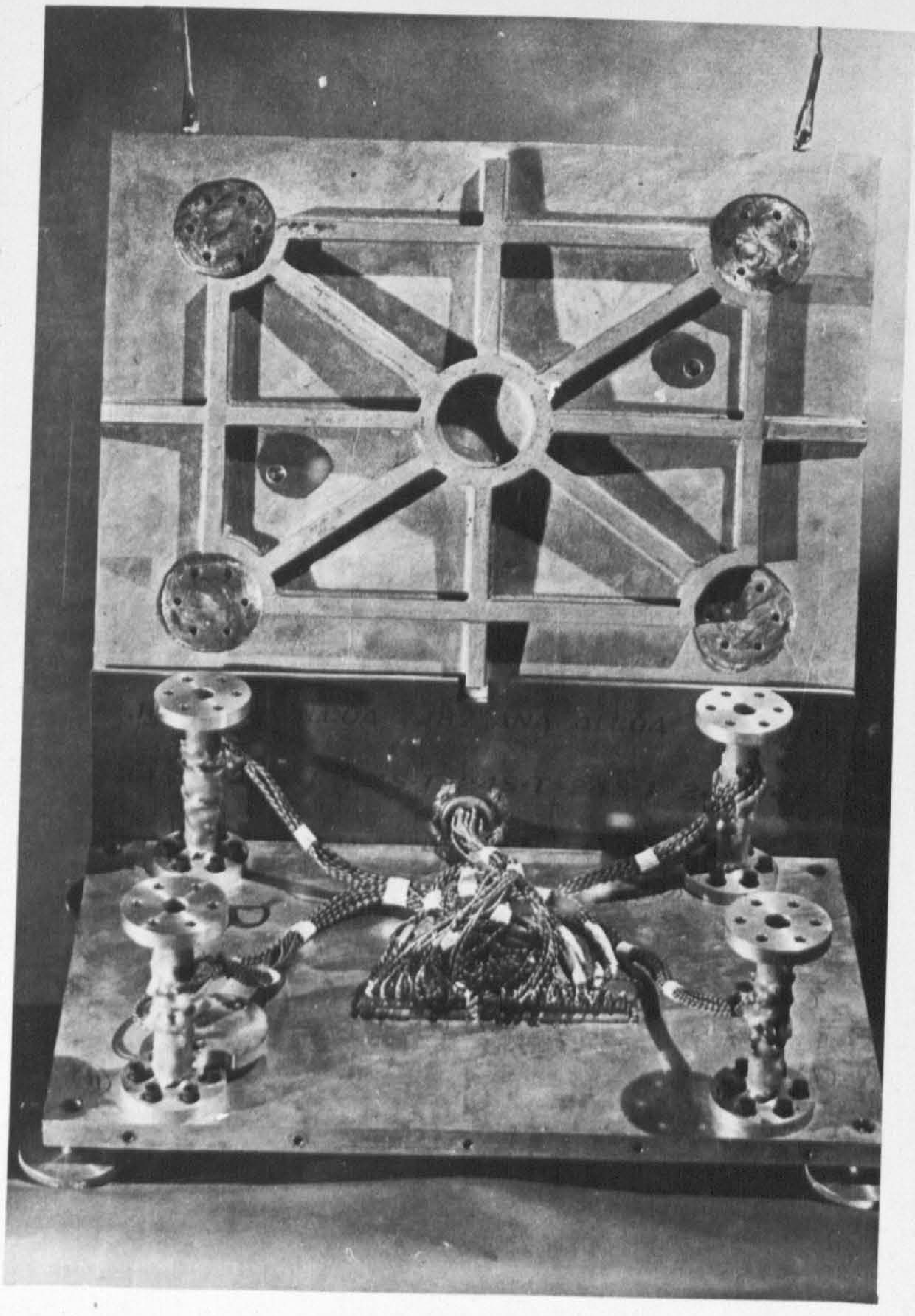
The angular displacement of the upper and lower leg centres of gravity were analysed harmonically up to the fifth harmonic, i.e. the constants A_{0-5} and k_{1-5} were found in the following expression

$$\emptyset = A_0 + A_1 \cos(\frac{2\pi t}{T} + k_1) + A_2 \cos(\frac{4\pi t}{T} + k_2) \text{ -- } A_5 \cos(\frac{10\pi t}{T} + k_5) \qquad \text{-------} (6)$$

where $\emptyset$ is the inclination of the axis of the limb segment to the vertical at time t and T is the time for one complete cycle. The variation in the moment about the knee axis during the swing phase was also obtained for one subject using Fischer's and Weinbach's coefficients for mass distribution. In the analysis of displacements no mention is made of the correction for parallax in the measuring system which for the dimensions quoted corresponds to errors exceeding 10% in displacement, velocity and acceleration, when the subject is at the end of the field of vision.

Glass walkway studies were performed in which the subject is viewed in two directions by one camera and in one direction by a third. The camera at the side records both the lateral view and the vertical view from below by means of an inclined mirror. The subjects wore a pelvic girdle with a long anterior extension to allow recording of pelvic movements and an ankle brace on which were mounted pins which penetrated the malleoli allowing recording of tibial rotation. Foot rotation was obtained from a white line on the sole of the shoe. Analysis was performed as previously described but because of the labour involved these records were generally used for visual rather than analytical comparison.

One of the principal advances of the California Group was the development of a force plate to measure the six quantities necessary to define the resultant force actions transmitted between ground and foot during walking. This device is described in their 1947 report and also in more detail by Cunningham and Brown (1952). Their force plate is illustrated in Fig. 10 and comprises a top plate held to a base plate by four tubular columns. Electrical resistance strain gauges mounted on the column and connected in bridge circuits gave output signals to six pen recorders. The measured quantities were the components of the resultant force in three reference directions at the centre of the top plate and the component moments



Exploded View of Force Plate of

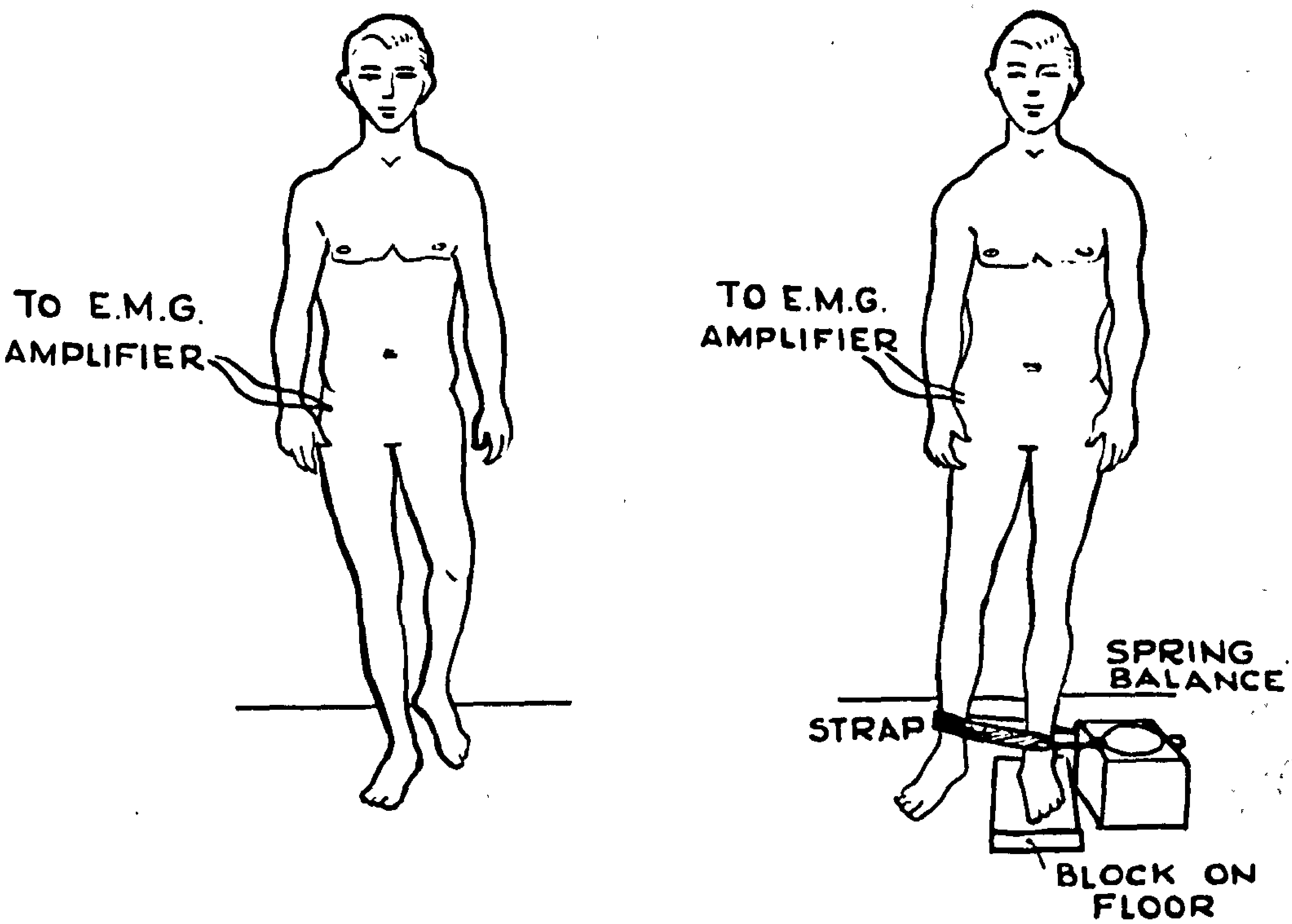
Cunningham and Brown (1952)

Fig. 10.

about these axes. Difficulty was experienced in interpreting the records of three quantities due to vibration of the top plate in the horizontal plane corresponding to the bending flexibility of the columns. The amplitude of this vibration was reduced by the use of viscous damping units mounted between the columns. The signals from the three bridge networks which depended on the resultant axial force in the columns required amplification before transmission to the pen recorders. Consistent linear static calibration curves are presented for each channel but no dynamic calibration is reported of the connected system comprising the force plate, strain gauge, amplifier and recorder. Cunningham (1950) reported additional information on gait obtained using this force plate. Tests were performed on ten normal subjects, seven below-knee amputees and eleven above-knee amputees walking on a straight level path: and on four normal subjects and three below-knee amputees proceeding in both directions on stairs and an inclined ramp. The results are reported as curves of components of resultant force and component moment about a vertical axis through the ankle. Graphs of variation of centre of pressure between ground and foot are also shown. No limb configuration records were taken and the results cannot therefore be used for estimation of joint forces and moments.

Using the University of California force plate Bresler and Frankel (1945) performed a full three-dimensional analysis of the periodic variation of resultant force and moment transmitted at ankle, knee and hip during walking on a straight level path for four young male subjects. Cine camera records were taken along the axis of progression and on a horizontal axis perpendicular to it. Marks on the skin surface were used to locate joint centre positions. No measurements were taken of angular rotations of the segments about their long axis and the couples corresponding to angular acceleration about these axes are shown to be small by comparison with other force actions. The paper makes no mention of the distances from camera to subject or of corrections for the errors due to parallax and perspective in taking measurements from the images projected from the negatives. Accelerations were obtained by graphical differentiation of the curves of displacement and Fischer's coefficients were used to calculate the force actions due to gravity and acceleration. The calculations were performed by desk calculators and the calculations for the first subject occupied 500 man hours, although this was improved so that the calculations for the fourth and final subject occupied half of this time.

Although Inman, who was active in the California Group, published a paper, Inman (1947), presenting a force analysis for the muscles of the hip, there appears to have been no attempt to relate the dynamic measurements of Bresler and Frankel to an analysis of the type presented by Inman. This analysis assumed the forces of Gluteus Medius, Gluteus Minimus, Tensor Fascia Lata, the Ilio-tibial tract and the weight of the body to act in the same vertical plane containing the centres of the femoral heads. The lines of action of these forces were determined from X-ray measurements on cadavers having metal rods inserted along the length of the muscles and fascia. Since the three muscles concerned are synergistic, the forces they exert were assumed to be in proportion to their weights as determined from muscles removed from cadavers and dried. Tests were performed on 35 subjects using an integrated electromyography technique to ascertain the contribution to abduction action of the above muscles in standing on one leg. The subject stood on the left leg only and attempted to abduct against a known horizontal force at the ankle as shown in Fig. 11. Integrated EMG values were read corresponding to different force values. Since the applied force exerted a moment about the hip this



TEST OF SUBJECT STANDING ON ONE LEG

CALIBRATION OF E.M.G. SIGNALS

FIG. 11

INMAN'S TEST PROCEDURE

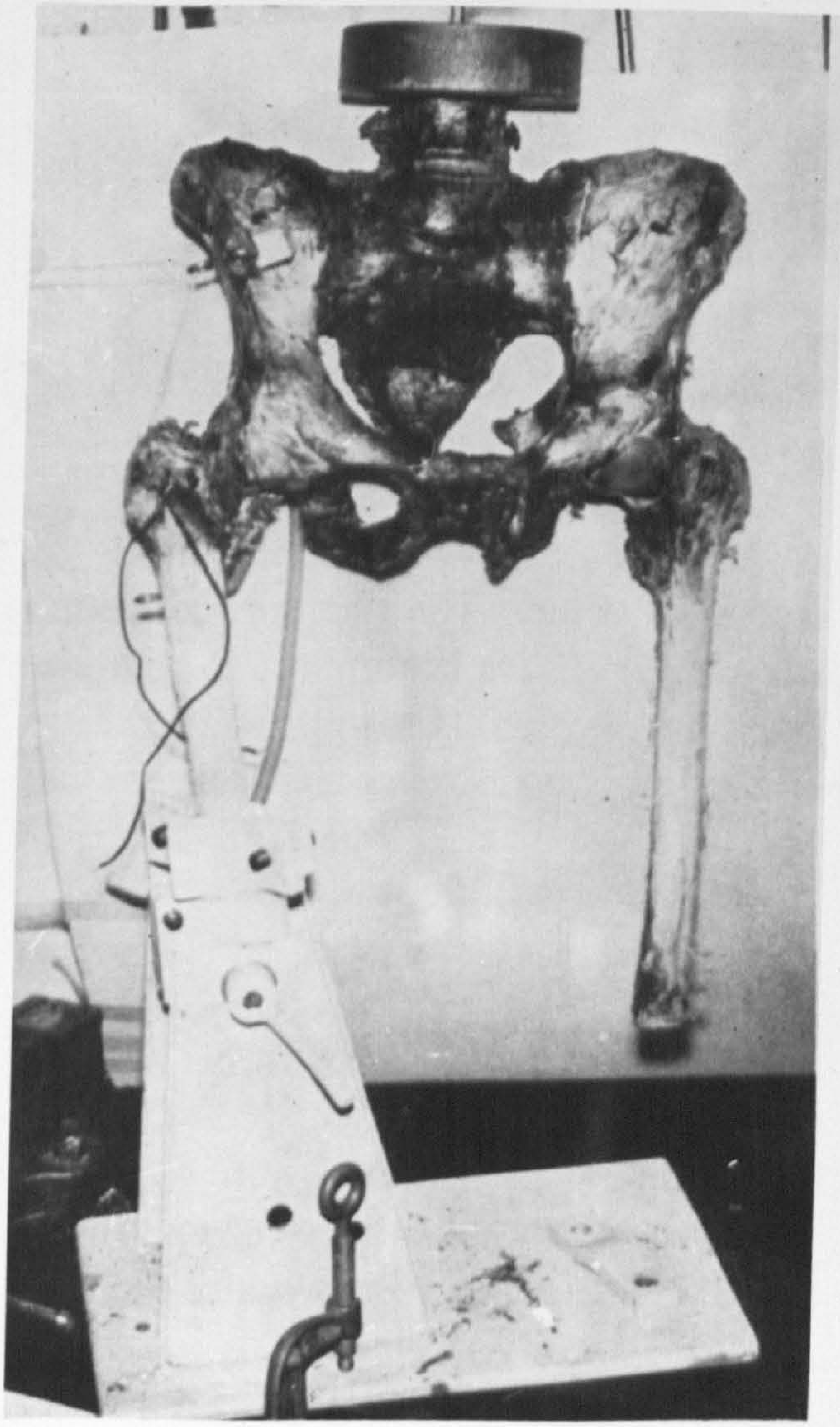
procedure was taken to give a calibration of the EMG record against hip abduction moment. Readings of the integrated EMG values were then taken when the subject stood on the right leg. Tests were performed at three angles of pelvic tilt. With the pelvis raised on the unsupported side the EMG values amounted to 91% of the calculated value. With the pelvis level the EMG values totalled 51% of the calculated and with the pelvis sagging 15 - 20 degrees to the unsupported side the percentage fell to 7. Inman concluded that a considerable amount of the necessary abduction moment was contributed by "passive tension" in the ilio-tibial tract. On this basis the resultant joint force was calculated to have a value of 2.4 to 2.6 times body weight.

A number of questions arise from Inman's analysis, however. If the ilio-tibial tract carries 49% of the hip moment when the pelvis is level, how is this force relaxed when normal standing is resumed or the subject lies down? A tension of this magnitude would require adductor activity to balance it in these situations and this is generally not present. Certainly, to move from relaxed standing on two feet to single support on the right leg requires horizontal movement of the pelvis to bring the body centre of gravity over the right foot and this corresponds to adduction of the right femur: i.e. when the pelvis is level during standing on the right foot alone, the right leg is adducted relative to its position in normal standing but the angular displacement is only about 6 degrees. The assumption that muscle force is proportional to mass, is also of doubtful validity. Generally the maximum force which a muscle can develop will be proportional to its effective cross-sectional area and not to its mass. Inman offers no argument, either, to justify his assumption that, if the maximum forces which muscles groups can exert are in a certain ratio, the muscles will, in a particular situation, exert forces in the same ratio. Inman's values of joint force must also be treated as conservative since the problem is treated as two dimensional. In fact the centre of gravity of the body does not~~e~~ lie in the vertical plane through the hip joints and as shown by Williams (1964) additional joint force will be developed corresponding to this moment action.

The analysis of Inman has been developed and varied in a number of papers such as Blount (1956), Backman (1957), Denham (1959), Frankel (1960) and Strange (1965) but none of these papers has proceeded to the essential analysis of the three-dimensional dynamic system of forces during activity.

Studies of the analysis of the gait of the normal individual and the amputee are being performed at New York University. The methods of analysis used are reviewed by Drillis (1955), (1959) and (1950). These authors are concerned only with movement parallel to the plane of progression and recommend therefore the use of interrupted light photography on a single plate camera. The use of accelerometers to study features of gait is also described. Drillis (1961) reports the alteration of the gait characteristics with age but only stride length and cadence are measured.

Frankel (1960) and Hirsch and Frankel (1960) reported the results of tests on cadavers to determine the change in hip joint pressure due to alteration in support conditions from single to double support and due to increase in total body weight. The tests involved preparations of the upper half of the femur and the pelvis in which the joint capsule remained intact but all soft tissue was removed. The abductor muscle action was represented by a cable attached to the trochanter passing over a pulley fixed to the pelvis and carrying a weight at its free end as shown in Fig. 12. A hole was



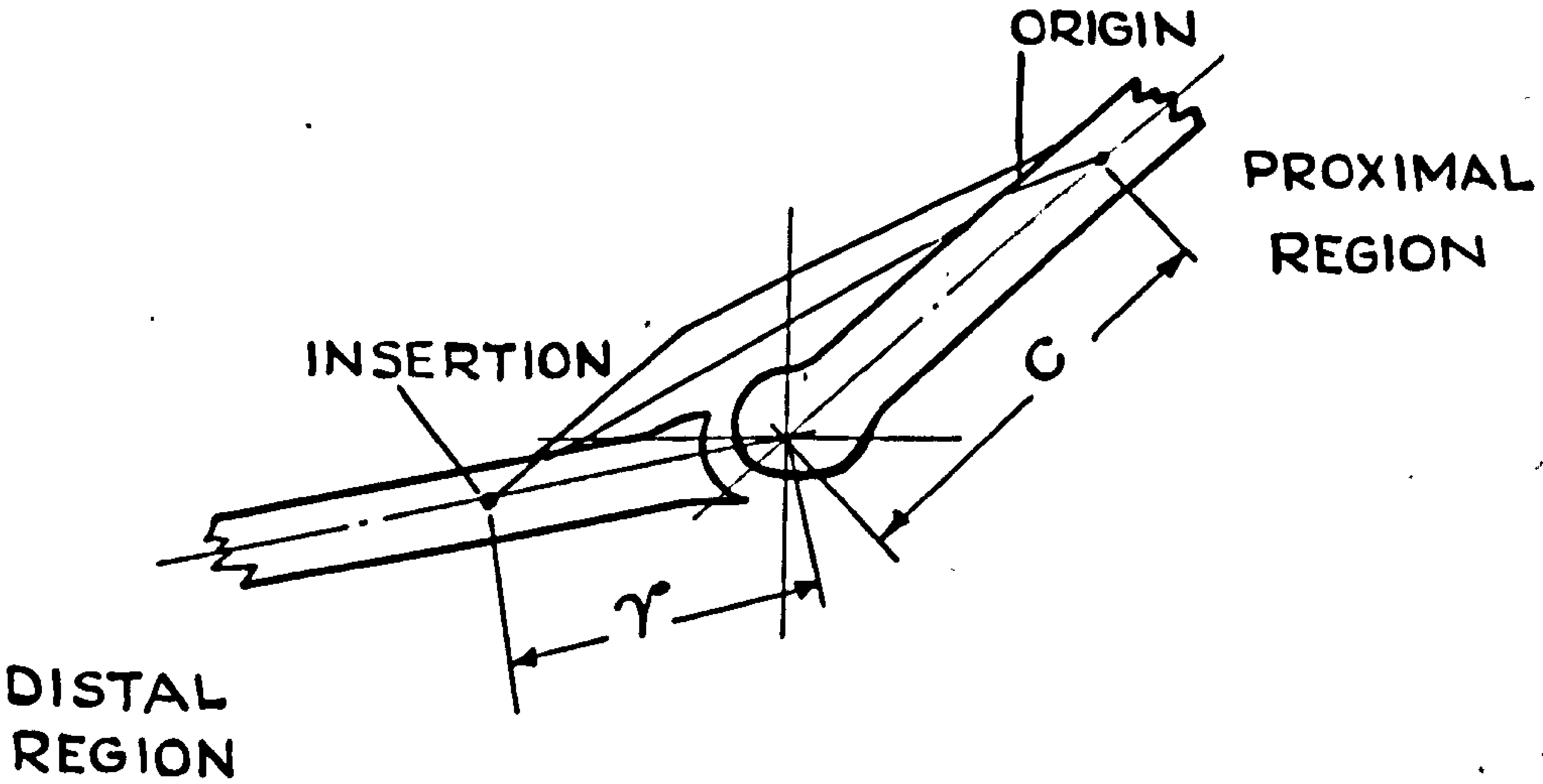
Experimental Layout of Hirsch and Frankel (1960)

Fig. 12.

drilled from the trochanter along the neck to pass through the head of the femur and penetrate the joint capsule. A transducer was introduced through this hole and fixed to the femur to indicate the force sustained by a limited area of the head of the femur when load was applied to the pelvis to correspond to the weight of the trunk. No tests were performed which simulated the full anatomical system of musclature of the three dimensional system of force actions present at this joint.

To assess the suitability of different types of floor covering, the Building Research Station conducted tests on a walkway with force plate to determine the normal pressure and the tangential components of force between floor and foot. The force plate was similar to that described by Cunningham and Brown (1952) and cine films were taken from below by means of a mirror mounted below a glass section of the walkway. The test results are reported by Harper, Warlow and Clarke (1961) and cover tests performed on 48 subjects walking in a straight line and 44 walking round a double right-angled U corner.

MacConnaill (1962) summarises the work of his previous papers and presents a version of the force analysis relating to the action of muscles at joints. Muscles are classes as "spurt" or "shunt" in their action. A "shunt" muscle at all times develops a component of force, tending to prevent the separation of the joint surfaces. If a muscle (and its tendinous connections if present) has a connection to bone distant c from the joint on the proximal side and distant r on the distal side as shown in Fig. 13, the muscle will be a "shunt" muscle if c is greater than r. All muscles having a value of c less than r are classified as "spurt" muscles. It is explained that in general action spurt muscles are best adapted for developing moment actions at a joint whereas "shunt" muscles may frequently be called into play to preserve the integrity of the joint when the action of external force and the spurt muscles would cause separation at the joint. There is however a paradox in this definition in that, for example, if a muscle is a spurt muscle as regards movements of the thigh relative to the trunk, the same muscle is a shunt muscle as regards movements of the trunk relative to the thigh. MacConaill also states the "law" of minimal muscle number :- "that no larger number of muscles is used for a task than is both necessary and sufficient for it". Electromyographic studies by Travill and Basmajian (1961) and De Sousa, De Morais and Ferraz (1957, 1958) are cited in support of this law. These studies show that in slow movements



MUSCLE ACTION AT A JOINT

FIG. 13

against light load one muscle shows electrical activity but that synergistic muscles are recruited as the load and/or speed are increased.

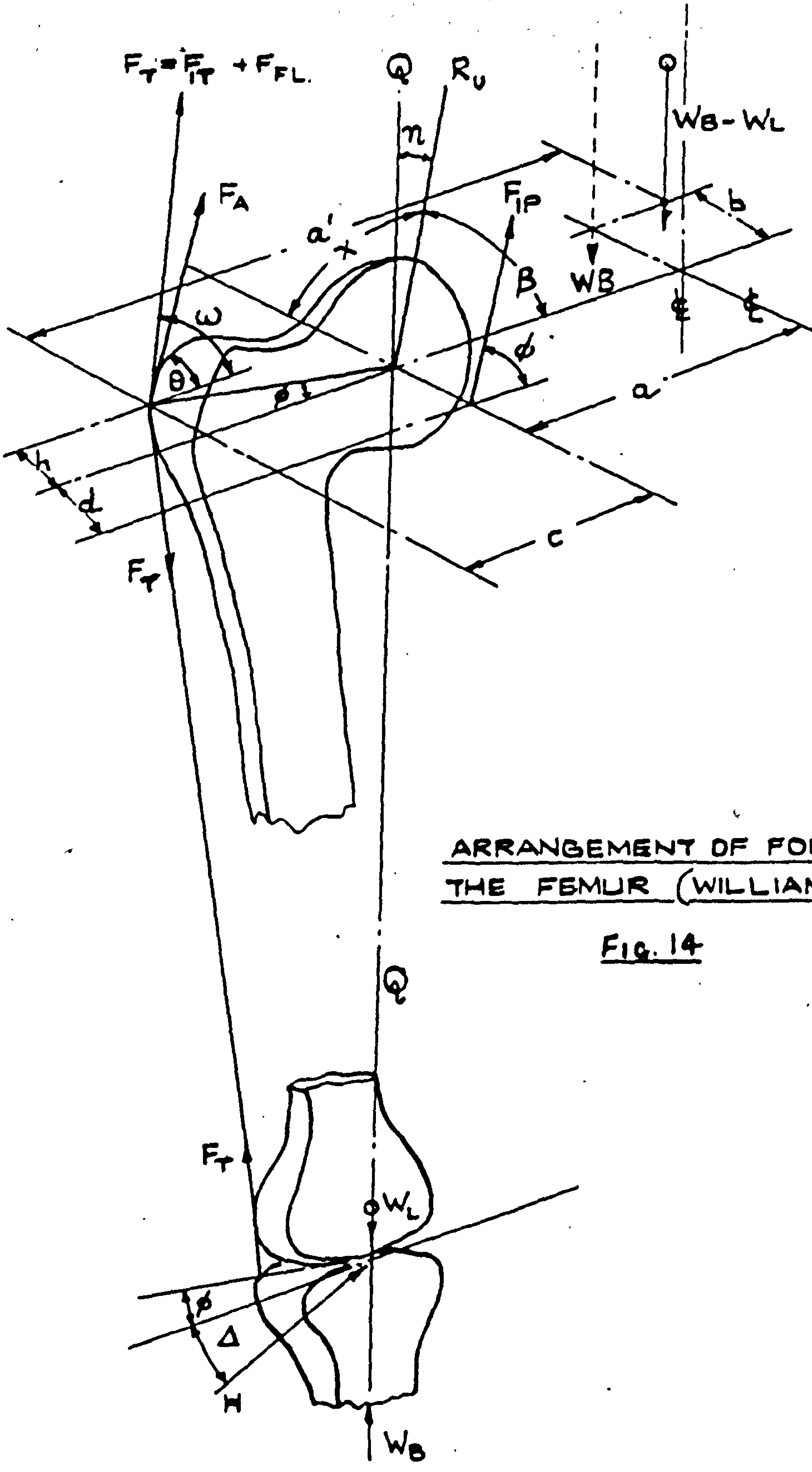
Beebe (1964) reports the results of tests performed on a walkway, incorporating force plates, which was installed in an aircraft. The subjects walked along the path and a record was taken while the aircraft performed a manoeuvre which resulted in the reduction of the gravitational field experienced. Tests were performed with effective gravitational fields in the range 0.1 g to 1.0 g. Graphs are shown of the normal and the two shear components of the resultant ground to foot force. No records were taken of the spatial configuration of the subject and analysis of joint forces is therefore not possible. In some tests accelerometers were fitted to the test subject.

In the course of a thesis concerned with the analysis of the stress in the neck of the femur Williams (1964) performs a three-dimensional analysis of the force actions of the femur during standing on one leg. Inman's values of the inclinations of the gluteal muscles and his ratio of forces developed are used but the value of force in the ilio-tibial tract obtained by the use of electromyography is not used. It was assumed that ilio-psoas, rectus femoris and the ilio-femoral ligament all contributed to resisting the extension moment at the hip and that the force in these three members was the same. The necessary dimensions of the pelvic/femoral region were obtained by measurement from a part-skeleton and the proportions were compared with values quoted by Williams and Lissner (1962). From the equilibrium equations for the parts of the body supported by the one hip joint, and for the supporting femur, Williams obtains a value of total hip joint force of 1090 lb. for a subject of 180 lb. weight, a ratio of 6.05:1. It is suggested that this figure is in error for the following reasons. Williams considers the whole body weight W_B acting on the body centre line as shown in Fig. 14 to be supported by force actions W_B and H transmitted through the single supporting knee joint. This is impossible. The moment of the force F_T in the ilio-tibial tract at the knee is not balanced by any moment acting on the femur from below.

Williams equation A.09 reads.

$$\left[F_T \sin \omega + F_A \sin \theta\right] . h + HL \sin \Delta = F_{IP} . d . \sin \psi \quad \text{----} (7)$$

This is the equilibrium equation for moments about the vertical axis 'QQ' through knee and femoral head and is correct for the force actions shown in Fig. 14. The moments of these forces about 'QQ' are, however small and it is suggested that these



ARRANGEMENT OF FORCES ON THE FEMUR (WILLIAMS 1964)

FIG. 14

moments will be comparable with these exerted by the rotator muscles which are not included in this analysis. For example if the equilibrium of the trunk and pendant leg is considered, and Williams equations are altered accordingly, the use of A.09 in some solutions indicates a negative tension in one muscle group, which is obviously impossible.

In addition, some of the values assumed by Williams are at variance with those used by other workers e.g.

Ø :	Williams 25° :	Bachman (1957)	13.54 $\pm$ 1.14°
b :	Williams 1.52 in :	Akerblom (1948)	0.71 in mean.

The dimension d corresponds to the moment arm in flexion of the combination of the rectus femoris and ilio-psoas muscles and the ilio-femoral ligament. Williams value of 0.9 in seems small. If $d = 1.2$ in, $b = 0.7$ in., $Ø = 12°$, and the errors in the equations are corrected, the value of the resultant joint force becomes 672 lb which is 3.73 times the assumed body weight.

The only direct experimental evaluation of hip joint force known to the author is described by Hirsch and Rydell (1965) and Rydell (1966). A male subject 51 years of age injured in a motor accident and a female subject 56 years of age injured in a home accident sustained injuries to the head of the femur. They were treated by the fitting of a specially designed Austin Moore prosthesis. The prosthesis had a hollow neck of reduced diameter and increased length to allow electrical resistance strain gauges to be fitted inside. The strain gauges were arranged to give signals corresponding to the following load actions for both subjects.

1. Axial force along the neck of the prosthesis.
2. Bending moment in one plane (A) at the centre of the head.
3. Bending moments in plane A 20 mm from the centre of the head.
4. Bending moment in plane B - perpendicular to A at the centre of the head.
5. Bending moment in plane B 20 mm from the centre of the head.

The prosthesis fitted to the female subject had gauges connected to a sixth channel to measure torsion of the neck. The gauges were connected to a sealed socket which was buried under the skin and fascia. Six months later, when the subjects were again ambulant and walking normally, tests were performed. The socket was brought out to the surface and connected to a bridge circuit whose output was fed to a galvanometer recorder.

During and after final assembly of the prosthesis, load calibrations were performed under combined static load actions. Prior to installation the prosthesis was sterilised at a temperature of 115°C and low relative humidity. It appears that calibration was not performed after sterilisation or when the prosthesis was heated to body temperature: this calibration would appear to be desirable even though the effect of temperature on the sensitivity of the gauge were expected to be small.

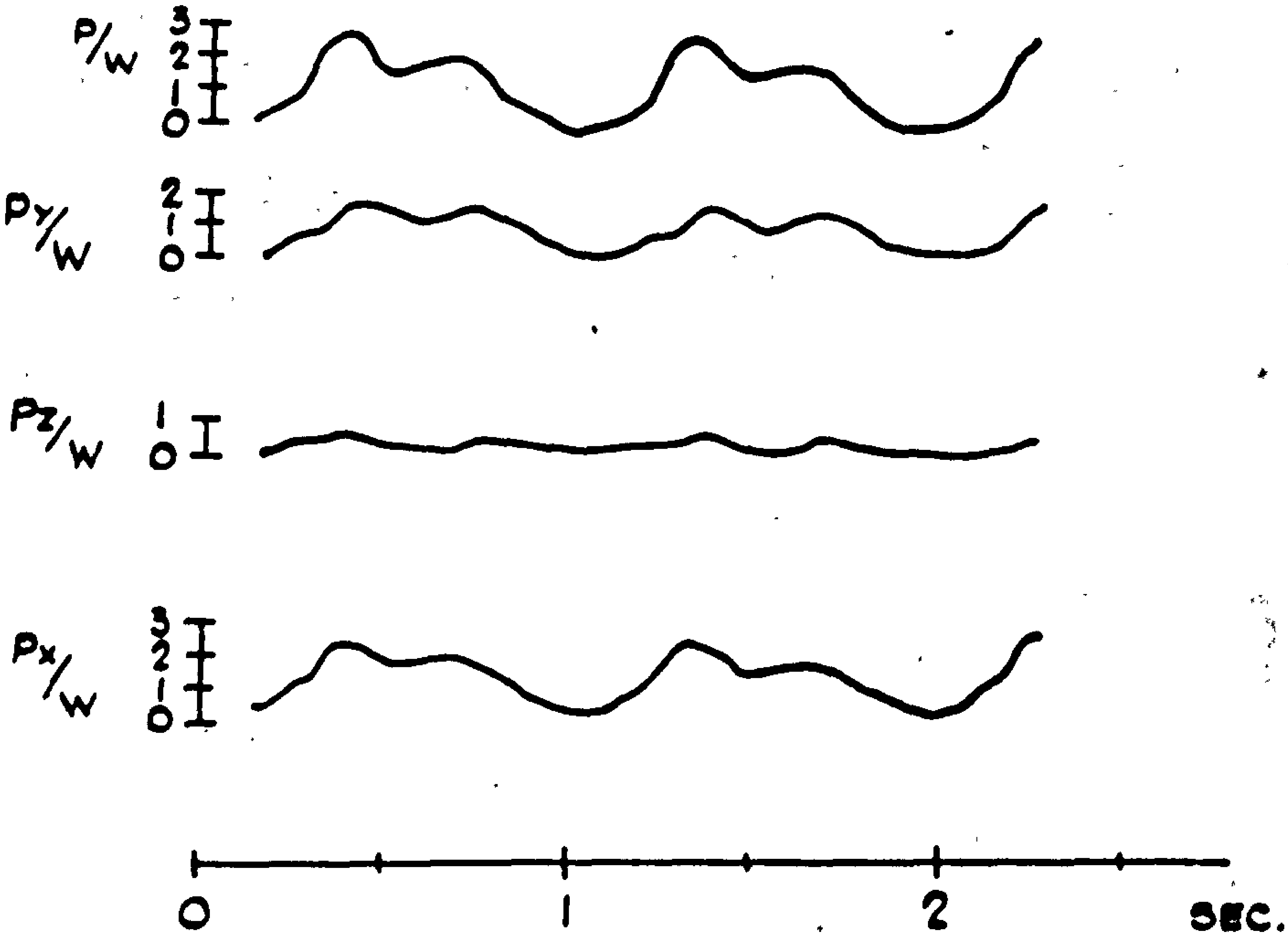
The following tests were performed by the subjects: static and dynamic flexion and extension of the thigh: standing on one leg: walking in a straight line on a level measuring walkpath : walking up and down stairs: one subject performed a running test. After tests the lead wires were severed at the prosthesis and removed.

During the level walking tests cine photographs and records of ground to foot force were taken. The cine records were taken by a single camera running at 64 frames/sec. mounted on a trolley which was moved by an operator to keep abreast of the subject. From the film, values of the angles of inclination of the thigh to the vertical were measured. A clock in the field of view of the camera was also connected to a galvanometer recorder to allow synchronisation with the forces recorded by the walk path.

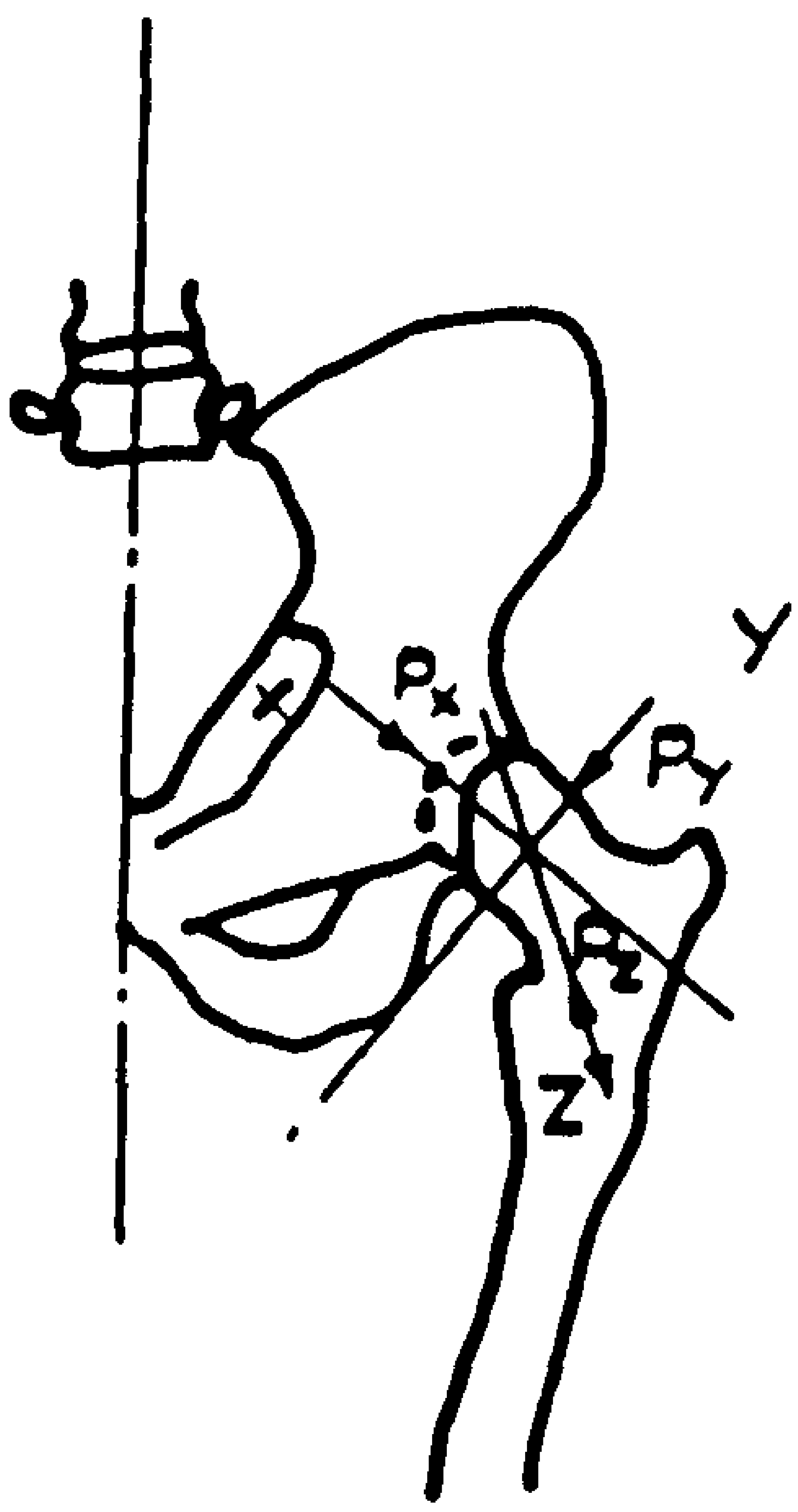
The walk path comprised two structures 5 metres long 20 cm. wide placed side by side 2 mm apart to measure the successive forces exerted by the left and right feet. Each structure was instrumented to measure vertical force and force along the line of progression only.

For each test values are quoted for:- the values of the three components of resultant joint force: the value of the resultant force and its angles to two reference directions. The greatest value of joint force recorded was that for the female subject during running when the joint force was 4.33 times body weight. In level walking the greatest ratio was 3.27 times and in standing on one leg 2.9 times, both of these being for the female subject. A typical graph from one of the subjects is shown in Fig. 15

Merchant (1965) reports the result of tests performed on the assembly of a dried adult male pelvis and the right femur. Certain muscles were represented by chains connected to strain gauged force transducers and secured to act along the mid lines of the appropriate muscles. Five such connections were made to represent gluteus medius , gluteus minimus, the joint action of the ilio-tibial tract and tensor fascia lata, the joint action of the adduction muscles and the joint action of the hamstrings. Vertical load was applied by a cable attached to the spine. Tests were performed with the femur in different degrees of abduction/adduction and internal and external rotation. The values of the



VARIATION WITH TIME OF THE COMPONENTS OF FORCE, P, ON THE HEAD OF THE FEMUR (RYDELL 1966)
W = SUBJECT'S BODY WEIGHT.



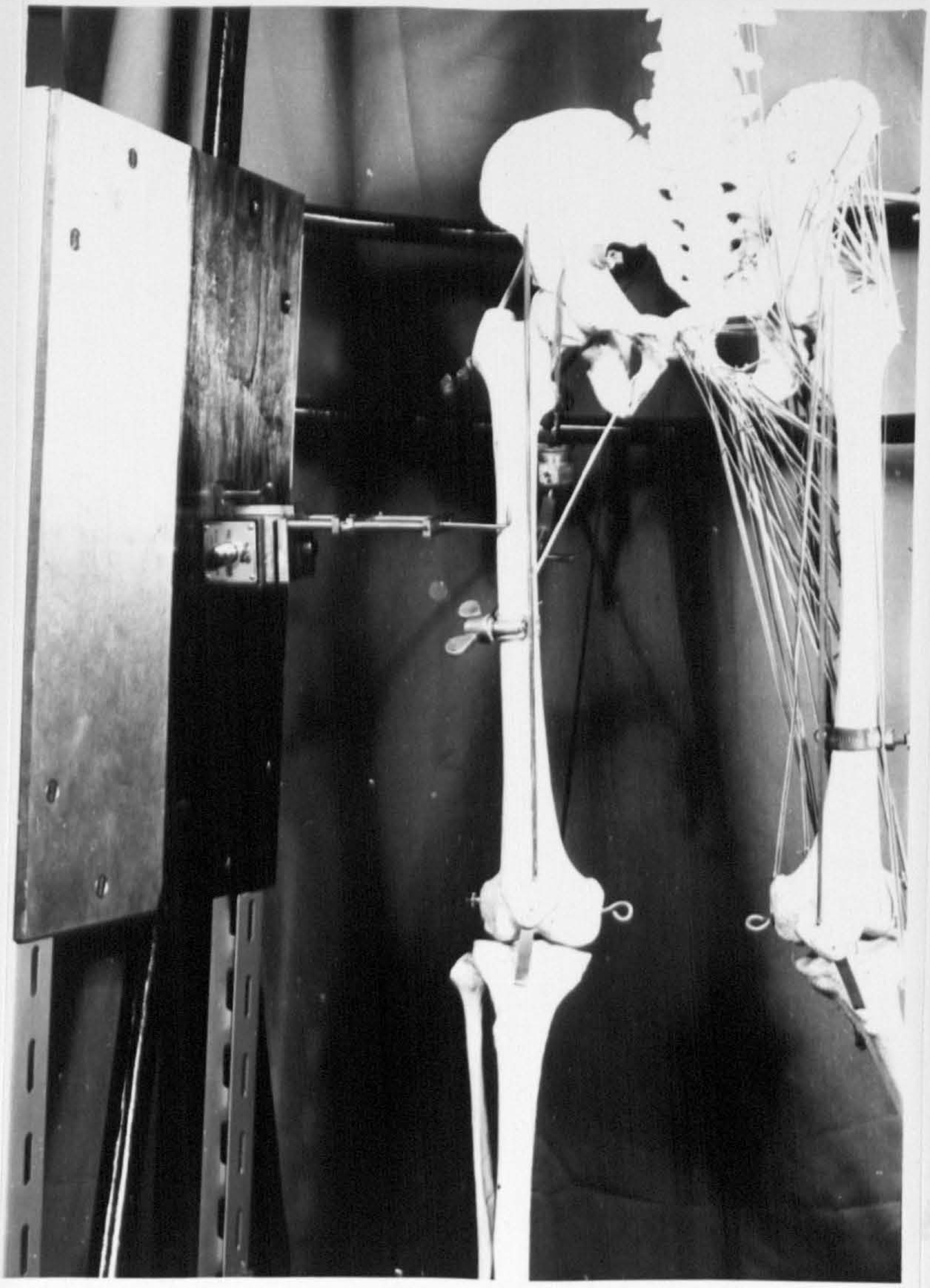
DIRECTIONS OF FORCE COMPONENTS
THE AXES ARE FIXED RELATIVE TO THE FEMUR., X LYING ALONG THE AXIS OF THE NECK OF FEMUR.

FIG. 15

forces measured by the three transducers representing abduction action are tabulated for each case. It is submitted, however, that this experimentation is invalid.

At any instant in time three muscles forces only are required to maintain the pelvis in equilibrium on the femur. Depending on the directions of the internal or body forces, these are, one abductor or adductor, one flexor or extensor and one rotator, internal or external. For three such muscles the forces may be calculated by the relevant equations for equilibrium, or a corresponding laboratory experiment can be set up to give the same result as modified by the effect of joint friction. If, as in the living body, more muscles are present and active, the equations of statics must still be satisfied, but these equations are no longer sufficient to obtain a solution. In the living, the values of the forces are determined by the degree of energisation transmitted to the muscles from the central nervous system and on the effective cross-sectional area of the muscles. In a test on a dissecting room specimen the values of the forces in the 'muscles' depend instead on the elastic stiffness of the connections representing them, generally the less extensible the connection, the greater will be the share of load taken. In four of the seven tests reported it appears significant that gluteus minimus, which is represented by the shortest and least flexible chain, is shown as developing the greatest tension. In five of the tests it develops forces greater than gluteus medius, although anatomically the latter is a muscle of greater bulk. Merchant shows for one loading case that the sum of the forces in the elements of the abductor muscles corresponds to the value calculated by statics for a single member performing the same function. This cannot be held to validate the values obtained for the individual muscles. The experimental technique is also incorrect in that, to represent the moment about the supporting hip of a man of 150 pounds weight, the line of action of weight of body + one leg is taken to be situated on the centre line of the trunk, instead of in the displaced position as shown by Pauwels (1935) and Inman (1947).

The first results of the investigations carried out at the University of Strathclyde into the value of hip joint forces were reported by Paul (1965). Force plate and cine camera records were taken in a similar manner to that of Bresler and Frankel and used to obtain curves of resultant force and moment transmitted between the leg and the trunk. Lines of action of the principal muscle groups acting at the hip were delineated on a skeleton as shown in Fig. 16 and these were combined into six major groups.



Skeleton with cords representing muscles and muscle groups.

Fig. 16.

1. The long flexors, comprising Rectus Femoris and Sartorius.
2. Psoas/Iliacus.
3. Gluteus Maximus.
4. The long extensors, comprising Biceps Femoris (Long head), Semimembranosus and Semitendinosus.
5. The Abductors, comprising Gluteus medius, Gluteus Minimus and Tensor Fascia Lata.
6. The Adductors, comprising Adductor Magnus, Adductor Longus, Adductor Brevis, Pectineus, and Gracilis.

The lines of action of these groups were taken to extend from the centre of their joint area of origin to the centre of their joint insertion or to any bony surfaces over which they passed. With the femur of the skeleton set to prescribed angles of flexion and extension, measurements were taken to define the position of the lines of action relative to a set of orthogonal axes through the centre of the head of the femur. From these measurements the moments about the two horizontal axes due to unit force in the muscle groups were calculated. The dimensions of the skeleton were related to these of the test subject by a single proportionality factor compounded from five measurements of the pelvis and femur of each.

Using this information the forces in the muscles and the corresponding joint forces were calculated. Since two groups of muscles resist extension of the femur on the trunk, two solutions for joint force are possible, depending on which group is presumed to carry the moment action exclusively. A similar situation exists for the extensor muscles. For any instant in time the value of the true joint force will lie between these two values. The results presented for one subject of 180 lb. weight show a value of joint force of 1050 lb. on the most optimistic selection of muscle action. This is a joint force/weight ratio of 5.8.

To eliminate the labour of calculations the first part of the procedure had been programmed for digital computer solution, and graphical smoothing of displacement curves was replaced, by a nine point double differentiation formula which involved a least square fit of the experimental points.

Hunt (1965) reported the results of an investigation into the accuracy of this procedure, using accelerometers mounted on the shank of the subject. The results showed that generally the differentiation procedure was satisfactory and that the use of the accelerometers would involve excessive complication and constriction of the test subject.

Further results of the test procedure described above was reported by Paul (1966) and improvements in the experimental technique were described in Paul (1967).

Determination of Mass Properties of Body Segments.

Braune and Fischer (1890, 1892) performed a series of tests on four frozen cadavers to determine the mass properties of the human body. Tests were performed on the intact bodies to determine the centre of gravity. There is some doubt, however, whether these values obtained from cadavers frozen on a horizontal table are relevant to the upright subject. The most valuable part of the work, however, relates to three of the cadavers which were cut into segments at the principal joints of the body. These segments were weighed: their C.G. positions were determined by two point support and their second moments of mass about their centre of gravity were determined by measurement of their period of oscillation when suspended vertically from pins driven through the bone. The results quoted show:-

1. the mass of a part of the body as a fraction C_1 of the total body weight.
2. the centre of gravity position as a distance C_2L from the proximal joint where L is the length of the part.

3. the radius of gyration about an axis perpendicular to its length as a fraction C_3 of the length of the part.

4. the radius of gyration about its longitudinal axis as a fraction C_4 of the diameter of the part.

The coefficients are averages of the results of the three dissected cadavers. The centre of gravity in each case is taken to lie on the line joining the joint centres. The radius of gyration is taken to be the same about both principal axes perpendicular to the line of joint centres. The values of the coefficients are quoted in table 3 and have been used by many subsequent investigators. It should be noted that the figures quoted here are average values from the three subjects measured, not the values from the third subject only, which are quoted in several papers.

The first significant study after Braune and Fischer was that of Amar (1914) who assumed simple geometrical forms for parts of the human body in order to calculate moments of inertia from simple body dimensions. The trunk was assumed to be a right cylinder and the thigh, shank, upper arm and forearm were all assumed to be frustra of cones. Zook, (1930) studied volume changes of bodysegments with age for males between the ages of 5 and 19 years, using an immersion technique which was

PART OF BODY.	C1	C2	C3	C4
HEAD.	0·069	0·74	0·43	—
TRUNK & NECK.	0·462	0·39	0·3	—
ONE UPPER ARM.	0·0332	0·47	0·3	0·35
ONE FORE ARM.	0·0209	0·42	0·3	0·35
ONE HAND.	0·0084	—	0·3	0·35
ONE THIGH.	0·1072	0·44	0·3	0·35
ONE SHANK.	0·0478	0·42	0·3	0·35
ONE FOOT.	0·0169	0·35	0·3	0·35

WT. OF PART = C1 x TOTAL BODY WT.

DIST. OF C.G. FROM PROXIMAL JT. = C2 x SEGMENT LENGTH.

RAD. OF GYRATION ABOUT TRANSVERSE AXIS THRO' C.G. = C3 x SEGMENT LENGTH.

RAD. OF GYRATION ABOUT LONGITUDINAL AXIS THRO' C.G. = C4 x SEGMENT DIAMETER.

BRAUNE & FISCHER'S COEFFICIENTS

TABLE. 3

claimed to allow the direct determination of specific gravity. This last statement is challenged by Drillis, Contini and Bluestein (1964).

Salzgeber (1949) reported that in the period beginning 1930 Bernstein performed tests on a single cadaver whose extremities were dissected into segments of 2 cm. height and weighed. The conclusion reached was that the centre of volume and centre of mass of the extremity were coincident and that volume determinations by immersion or other techniques could be used to obtain mass, centre of gravity and second moment of mass. Bernstein, Salzgeber, Pavlenko, and Gurvich (1936) at the Russian All-Union Institute of Experimental Medicine in Moscow performed an extensive investigation into body segment parameters of living subjects. These investigations were reported in a monograph which the writer has been unable to obtain in this country. Drillis et al quote extracts from the monograph which show that, whereas, for the range of 152 subjects studied by Bernstein, the average mass properties are in fair agreement, the extremes of the range differ considerably from the values of Braune and Fischer. Bernstein claims that his technique allowed the determination of mass centre to an accuracy of $\pm$ 1 mm. Considering a typical femur of

16.5 in. length this is an error of only $\pm$ 1/4 per cent. Unfortunately no details are available of the test procedure used.

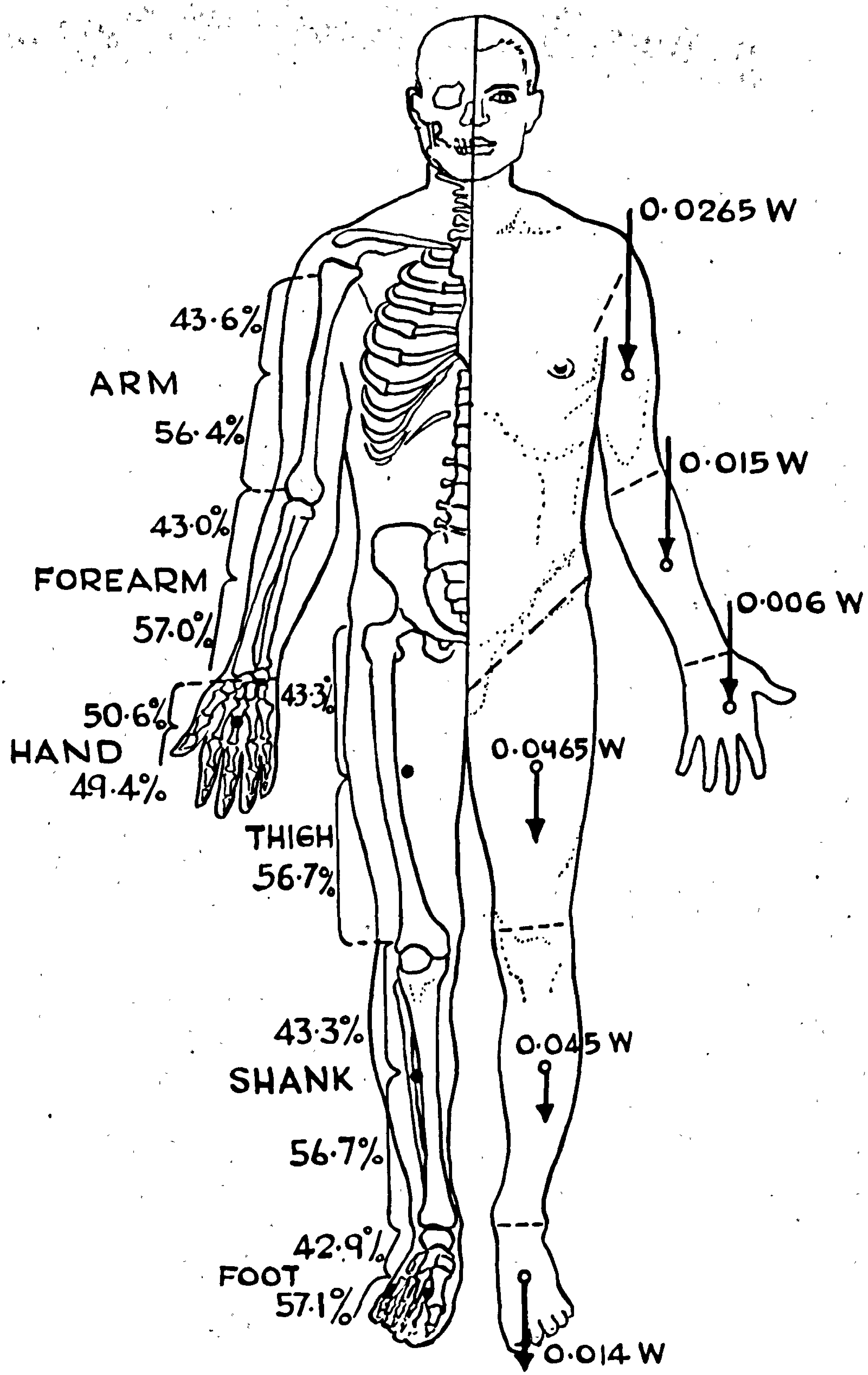
Weinbach (1938) propounded a method for calculation of body segment mass properties based on a series of measurements of the subject taken perpendicular to the long axes. The analysis was performed by producing for the standing subject a curve showing for any horizontal level the area of the cross-section of the body at that level. This curve was obtained from the measured dimensions assuming that each cross-section of the body was elliptical in cross-section. The volume curve was then integrated twice, graphically, to allow the obtaining of volume, centre of volume and second moment of volume. With the further assumption of uniform body density of 1.00 gm/cm^3 the mass properties were obtained. The assumption of elliptical form of cross-section is certainly in error for regions with bony prominences such as the foot, the knee and the hip. To evaluate this error, in other regions the author performed certain calculations based on typical cross-sections of the human leg illustrated by Morton (1944). The areas of these were measured by planimeter and compared with the values based on Weinbach's assumption. Two cross-sections of the shank and two of the thigh were investigated. The percentage errors ranged from 2 - 10% averaging 5.5%. In each case Weinbach's method gave an overestimate.

The validity of the density figure quoted by Weinbach is doubtful, since in the foot and lower shank the proportion of bony to soft tissue is vastly different to the value at mid-shank and mid-thigh. Bashkirew (1958) quotes a variation in the specific gravity of different regions of the body from 0.978 to 1.109 with a mean value of 1.044. Thus the assumption of uniform specific gravity of 1.00 will give a probable mean error in total mass of segment of 4.4% and an extreme error of 10.9%,with corresponding errors in centre of gravity position and second moment of mass.

Dempster (1955), who had then no information about Bernstein and co-worker's studies, repeated and extended the range of Braune and Fischer's studies using cadavers. Dempster obtained the volume, mass, centre of gravity position and second moment of mass for the segments of eight cadavers. The volume was obtained by immersion. The values of the coefficients C_1, C_2, C_3 referred to by Braune and Fischer differ from those of Dempster particularly in respect of C_1 the weight of the segment as a fraction of total body weight. The average of Dempster's C_1 values for head, neck and trunk without limbs is 0.565 compared with 0.497 from Braune and Fischer. This is probably accounted for by the difference in subjects. Dempster's cadavers were in the age range 52 - 83 at death and had an average weight of only 131 lb. whereas Braune and Fischer's cadavers of average weight 141 lb and aged 45 - 50 were close in height and weight to the average German soldier and presumably well developed in the muscles of the limbs. Dempster analysed his results statistically and quotes regression equations for the appropriate quantities. Williams and Lissner (1962) illustrate Dempster's results in the diagram reproduced as Fig. 17.

Studies are currently proceeding at New York University to determine and correlate values for the volume, mass, centre of gravity and second moment of mass for segments of the living human body. Drillis et al describe the following experimental techniques although no results have yet been published:-

1. Volume measurement by displacement.
2. Volume measurement by photogrammetry.
3. Change of reaction measurement.
4. Quick release technique.
5. In vivo pendulum test.
6. Model pendulum test.
7. Torsional pendulum test.



CENTRE OF GRAVITY AND FRACTIONAL WEIGHTS OF BODY SEGMENTS

FIG. 17

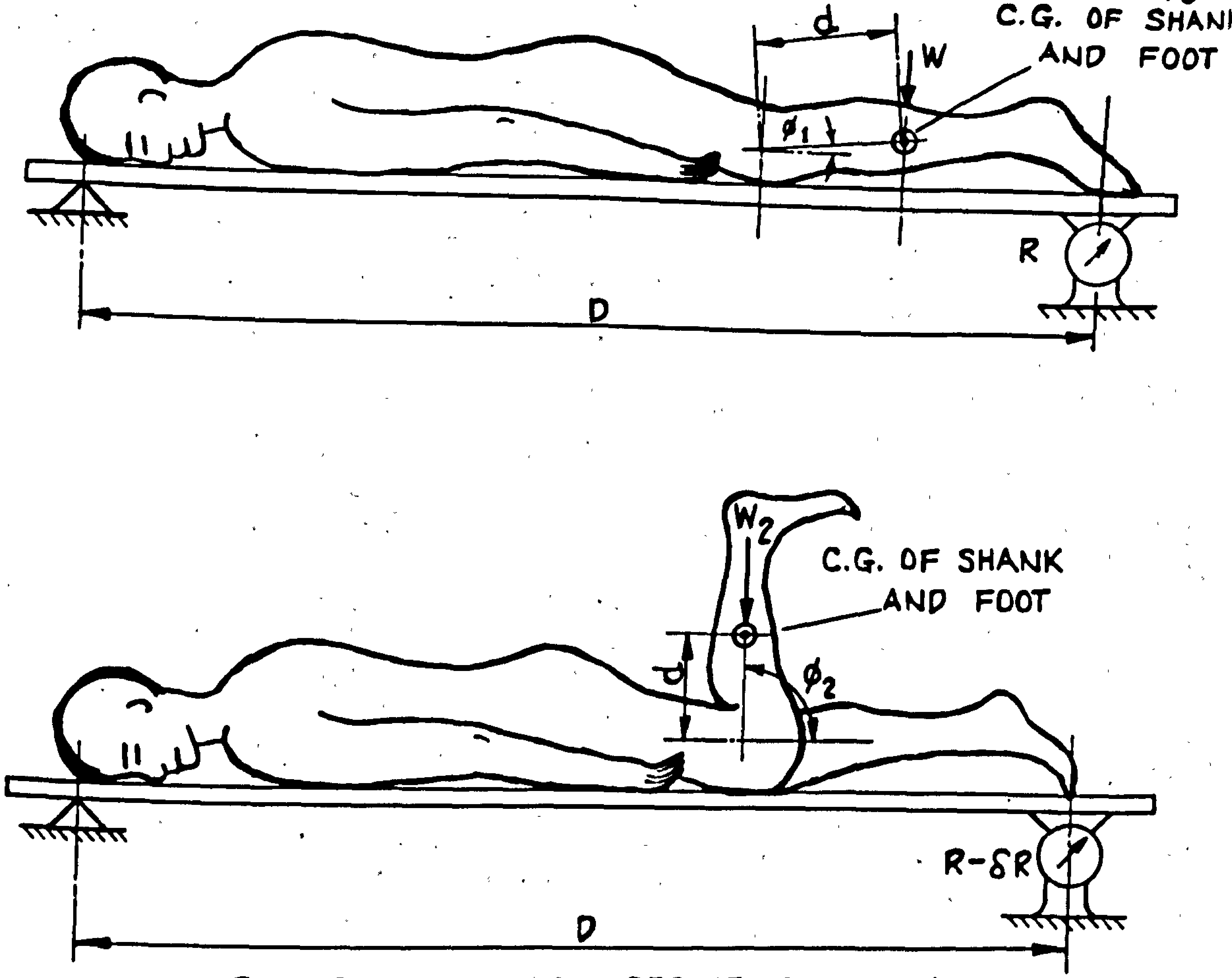
In (1) the displacement procedure, the segment is suspended in a regular containing vessel of cross-section not much greater than that of the segment. Measured volumes of liquid are added and the corresponding changes of liquid level are a measure of the local cross-sectional area of the segment. The curve of area to liquid depth is integrated to obtain total volume and centre of volume. In (2) the photogrammetric method, the segment of interest is photographed and the resulting picture is treated as an aerial photograph of terrain upon which contour levels are applied. The portions of the body part between successive contour levels form segments whose volumes can be found by the use of a polar planimeter.

In (3), the reaction change technique, the subject lies on a board which is supported on two axes, distance D apart as shown in Fig. 18.

The reaction R at one of the supports is measured with the segment in one position say $\emptyset_2$ to the horizontal. When the segment is rotated to position $\emptyset_2$ the reaction changed by δR. If the weight of the body part distal to the joint at which rotation occurs is w and the centre of mass is d from the joint then

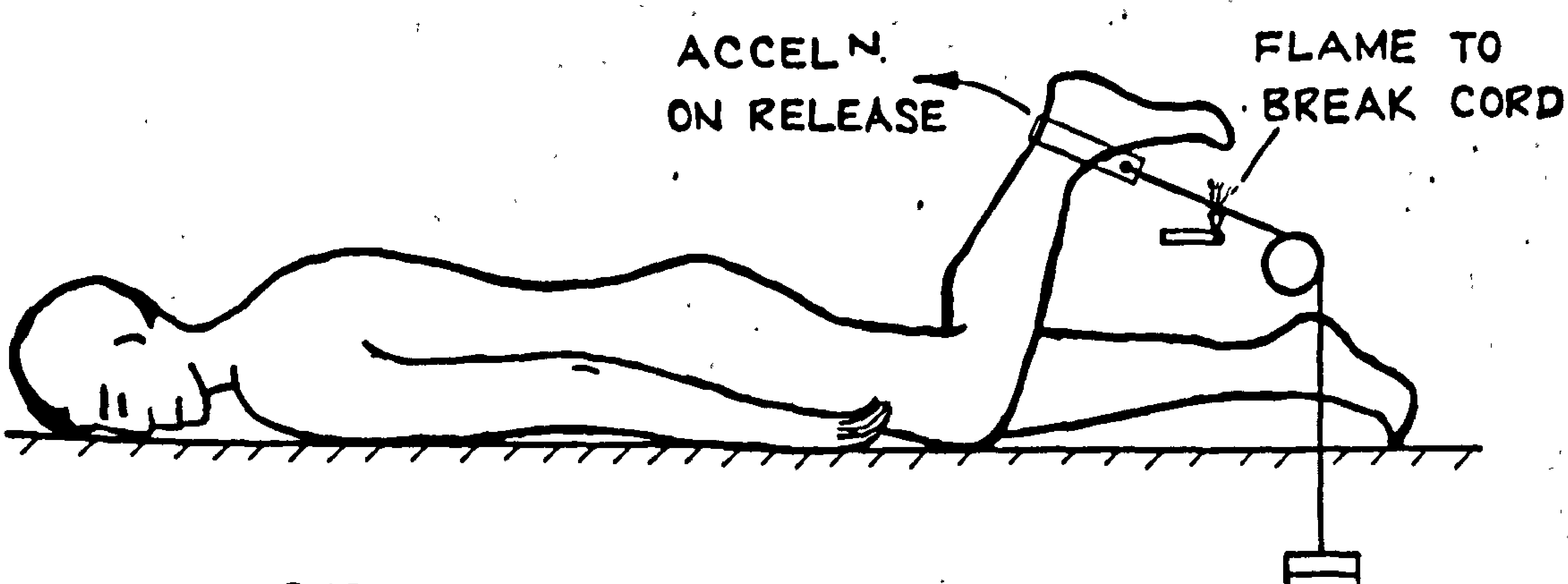
$$wd\,(\cos \emptyset_1 - \cos \emptyset_2) = \delta R.D. \quad \text{---- (9)}$$

If the centre of gravity position is known i.e. d, $\emptyset_1$ and $\emptyset_2$



DETERMINATION OF SEGMENT WEIGHT

FIG. 18



SUDDEN RELEASE METHOD

FIG. 19

are known, the technique can be used to obtain mass of the segment.

In (4), the quick release method, the subject exerts a moment about a joint against a measured restraint as shown in Fig. 19. When the restraint is suddenly released the part distal to the joint is subjected to an accelerating torque whose initial value will be that produced by the restraint. If the resulting acceleration is measured, the second moment of mass of the parts distal to the joint can be calculated.

In (5), the in vivo compound pendulum method, the segment is made to oscillate about its proximal joint

(a) alone

(b) with a known mass fixed at a definite location

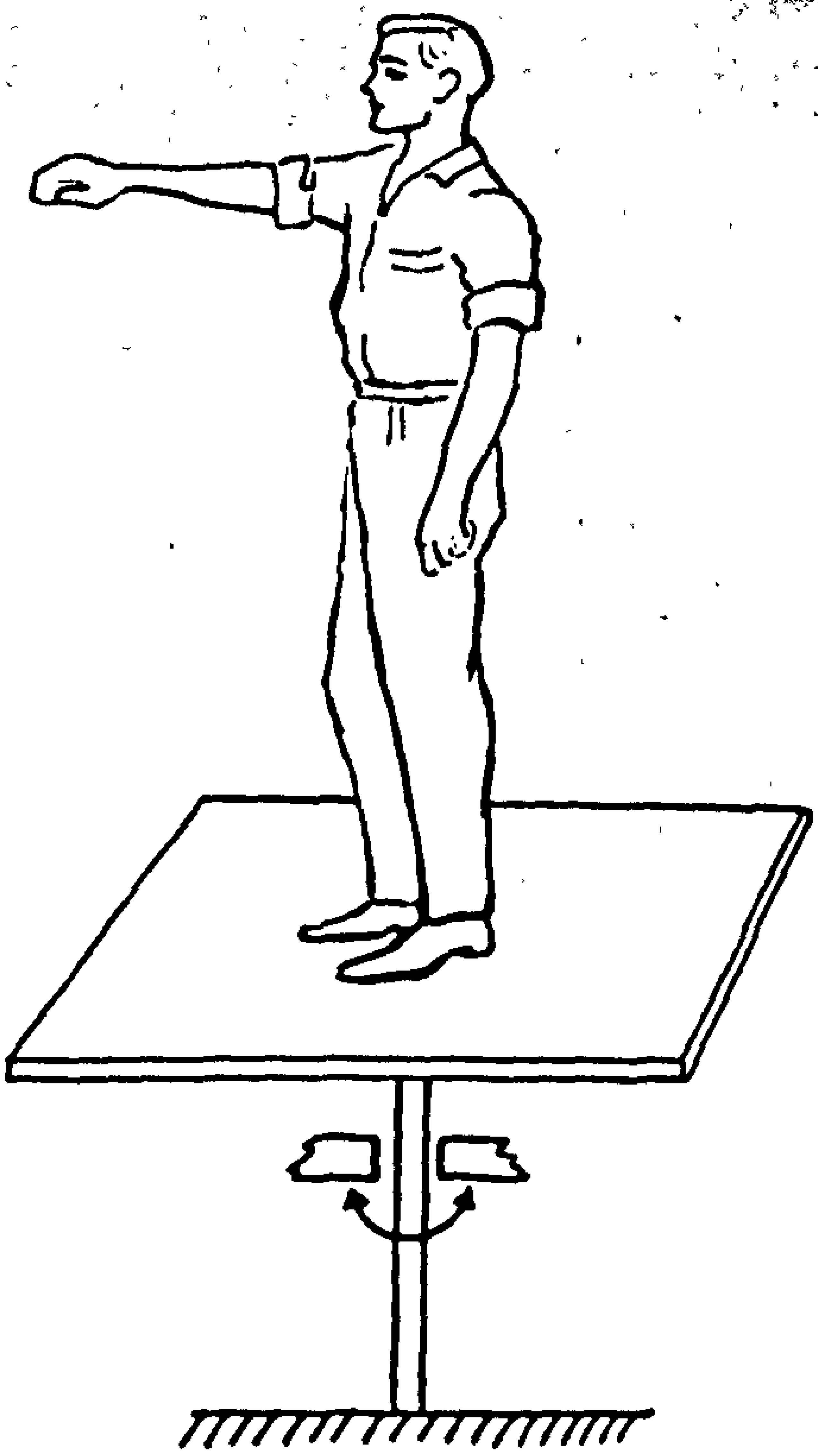
(c) with another known mass fixed at the same location.

The periods of oscillation in these three tests allow the calculation of the effective centre of rotation, the centre of gravity and the second moment of mass.

In (6), the model compound pendulum method, a plaster of Paris casting is made of the segment, and this is used to determine centre of gravity and radius of gyration by suspending it as a compound pendulum from two axes in succession and measuring the periods of oscillation.

The period of oscillation of a plate supported in a horizontal plane and oscillating about a vertical axis against the elastic restraint of a bar loaded in torsion can be measured. The alteration in period where the mass of a test subject is supported on the plate is the basis of the "torsional pendulum" method(7) as shown in Fig. 20. If the segment of interest is placed successively in three different positions in the plane of the plate as shown in Fig. 21 the corresponding measured periods of oscillation allow the calculation of the mass of the segment and the position of the centre of gravity. The authors do not state this, but if the segment of interest was moved so that its axis was inclined to the plane of the plate information could also be obtained about the second moment of the mass of the segment.

In method (1) since the limb is held below trunk level it appears that for the hand and arm the values obtained will be in error due to engorgement. This effect is obvious in Fig. 22 which is reproduced from the paper. Methods (1), (2) and (6) assume uniform density of known value throughout the segment. Method (3) allows the direct determination of the first moment of mass about the proximal joint of the segment (md). Method (4) allows the direct determination of the second moment



TORSIONAL PENDULUM
FIG. 20

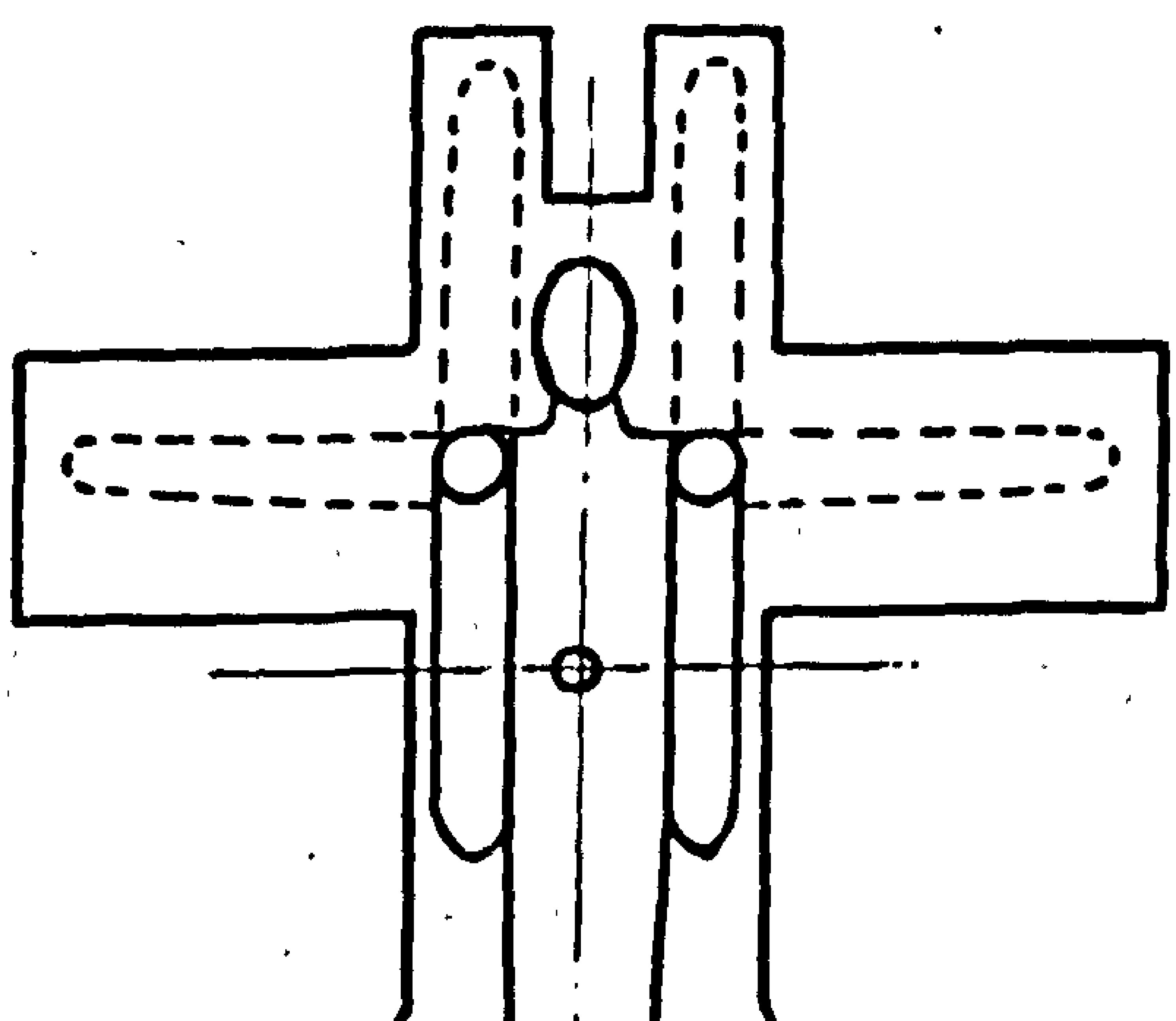


FIG. 21 POSITIONS OF SUBJECT LYING ON TABLE TO DETERMINE ARM MASS PROPERTIES

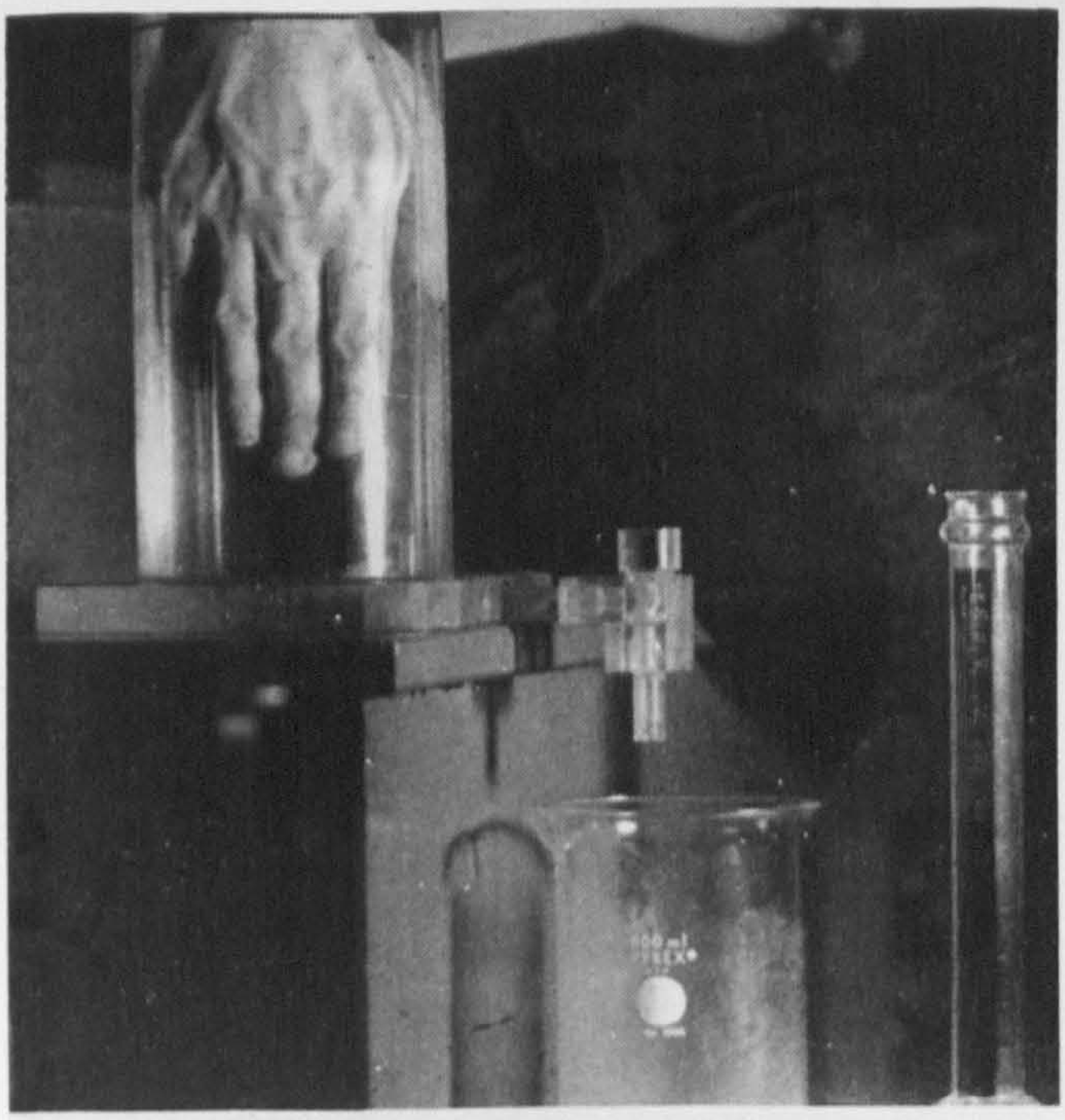


Fig. 11. Det rmination of the hand volume.

Determination of Segment Volume by Displacement

(Drillis et al (1964)).

Fig. 22.

of mass about the proximal joints $m(d^2 + k^2)$. Method (5) appears to offer the most information directly measured but it should be realised that the quantities depend on the differences in the squares of the periods of oscillation of the segment alone and with added weights. Since the number of "free" oscillations of "small" magnitude performed before coming to rest will be very small, the accuracy of determination will be poor. These comments refer also in part to method (7), although in this case the number of oscillations before the plate comes to rest will be much greater, yet the change in second moment of mass about the centre of rotation due to movement of one body segment will be very small compared to the total second moment of mass of the whole body and the supporting table and the accuracy is therefore open to question. Another serious objection to methods (3), (4), (5), (6), and (7) is that if separate values are required for the properties of all the segments of a limb, for example foot, shank and thigh, successive tests must be made to determine the properties a for the foot alone b for the foot and shank together and c for the shank and thigh together. During tests b and c particular care must be taken to maintain the further part or parts in the same position relative to the segment of interest. The accuracy of determination for the nearer parts will also be prejudiced since in the values obtained for the whole leg the greatest fractions will correspond to the further parts.

It is apparent that no ideal method has been cited to determine the mass properties of segments of the human body. Drillis et al state that an extensive survey is in progress to relate the mass properties as determined and correlated by the above tests to the subjects' weight and build as described by "somatotype" (Sheldon Stevens and Tucker (1949), Sheldon, Dupertuis and McDermott (1954).) The results are awaited with interest.

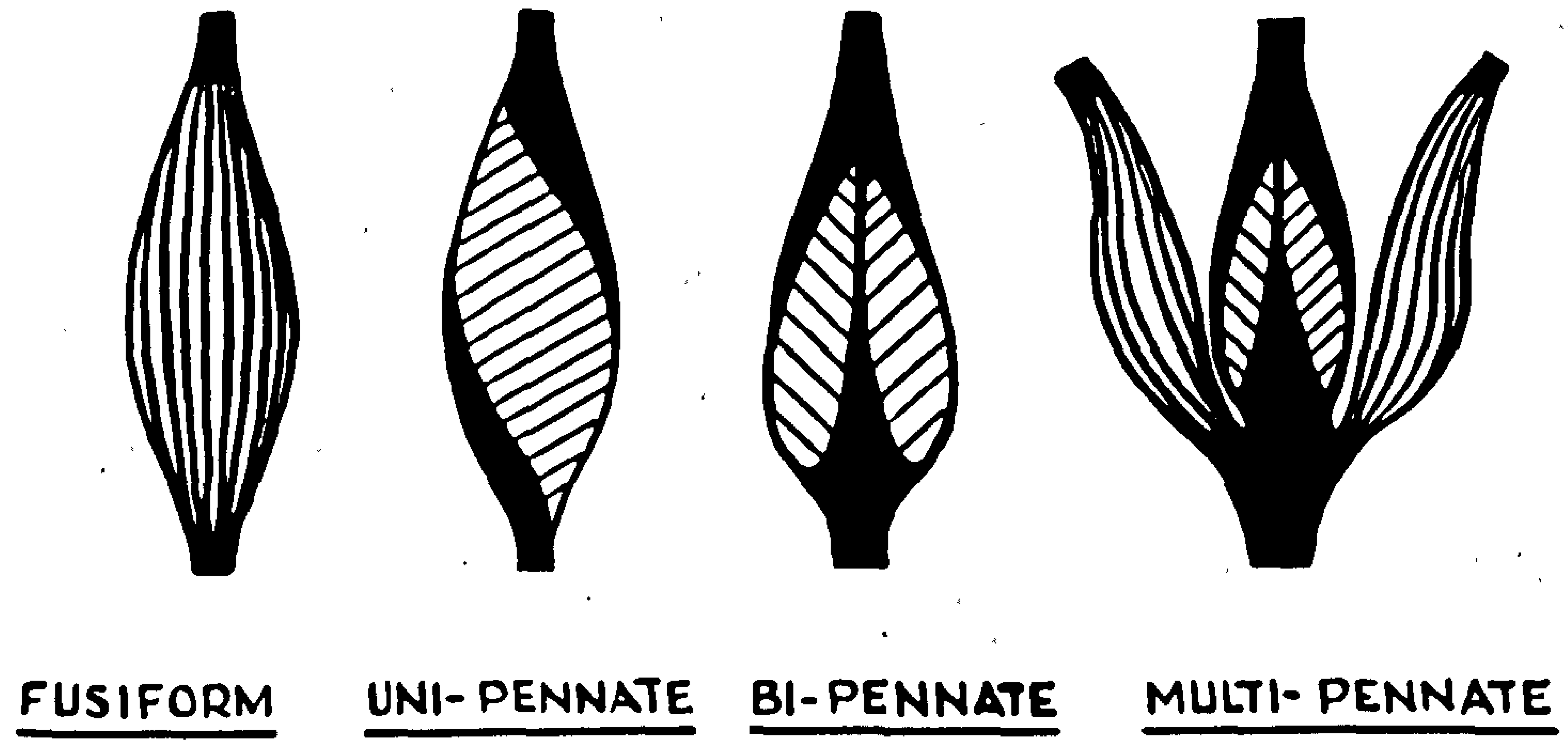
Physiology and Dynamics of Muscle Action.

Anatomy of Muscle.

Three basic types of muscle exist in the human body. Smooth muscle forms the walls of the stomach, bladder and the tubes of the circulatory system. It is involuntary in action but contains pain nerve endings and is more sensitive to thermal and chemical stimulae than other muscle types. Smooth muscle is generally slow in action. Where rapid movements are required as in the musco-skeletal system, striated muscles are found. Striated muscles are composed of thread like fibres displaying alternate dark and light bands. These muscles have nerve connections to the cranial or spinal nerve centres and are under voluntary control. In addition to the nerves controlling the action of the muscles there are nerve endings sensitive to pain and proprioceptors which transmit signals corresponding to the instantaneous length of the muscle. Cardiac muscle displays structural and functional resemblances to both skeletal and smooth muscle. Its contractile elements are transversely striated but adjacent fibres have nerve cross connections with the result that all fibres are energised simultaneously. In the present study skeletal muscle alone is of interest.

In striated muscle, the elongated cell which comprises a muscle fibre is enclosed in a thin, structureless, selectively permeable membrane adhering to an outer network of fibres. This keeps the adjacent fibres from merging into a single jelly-like mass and isolates them so that they can function as separate units. Units of 100 to 150 muscle fibres are bound with connective tissue into bundles called fasciculi and several fasciculi are bound together in turn by a sheath to form a larger unit. Bundles of these are enclosed in a further network of fibres to form muscle. The various sheaths merge together to form the tendon which connects the muscle to the bony surface of origin or insertion. The thickness and strength of the external sheath varies greatly with the location of the muscle, being greatest near the distal end of a limb where the muscle might be near the skin surface and subject to blows and abrasions. These sheaths are elastic up to strains of 40 per cent. The relative amounts of connective and contractile tissue vary greatly from muscle to muscle and this leads to difficulty in comparing the results of experiments on muscle physiology.

In gross structure, muscles can be classified in two main types, fusiform or penniform, although there are variations from each main form - as shown in Fig. 23. In the fusiform or longitudinal muscle,parallel fibres extend along the full length of the muscle from one tendon to the other. Muscles of this type are capable of contraction to about 70% of their maximum anatomical length. Where less contraction and greater force are required the pennate form is found. In the unipennate form, muscle fibres extend from one side of one tendon to one side of the other tendon. Bipennate muscles have one tendon split into two parts from which muscle fibres extend to the other tendon which is situated centrally. As shown in Fig. 23, more complex arrangements are generally described as multipennate.



ARRANGEMENT OF FIBRES OF SKELETAL MUSCLES

FIG. 23

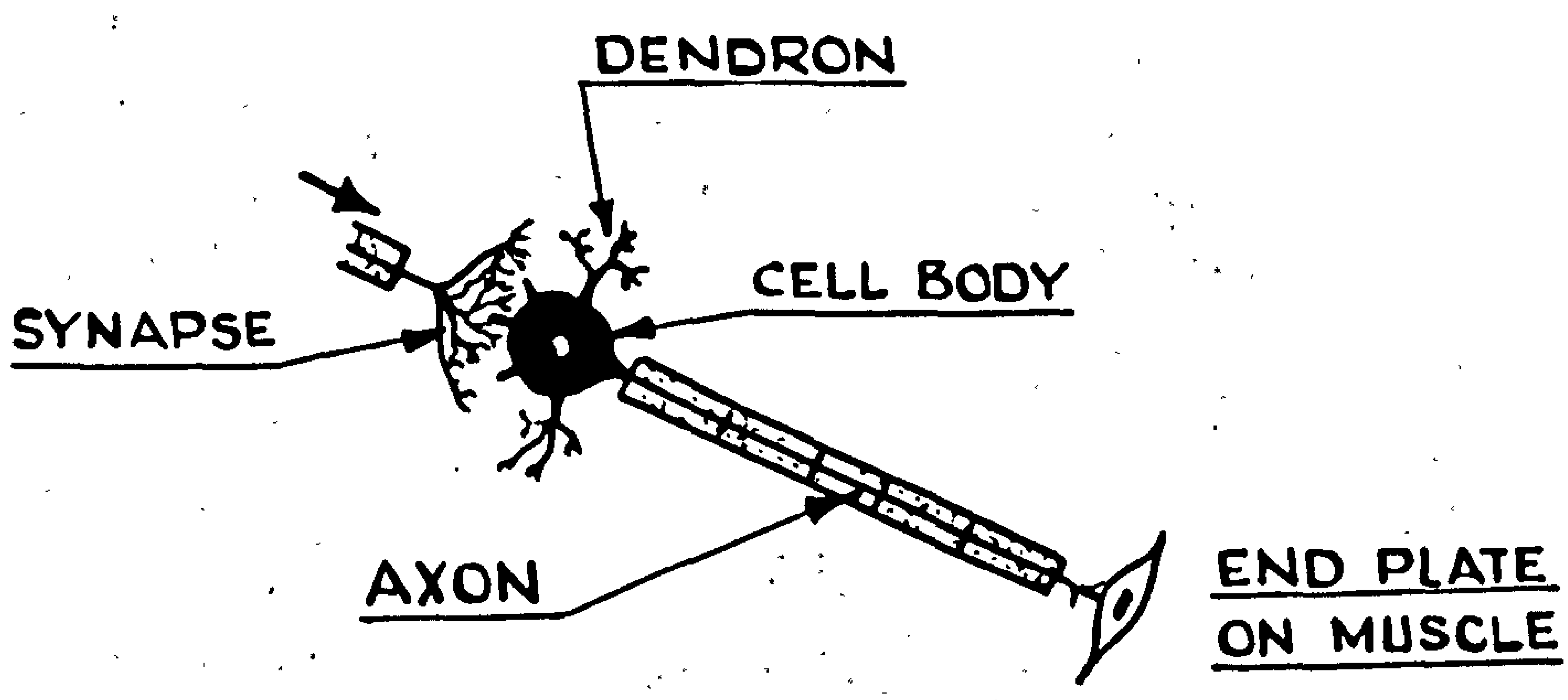


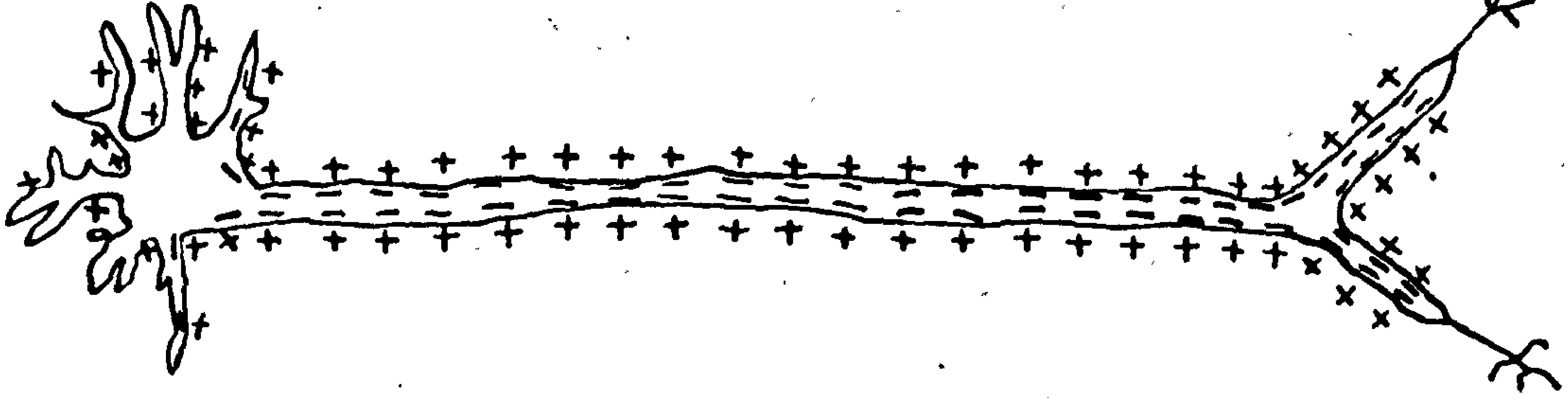
DIAGRAM OF A MOTOR NEURON

FIG. 24

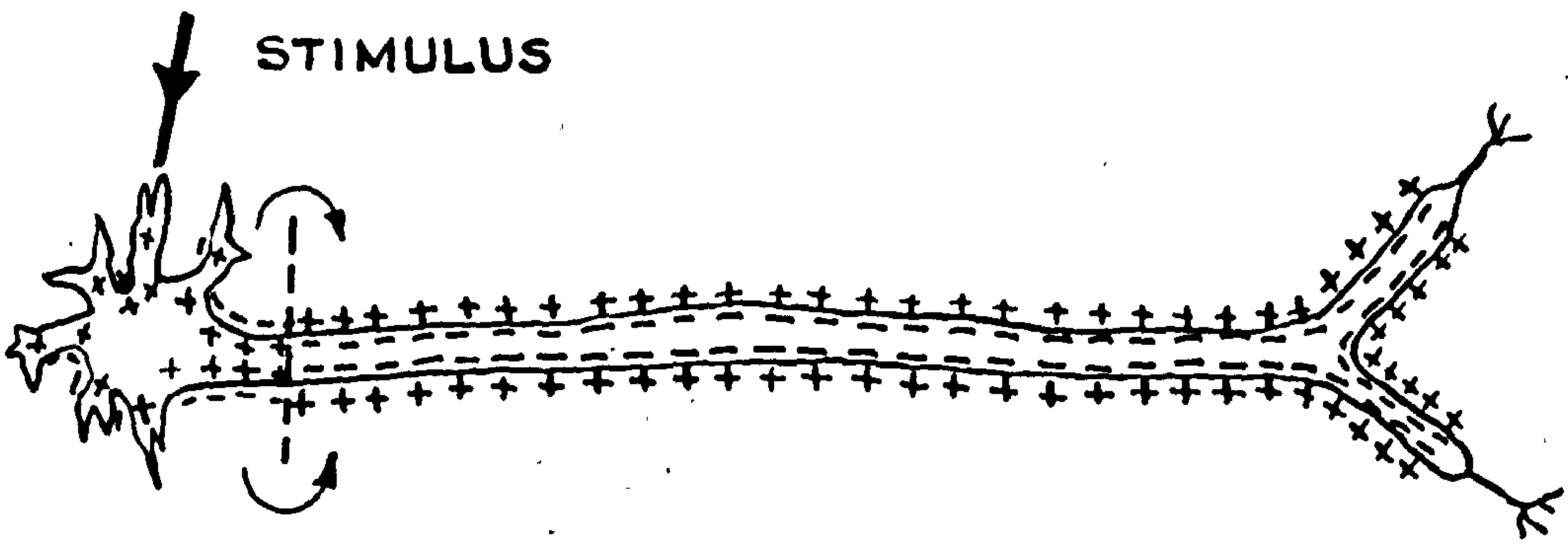
<u>Nervous System.</u>

The basic characteristics of nerves are generally described as excitability and conductivity and the conductivity must further be limited to unidirectional transmission. A nerve cell including all its parts is generally described as a neuron. The nerve cell body has numerous short branched connections known as dendrites conducting impulses toward the cell body and a long slender connection called the axon along which the nerve stimulus is transmitted. At its end the axon frequently divides into numerous branches as shown in Fig. 24. In the case of 'motor' neurons these branches are attached to muscle, one branch per muscle fibre. The dendrites receive impulses from the terminal branches of the axons of other nerves, the junctions being called synapses.

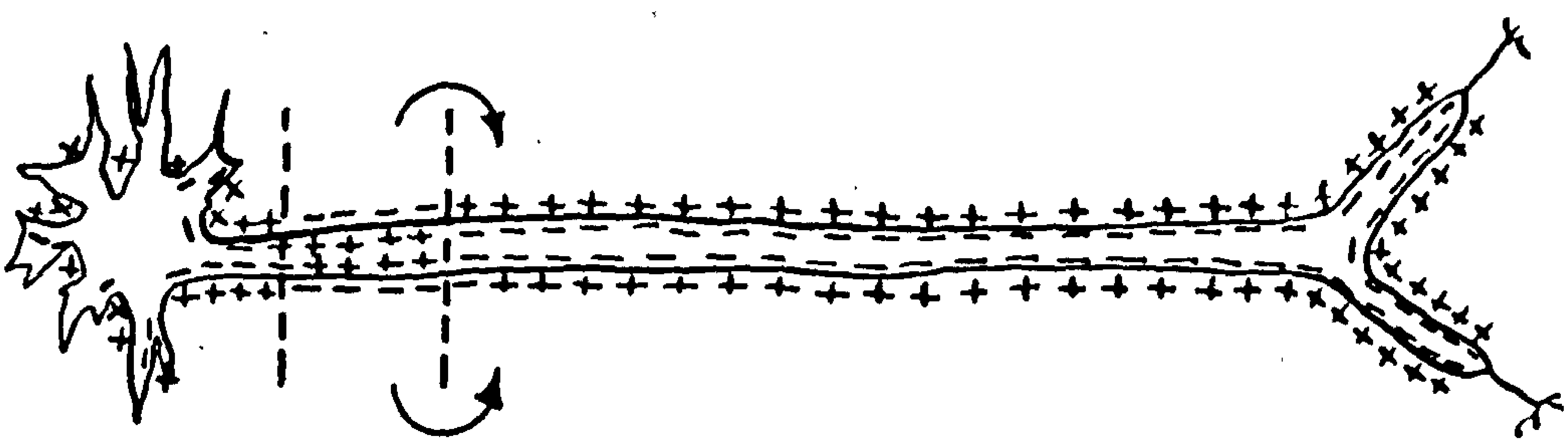
The transmission of an impulse along a nerve is not by conduction as known in metals or electrolytes. Controversy currently exists on the precise electro-chemical mechanism of the transmission, but it is generally agreed that an unstimulated nerve has standing potentials, positive externally and negative internally, as shown in Fig. 25. On stimulation, these potentials are reversed over the stimulated region and this system of reversed potentials travels down the nerve at a speed between 6 metres/sec



NON-STIMULATED NERVE FIBRE



NERVE DURING INSTANT OF STIMULUS



NERVE DURING PASSAGE OF IMPULSE ALONG AXON.

NEURAL TRANSMISSION OF IMPULSES

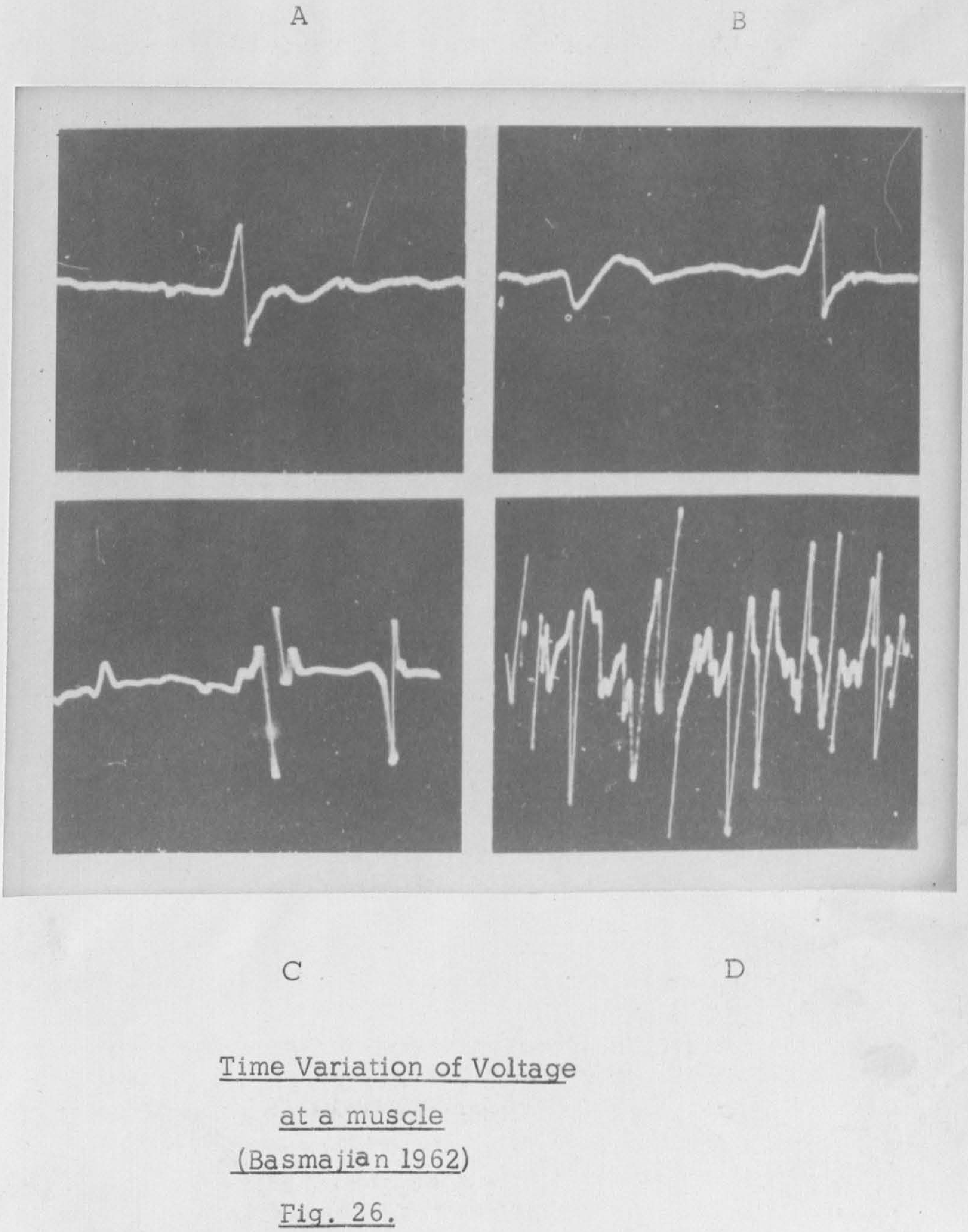
FIG. 25

and 120 metres/second depending on the diameter of the fibre. After the impulse passes a segment of the nerve fibre, the segment recovers to its original polarised state and is then ready to transmit a new impulse. If a second stimulus is applied before the recovery of the stimulated segment has occurred no impulse is transmitted. The order of magnitude of the recovering time may be about 1 m.sec. A nerve fibre will not respond unless the stimulus is of a certain minimum threshold value and a single threshold stimulus will start one impulse travelling along the nerve. Increase in stimulus strength will not affect the impulse magnitude or velocity, and control of the response of the motor unit is achieved by adjustment of the repetitive frequency of stimuli.

Each nerve branch is attached to many muscle fibres but these fibres are not necessarily in the same bundle, the fibres in each bundle are separately innervated by branches from several nerves. The signal transmitted by the nerve is fed centrally into the muscle fibre at the "motor end plate". One nerve and all the muscle fibres connected to it are collectively designated a motor unit. It is reported by Feinstein et al (1955) that in the human leg muscle there are 2,000 muscle fibres per motor unit. Motor units contract sharply upon the arrival of a nervous impulse, provided that the impulses do not repeat at a frequency greater than 50 per second. A strong muscular contraction requires the action of many motor units and in normal healthy muscle the necessary impulses transmitted by the many axons are arranged to be asynchronous so that the sum of all the separate contractions may be a smooth exertion of force as described later.

The voltage from a single impulse transmitted by a nerve to a muscle fibre varies with time as shown in Fig. 26 A. Conventionally a negative pulse is shown upwards. This single pulse occupies a time of 7 m. sec. and has an amplitude of 0.8 mV. Figures 26 B, C and D show the patterns recorded as firing occurs in increasing numbers of motor units. If electrodes are situated in the neighbourhood of the muscle they pick up these signals and the pattern recorded, after suitable amplification is referred to as an electromyogram. The occurrence of numbers of potential 'spikes' is evidence that the muscle is contracting or exerting force.

Various types of electrodes used to pick up myo-electric signals are described by investigators, Close (1963), Basmajian (1962) Joseph (1960) Bechtal (1957). Inserted electrodes are either of needle form or, if composed of flexible wires, are introduced into the muscle by a hypodermic needle. Inserted



Time Variation of Voltage
at a muscle
(Basmajian 1962)
Fig. 26.

electrodes are sometimes unipolar and require to be 'paired' with another electrode either surface or inserted. In a bipolar electrode two insulated leads are introduced simultaneously generally through the base of a hypodermic needle and lie in the same localised region of muscle. Inserted electrodes are essential in clinical use where velocity of impulse transmission and synchronisation of motor units are under investigation. It is reported however that in all cases the subject experiences discomfort and in some cases pain, when muscular contraction occurs while the electrodes are installed. Where large muscle movements are involved as in walking their use is particularly circumscribed.

Surface electrodes comprise pairs of discs, generally of silver, mounted some distance apart, on the skin, over the location of the muscle of interest. Conduction to these discs is generally improved by (i) removal of surface grease from the skin using a solvent: (ii) removal of dead skin by abrasion, and (iii) application of an electrolyte jelly to the skin/disc interface. Generally an additional grounding electrode is secured to a distal part of the region investigated and a lead to this is used as a screen for the leads of the working electrodes. The disadvantages of surface electrodes are reported to be as follows.

1. Signals from more than one muscle may be picked up.
2. Only surface muscles can be investigated.
3. The signals picked up are smaller, require more amplification and therefore pick up more interference.
4. The magnitude of the signals picked up depends on the thickness of the layer of subcutaneous fat and varies therefore with the test subject.

The advantages of surface electrodes are their ease of application and the extended area of muscle from which signals are picked up, allowing a more representative picture to be obtained of the gross muscle action.

The magnitude of the signal received depends on the type of electrode employed. In mild muscular activity an inserted electrode may pick up spikes of activity of 100 μV peak to peak where a surface electrode would pick up 40 μV. The signals cannot be described as having a frequency of occurrence since this implies regular repetition. The peaks may occur at rates of 40 - 50 per second. To record the peaks an instrument reproducing uniformly over a frequency band of 10 - 2,000 cycles per second is recommended. To drive a recording instrument considerable amplification is obviously required and the amplifiers must be designed to reject 'common mode' signals e.g. signals due to interference from other electro magnetic fields which will be common to both pick up leads. An extensive range of instruments is available with outputs either to cathode ray oscilloscopes whose trace may be photographed or to galvanometer recorders of the inking or photographic recording types. Generally the electrodes are connected to the amplifiers and these to the recorder by screened cables. Battye (1962) describes a radio telemetry system to transmit electromyograph signals from electrodes on a moving subject.

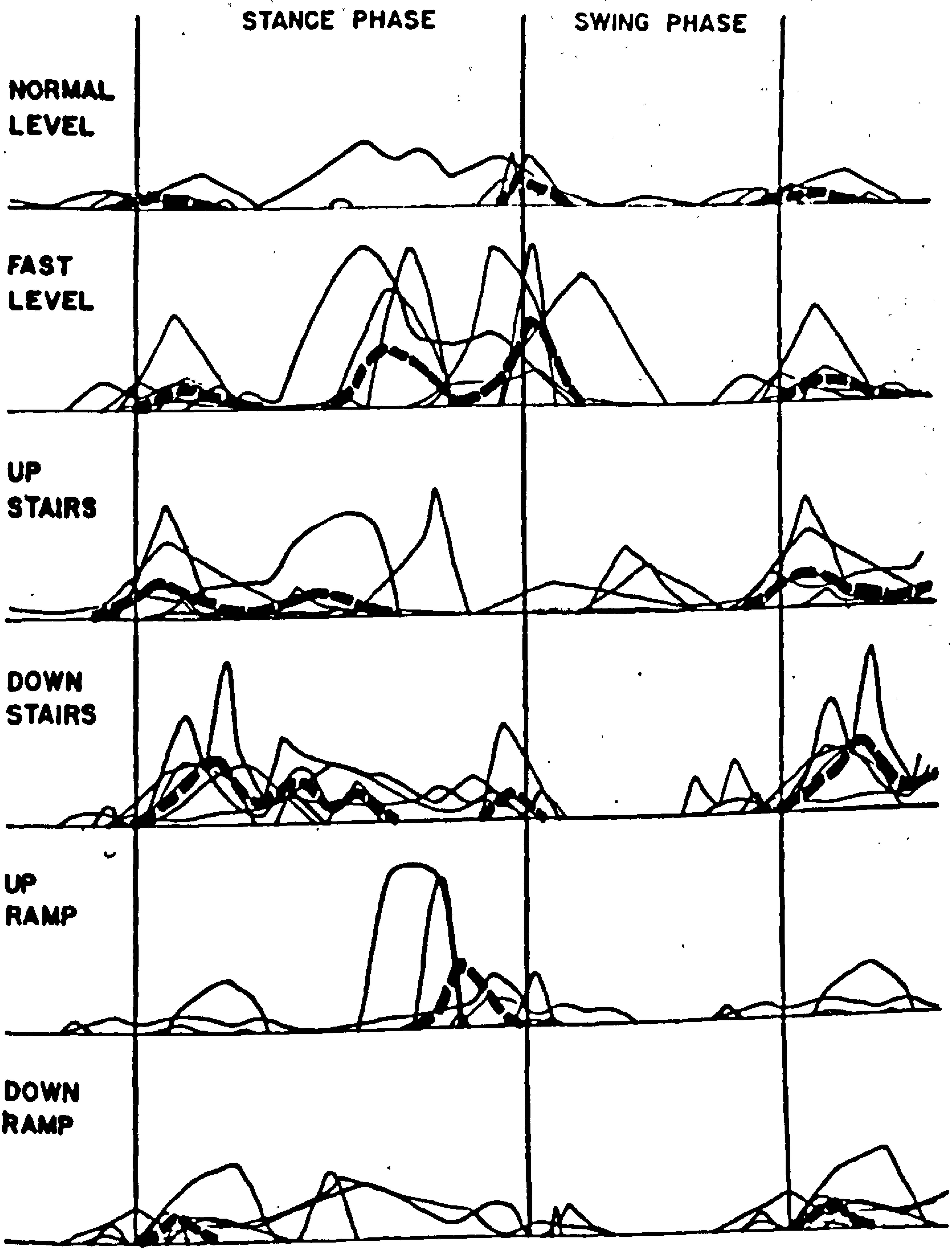
The electromyogram (EM G) has been used by many investigators to indicate activity and the periods of activity of muscle. Studies on the leg muscles involved in maintaining the upright human posture are reported by Joseph and Nightingale (1952), Joseph and Williams (1957), Joseph (1963), Basmajian (1962) and Close (1963). Studies of the phasic activity of the muscles of the leg in walking have been reported by Hirschberg and Nathanson (1952), University of California (1953), Hardy (1959), Close and Todd (1959), Houtz and Fischer (1960) Sorbie and Zalter (1965), Joseph and Battye (1966). The University of California publication reports the results of tests on six normal subjects (5 male, one female) in which electrodes were inserted into 28 muscles, situated between the ankle and the lumbar region.

After insertion of the electrodes galvanic stimulation was applied to them so that it could be ascertained that the electrodes were in the desired muscle. This was repeated at the conclusion of the tests. Records were taken for each muscle when the subject walked as follows:

1. At normal speed on the level.
2. Rapidly on the level.
3. Upstairs.
4. Downstairs.
5. Up a ramp.
6. Down a ramp.

The EMG signals were rectified and smoothed before recording to give continuous curves whose magnitude bore some relation to the muscle force. The correlated results of the tests on rectus femoris of six subjects are shown in Fig. 27. The report states that there is a time delay between the peak of electrical activity and the peak of force tension. The force peak lags by about 0.08 seconds. This may be due to the electrical characteristics of the rectification and smoothing circuit or to the physiological performance of the muscle.

The records were taken for one muscle at a time. It was stated that mutual interference made it impossible to investigate



INTEGRATED E.M.G. CURVES FOR RECTUS FEMORIS
(UNIVERSITY OF CALIFORNIA 1953)

FIG. 27

more than one muscle at a time. No such difficulty was reported by Sorbie and Zalter. The given figures imply that each subject had to perform a total of 168 test runs, disregarding those which had to be rejected. It appears likely that there will be significant variations in gait between test runs.

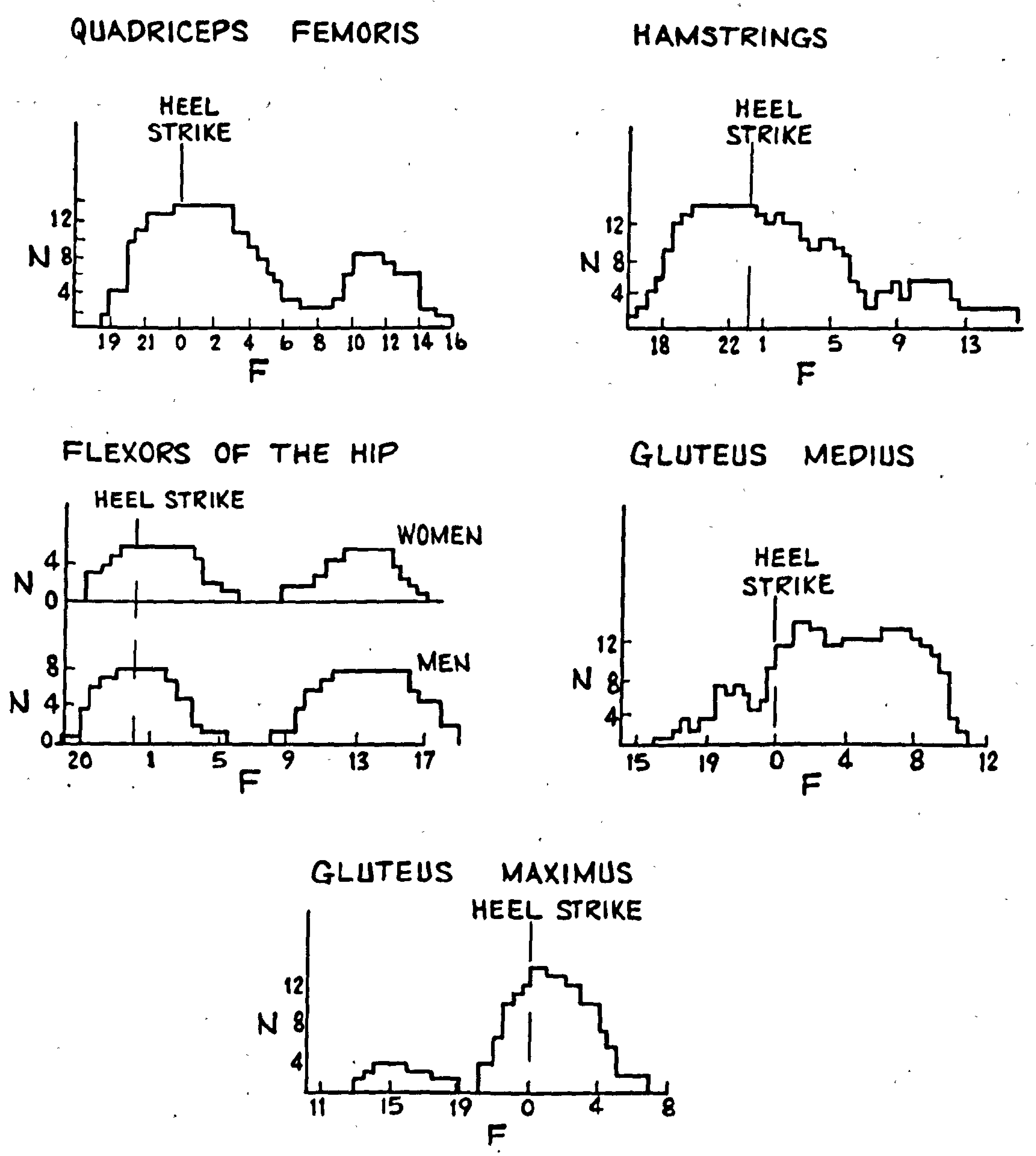
Joseph and Battye reported the results of walking tests on 14 normal subjects, 8 male and six female, in the age range 14 - 48 years. Surface electrodes were used and the signals from one muscle at a time were transmitted to the recorder by radio telemetry. A single photograph was taken during each test, and the instant of firing of the flashgun for this was recorded together with the EMG. To analyse the phasing of muscle action relative to heelstrike and toe-off this single photograph was compared with a series of 23 cine photographs of the gait of another test subject, shown in Fig. 28. It appears that this sytem will be unreliable where test subjects display major gait differences. Results are reported for electromyograms taken from eight sites in the leg and hip and the results are analysed statistically. Typical results are shown in Fig. 29.

In the analysis of body dynamics any method of obtaining the instantaneous value of the force developed by a muscle would be advantageous. If a muscle is connected to a non-deflecting

Cine Record of Phases of Walking

(Joseph and Battye (1966)

Fig. 28.



N = No. of subjects showing activity
F = Frame number

HISTOGRAMS OF MYO-ELECTRIC ACTIVITY IN NORMAL WALKING (BATTYE AND JOSEPH 1966)

FIG. 29

measuring instrument, and a single impulse is transmitted to it by a motor nerve the force time behaviour of the muscle is as shown in Fig. 30. The effect on muscle force of the time interval between two successive stimuli is shown in A, B and C of Fig. 31. The curves B-E of Fig. 32 show respectively the effect of increasing the frequency of a succession of impulses to a muscle. When the frequency is sufficiently rapid the smooth curve of force E is attained: this situation is called tetanus. It is shown also in Fig. 33 that a smooth muscular pull can be developed by two muscle fibres, each of which is in a state of partial tetanus, provided the nerve impulses are asynchronous. The individual muscle fibres display an all-or-none behaviour in that if the stimulus is below a certain threshold magnitude no contraction will occur and if contraction does occur it will be a maximal effort regardless of the amount by which the stimulus exceeds the threshold. Stronger stimuli to the nerve will increase muscle force, however, since more muscle fibres will experience supra-threshold stimuli.

The foregoing accounts stem from typical text-books on Physiology (Samson, Wright (1961), Fulton (1955) etc.). Whitney (1958) contends that the concept of tetanus is never realised in living muscle and that all muscular force action has an oscillating

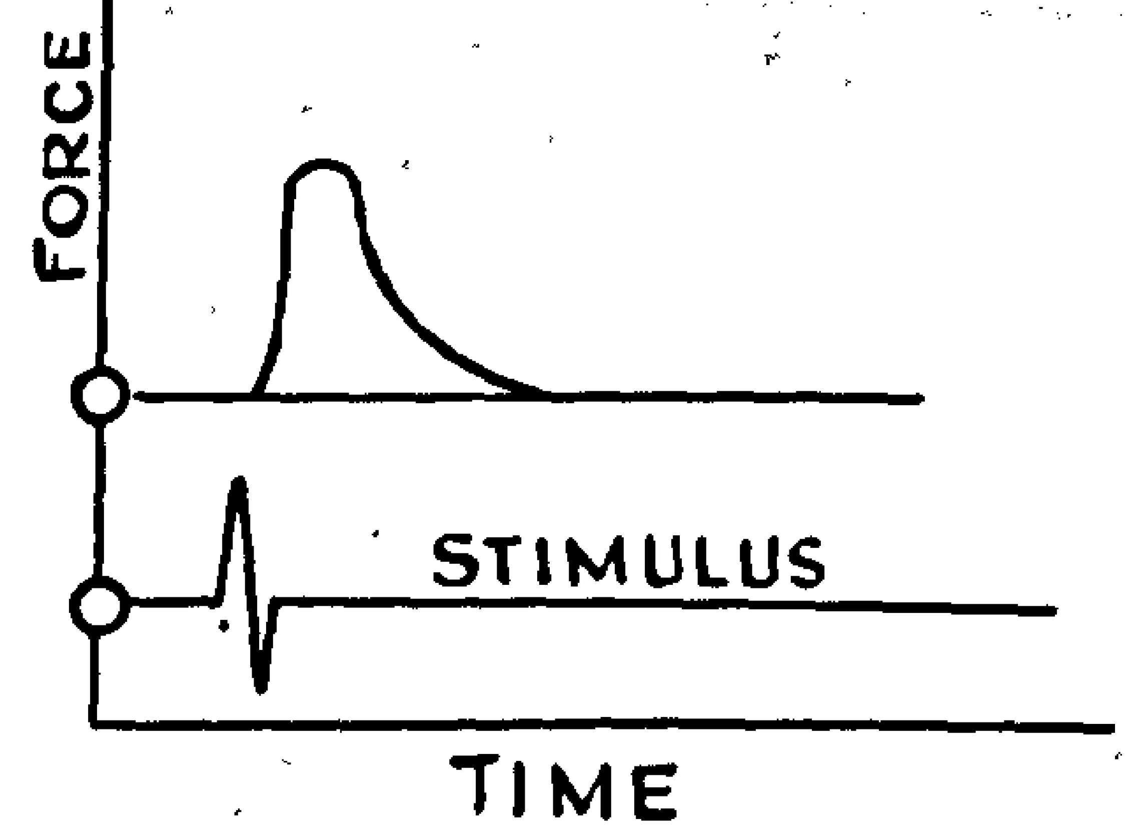


FIG. 30

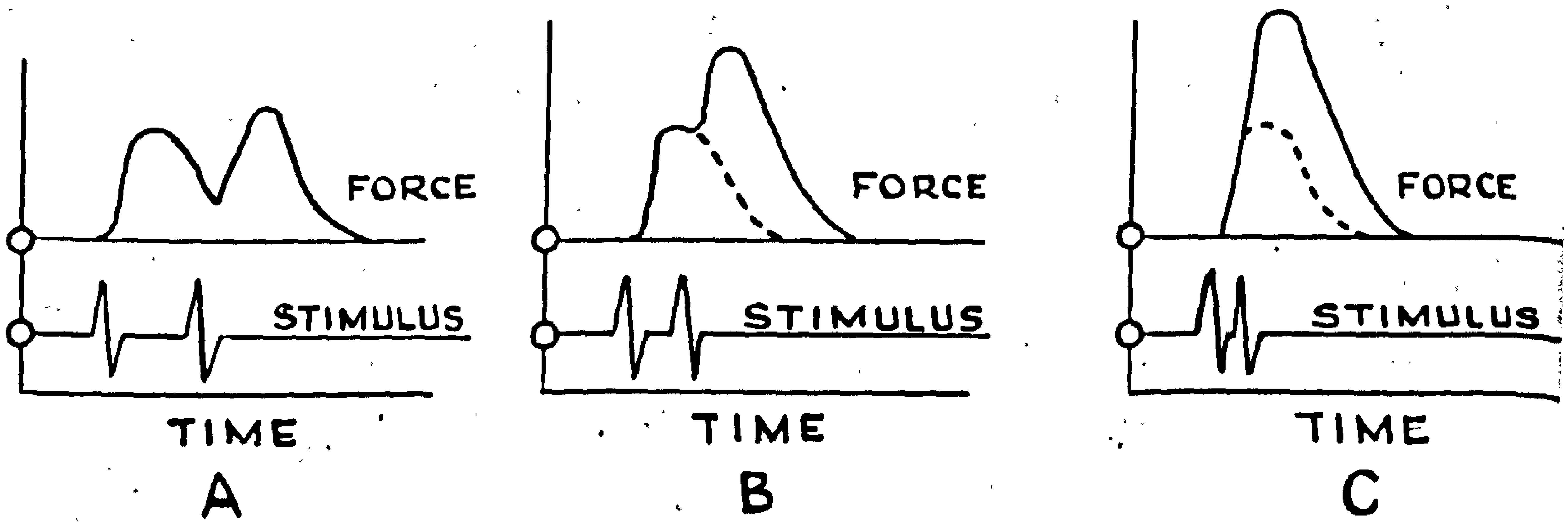


FIG. 31

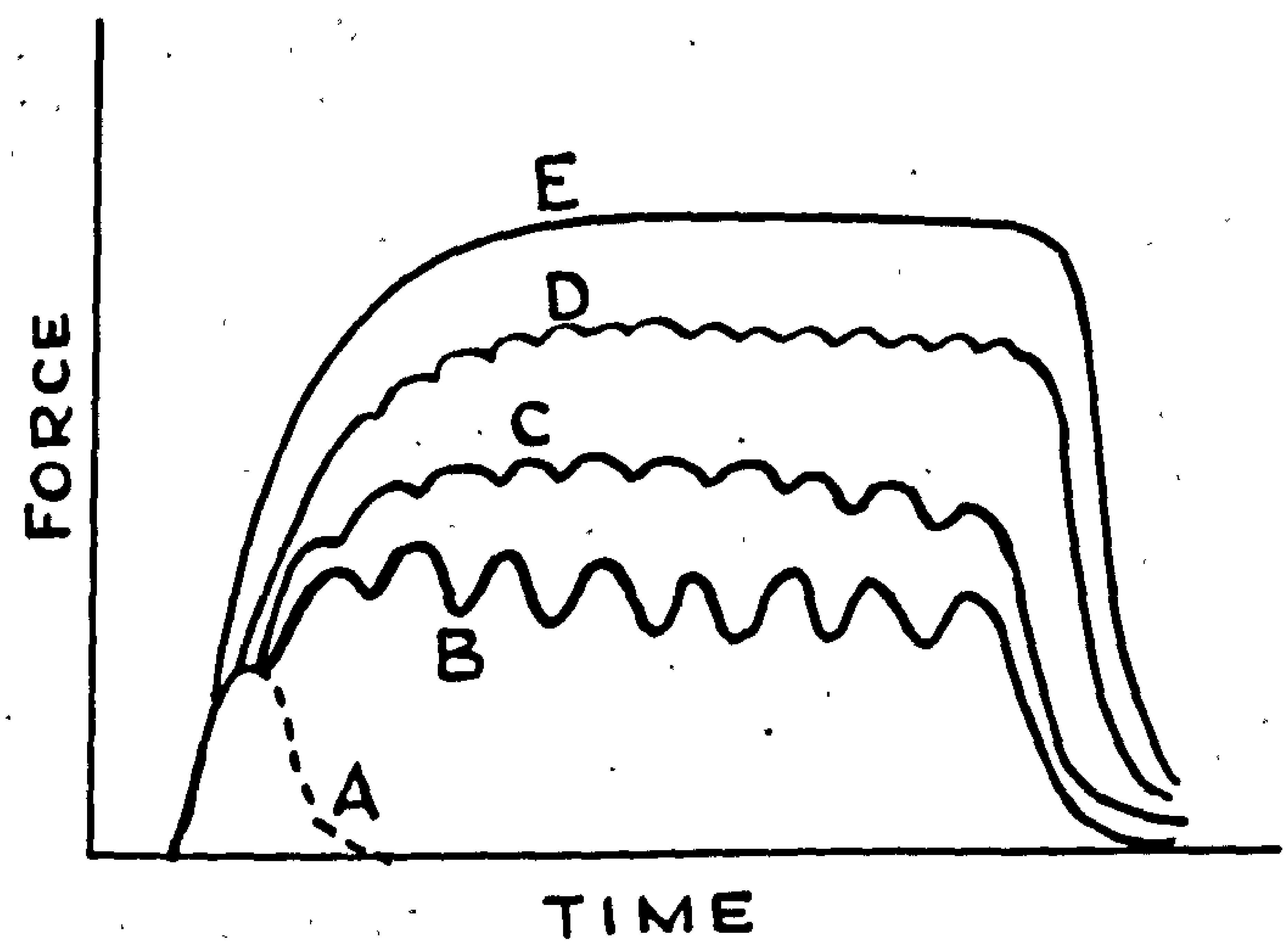


FIG. 32

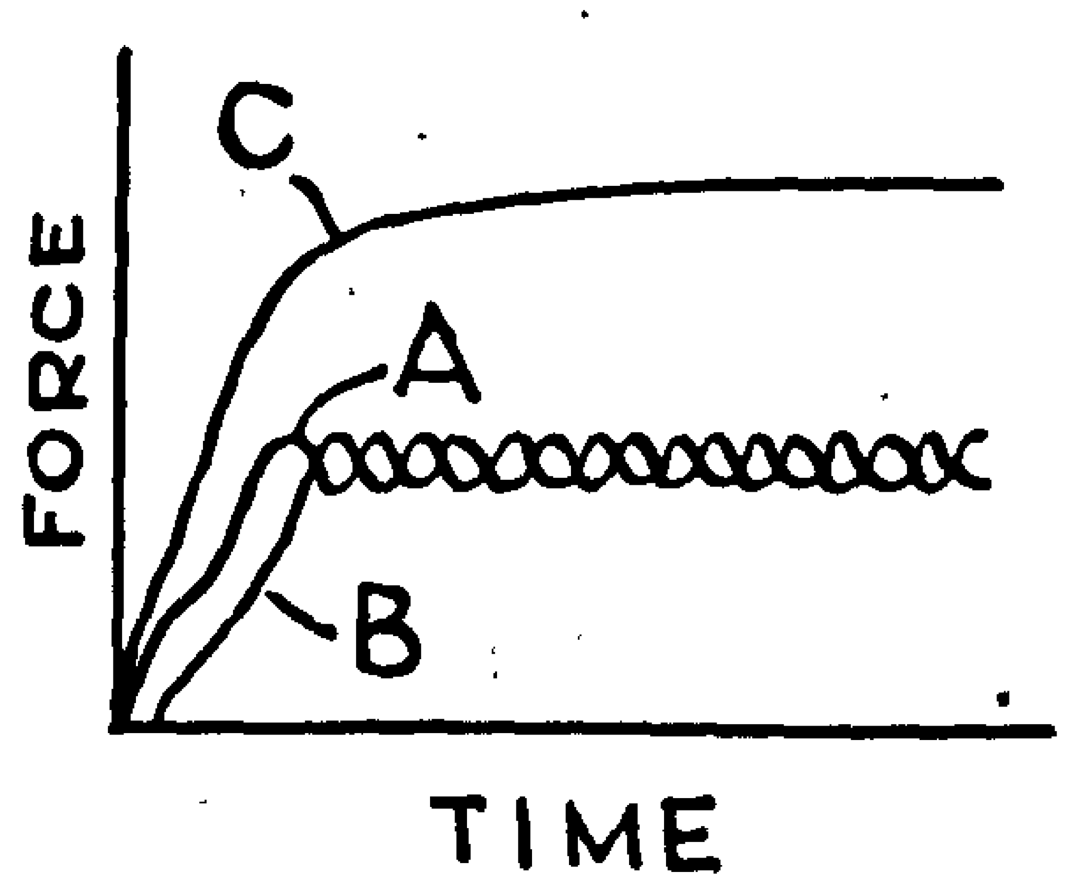


FIG. 33

DIAGRAMS ARE TAKEN FROM SAMSON WRIGHT (1961) component of from 10 to 35 cycles per second superimposed on a mean value and that this component causes cyclical tremors particularly during great exertion. This appears likely since the maximum frequency of occurrence of impulses is reported by electro-myographic investigations to be about 50 per second whereas tetanus is reported to require frequencies of over 100 per second in the human.

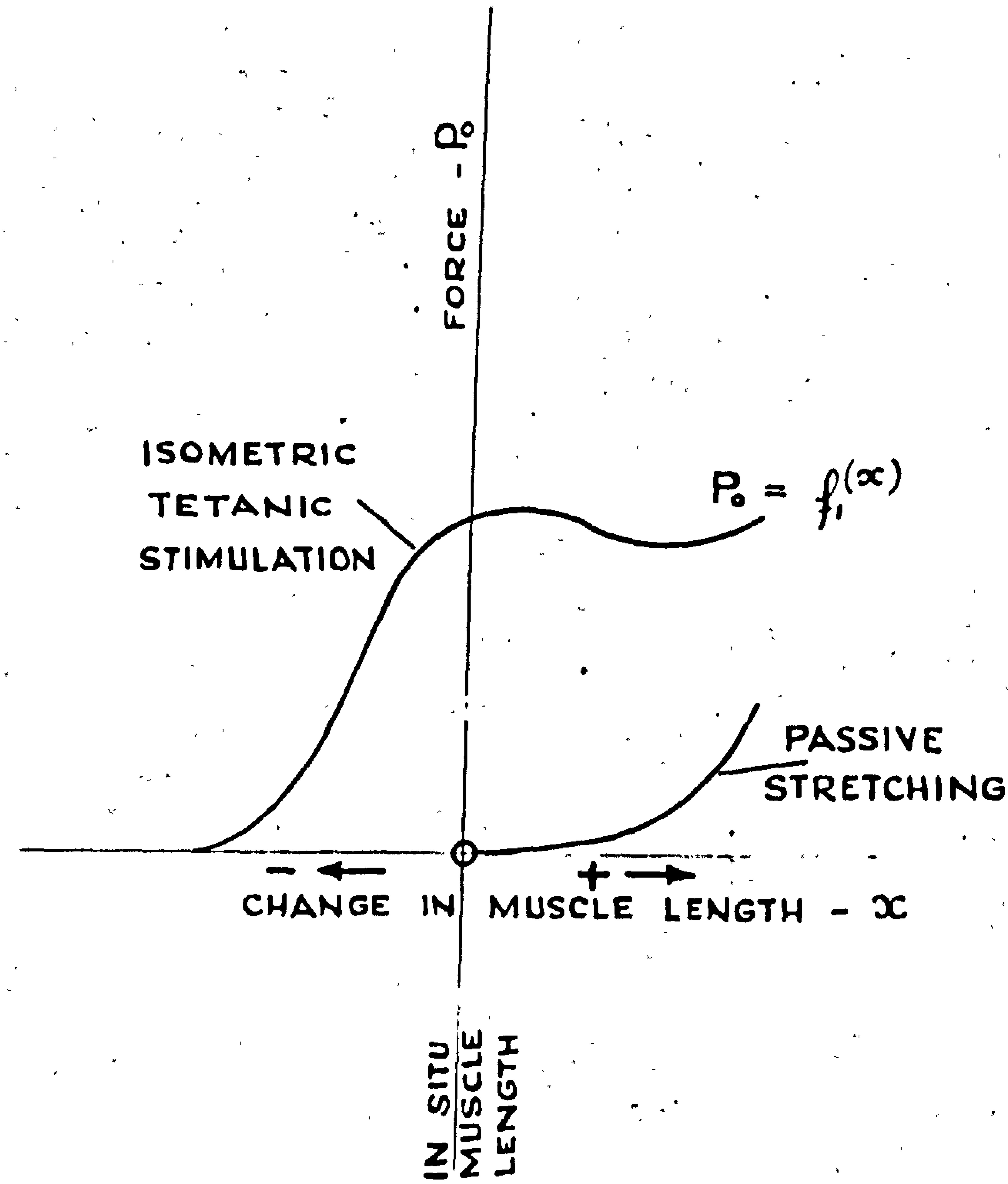
All the above relationships depend on the muscle being tested isometrically. If the length of the muscle changes during stimulation, the force P, and the velocity of shortening V are related by the following equation due to Hill (1938)

$$(P + a)(V + b) = \text{constant} = (P_o + a)b. \quad \text{---} \ (9)$$

where P_o is the maximum force developed in isometric tension and a and b are constants. According to Wilkie (1956) this equation is valid only in a limited region of stretch of the muscle since the maximum force developed in isometric tetanic stimulation corresponds to the amount of stretch imposed as shown in Fig. 34. Wilkie suggests that the equation should be modified to read:-

$$V = \frac{dx}{dt} = \left[(f_1(x) - P\right].b/(P + a) \quad \text{----} \ (10)$$

where $f_1(x)$ represents the P - x curve of Fig. 34.



LENGTH - TENSION CURVES FOR MUSCLE

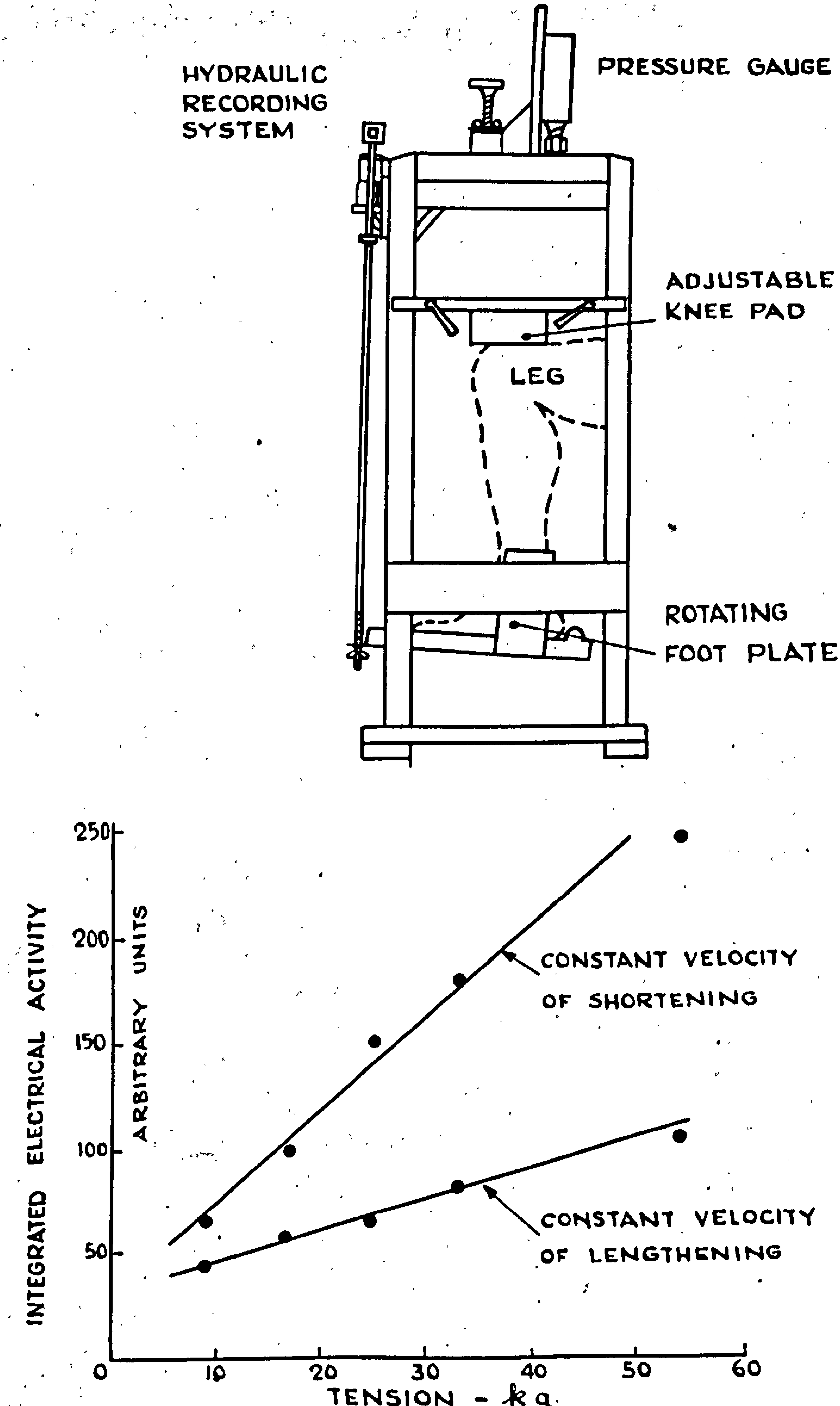
FIG. 34

(FROM WILKIE (1956))

Unfortunately this well-substantiated work on the force/velocity equations for muscle was derived from tests on muscle in vitro to which full tetanic stimulation was applied. In the living body all fibres are not stimulated at any instant and the fibres which are stimulated are not in full tetanus. The relationship between stimulus as manifested by electromyogram and muscle force has therefore been the subject only of experimental investigation. Dern, Levene and Blair (1947) performed tests to investigate the force velocity relationship for human arm muscles using an electromyograph to record the occurrence of stimulation in the muscles. In these tests, however, the subject was in all cases instructed to perform maximum voluntary movements and the results are not therefore applicable to sub-maximal contractions.

Inman (1947) in his study of the abductor muscles of the hip obtained a curvilinear relationship between E, the "voltage integrated count per second" signal from skin and needle electrodes applied to the abductors, and M, the adduction moment applied to the leg. It is apparent however that no attempt was made to maintain the length of the muscles at a constant value. Inman, Ralston, Saunders, Feinstein and Wright (1951), working with the University of California group, performed tests to relate electromyogram output and muscle tension. The EMG output was rectified and smoothed: this process was described as "integration". The results show that, regardless of the type of electrode employed, curves of force and integrated EMG are similar provided isometric conditions are maintained but the EMG/force relationship is not plotted. A discrete time interval of 0.08 ± 0.02 sec. elapses between the occurrence of a maximum in the EMG record and the corresponding maximum in the force record. This cannot be ascribed exclusively to the smoothing circuit since the charge time for it is only 0.01 sec.

A series of controlled experiments to correlate integrated myo-electric potential signals to muscle force are reported by Lippold (1952) and Bigland and Lippold (1953). The experimental layout of Lippold is shown in Fig. 35 and needs no further explanation. Surface electrodes were applied over the calf muscle gastrocnemius. The EMG records were integrated by planimeter and all tests on thirty subjects indicated a linear relationship between integrated activity and force sustained. The test action is, however, produced also by the Soleus muscle and no information is supplied on whether Soleus was active during all or any of the tests. There is



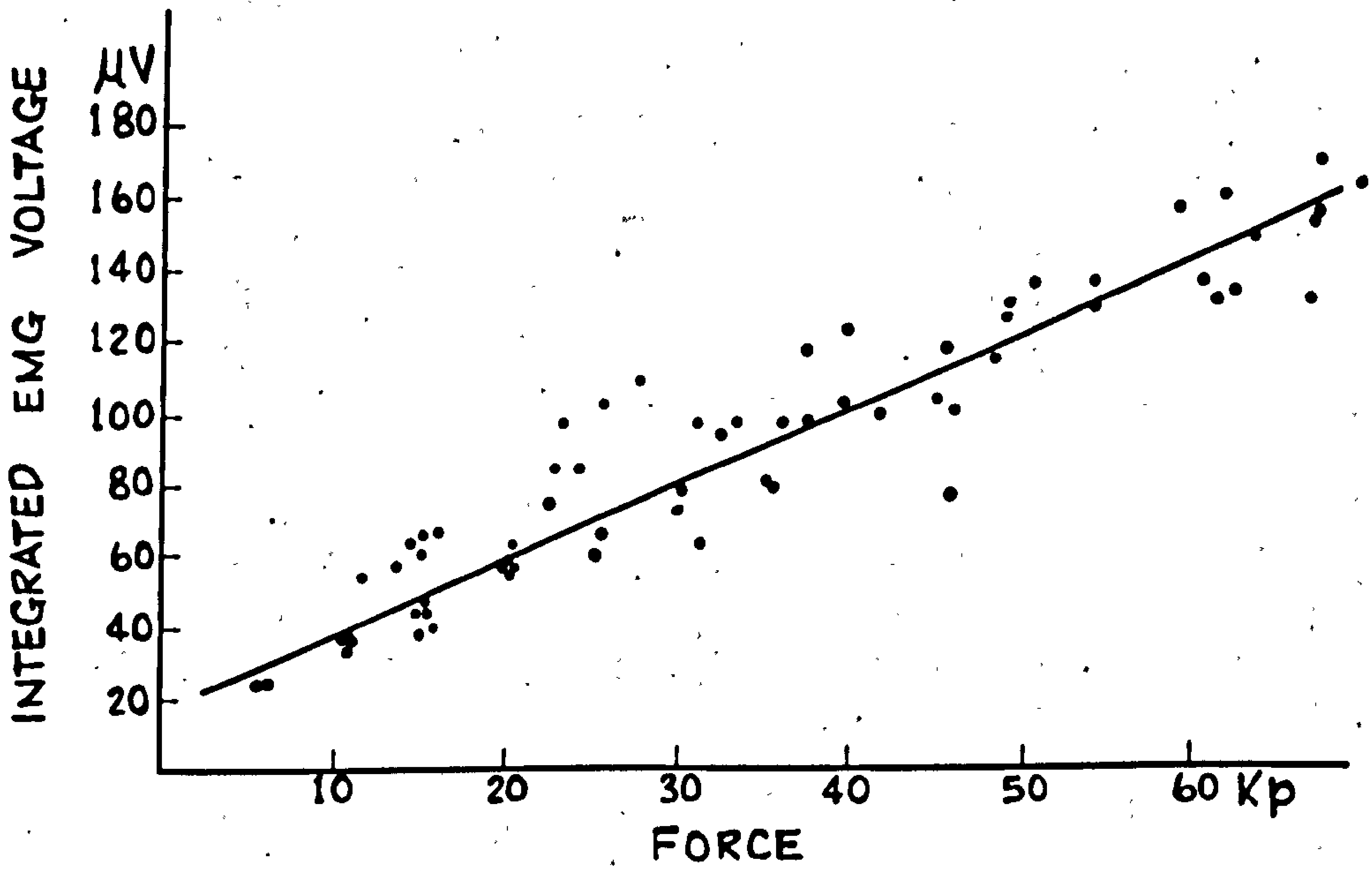
APPARATUS AND RESULTS FOR TENSION / EMG. TESTS. (LIPPOLD 1952).

FIG. 35

evidence, Travil and Basmajian (1961), De Sousa et al (1958), that synergistic muscles are successively recruited as the intensity of loading increases. If this obtained here, the linear relationship no longer holds between the muscle force of Gastrocnemius and integrated EMG.

The previous apparatus was modified to allow dynamic tests, either at constant velocity with varying degrees of force or at constant force with tests performed at different velocities. It was found that at any velocity the force/electrical activity graphs remained linear but that the slope of the graph varied with velocity. It was suggested but not established that the slope varied linearly with velocity. It was found that corresponding results were obtained whether surface or implanted electrodes were used.

Tests to relate integrated EMG values to force in the back muscles are reported by Asmussen, Poulsen and Rasmussen (1965). Readings were taken from electrodes on the erector spinae muscles as the subject, with back straight and inclined at 45° to the vertical, attempted to lift the shoulders against the restraint of a strain gauge dynamometer. The results obtained on one subject on seven different occasions within one month are shown in Fig. 36. The scatter in results is probably due



TESTS ON ERECTOR SPINAE MUSCLE.
ASMUSSEN ET AL (1965)

FIG. 36

to small differences in electrode placement, alteration in skin resistance, and varying amonnts of muscular contraction from test to test.

Isidor and Nicolo (1966) have made a comparison of various electrical circuits used to "integrate" the myo-electric signals to obtain a function corresponding to muscle forces. Six mathematical functions and the corresponding electrical circuits are described and the results from a test on the arm muscles of one subject are described. No details of the test procedure or of limitation of arm movement are quoted.

Three analytical papers are on file from Nubar at New York University. These attempt to fit mathematical models to the dynamic behaviour of biological material. In Nubar and Contini (1961) an attempt is made to relate the instantaneous or static position adopted by an individual to a function referred to as muscular effort.

Muscular effort, E, is defined as:-

$$E = \sum C_m \ M^2_m \ \Delta t \ + \ A_o \qquad \text{----- (11)}$$

where C_m is a numerical factor,

M_m is the moment transmitted by a joint,

Δt a short interval of time,

and A_o an initial constant.

This is justified by saying that for one muscle its effort may be assessed by the product of muscular force and time of operation. Moment at a joint, M, is however muscle force multiplied by lever arm and muscular effort can therefore be taken to be $CM\Delta t$. Since the moments at joints may be positive or negative and all muscular efforts are defined as additive, this term is arbitrarily modified to $CM^2 \Delta t$.

A minimal principle is then postulated as follows:-

"A mentally normal individual will, in all likelihood, move (or adjust his posture) in such a way as to reduce his total muscular effort to a minimum, consistent with the constraints."

Equations of force equilibrium are then set up relating the angular displacements and acceleration for a body part with the joint moments. It is stated that if the minimising procedure is applied to the muscle effort equation a sufficient number of times, then a solution can be obtained for the muscle forces. Only one example is quoted, however, for a static configuration of a five segment plane figure representing a human body standing on one leg with the supporting foot 20 ins. behind the free foot. The writers suggest that tests should be performed on a range of subjects to see if they adopt a configuration corresponding to the solution obtained using arbitrary coefficients.

It appears that this work is not readily applicable to the analysis of gait since it is not established that $CM^2 \Delta t$ is a measure of "muscle effort". In addition the paper notes that due to the physical structure of the body, other limiting factors will exist, expressing the greater moment carrying capacity of one joint than another and the conscious or unconscious "favouring" expressed by the individual in response to pain or discomfort stimuli from weaker joints. It is suggested also by the present author that the analysis is not sufficiently basic since it neglects the possibility of antagonistic muscle activity and the division of load bearing by synergistic muscles.

The force deformation relationships in muscle are analysed in Nubar (1962). Muscle is assumed to consist of an indefinite number of single fibres which are postulated to exhibit a cubic stress/strain relationship defined by three elastic moduli and three generalised 'Poisson' ratios. The change in fibre thickness is expressed as a cubic function of the longitudinal strain by four coefficients dependent on the stimulation. It is assumed that the elongation of each fibre is proportional to the square of its distance from the centre line of the muscle. No justification is made for this basic assumption. Arbitrary values are given to nine of the ten assumed coefficients and the shape of the length/tension curve for stimulated muscle is compared with a corresponding curve obtained by Ramsey (1947) from tests on isolated frog muscle fibres in vitro. In view of the number of coefficients to which values are to be assigned, it would appear that it would be possible to fit quite widely varying curves by the expressions obtained.

The thermodynamics of muscular force transmission are considered by Nubar (1963). For a series of muscles developing force F, and experiencing a length change of dl, the heat liberated in time dt is expressed as

$$\text{(Heat Output of Muscles)} = \sum(mFdt - ndl) \quad \text{----} \quad (12)$$

where m is a proportionality factor of dimension velocity and n is a proportionality factor of dimensions force m and n are referred to as heat coefficients of the muscles.

The first term in this equation is derived from Hill (1960) and the second from an analogy of Nubar's to an electromagnet. The work equation is expressed in the following manner:-

(Heat Input to Segment) + (Decrease in chemical energy of system)

= (H + C)dt

= Heat Output of Muscles + Increase in K.E. of system + work done by system.

In the case of the human forearm rotating about a fixed elbow joint A as shown in Fig. 37:-

The kinetic energy of the segment $= \frac{1}{2} I_A \dot{\theta}^2$ or the change in K.E. in time dt $= I_A \dot{\theta} \ddot{\theta}$ dt.

Work done by the system in time dt $= Mg\rho L \dot{\theta} \cos\theta \; dt + PL \dot{\theta} \cos\theta \; dt$

Substituting and dividing by dt

$$H + C = \sum(mF - n\frac{dl}{dt}) + L\dot{\theta}\cos\theta\,(Mg\rho + P) + I_A\dot{\theta}\ddot{\theta} - \sum Fh\dot{\theta} \qquad \text{----- (13)}$$

The equilibrium equation for moments about A is:-

$$\sum Fh - Mg\rho L\cos\theta - PL\cos\theta - I_A\ddot{\theta} = 0 \qquad \text{---- (14)}$$

It follows that $H + C = \sum(mF - nh\dot{\theta}\cos\theta) + \sum Fh\dot{\theta}$

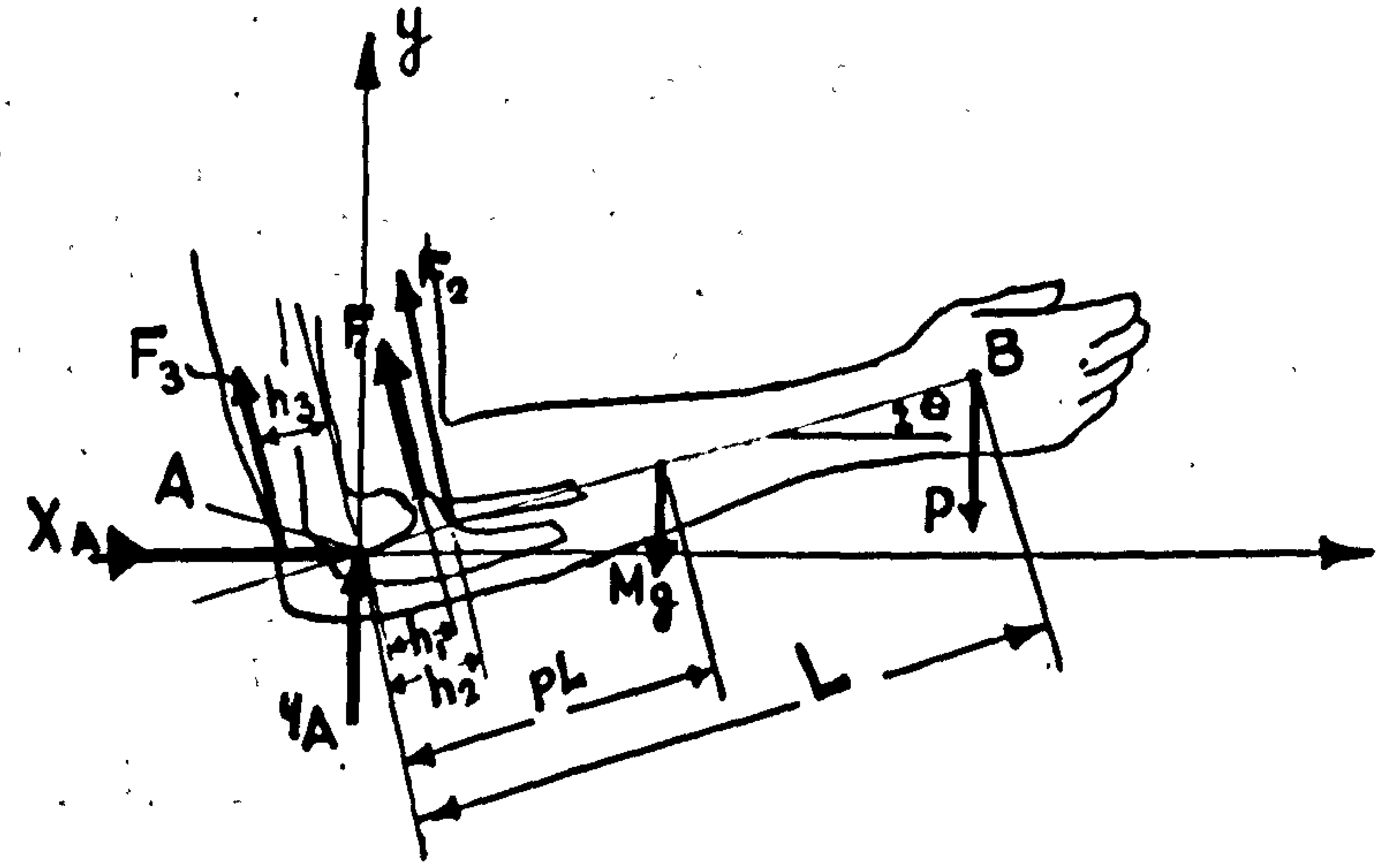
$$= m_1F_1 + m_2F_2 + m_3F_3 + (n_1h_1 + n_2h_2 - n_3h_3)\dot{\theta}$$
$$+ (F_1h_1 + F_2h_2 + F_3h_3)\dot{\theta} \qquad \text{------ (15)}$$

Nubar states that it is hoped that the measurement of the metabolic rate (H + C) is not beyond present experimental skill, and suggests that work is being done which might lead to this possibility.

If only one muscle, designated 1 is active over a given range of activity,

$$H + C = m_1F_1 + n_1h_1\dot{\theta} + F_1h_1\dot{\theta}$$

From determination of H and C at two different values of θ the coefficients m_1 and n_1 could be found. This procedure



FORCE ACTIONS IN THIS FOREARM

NUBAR (1963)

FIG. 37

might possibly then be repeated successively for situations where only muscles 2 and 3 are active and the remaining coefficients m_2n_2 m_3n_3 could be obtained. Then in a general situation where all muscles might be active the forces in them might be obtained from the statics equation (14) and by minimising the total energy input to thesystem, i.e.

$$\frac{\partial}{\partial F_1}(H + C) = \frac{\partial}{\partial F_2}(H + C) = \frac{\partial}{\partial F_3}(H + C) = 0 \text{ ---- } (16)$$

This procedure would have the additional stipulation that F_1 F_2 and F_3 were all non-negative.

It will be seen that the proposed procedure is not yet possible since the metabolic input (H + C) cannot be determined and also since it presumes that it will be possible to obtain calibration conditions in which each muscle group in turn is acting alone attwo different velocities of stretch.

EXPERIMENTAL WORK.

Design of Equipment.

a) Force Plate.

In normal activity the major loads on the lower limb arise due to the forces exerted on the foot by the ground. It was necessary in this investigation to measure the six component quantities necessary to define the magnitude direction and position in space of the ground to foot force actions. The force plate of Cunningham and Brown (1952) comprised a top plate connected to its base by four tubular columns, with end flanges bolted to the plates, so that each column could transmit three component force and three component moment actions. Since the experiment was to measure 6 quantities only, the construction is seen to be statically indeterminate to the 18th degree. It was decided to investigate the possibility of designing a statically determinate force plate. It is essential that the deflection of the test surface be so small as to be unnoticed by the test subject. It was therefore decided to support and restrain the force plate by spring steel strips loaded in tension and to measure forces by electrical resistance strain gauges mounted on these members.

The limits on the design were placed by:-

1. Allowable stress in the steel strips.

2. The necessity for sensititivy in force measurements requiring large strains in the strips.

3. The desirability that the calibrations for each measured quantity be independent of the values of other load components.

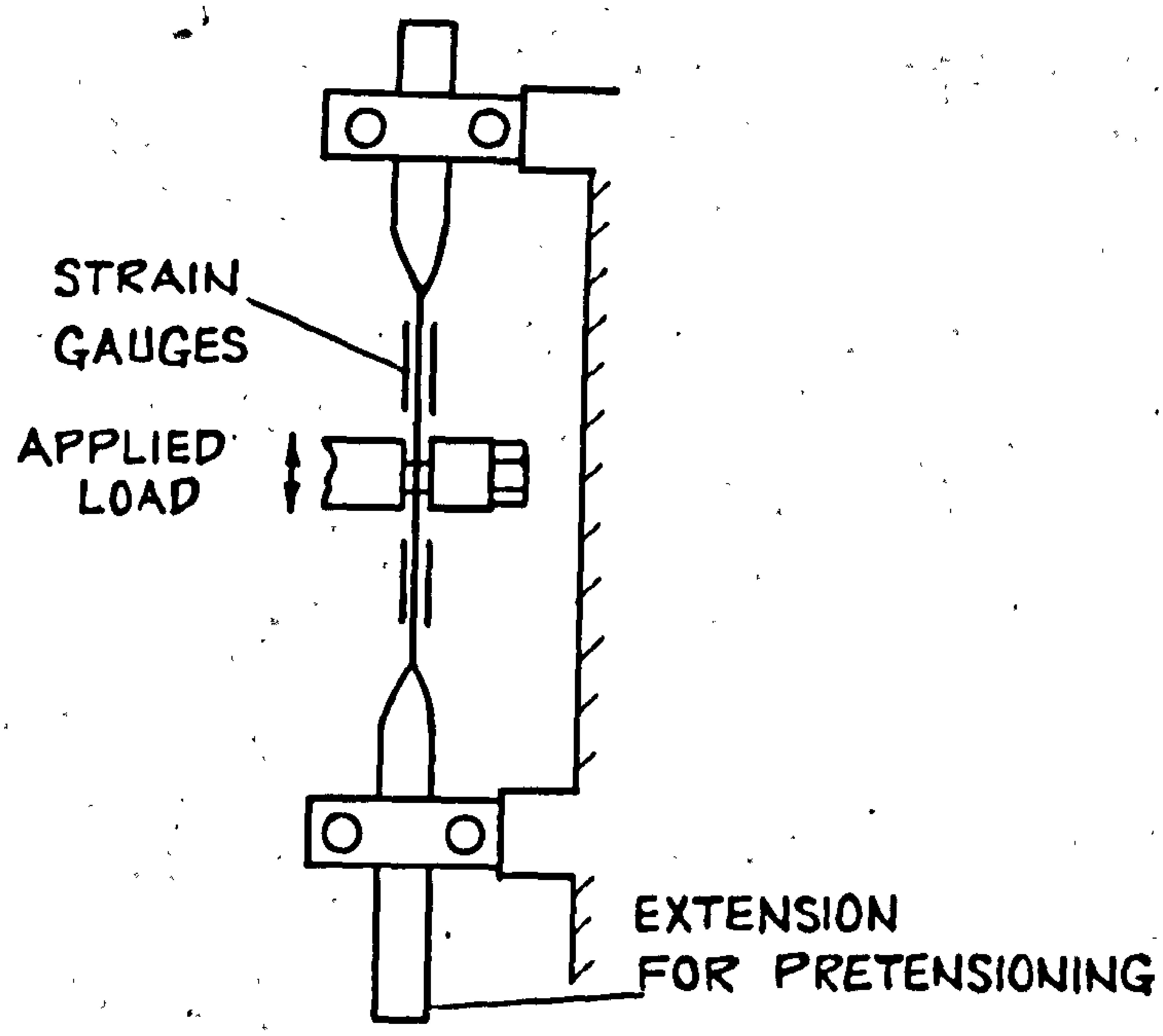
The dynamometer elements were pretensioned in the supporting frame and the measuring plate was attached to the centre of each element. Initially the strips were twisted as shown in Fig. 38 so that their effective bending stiffness in both transverse directions was low. This procedure was later discontinued since the greater length of strip necessary was allowing lower natural frequencies of vibration and the twist in the strip was causing local failure due to stress concentration. It was found that satisfactory calibrations could be obtained for five of the six channels when straight strips were employed. An analysis of the strain gauge network is given in Appendix III.

The force plate was restrained by six strip dynamometers arranged as shown in Fig. 39. If the forces measured by the six dynamometers are P1 - P6 as shown the general three dimensional force system acting on the plate is defined by:-

$$P_x = P_2 + P_3$$

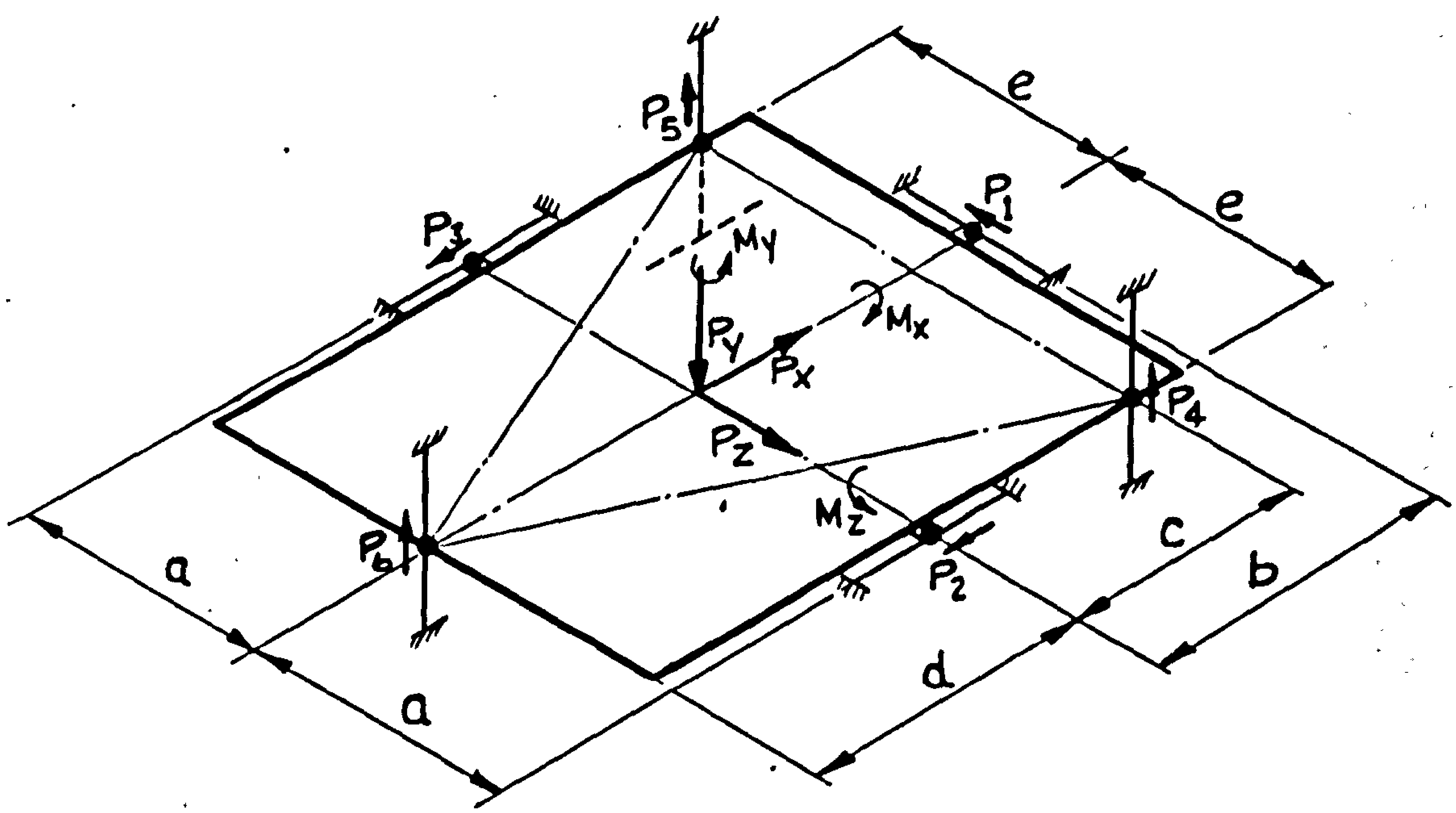
$$P_y = P_4 + P_5 + P_6$$

$$P_z = P_1$$



TWISTED STRIP DYNAMOMETER FOR SINGLE FORCE MEASUREMENT

FIG. 38



DIAGRAMMATIC ARRANGEMENT OF FORCE PLATE

FIG. 39

$$M_x = (P_4 - P_5)e$$

$$M_y = (P_2 - P_3)a - P_1 b$$

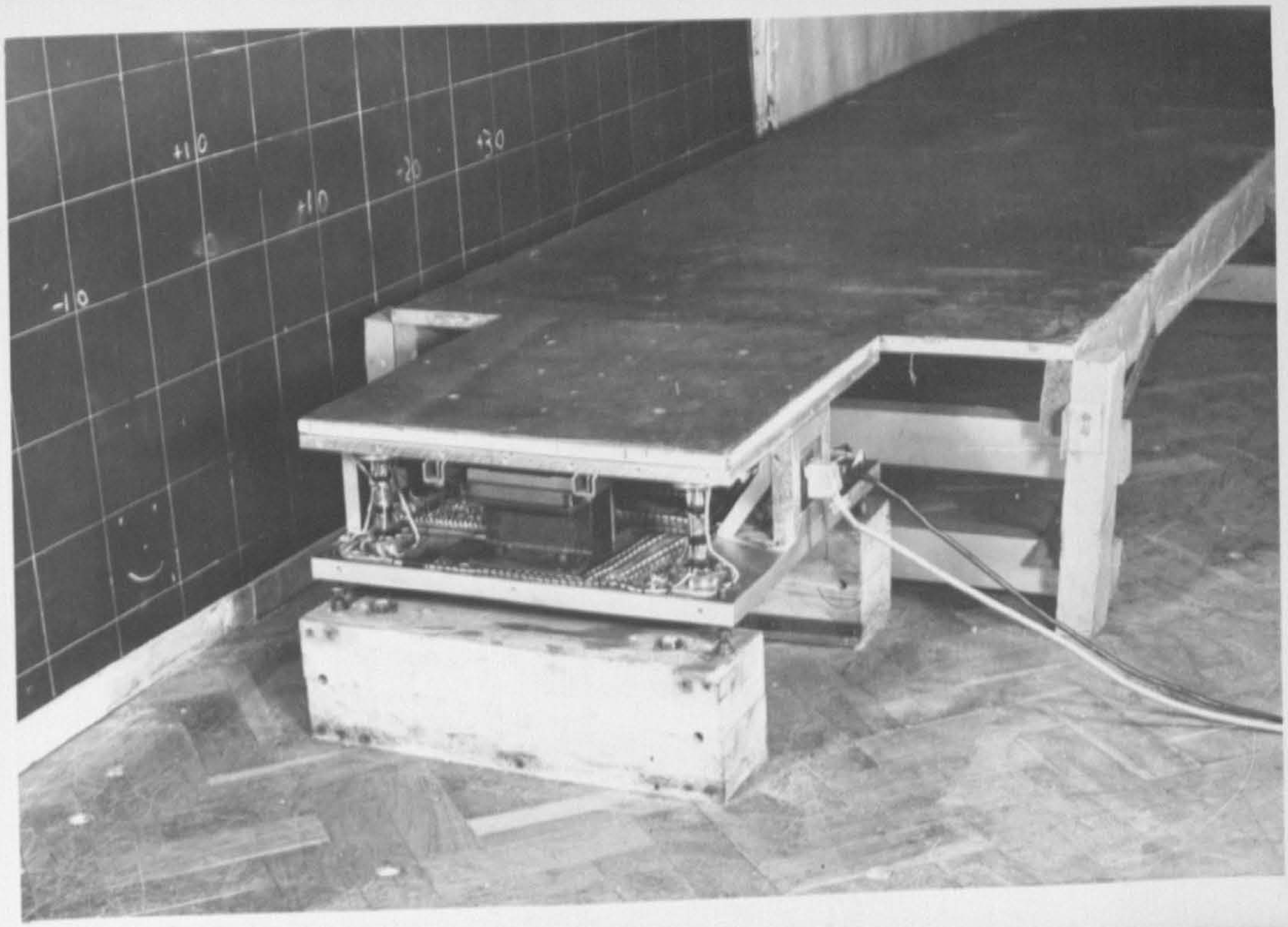
$$M_z = P_{6d} - (P_4 + P_5)C.$$

This force plate was calibrated and used for several tests but difficulty was experienced with the calibration of dynamometer. It appeared that the calibration varied with the position of application of the calibrating force which was eventually ascribed to the transverse load carrying capacity of the other dynamometers. In addition it was found necessary to align the dynamometers particularly carefully on assembly otherwise cross sensitivity between channels became critical due to the varying relative magnitudes of the quantities of interest. The maximum vertical force component transmitted in walking may be 200 lb, the horizontal component in the walking direction 30 lb and the transverse component 12 lb. To measure the lateral force to an accuracy of 0.1 lb, regardless of the other force values implies that the alignment of the dynamometers 4,5 and 6 must be within 1/2,000 of the vertical, i.e. 0.03 degrees or 0.001 in. allowable offset on the 2 in. half-length of the dynamometer. To maintain this accuracy a frame considerably more rigid than that of the original would have been required and it was decided instead to construct a force plate similar to that of Cunningham and Brown

(1952) as shown in Fig. 40.

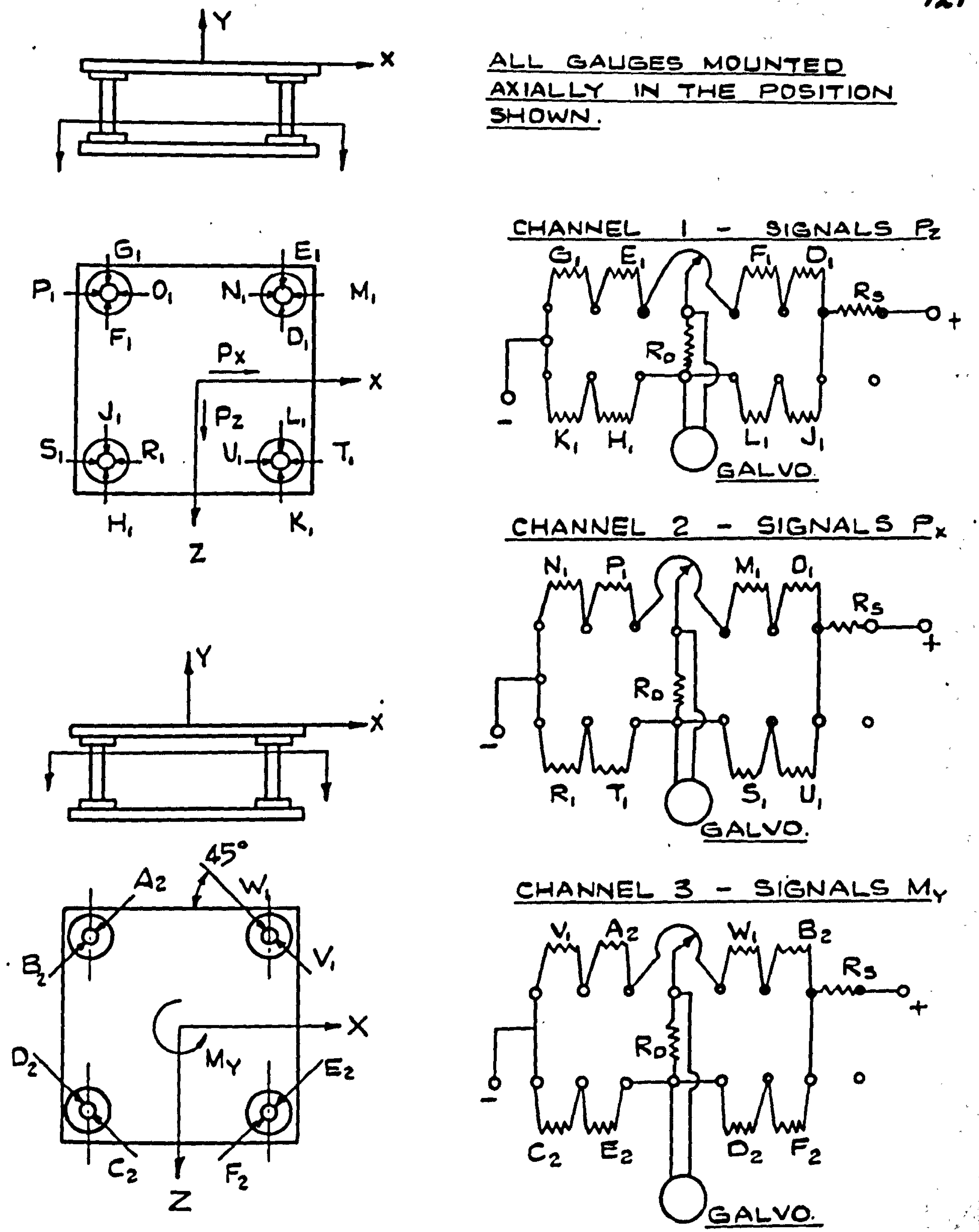
The critical part of the design was that of the supporting columns. The columns of Cunningham and Brown's force plate were turned from solid bar and since the flanges are 2.5. in. diameter and 5 in. apart and the tubular stem was 5/8 inch bore and 1/32 in. thick it is evident that the machining must have been both difficult and expensive. It was therefore decided to fabricate the columns for the Strathclyde force plate from drawn seamless tube to which the flanges were welded before final machining on a mandrel. This eliminated the major production difficulty of producing a uniform concentric bore 5.1/4 in. long. The arrangement of the strain gauges used to measure the applied load actions was not changed, but with the availability of silicon semi-conductor gauges it was found possible to transmit the signals directly from all channels to a multi-channel ultra violet light galvanometer recorder without the necessity for amplification. The circuits used are shown in Figs. 41 and 42.

The structural analysis for various forms of loading on the force plate is given in Appendix IV. It is seen that although the ratios of the stiffnesses of the top and bottom plates to that of the columns are high, being 58 and 150 times respectively, yet bending actions due to vertical load on these plates are



General View of Force Plate.

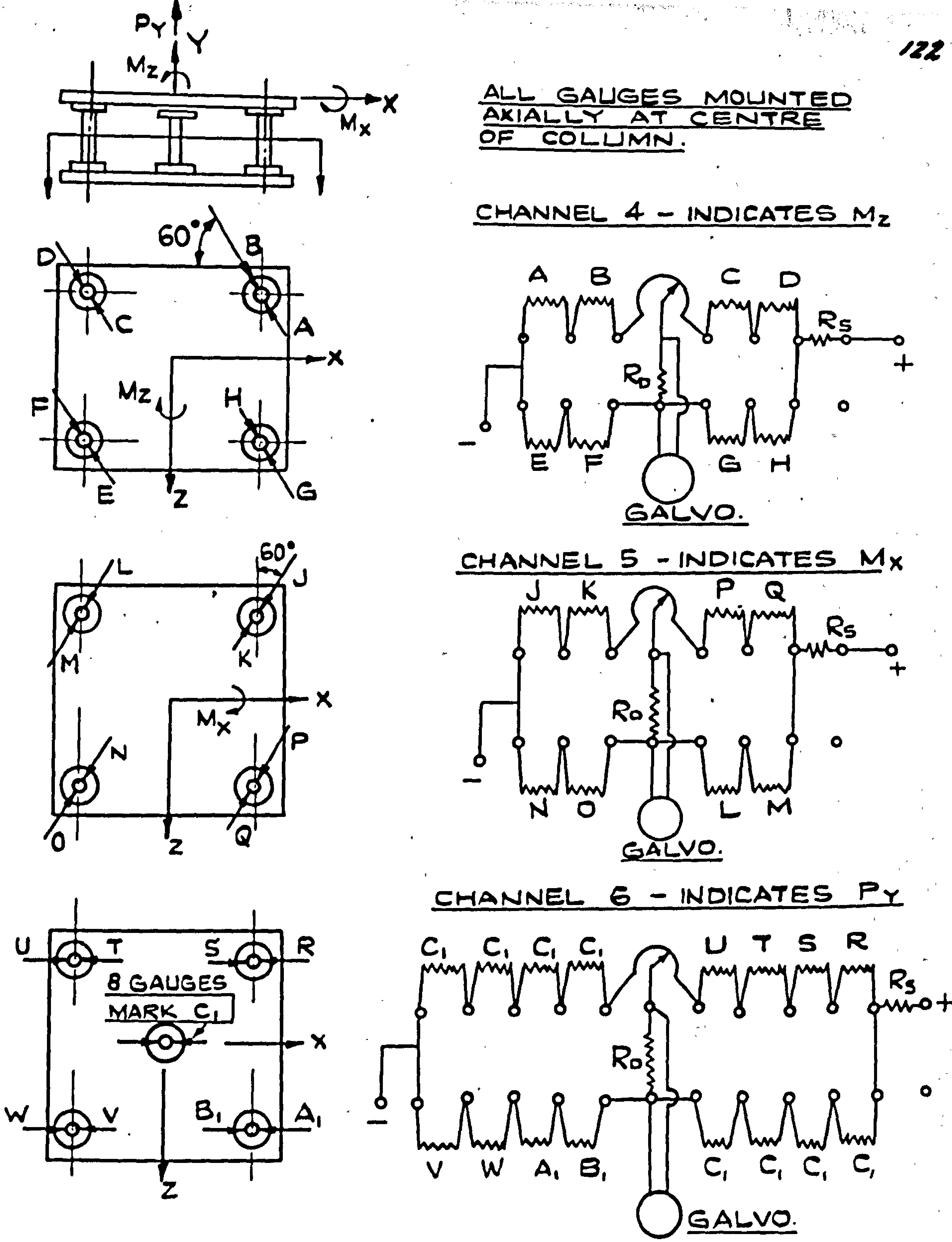
Fig. 40.



FORCE PLATE

GAUGE POSITIONS & CIRCUITS FOR THREE MEASURING CHANNEL

FIG. 41



FORCE PLATE.

GAUGE POSITIONS & CIRCUITS FOR THREE MEASURING CHANNELS.

FIG. 42.

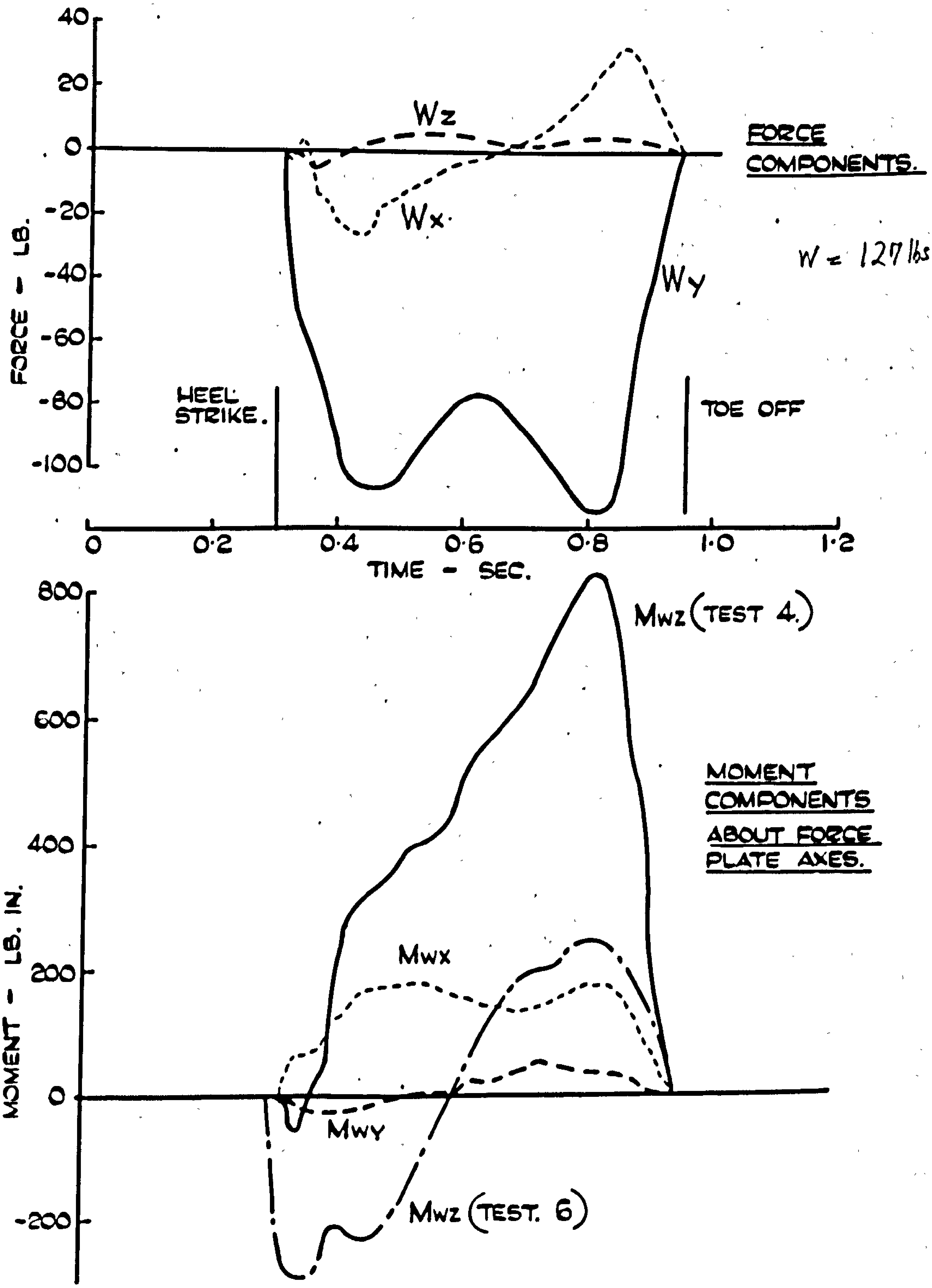
transmitted to the columns. This could cause cross-sensitivity between channels.

Generally it was found that linear calibration curves could be obtained for the six measuring channels, but there was a small amount of cross-sensitivity between some channels. This was dealt with by using cross-sensitivity calibration factors where necessary.

The silicon semi-conductor strain gauges were sensitive to temperature. This caused drift of the signals. Zero adjusting resistors were introduced into each circuit and the zero position of the traces could be adjusted as required. The rate of drift was slow and there was no detectable change during the time of the stance phase for one foot, which occupied about 0.6 seconds. A more serious thermal effect was, however, change in gauge sensitivity with temperature, which resulted in variation of the calibration factors from test to test. This necessitated calibration on the occasion of each test.

Since the force plate was being used to measure dynamic force actions with wave forms generally as shown in Fig. 43, it was necessary to consider the dynamic response of the force plate and the recording system. The fundamental part of the applied force can be considered to be a half sine wave of full

SUBJECT Nº 4. TEST 2.



GROUND TO FOOT FORCE ACTIONS

FIG. 43.

wave period approximately 1.2 second i.e. a frequency of 0.8 cycles per second. An analysis of the modes of vibration, natural frequencies and damping coefficients for the force plate is given in Appendix V and this shows the lowest natural frequency of the force plate to be 56 cycles/sec.

This frequency occurs for the linear vibration of the top of the force plate in the lateral direction.

b) Walkway.

The force plate was installed above floor level and situated between sections of timber walkway. Each section was 6 ft long and 3 ft wide, and was secured to the adjacent sections. The top surface was covered with a black sheet rubber covering. Since the force plate was only 20 in. broad, two filler pieces in the form of separate stools, were arranged to make up the breadth to 3 ft. The general arrangment is shown in Fig. 44.

c) Galvanometer Recorder.

A 25 channel galvanometer recorder model SE 2,100 produced by Messrs. S.E. Laboratories Ltd., was used to record the force plate and EMG records. This recorder had an ultra violet light source and the records on 12 in wide paper could be inspected immediately after a test to check that a satisfactory record had been produced. The paper speed could be varied in

General View of Test Area.

Fig. 44.

steps from 2 mm/sec to 2,000 mm/sec. A timing unit tied to mains frequency gave transverse lines across the record at intervals of 10, 1, 0.1 or 0.01 sec. The galvanometers used for the force plate records were type B.100 having the following nominal characteristics:-

Natural frequency	100 cycles/sec.
Sensitivity	0.0025 mA/cm.
Resistance	80 ohm.

The bridge circuits of the force plate were arranged to provide the 250 ohm resistance across the galvanometer required to give electromagnetic damping of 0.65 times critical damping. This ensures a frequency response flat within $\pm$ 3% up to 60 cycles/sec. The results of a Fourier series analysis of a force record are shown in Table 4 and it is apparent that the significant terms in the wave form lie within the capabilities of the galvanometers. The attenuation of harmonics up to 150 cycles/sec will not be more than 50 % and the galvanometer records would therefore record any such terms clearly if they existed.

d) Cameras.

The records of the spatial configuration of the test subject's limb segments were taken by two Paillard Bolex H16 Reflex Cameras

TERM	FOURIER COEFFICIENT FOR CHANNEL Nº					
	1	2	3	4	5	6
	W_z	W_x	M_{wy}	M_{wz}	M_{wx}	W_y
1	13·98	-1·57	-2·03	0·79	27·41	-46·87
2	-4·4	-30·18	-29·45	-31·51	-4·03	0·27
3	-0·18	0·79	2·94	-3·44	8·04	-20·19
4	-2·05	-15·13	-5·49	-9·21	3·30	-7·61
5	-9·94	-1·89	-2·85	-5·13	0·14	-3·77
6	-7·44	-3·08	2·43	-2·76	0·31	-3·99
7	-8·01	0·52	-0·80	-3·31	-0·70	0·28
8	-7·18	0·44	2·03	-0·77	-0·50	-2·41
9	-4·97	0·76	-0·89	-2·10	-0·84	-0·18
10	-0·30	1·60	0·70	-0·56	0·25	-1·64

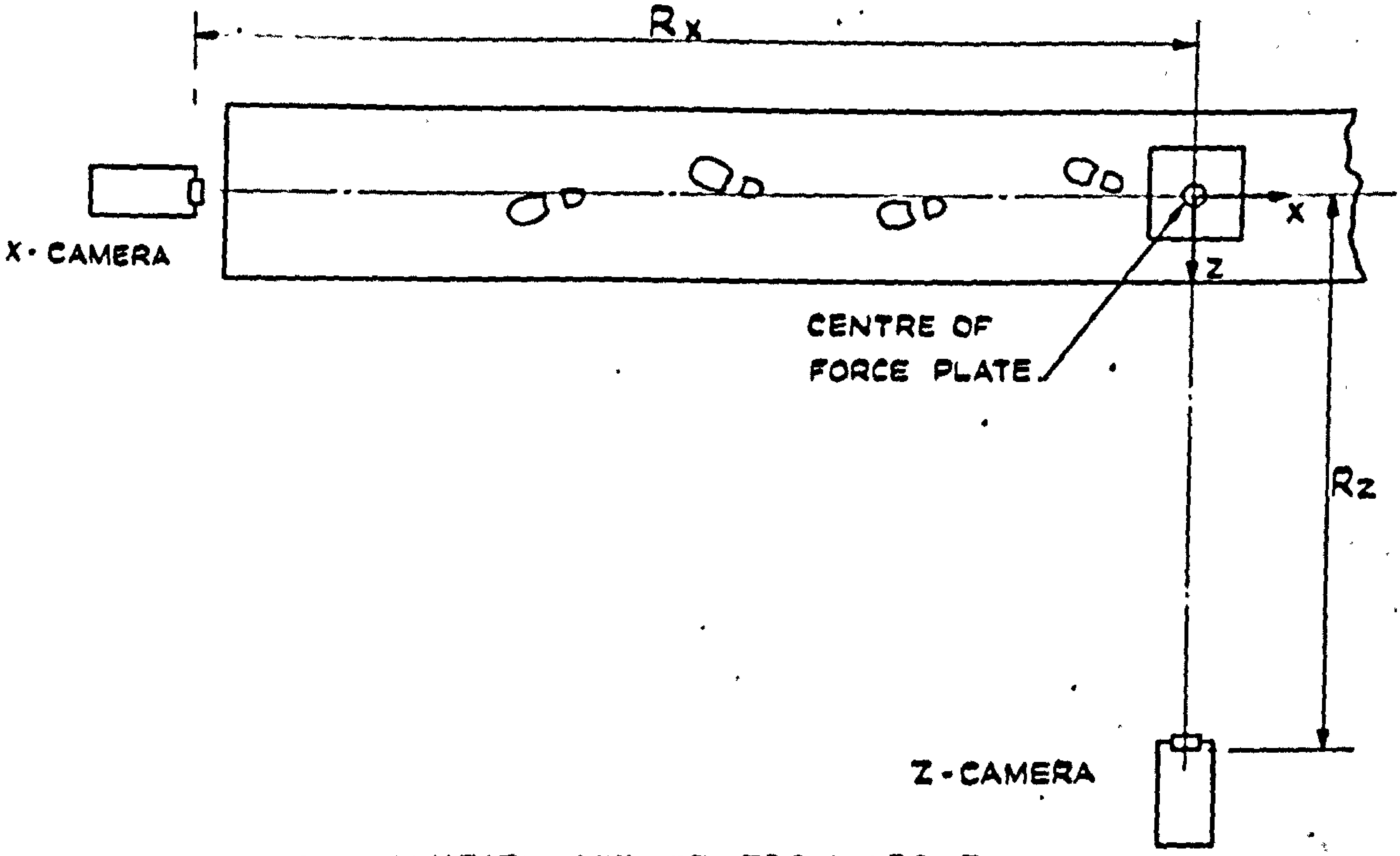
FOURIER ANALYSIS OF GALVANOMETER RECORD OF GROUND TO FOOT FORCE ACTIONS.

UNITS ; MILLIMETRES OF TRACE DEFLECTION.

FUNDAMENTAL FREQUENCY FOR THIS SUBJECT 0·83 c/SEC.

TABLE 4

using 16 mm. film. The cameras were arranged as shown in Fig. 45 to allow the three co-ordinates of the centre of each reference level of the subject to be obtained. Since the displacement records of the two cameras were to be correlated it was desirable that the cameras record at the same time intervals. Since the displacement records were to be used to obtain accelerations it was desirable that they record at equal controlled time intervals. Two 1/75 horsepower synchronous electric motors running at 3,000 rev/min were used to drive the cameras through a worm reduction drive of ratio 8:1. Since one revolution of the external drive shaft of the cameras corresponded to the exposure of 8 frames of film, this arrangement gave a film speed of 50 frames per second for each camera. There was in this area no history of electric mains frequency variations of more than 0.2 cycles per second and the camera drive speed was therefore constant within 0.4%. There remained the possibility of intermittent action by the camera and drive assembly due to vibration of the elastic systems, pitch errors in the gears and obliquity of drive shaft. This was investigated by filming a crystal controlled timing unit giving its output on decatron display tubes having complete revolution times of 100, 10, 1, 1/10 and 1/100,



PART OF WALKPATH VIEWED FROM ABOVE.

FIG. 4.5

seconds respectively. When the film was developed and projected, the 1/10 second index was observed to progress uniformly and the 1/100 the second index was in a consistent position. It was therefore concluded that the camera timing devices were accurate to within the 0.4 % possible mains frequency variation.

e) Phasing

The two cameras and the galvanometer recorder comprised three channels of a data recording system whose speed was tied to the electric mains supply. Ideally the cameras should have been arranged to have their shutters opening simultaneously. The speed of variation of height, y, and lateral positions, z, of the pelvic markers was at most 0.5 in per frame, however, and it was attractive for mechanical simplicity to have the cameras independently driven. It was decided therefore to relate the systems in time by firing a flash bulb in the field of view of both cameras and arranging for the firing of the flash to operate an event marker in the galvanometer recorder. The operation of the cameras was such that the aperture was closed for 12 m.sec and open for 8 m.sec of every 20 m.sec cycle. It was not therefore possible to use an electronic flash gun since the flash time of these was shorter than 0.012 sec and thus might not have been recorded by one or other camera. The galvanometer recorder carried an event marker triggered by the connection of the flash bulb circuit. The rise time for the flash bulb light intensity was stated by the makers to be of the order of 10 m. sec for half peak illumination but the firing was visible as a faint glow before full ignition occurred.

If the flash becomes visible in camera Z just before the shutter closes and the shutter in the X camera closes 0.001 sec. later, a trace of the flash will be seen on the Z camera film frame. This frame would then be related to the subsequent frame on the Z camera given a maximum error of approaching 1 frame.

Fortunately the z displacements viewed on the X camera are slow relative to the x displacements and this error would have a possible maximum value of 0.15 in for the pelvic marker of a typical subject.

The flash bulb was arranged in parallel with the event marker of the galvanometer which is energised from the 26 V line of the recorder by push button. The times for the event marker to reach its maximum indication with the flash bulb in and out of circuit were about 1 and 4 m. sec. respectively. Assuming that illumination of the flash bulb will not occur until after this delay and taking the time reference as occurring when the event marker reaches the extreme position the worst phasing error E_p is given by

$$E_p = \text{flash delay} - 4 \text{ m.sec} + \text{camera closed shutter time}$$
$$= 10 - 4 + 12$$
$$= 18 \text{ m.sec.}$$

The effect of this possible error is discussed later.

The central area of the walkpath was illuminated by four studio lamps of fixed output and four Colourtrans lights of variable output arranged to give a uniform intensity of illumination of the front and the side of the test subject within a stride's length on each side of the force plate. A piece of 1 in. thick blockboard was painted with a matt finish black paint and lined off in 5 in. squares in white drawing ink for use as a reference grid from which to take measurements of space co-ordinates. The grid board was superimposed on the film of the subject's walk by a double exposure technique described later.

Analysis of the film record was performed using a 'Specto' motion analysis projector having the facility of single frame advance by remote push button control. For the repetitive routine calculations for each record a Ferranti Sirius digital computer was used.

f) Electromyography.

E.M.G. signals were taken from selected areas of the hip and leg of the test subject using surface electrodes in the form of nickel silver discs, 1/2 in. diameter soldered to the connecting cables. The test subject wore a leather harness carrying a junction box and a multicore screened cable led from this box via an overhead swinging boom to the E.M.G. set. The bulk of the E.M.G. records reported in this thesis were obtained using a standard "Stanley Cox" two channel recording electromyograph. The amplified signals from the subject were visible on two C.R.O. screens, one with a variable time base and one without time base. The signals on the latter was recorded by a 'Cossar' oscilloscope camera. This formed a fourth channel of data to be phased with the cameras and galvanometer recorder. There was no convenient method available for injecting a timing mark into the E.M.G. record, since the apparatus was on loan and it was not wished to modify it. Eventually after trying indicators within the camera and step signals applied to the amplifier input, the output from the amplifiers, which was available through a standard socket was taken to two additional galvanometers mounted in the recorder. If the galvanometers were directly connected, the amplifiers were overloaded and no signal was obtained. External power amplification was attempted but spurious signal uptake developed. Eventually the output was taken to two 100 cycle/sec galvanometers through an adjustable high resistance. It was found possible to obtain a signal on the galvanometers. Due to their low natural frequency the galvanometer records were not reliable indications of the multiple burst of activity from the electrodes, but their shape corresponded generally to the form of the 'true' signal recorded by the oscilloscope camera and this allowed the E.M.G. records to be 'phased' in time with the other signals.

Subsequently a six channel amplifier set was obtained whose output could be fed directly to high frequency galvanometers and this shortened the experimental test time considerably.

The E.M.G. leads were carried from a harness carried by the subject through a flexible cable suspended from an overhead beam which can be seen in Fig. 44. The beam swung freely to follow the movement of the subject.

Test Performed.

Tests were performed on three female and nine male subjects wearing normal footwear. Each female subject performed tests using low and high-heeled shoes. Each subject walked at a gait selected by themselves after being requested to walk at normal speed. The results of a total of 18 tests walks were fully analysed and are presented here. In every case the analysis is performed for the left leg of the subject.

The ranges and averages of physical measurements of the subjects and parameters of their walks are shown in Table 5.

Experimental Procedure.

a) General Preliminaries.

The walkway was set up with the force plate in position and energised at least 1 hour before test to allow warming up. The cameras were clamped to fixed tables, and loaded. Kodachrome IIA film was used to get a contrast between the flesh colour of the subjects and the superimposed white grid line markings. The choice of film involved a compromise between "speed" of the films, the grain size, and the intensity of illumination. The limiting factor in the lighting was the accompanying heating. The subject was in the field of 8 light sources developing some 8 KW and spread to cover an area 6 ft long by 4 ft high. The exposure meter used indicated an illumination of approximately 260 foot candles. At the speed of exposure of 50 frame/sec

PARAMETER		MAXIMUM	MINIMUM	AVERAGE
SUBJECT'S HEIGHT -	IN.	72.5	62.5	68.2
WEIGHT -	LB.	180	127	140.7
AGE -	YR.	36.9	18.5	21.6
DOUBLE STRIDE LENGTH -	IN.	88.3	50.1	63.5
STRIDE TIME -	SEC.	1.24	1.02	1.11
MEAN SPEED -	IN./SEC.	81.7	40.9	57.6

RANGE OF PARAMETERS

IN WALKING TESTS ANALYSED.

TABLE 5.

which involved a shutter opening of 8 m.sec an aperture of f 1.8 or f 2 was necessary. Comfort of the test subjects required that no additional lighting be used and the cameras could not be 'opened up' any further. The grain size in the film under these conditions transpired to be just tolerable for the measurements required.

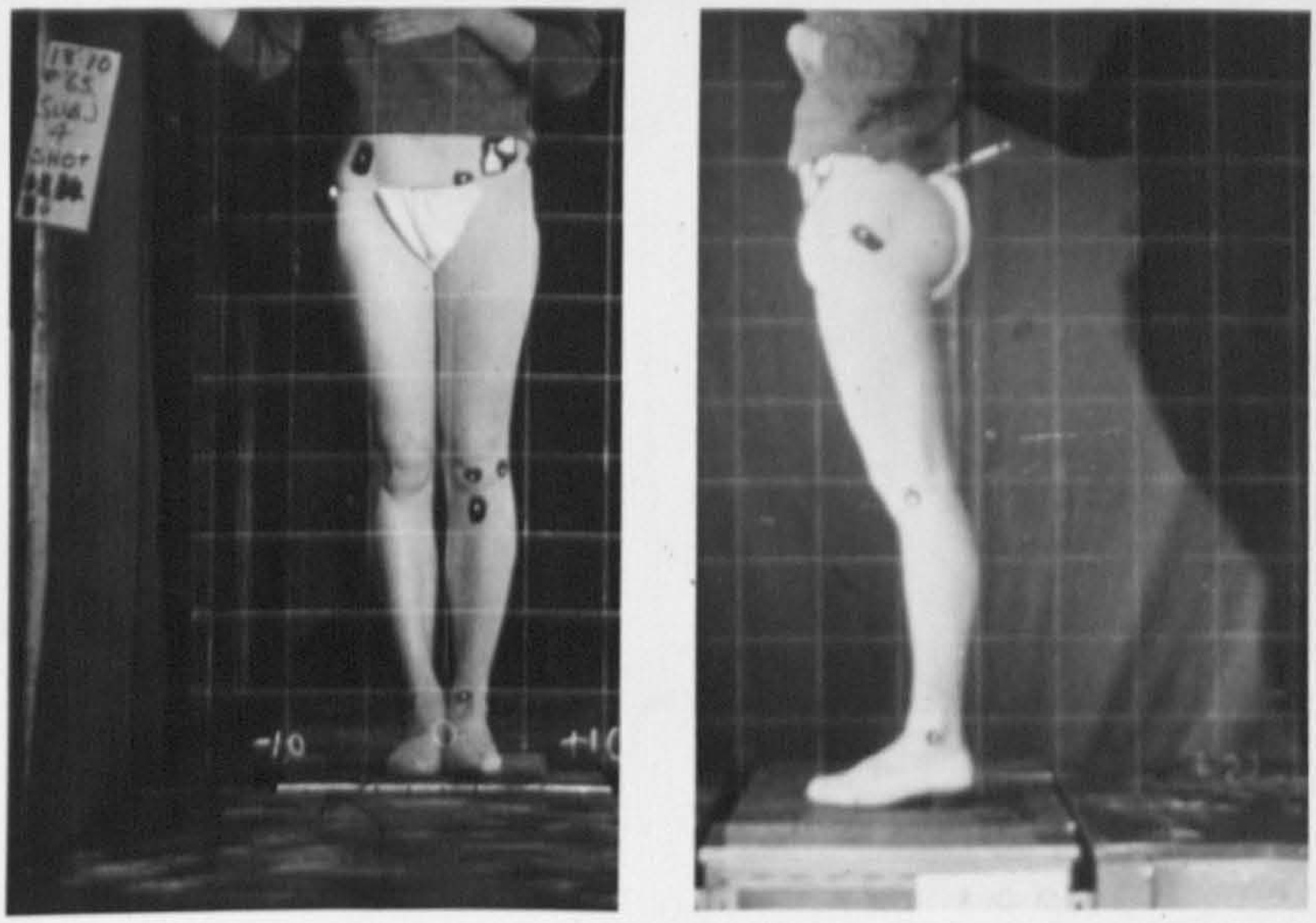
The distances of the cameras from the centre of the force plate were measured. The operation of the cameras, force plate and galvanometers and the flash equipment was checked.

b) Subjects.

The test subject wore no lower leg covering in addition to his normal shoes. The male subjects wore 'jock'straps' and the female subjects wore 'modesty briefs'. Loose clothing and watches were removed from the trunk and arms. Generally the male subjects wore no upper covering. A sweater was found most convenient for the females. The subjects were weighed and their shoe weight noted.

Markers were now placed on the subject to indicate the positions of the pelvis and the leg joints. An area of the skin approximately one inch square centred on the required regions was blackened with a spirit solvent ink applied through a felt tip - the product with the trade name "Magic Marker' was usually employed. Initially black 'Sellotape' had been used but trouble was experienced in the film analysis with reflections from the glossy surface. At the centre of the blackened area a self adhesive circular white paper 'spot' 1/4 inch diameter was stuck. This comprised the target from which measurements were taken.

Markers for the pelvis were placed on B, the left, and D, the right, anterior superior iliac spines, as shown in Fig. 46. The marker on the left spine was viewed from the side as well as the front and guide arrows were stuck on to define the position of this marker more clearly. The third reference point on the pelvis, C, was defined by a balsa wood 'tail' stuck centrally to the sacral region of the back using a commercial rubber based adhesive. The position of the centre of the femoral head was established by inspection and palpation and measurements were taken of its position relative to the external markers. The distances between the external markers were also measured. In the middle of the stance and swing phases the marker on the left anterior superior spine was obscured by the left arm swinging past. To allow readings to be taken at this time an auxiliary marker was



Positions of Markers on Test Subject.

Fig. 46.

situated at approximately the same height as the anterior superior spine and about 1.5 times the wrist breadth behind it. This marker had a fleshy muscular region beneath it and its readings were only taken when the other was obscured.

Markers for the knee (K) and ankle (A) were placed at the level of the joint surface in the lateral view and in the frontal view on the tibial shaft as close to the joint as would allow reliable readings. For the knee this was governed by the mobility of the patella and its ligament. For the ankle this was governed by the amount of dorsiflexion,which could allow the foot to obscure this marker. A marker, P, was also placed on the lateral surface of the shoe over the centre of the fifth metatarsal- phalangeal joint.

c) Electromyography.

The areas of application of the electrodes were selected generally near the centre of the muscle of interest if this was accessible, otherwise where the muscle was accessible without overlying muscles. The muscles selected were rectus femoris,gluteus maximus, gluteus medius, biceps femoris, adductor magnus and ilio-psoas. For the last named the electrodes were placed near the upper lateral border of the femoral triangle, keeping clear of the line of sartorius. The areas selected were successively shaved, rubbed lightly with abrasive paper and degreased by swabbing with ether. The electrodes were 1/2 in diameter nickel silver dished discs. These were filled with proprietary electrode jelly and fixed in position 1.1/4 inches apart along the line of the muscle wherever possible. The electrodes were secured by surgical adhesive tape. Particular care was taken to avoid the possibility of excess electrode jelly forming a conducting bridge between the electrodes.

After the electrodes were in position the leads were connected to the junction box on the patient's harness and the signals on the electromyograph were viewed as the patient performed prescribed movements of abduction, adduction, flexion and extension against resistance. For ilio-psoas the hip was flexed with the shank hanging relaxed from the knee. In this way the two joint muscles were retained in a relaxed condition. It was possible that part of the signal obtained thus was due to pectineus, but it certainly was not due to the two-joint muscles. If weak or intermittent signals were obtained or if non typical shapes appeared on the screen the electrodes and jelly were removed, the skin again swabbed with ether and the electrodes were re-applied. In some subjects, particularly the females and the plumper males it was found difficult to obtain large signals from the gluteal muscles, presumably due to the extent of the subcutaneous fat.

The subjects were next asked to walk to ensure that no intermittent faults developed on movement and the gain of the amplifiers was adjusted to give a maximum signal of amplitude between 1 and 2 cm. With the generally low level of activity in walking, the E.M.G. signals were small and noise was frequently a problem. This was much improved if the subject was earthed through an electrode mounted near the ankle and attached to the common earth of the instruments through the screen of the cable used. It was necessary to lay out the various instruments, and the cables for them and the lighting in such a way as to minimise interaction.

The actions of the six muscle groups selected were investigated in pairs since a two channel E.M.G. was initially available. The same procedures were adopted at each change of electrode.

d) Test walks.

The walking tests to assess the E.M.G. connections were used to establish the natural walking pace and stride length of the test subject and a starting line was placed on the walkway to guide the subject so that the required foot landed clearly on the force plate. The subject took some further practice walks to check that he was neither breaking stride or stretching out to land on the force plate. When this was achieved a recorded run was taken. As the subject started to walk the cameras and the galvanometer recorder were switched on by the operator. The subject and the E.M.G. were watched for abnormalities in gait and signals and when the subject had passed the force plate the flash bulb/event marker control was operated. If all traces were present in reasonable shape on the recording paper the subject was prepared for the following test.

Meanwhile the camera lens covers were fitted and the film was wound backwards to its starting position. The grid boards were placed on their reference lines, the lighting adjusted and the film was re-exposed on the grid boards for each camera in turn.

The second exposure was per formed with the film moving forwards so that it might be registered in the same position in the camera on each occasion. Certain reference marks were in the field of view of the camera during both exposures and no relative movement of these was observed which exceeded the chosen sensitivity of measurement of 0.05 in. Generally each subject made three or four test walks and it was found possible to test two subjects in an extended afternoon session.

After the session a check calibration was performed on the force plate.

e) Anthropometric Measurements.

The height of the subjects was measured, and the subject was placed supine on a table which was fitted with pads of variable height at the head, neck, shoulder and hip. These pads were set in positions to reproduce the subject's configuration in normal standing. The left leg was placed so that the centre of the knee was in the cranio-caudal plane through the left femoral head. Taking an origin at the left anterior superior iliac spine, measurements were then made of the three dimensional co-ordinates of the following positions:-

Right anterior superior iliac spine,

Adductor Origin, Inferior Ischial Brim;

Pubic Tubercle, Greater Trochanter,

Patella Centre, Knee Centre, Adductor Tubercle,

Posterior superior iliac spine, Sacral Tip.

These quantities were used in the assessment of the position and direction of the lines of action of muscle forces.

f) Analysis of test records.

The film records were projected on to a screen which was positioned so that in the area in which measurements were being taken the 5 in. grid squares appeared as 5 cm. squares and measurements of the co-ordinates of the required points were made by an operator using a rule graduated in millimetres. All measurements were taken relative to the enclosing grid square and were read to the nearest 1/2 mm. corresponding to 0.05 in at full size. By this means errors due to film distortion and geometrical irregularly in the lenses of the cameras and projectors were confined to the amount occurring in a 5 in square. The field of view of the Z cameras was approximately 140 in broad along the X -axis at the range at which the camera was set. This range was limited to about 130 in by the space available in the test area.

The readings were called out to a second operator who wrote them in a prepared table, graphed them and also adjusted the focus of the projector as necessary. The times at which heel strike and toe off occurred, and the flash bulb fired were also recorded. Generally the x, y, co-ordinates measured from the Z-camera record were analysed first and the Z-camera record was then projected. From the time of occurrence of the time marker flash the projector from counter was set to correspond to the numbers recorded for the Z-camera record. The required co-ordinates were then measured frame by frame and entered in the general table. Where the points appeared irregularly in the graphs, the relevant region of the film was re-measured. The most frequently occurring errors were $x \pm 5.0$ and $x \pm 1.0$ where x was the correct reading. These were generally noted by the recorder as called out, but since communication was verbal other slips occurred.

In the records taken initially it was usual to analyse the record over the period including the swing phase before heel strike on the force plate together with the stance phase on the force plate. Due to difficulties in deciding on the occurrence of heel-strike and toe off from the film record alone this procedure was altered. A complete cycle was analysed from the time at which the left ankle swung past the stationary right ankle prior to the stance phase on the force plate until the next occurrence of this apposition of the ankles (A.A.). This allowed accurate timing of the cycle and made the measured region approximately symmetrical about the axis of the camera.

For each pair of frames from the two cameras the following

15 quantities were recorded:-

Frame number, x,y and z for left anterior superior iliac spine (B), knee (K), and ankle (A), x and y for the tail marker (C) and the foot (P), y for the right anterior superior iliac spine (D). These quantities were punched on to computer tape, followed for each frame by a recognition character 999, 888 or 777, indicating respectively end of normal data set, end of last data set, and end of data set at heel strike.

The heel strike character was used so that the force plate record would only be read when it had non-zero values, i.e. in the stance phase. The galvo-recorder record was measured at time intervals corresponding to the film frame using a millimetre rule and measuring to the nearest 0.2 mm from each base line. Typical maximum signals recorded were 60 - 70 mm.

Since the digital computer available for the calculations, a Ferranti Sirius, had only a 4,000 word store it was not possible to perform all the calculations required with one programme. The calculations were arranged therefore to obtain first from the experimental quantities, the resultant force and moment at the hip/trunk section and the relative angular positions of the limb segments. The output tape from the computer giving these quantities was subsequently used as a data input for the next programme which calculated muscle and joint forces.

The measurement of the film and force plate records for one test on one subject occupied two operators for approximately 9 hours. The tape preparation and checking took 5 hours more. The computer normally took 160 secs per frame of record and these generally were 60 - 70 frames in length. Unfortunately the computer was approaching the end of its working life - it is not now operational - and interruptions and malfunctioning frequently trebled the nominal operation time.

Fromthe printed output of the computer, graphs were drawn of the required quantities.

A larger more sophisticated computer is now available and this has graph output facilities. The programmes are presently being re-written in Algol language for this computer.

THEORETICAL ANALYSIS.

The principal aim of this investigation is the use of equilibrium equations for one leg to determine the force transmitted at the hip joint of a walking subject from readings of displacement of body markers and ground/foot force actions.

The steps in the analysis can be summarised as follows:-

1. Elimination of parallax errors in camera observations
2. Determination of unknown position co-ordinates.
3. Determination of gravity and inertial force actions on body segments and the corresponding force and moments about the reference axes through the hip joint.
4. Calculation of ground/foot force actions and the corresponding force and moment actions at the hip joint.
5. Summing for the resultant leg/trunk force actions.
6. Determination of lines of action of muscle groups.
7. Selection of relevant muscle groups for the transmission of force action present and calculations of muscle forces.
8. Calculation of joint forces.

Procedures 1 - 5 were programmed in Sirius Autocode for the first programme and 6 - 8 for a subsequent programme. Had a larger computer been available all calculations would have been performed in one programme.

1. Measurements and parallax errors.

Co-ordinates of all joints are measured relative to a set of orthogonal axes x, y, z, with origin at the centre of the force plate as shown in Fig. 47. Grid boards are set up along these axes as shown in Fig. 44 and hence a point P(x, y,z) has apparent co-ordinates x', y', z' when measured from the projected film. From the geometry of Figure 48, it can be seen that:-

$$y = y' - (y' - h)z/R_z \qquad (17)$$

$$x = x' - x'z/R_z \qquad (18)$$

$$z = z' + z'x/R_x \qquad (19)$$

From (18) and (19)

$$x = x'(1 - z'/R_z) / (1 + x'z'/R_xR_z) \qquad (20)$$

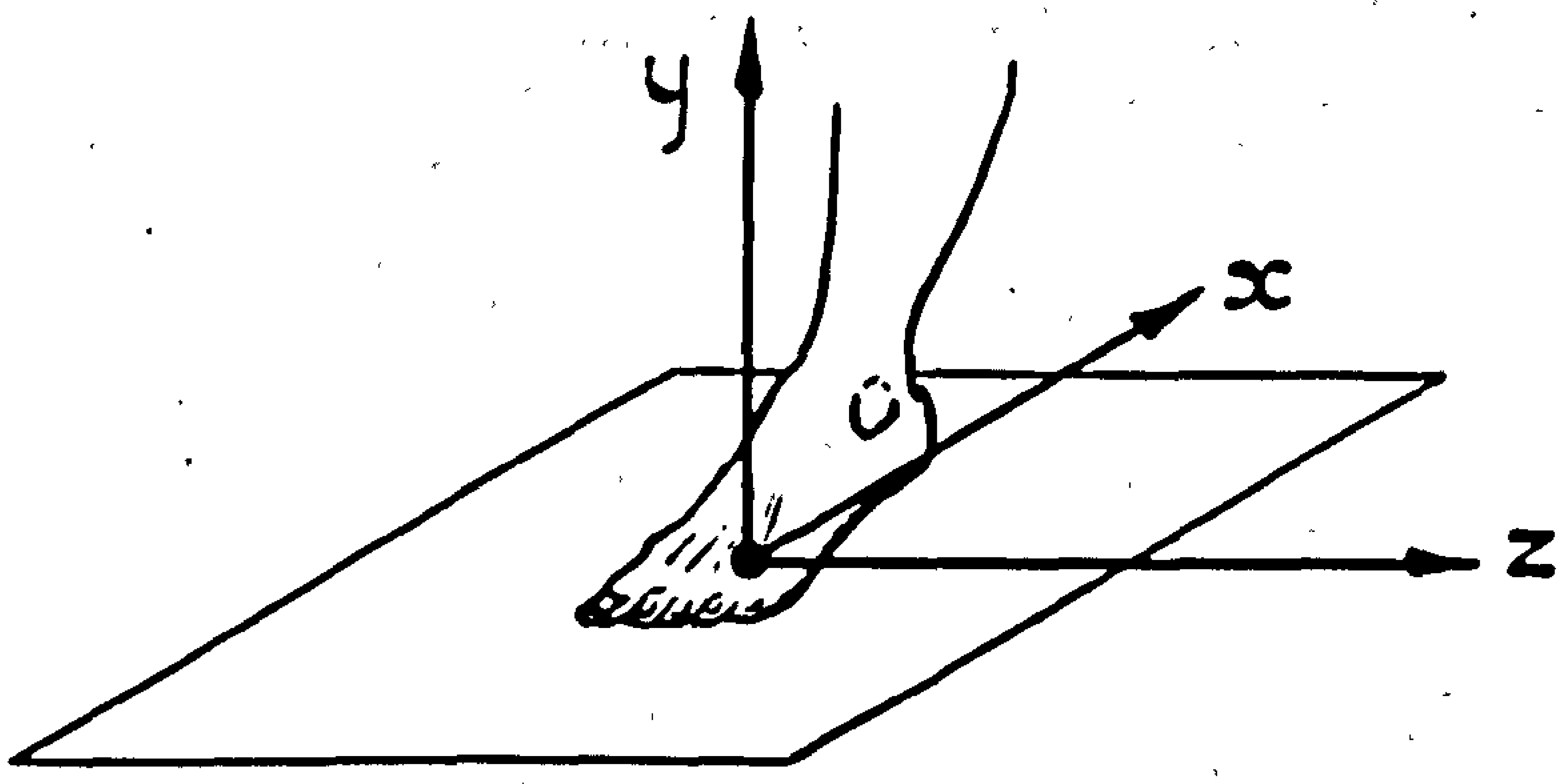
$$z = z'(1 + x'/R_x) / (1 + x'z'/R_xR_z) \qquad (21)$$

In one case the y co-ordinate is measured from the record of the X-camera and the corresponding equation is:-

$$y = y' + (y' - h)\,x/R_x \qquad (22)$$

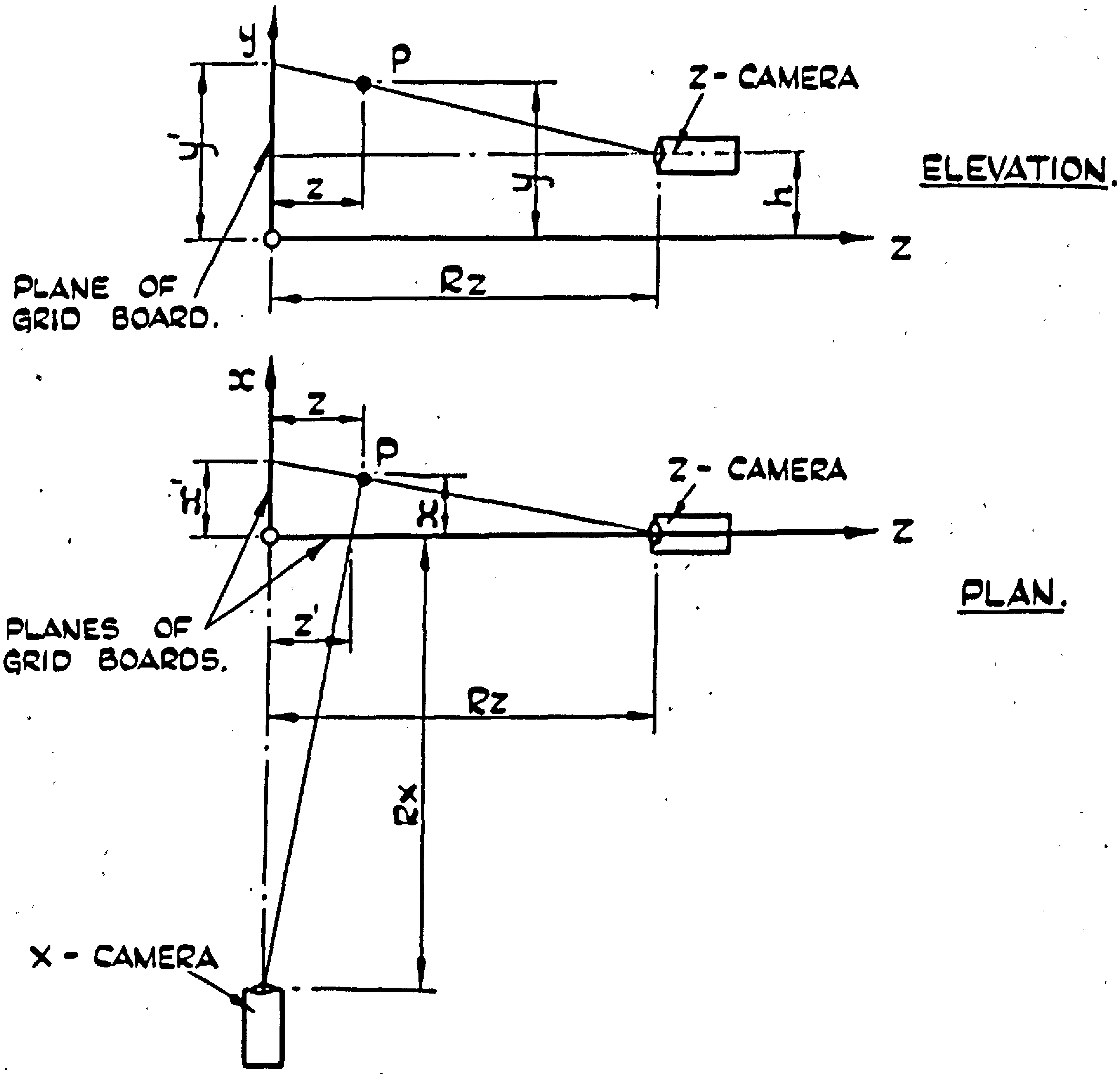
These corr ection equations can obviously be applied only where all three co-ordinates could be measured.

At the knee and the ankle, different markers are viewed in the X and Z cameras, and errors exist since the point required, namely the centre of the joint is offset from both markers. For Parallax corr ection the side marker is taken to be at the z-position



ORIGIN OF CO-ORDINATES.

Fig. 4.7



PARALLAX ERRORS IN FILM ANALYSIS

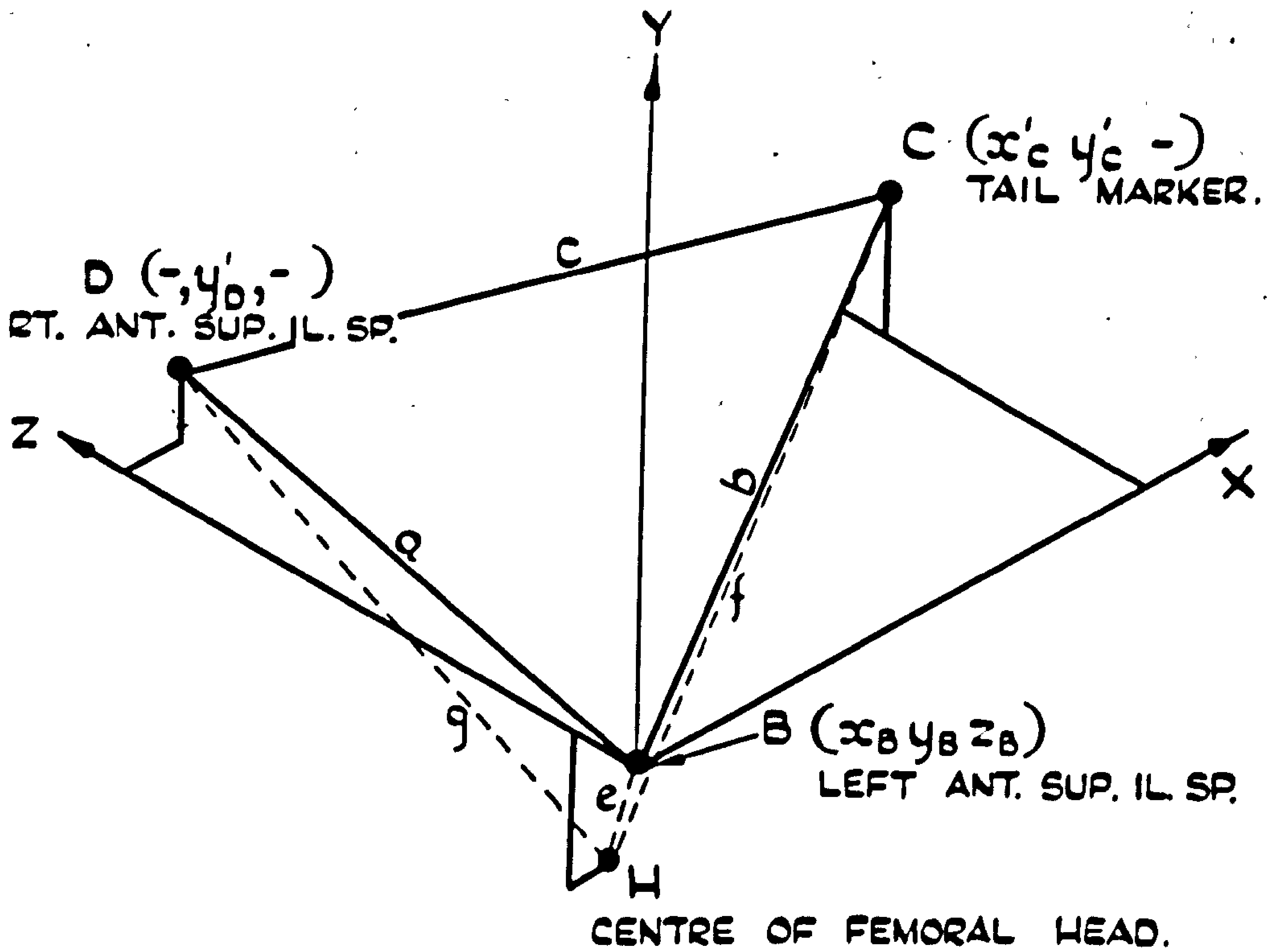
Fig. 4.8

indicated by the front marker and vice-versa. Since the dimensions of the knee are small by comparison with the offsets x and z this is a second order error. Rotation of the leg about its long axis introduces a further error. No convenient means of measuring this rotation was found. The radius of the markers from the centre of the knee is about 2 in. and the University of California (1947) studies showed that the average range of rotation for 11 subjects was 15.16°. The order of error involved is therefore possibly 0.26 in.

For the foot marker the Z position is taken to be the same as that for the ankle. This again involves a second order effect in the parallax correction but also neglects the effect of gravity and inertial force actions corresponding to rotation, and inversion or eversion at the ankle. Due to the small mass of the foot and the small amount of these movements in walking this is a reasonable approximation.

2. Unknown Co-ordinates.

If the pelvis is considered as a rigid body, six quantities are required to define its position in space. Since point B in Fig. 49 is visible in both cameras its three co-ordinates can be measured and corrected for parallax error immediately. Point C is visible in the Z camera only and its apparent co-ordinates x'_C, y'_C



REFERENCE POINTS ON PELVIS.

Fig. 4·9

are measured. Length b between B and C is measured at test and hence z_c is obtained as follows:-

$$(x_c - x_b)^2 + (y_c - y_b)^2 + (z_c - z_b)^2 = b^2 \quad \text{-----} \ (23)$$

Substituting (17) and (18) in (23) gives

$$\left[x'_c\left(1 - \frac{z_c}{R_z}\right) - x_b\right]^2 + \left[y'_c - (y'_c - h)z_c/R_z - y_b\right]^2 + \left[z_c - z_b\right]^2 = b^2$$

which can be re-arranged to

$$z_c^2\left[(x'_c/R_z)^2 + (y'_c - h)^2/R_z^2 + 1\right] + (2z_c/R_z)\left[x'_c(x_B - x'_c) + (y'_c - h)(y_B - y'_c) - z_B R_z\right] + (x'_c - x_b)^2 + (y'_c - y_b)^2 + z_b^2 - b^2 = 0 \quad \text{-----} \ (24)$$

This is a quadratic in z_c which must have two real roots one greater and one less than z_b. The algebraically smaller root it taken since only on large rotation about the y axis will z_c exceed z_b. In a typical case where a = 9.11 in. b = 13.36 and c = 13.16 the pelvic rotation would require to be 22 degrees to cause this condition. University of California (1947) studies show an average for this rotation of about ± 4 degrees.

Using this value of z_c, x_c and y_c are obtained using (17) and (18).

Marker D is visible continuously in camera X only and its apparent y and z positions can be measured. In fact only one of these is required and to reduce the number of readings taken only y'_c is measured. The co-ordinates of D are then calculated as follows using the lengths a and c in Fig. 49 which are measured on the occasion of the test:-

$$(x_d - x_b)^2 + (y_d - y_b)^2 + (z_d - z_b)^2 = a^2 \qquad \text{-----}\ (25)$$

$$(x_d - x_c)^2 + (y_d - y_c)^2 + (z_d - z_c)^2 = c^2 \qquad \text{-----}\ (26)$$

Substituting (22) into (25) and (26)

$$(x_d - x_b)^2 + (y'_d + (y'_d - h)x_d/R_x - y_b)^2 + (z_d - z_b)^2 = a^2 \quad (27)$$

$$(x_d - x_c)^2 + (y'_d + (y'_d - h)x_d/R_x - y_c)^2 + (z_d - z_c)^2 = c^2 \quad (28)$$

Subtracting and re-arranging gives

$$\left.\begin{aligned} &Ax_d + Bz_d = C \\ \text{where}\quad &A = 2\left[x_c - x_b + (y_c - y_b)(y'_d - h)/R_x\right] \\ &B = 2(z_c - z_b) \\ &C = a^2 - c^2 + z_c^{\,2} - z_b^{\,2} + y_c^{\,2} - y_b^{\,2} + x_c^{\,2} - x_b^{\,2} - 2y'_d(y_c - y_b) \end{aligned}\right\} (29)$$

Eliminating x_d between (25) and (29) gives:-

$$\left.\begin{aligned} &Ez_d^{\,2} + Fz_d + G = 0 \\ \text{where}\quad &E = ({}^B/A)^2\left[1 + (y'_d - h)^2/R_x^{\,2}\right] + 1 \\ &F = -2({}^C/A - x_b)B/A - 2({}^B/AR_x)(y'_d - h)\left[y'_d - y_b + C(y'_d - h)/AR_x\right] - 2z_b \\ &G = ({}^C/A - x_b)^2 + \left[y'_d - y_b + (y'_d - h)C/AR_x\right]^2 + z_b^{\,2} - a^2 \end{aligned}\right\} (30)$$

(30) is a quadratic equation giving two values of Z_d one to the right of line BC and one to the left. Obviously the algebraically smaller solution is relevant

Successive substitutions in (29) and (22) give the values of x_d and y_d.

At test, the position of the head of femur H is measured relative to the B marker and the relative positions of the B, C and D markers are measured in the normal standing position. The position of the head of femur relative to B, C, D is then defined by

$$\overline{BH}^2 = e^2 = ({}_ox_B - {}_ox_H)^2 + ({}_oy_B - {}_oy_H)^2 + ({}_oz_B - {}_oz_H)^2 \qquad (31)$$

$$\overline{CH}^2 = f^2 = ({}_ox_c - {}_ox_H)^2 + ({}_oy_c - {}_oy_H)^2 + ({}_oz_c - {}_oz_H)^2 \qquad (32)$$

$$\overline{DH}^2 = g^2 = ({}_ox_d - {}_ox_H)^2 + ({}_oy_d - {}_oy_H)^2 + ({}_oz_d - {}_oz_H)^2 \qquad (33)$$

These values of e^2, f^2 and g^2 determined initially are used to calculate the co-ordinates of H for each instant in time defined by one frame of cine film using equations (31), (32) and (33) without the 0 subscript on the co-ordinates. To solve for x_H, y_H, z_H, (31) - (32) gives:-

$$\left.\begin{array}{l} 2x_H(x_c-x_b) + 2y_H(y_c-y_b) + 2z_H(z_c-z_b) = H \\ \text{where } H = e^2 - f^2 + x_c^2 - x_b^2 + y_c^2 - y_b^2 + z_c^2 - z_b^2 \end{array}\right\} \qquad (34)$$

$$\left.\begin{array}{l} \text{(32-33) gives:-}\ 2x_H(x_c - x_d) + 2y_H(y_c - y_d) + 2z_H(z_c-z_d) = J \\ \text{where } J = f^2 - g^2 + x_d^2 - x_c^2 + y_d^2 - y_c^2 + z_d^2 - z_c^2 \end{array}\right\} \qquad (35)$$

$$\left.\begin{array}{l} \text{(33-31) gives :-}\ 2z_H(z_d-z_b) + 2y_H(y_d-y_b) + 2z_H(z_d-z_b) = K \\ \text{where } K = g^2 - c^2 + x_b^2 - x_d^2 + y_b^2 - y_d^2 + z_b^2 - z_d^2 \end{array}\right\} \qquad (36)$$

Eliminating x_d between (34) and (35) gives:-

$$z_H = Ly_H + M \qquad (37)$$

$$\left.\begin{aligned} \text{where } L &= \frac{(y_c - y_d)(x_c - x_b) - (y_c - y_b)(x_c - x_d)}{(z_c - z_b)(x_c - x_d) - (z_c - z_d)(x_c - x_b)} \\ M &= \frac{H(x_c - x_d) - J(x_c - x_b)}{2\left[(z_c - z_b)(x_c - x_d) - (z_c - z_d)(x_c - x_d)\right]} \end{aligned}\right\} \quad (37)$$

Eliminating z_H between (35) and (36) gives:-

$$\left.\begin{aligned} x_H &= N\,y_H + P \\ \text{where} \quad N &= \frac{(y_d - y_b)(z_c - z_d) - (y_c - y_d)(z_d - z_b)}{(x_c - x_d)(z_d - z_b) - (x_d - x_b)(z_c - z_d)} \\ P &= \frac{J(z_d - z_b) - K(z_c - z_d)}{2\left[(x_c - x_d)(z_d - z_b) - (x_d - x_b)(z_c - z_d)\right]} \end{aligned}\right\} \quad (38)$$

(37) and (38) are substituted in (31) to give:-

$$\left.\begin{aligned} Q\,y_d^2 + R\,y_d + S &= 0 \\ \text{where} \quad Q &= 1 + L^2 + N^2 \\ R &= -2N(x_b - P) - 2y_b - 2L(z_b - M) \\ S &= (x_b - P)^2 + y_b^2 + (z_b - M)^2 - c^2 \end{aligned}\right\} \quad (39)$$

(39) is a quadratic in y_d the algebraically smaller root of which corresponds to the true position of H.

x_H and z_H are then obtained by substitution in (37) and (38).

The co-ordinates of the centres of all the major joints in the left leg are now known and the co-ordinates of the centres of gravity of the limb segments are now calculated using Fischer's coefficients. For example for the thigh centre of gravity, T.

$$x_T = x_K - C_{2T}(x_H - x_K)$$

$$y_T = y_K - C_{2T}(y_H - y_K)$$
$$z_T = z_K - C_{2T}(z_H - z_K) \qquad (40)$$

where C_{2T} is the appropriate Fischer coefficient for the C.G. position of the thigh.

3. Gravity and Inertia Force Actions.

The weights of the limb segments were calculated from the subjects' body weight, W, using Fischers coefficients. For example for the thigh:-

$$W_T = C_{1T}W$$

The weight of the shoe was added to the weight of the foot and it was assumed that the C.G. position of the assembly was not significantly distant from that of the foot alone.

The accelerations of the limb segments were determined from their position in successive frames of the film. This procedure is known to be subject to large errors unless great care is taken. Felkel (1951) indicated that best results for acceleration determination could be obtained by graphical smoothing of the displacement data followed by numerical differentiation of the successive displacements using the expression $\dot{x}_{t+h/2} = (x_{t+h} - x_t)/h$. The curve of velocities obtained was integrated graphically at intervals to check for correspondence with the displacements. The same procedure was then applied to determine accelerations from the velocity curve. The procedure was reported to be time consuming and dependent on the skill of the operator.

It was decided for this investigation to use a numerical smoothing and double differentiation procedure described by Lanczos (1957). To obtain the linear acceleration component in the x-direction when the displacement is x_o, nine points x_{-4}, x_{-3}, x_{-2}, x_{-1}, x_o, x_1, x_2, x_3 x_4, all separated by equal time intervals h are used. A quadratic curve $y = Ax^2 + Bx + C$ is fitted to the five points x_{-4}, x_{-3}, x_{-2}, x_{-1}, x_o, using 'least squares'. The velocity at position x_{-2} is taken to be the slope of the quadratic at this point. The relevant expression is:-

$$\dot{x}_{-2} = (-2x_{-4} - x_{-3} + x_{-1} + 2x_0)/10h$$

$$\text{Similarly } \dot{x}_{-1} = (-2x_{-3} - x_{-2} + x_o + 2x_1)/10h \qquad (42)$$

The velocities at the five successive points x_{-2}, x_{-1}, x_o, x_1, x_2 are similarly treated to obtain the acceleration at the centre point.

$$\ddot{x}_o = (-2\dot{x}_{-2} - \dot{x}_{-1} + \dot{x}_1 + 2\dot{x}_2)/10h$$

or

$$\ddot{x}_o = (4x_{-4} + 4x_{-3} + x_{-2} - 4x_{-1} - 10x_o - 4x_1 + x_2 + 4x_3 + 4x_4)/100h^2 \qquad (43)$$

The factors governing this choice were:-

1. The desirability of a procedure amenable to operation by digital computer.

2. The knowledge that Bresler and Frankel (1950) had reported that the effect of gravity and inertial forces on the moments transmitted from leg to trunk was relatively small during the stance phase of movement.

Using displacement data from a test subject, accelerations were determined by four methods:- Lanczos' nine point method, a seven point method due also to Lanczos, double differentiation directly from a quadratic fitted to five adjacent points by 'least squares' and from three adjacent points assuming uniform acceleration. The results are shown in Fig. 50 and Fig. 51.

It appears that local peaks of acceleration are being lost by the 7 and 9 point formulae, but much obviously irrelevant noise is being rejected also.

The x, y and z components of the accelerations of the centres of gravity of the limb segments were determined using (43).

The projected inclinations of the limb segments relative to the vertical as viewed by the cameras are calculated for each frame. For the thigh for example:-

COMPARISON OF DOUBLE DIFFERENTIATION PROCEDURES

FIG. 50

SUBJECT N°4
TEST N°2
MARKER Y FOOT.
H.S.L.F.
T.O.L.F.
ALL ZERO
ILLUSTRATION OF DOUBLE DIFFERENTIATION
PROCEDURES
SYMBOL △ + × ⊙
N° OF TERMS 3 5 7 9
ACCELERATION - IN/SEC.²
TIME - SEC.
400
200
0
-200
-400
-600
0
0.2
0.4
0.6
0.8
1.0
1.2

FIG. 51

$$\theta_{xT} = \tan^{-1} \left\{ \frac{z_H - z_K}{y_H - y_K} \right\}$$
$$\theta_{zT} = -\tan^{-1} \left\{ \frac{x_H - x_K}{y_H - y_K} \right\} \qquad (44)$$

The angular accelerations $\ddot{\theta}_{xT}$, $\ddot{\theta}_{zT}$ are obtained by the use of (43).

The linear acceleration and gravity forces for the thigh are

$$I\,F_{Tx} = WC_{1T}\ \ddot{x}_T/g$$
$$I\,F_{Ty} = WC_{1T}\ (1 + \ddot{y}T/g) \qquad (45)$$
$$I\,F_{Tz} = WC_{1T}\ \ddot{z}_T/g$$

For the angular inertia force action the following approximate procedure is adopted. A more detailed analysis is outlined in Appendix VI.

Since the limb segments are inclined in space to the reference axes by angles θ_x and θ_z their effective radii of gyration k_x, k_z about these ones are reduced.

The reduced radii of gyration k' are taken to be

$$k'_x = k_x \cos \theta_z \qquad k'_z = k_z \cos \theta_x$$

The inertia moments for the thigh become therefore:-

$$M_x = WC_{1T}\,(C_{3T}\,L_T \cos \theta_z)^2\ \ddot{\theta}_x/g$$
$$M_z = WC_{1T}\,(C_{3T}\,L_T \cos \theta_x)^2\ \ddot{\theta}_z/g \qquad (46)$$

Due to the force actions on the thigh the resultant moments about reference axes X, Y and Z through the centre of the femoral head as shown in Fig. 52. are:-

$$\left.\begin{aligned} M_{Hx} &= (WC_{1T}\,\ddot{z}_T/g)(y_H - y_T) - WC_{1T}\,(1 + \ddot{y}_T/g)(z_H - z_T) \\ &\quad - WC_{1T}\,(C_{3T}L_T \cos\theta_z)^2\,\ddot{\theta}_{xT}/g. \\ M_{Hz} &= WC_{1T}\,(1 + \ddot{y}_{T}/g)\,(x_H - x_T) - (WC_{1T}\,\ddot{x}_{T}/g)\,(y_H - y_T) \\ &\quad - WC_{1T}\,(C_{3T}L_T \cos\theta_x)^2\,\ddot{\theta}_{zT}/g. \end{aligned}\right\} \quad (47)$$

4, Ground to Foot Force Actions.

The behaviour of the force plate is described principally by six calibration constant $K_1 - K_6$ which when multiplied by the galvanometer readings $G_1 - G_6$ gave the force actions acting by the ground on the foot shown in Fig. 43 as follows:-

$$\begin{gathered} W_z = K_1G_1 : W_x = K_2G_2 \\ M_{wy} = K_3G_3 : M_{wz} = K_4G_4 \quad M_{wx} = K_5G_5 \quad W_y = K_6G_6 \end{gathered} \quad (48)$$

In fact interaction between channels 1 and 4 and 2 and 5 was found on calibration. This effect is reported for a force plate of this type by Harper Warlow and Clarke (1961). It was found that this could be fully described as follows:-

$$\begin{aligned} M_{wz} &= K_4\,(G_4 - K_{14}G_1) \\ M_{wx} &= K_5\,(G_5 - K_{25}G_2) \end{aligned} \quad (49)$$

A small difference in the calibration of channel 4 was found to depend on the direction of the applied moment. If the reading G_4 was negative an alternative factor K_4' was used.

FREE BODY DIAGRAM OF LEG

FIG. 52

5. <u>Summation of Force Actions at the Hip Joint.</u> The resultant leg to trunk force actions transmitted across a section at the hip are defined by force components H_x, H_y, H_z and moment components M_{Hx}, M_{Hy}, M_{Hz} acting on the leg in the directions shown in Fig. 52 where:-

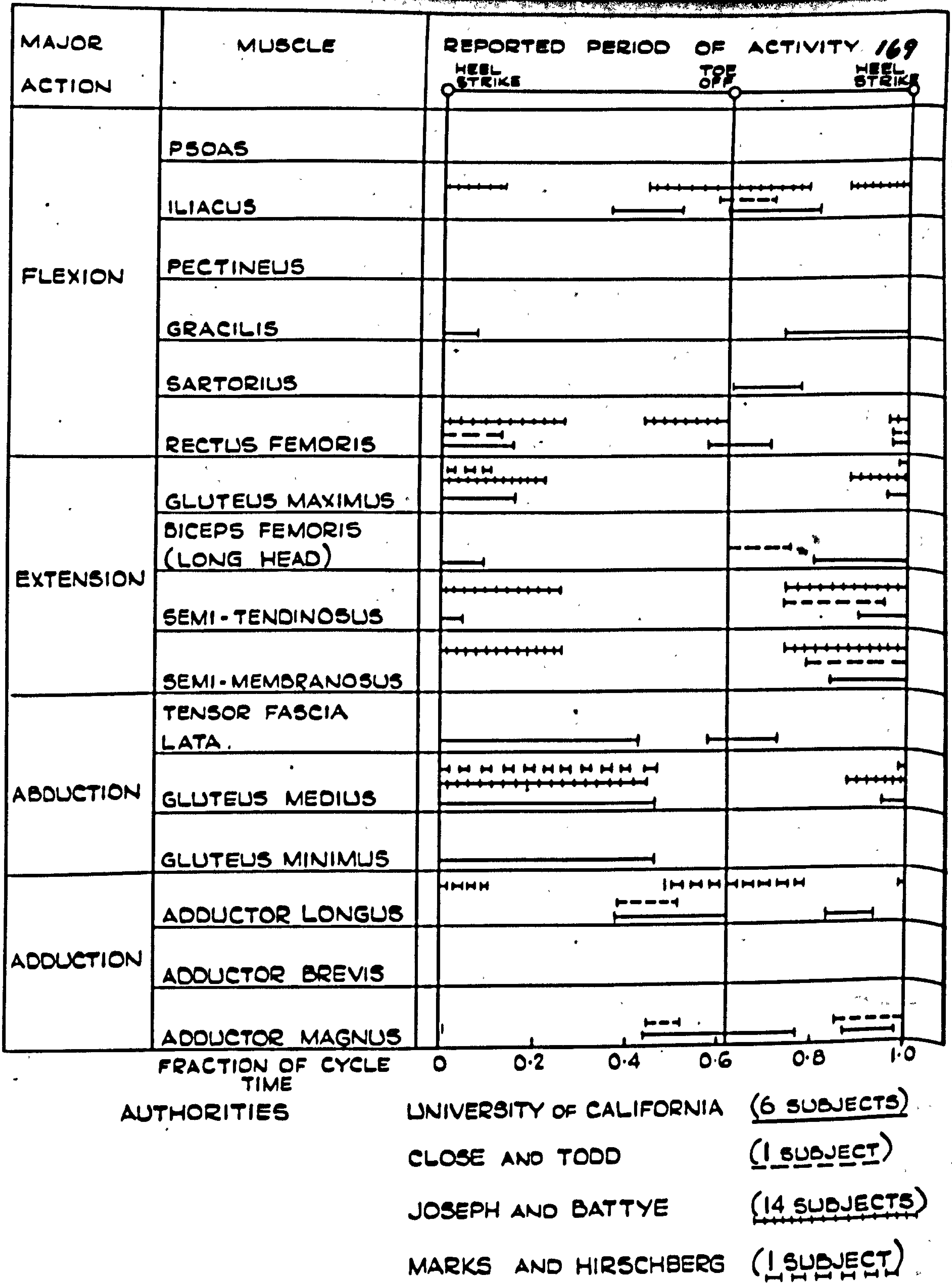
$$
\begin{aligned}
H_x &= -W_x - \sum WC_1\, \ddot{x}/g. \\
H_y &= -W_y - \sum WC_1\, (1 + \ddot{y}/g) \\
H_z &= -W_z - \sum WC_1\, \ddot{z}/g. \\
M_{Hx} &= W_z y_H - W_y z_H - M_{wx} + \sum (WC_1 \ddot{z}/g)(y_H - y) \\
&\quad - \sum WC_1 (1 + \ddot{y}/g)(z_H - z) - \sum WC_1 (C_3 L \cos \theta_z)^2\, \ddot{\theta}_x/g \\
M_{Hy} &= W_x z_H - W_z x_H - M_{Wy} + \sum (WC_1 \ddot{x}/g)(z_H - z) \\
&\quad - \sum (WC_1\, \ddot{z}/g.)(x_H - x). \\
M_{Hz} &= W_y x_H - W_x y_H - M_{wz} + \sum WC_1 (1 + \ddot{y}/g)(x_H - x) \\
&\quad - \sum (WC_1\, \ddot{x}/g)(y_H - y) - \sum WC_1 (C_3 L \cos\theta_x)^2\, \ddot{\theta}_z/g
\end{aligned}
\tag{50}
$$

6. <u>Determination of Lines of Action of Muscle Groups.</u>

The force actions transmitted between leg and trunk can be described in terms of the three components J_x, J_y, J_z of the resultant hip joint force and three tensions in connective tissue traversing the leg/trunk section. There are however twenty two muscles traversing this section. The experimental relationships between E.M.G. and muscle tension (Inman (1947) Inman et al (1951)

Lippold (1952), Bigland and Lippold (1954) have generally been obtained in situations where not more than two muscles were available to resist externally imposed calibrating forces. This is not possible in the complex system at the hip. In this investigation an analysis is presented by considering the lines of action of the forces corresponding to the action of groups of muscles. The muscles were grouped on the basis of their major function, their anatomical position and the phasing of their electrical activity as reported by Marks and Hirschberg (1958) Close and Todd (1959), University of California (1953) and Joseph and Battye (1966). Because of their small volume, the following muscles are excluded from the analysis:- Obturator Internus and Externus, Gemellus Inferior and Superior, Piriformis and Quadratis Femoris.

The major function and the periods of activity reported from E.M.G. investigations are shown in Fig. 53. The general conclusion drawn from this diagram is that of Marks and Hirschberg that muscles in the same group act synchronously. The results of the investigations differ in detail since Joseph and Battye using surface electrodes received signals from a greater area of muscle, and possibly "cross-talk" from other muscles; whereas the other investigations using needle electrodes may not



ACTION AND PHASING OF HIP MUSCLES.

FIG. 53

have recorded activity from motor units distant from the needles.

For analysis the muscles are arranged in the following groups:-

1. The short Flexors - Psoas, Iliacus:
2. The long Flexors - Rectus Femoris, Sartorius
3. The short Extensors - Gluteus Maximus.
4. The long Extensors - Biceps Femoris (Long Head), Semimembranosus Semitendinosus.
5. The abductors - Tensor Fascia Lata, Gluteus Medius, Gluteus Minimus.
6. The Adductors - Adductor Longus, Brevis and Magnus, Pectineus and Gracilis.

The areas of origin and insertion of these muscles are marked on a skeleton and the lines of action of the resultant forces developed by the muscle groups are taken to lie along the line joining the centres of the corresponding areas of origin and insertion as shown in Fig. 16. Since relative movement occurs between femur and pelvis the inclinations of these lines of force are calculated for each frame of film record.

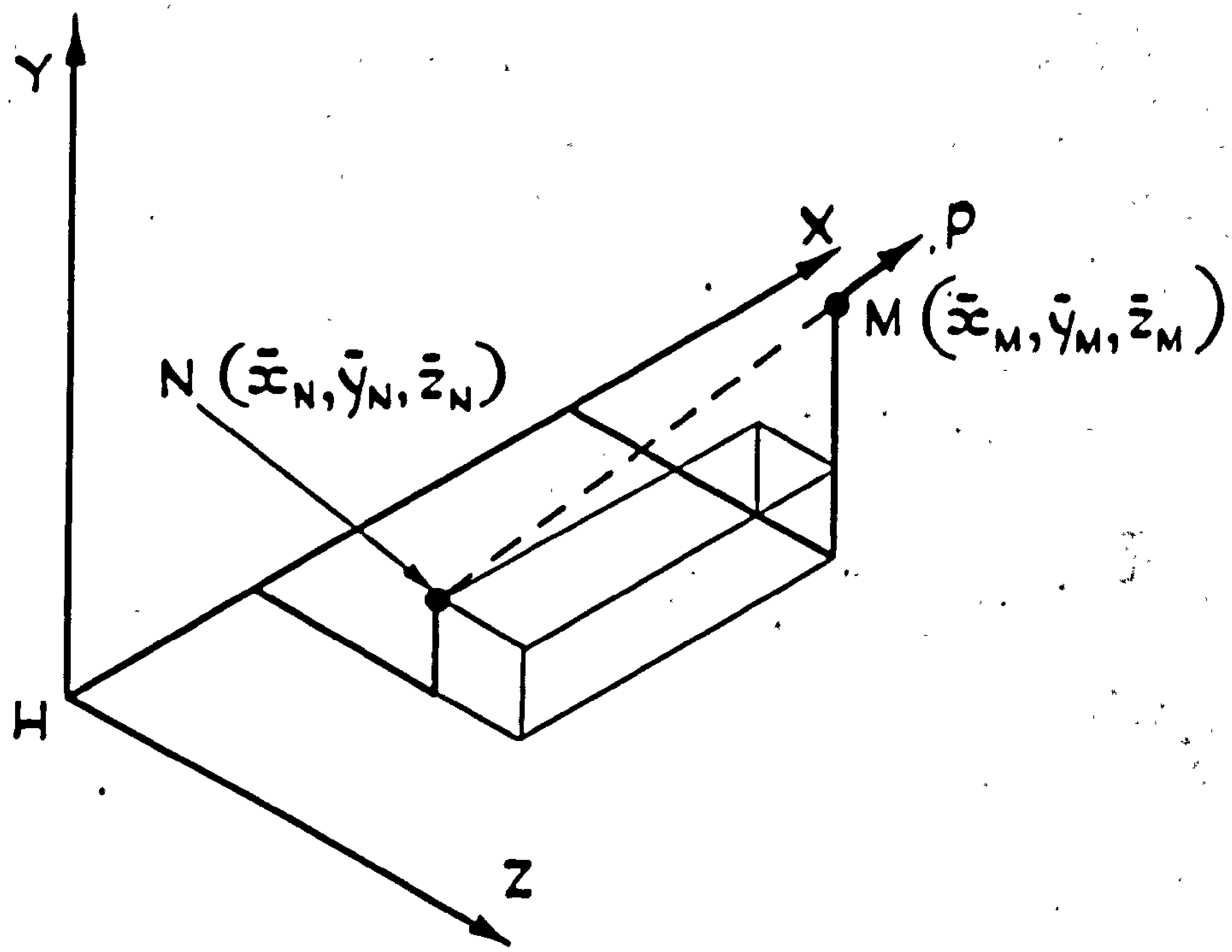
The co-ordinates of points on the line of action of the muscle groups near the origin and insertion were measured relative to axes X, Y, Z centred on the femoral head with the skeleton in a reference position. Three typical measurements of the skeleton pelvis and two of the femur are compared with the corresponding measurements of the test subject and the co-ordinates of the lines of action of the muscle groups are scaled accordingly.

Consider a typical muscle group shown in Fig. 54 having on the skeleton co-ordinates of the origin X_{Ms}, Y_{Ms}, Z_{Ms} and insertion X_{Ns}, Y_{Ns}, Z_{Ns}, scaling factors K_{xP}, K_{yP}, K_{zP}, K_{xzF} and K_{yF} for the X, Y and Z of the pelvis and XZ and Y of the femur, respectively.The corresponding co-ordinates for the test subject are:-

$$\begin{aligned} &\text{for the origin } X_M = K_{xP} X_{Ms} : Y_M = K_{yP} \; Y_{Ms} : Z_M = K_{zP} Z_{Ms} \\ &\text{for the femur } X_N = K_{xzF} X_{Ns} : Y_N = K_{yF} \; Y_N \; : Z_N = K_{zF} Z_{Ns} \end{aligned} \tag{51}$$

The pelvis is defined by B, C and D,and an origin of co-ordinates is taken at B. Axes X, Y and Z are taken through B parallel to x, y and z of the force plate when the subject is in the normal standing position. D is taken to have xyz co-ordinates (0,0,- a) and c (X_c Y_c Z_c). After a general rotation, let BX be defined by direction cosines t_{11}, t_{12}, t_{13} relative to x, y and z, BY by t_{21}, t_{22}, t_{23} and BZ by t_{31}, t_{32}, t_{33} respectively. Then:-

$$x_D - x_B = t_{11} X_D + t_{12} Y_D + t_{13} Z_D = t_{13}(-a)$$



REFERENCE DIRECTIONS FOR MUSCLE GROUP.

Fig. 54

$$y_D - y_B = t_{21} X_D + t_{22} Y_D + t_{23} Z_D = t_{23}(-a)$$
$$z_D - z_B = t_{31} X_D + t_{32} Y_D + t_{33} Z_D = t_{33}(-a)$$
$$x_C - x_B = t_{11} X_C + t_{12} Y_C + t_{13} Z_C$$
$$y_C - y_B = t_{21} X_C + t_{22} Y_C + t_{23} Z_C$$
$$z_C - z_B = t_{31} X_C + t_{32} Y_C + t_{33} Z_C.$$

The remaining three equations necessary to solve for the nine unknown direction cosines are given by the orthogonality conditions:-

$$\begin{aligned} t_{11} &= t_{22} \times t_{33} - t_{32} \times t_{23} \\ t_{21} &= -t_{12} \times t_{33} + t_{32} \times t_{13} \\ t_{31} &= t_{12} \times t_{23} - t_{22} \times t_{13} \end{aligned} \qquad (53)$$

Having solved these equations the displaced position of the origin of the typical muscle group is defined by co-ordinates X_n, Y_n, Z_m where for example:-

$$X_m = t_{11} X_m + t_{12} X_m + t_{13} Z_m \qquad (54)$$

The measurements of the positions of the femur do not suffice to describe its rotation about its long axis.Ryker (1952) shows that the average volume for this rotation obtained from seven subjects is 7 degrees measured from the stance position. The effect of this rotation is therefore neglected and it is assumed that the displaced position of the femur is obtained by rotation about its two other principal axes as described by angular rotations θ_x and θ_z defined by equations (44).

In the pictorial view in Figure 55 the axis KHY is shown displaced by θ_x and θ_z into position K'HY'. The appropriate direction cosines for HY' are therefore given by:-

$$\begin{aligned} t_{21} &= -KL/KK_1 = -HK \tan\theta_z / \sqrt{HK^2 + KM^2 + KL^2} \\ &= -\tan\theta_z / \sqrt{(1 + \tan^2\theta_z + \tan^2\theta_x)} \\ t_{22} &= HK/KK_1 = 1/\sqrt{(1 + \tan^2\theta_z + \tan^2\theta_x)} \\ t_{23} &= MK/KK_1 = \tan\theta_x / \sqrt{(1 + \tan^2\theta_z + \tan^2\theta_x)} \end{aligned} \qquad (53)$$

also angles δ and ϕ in this view are given by:-

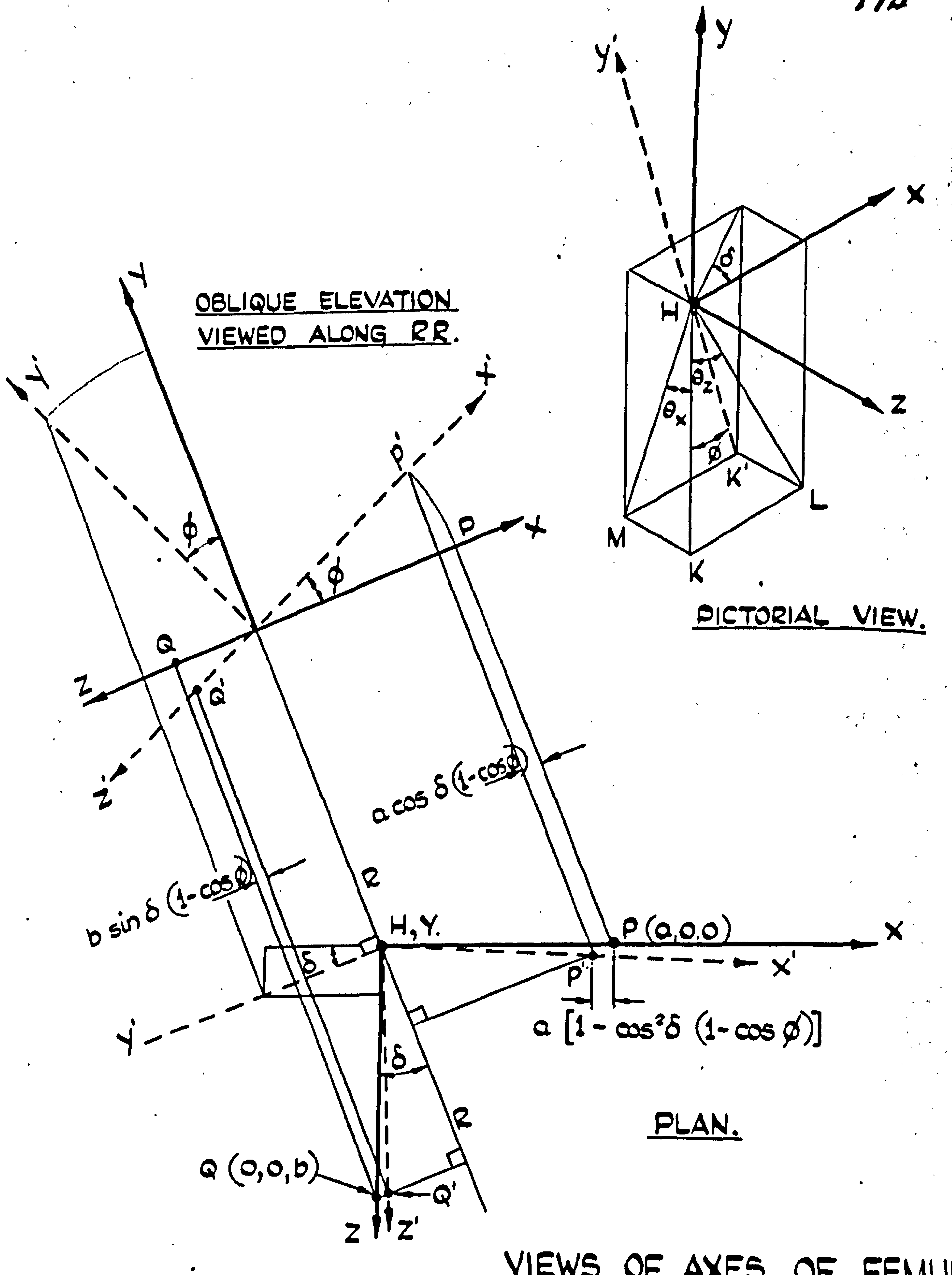
$$\begin{aligned} \tan\delta &= \tan\theta_x / \tan\theta_z \\ \tan\phi &= KK'/HK = \sqrt{(\tan^2\theta_x + \tan^2\theta_z)} \end{aligned} \qquad (56)$$

In the plan view of Fig. 54 the axis HY is displaced to HY'. If no rotation of the femur occurs about its long axis the rotations of HX and HZ must be about axis RR. From the geometry of this figure, it follows that point P(a,o,o) is displaced to P' (X_P, Y_P, Z_P) where:-

$$\begin{aligned} X_P &= a\left[1 - \cos^2\delta(1 - \cos\phi)\right] \\ Y_P &= a\cos\delta\sin\phi \\ Z_P &= a\cos\delta\sin\delta(1 - \cos\phi) \end{aligned} \qquad (57)$$

The direction cosines of OX' are therefore given by

$$t_{11} = 1 - \cos^2\delta(1 - \cos\phi) = 1 - \tan^2\theta_z(1 - \frac{1}{R})/S \qquad (58)$$



VIEWS OF AXES OF FEMUR

Fig. 55

$$\left.\begin{aligned}
&\text{where } S = \tan^2 0_x + \tan^2 0_z \text{ and } R = \sqrt{(1 + \tan^2 0_x + \tan^2 0_z)} \\
&t_{12} = \cos\delta \sin \emptyset = \tan 0_z/R = -t_{21} \\
&t_{13} = \cos\delta \sin (1 - \cos \emptyset) = (1 - \tfrac{1}{S}) \tan 0_z \tan 0_x/R \\
&\text{similarly} \\
&t_{31} = (1 - \tfrac{1}{R}) \tan 0_z \tan 0_x/S = t_{13} \\
&t_{32} = -\tan \theta_x/S = -t_{23} \\
&t_{33} = 1 - \tan^2\theta_x (1 - \tfrac{1}{R})/S
\end{aligned}\right\} \quad (58)$$

The co-ordinates X_N, Y_N, Z_N for the displaced position of the insertion of the typical muscle group can now be obtained, for example:-

$$X_N = t_{11} X_N + t_{12} Y_N + t_{13} Z_N.$$

If this muscle group develops a tension p as shown in Fig. 55 the components of force and moment relative to the x, y, z axes through H can be obtained in terms of the direction cosines T of the line of action where:-

$$T_X = (X_M - X_N)/V \quad \text{and} \quad V = \sqrt{\left[(X_M - X_N)^2 + (Y_M - Y_N)^2 + (Z_M - Z_N)^2\right]}$$

$$T_y = (Y_M - Y_N)/V$$

$$T_z = (Z_M - Z_N)/V.$$

Then the force and moment components of p relative to the reference axes are :-

$$p_x = T_x p : \qquad p_y = T_y p \qquad p_z = T_z$$

$$
\begin{aligned}
M_x &= p_z \bar{Y}_m - p_y \bar{Z}_m = p\,r_x \\
M_y &= p_x \bar{Z}_m - p_z \bar{X}_m = p\,r_y \\
M_z &= p_y \bar{X}_m - p_x \bar{Y}_m = p\,r_z
\end{aligned}
\qquad (60)
$$

7. <u>Selection of Muscle Groups and Calculation of Forces</u>. 50

The moments M_{HX}, M_{HY}, M_{HZ}, defined in equations (~~6~~ 2) are transmitted by tension in appropriate muscle or ligament groups and the corresponding compressive force at the joint surface. Moments about the Y axis are transmitted by the rotators and by X and Z components of the forces in other muscles, but the lines of action of these are not greatly inclined to the Y axis. The equilibrium equation for moments about the Y axis is not used, therefore, and this leaves 5 equilibrium equations. The unknowns at the leg/trunk section comprise:- Components J_x, J_y, J_z of the resultant joint force J: Tensions $p_1 - p_6$ in the six selected muscle groups. If no antagonistic action is present groups 1 and 2 <u>or</u> groups 3 and 4 will have zero force and group 5 or group 6 will have zero force. This reduces the number of unknown forces to six compared with 5 equilibrium equations. Solutions were therefore obtained assuming that, either group 1 or group 2 acted alone or that group 3 or group

4 acted alone. If in fact load sharing occurred between synergistic groups it was taken that the value of the resultant joint force would lie between the two bounds defined in this way.

For example, if, at an instant in time, M_{HX} is positive and M_{HZ} is negative, as occurs following heel strike, groups 3, 4 and 6 are taken to be inactive and the following equations obtained from (50) and (60) are solved:-

$$\text{either} \quad \begin{cases} M_{HX} + p_5 r_{5x} + p_3 r_{3x} = 0 \\ M_{HZ} + p_5 r_{5z} + p_3 r_{3z} = 0 \end{cases}$$

$$\text{or} \quad \begin{cases} M_{HX} + p^1_5 r_{5x} + p_4 r_{4x} = 0 \\ M_{HZ} + p^1_5 r_{5x} + p_4 r_{4z} = 0 \end{cases} \tag{61}$$

Then the resultant joint force components are obtained as:-

$$J_x = H_x + p5\ T_{x5} + p_3 T_{3x} \quad \text{etc.}$$

$$\text{and } J = \sqrt{(J_x^2 + J_y^2 + J_z^2)} \tag{62}$$

$$\text{or } J_x = H_x + p^1_5 T_{x5} + p_4 T_{4x} \quad \text{etc.}$$

If a negative p value is obtained for the force in a group, A, the solution is discarded and an alternative set of equations (61) is selected using groups antagonistic to group A .

The simplified block diagrams for the two computer programmes developed by the author for these calculations are shown in Fig.

56 and 57. The programmes were written in Sirius Autocode in which the most complicated arithmetical step in any instruction is $v_{no} = - v_{n1}/v_{n2}$ where v_n denotes the number in the store denoted by the number in store n. The programmes are consequently lengthy and not included here.

The programmes were checked on completion by artificial sets of data corresponding to uniform linear and angular accelerations of the limb for which direct solutions could easily be obtained

BLOCK DIAGRAM FOR FIRST PROGRAMME TO CALCULATE HIP FORCE AND MOMENTS.

Fig. 56.

BLOCK DIAGRAM FOR SECOND PROGRAMME TO CALCULATE HIP JOINT FORCE.

Fig. 57.

ANALYSIS AND DISCUSSION OF RESULTS.

1. Detailed Consideration of Typical Experimental Readings.

The available information for test No. 2 of test subject 4, a female subject wearing low heel shoes, is discussed in detail. Insert Figure 1 on transparent film may be superimposed on the graphs to obtain greater detail of the limb configuration and the salient points of the walking cycle. In each case the quantities refer to the left leg. The following abbreviations are used:-

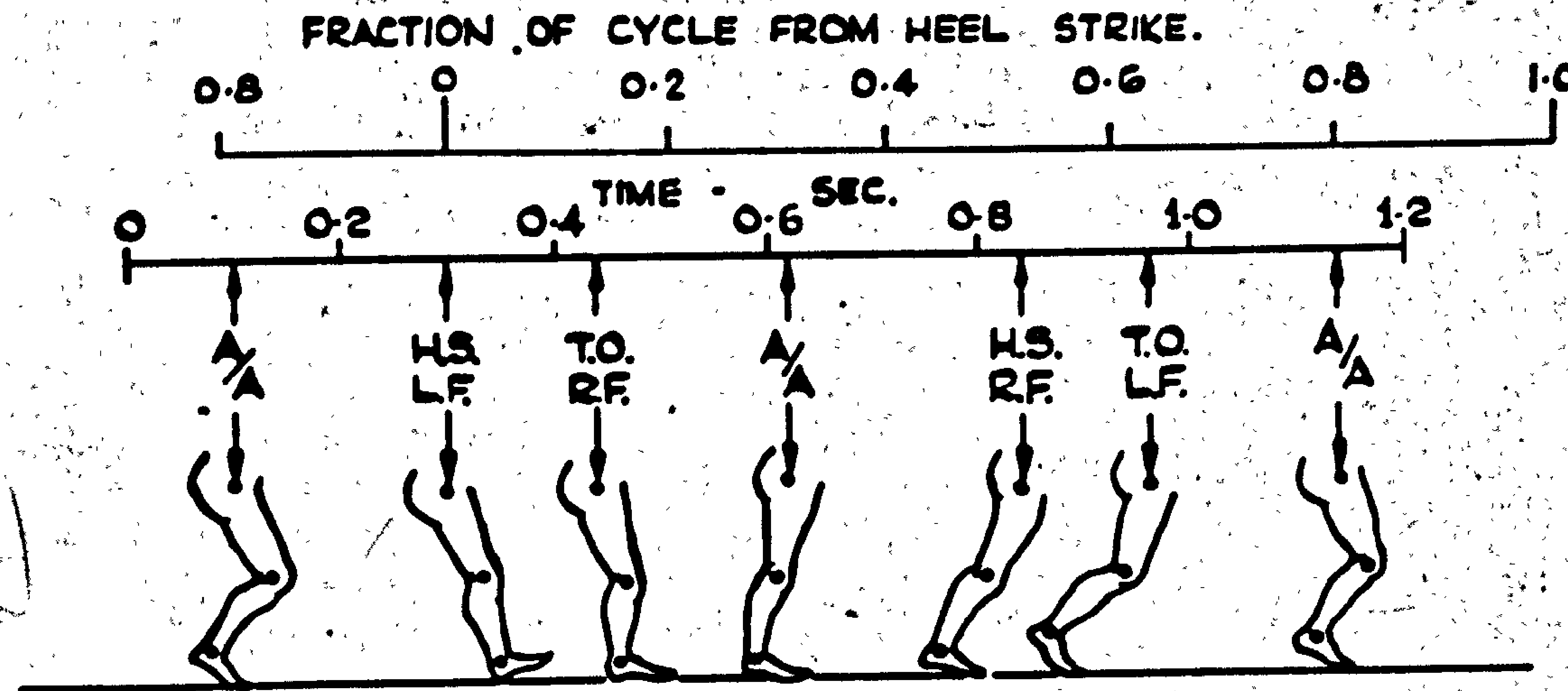
L.F. - left foot : R.F. - Right foot

H.S. - Heel Strike : T.O. - Toe Off.

A/A - Ankles in Apposition.

The cycle time extends from heel strike of the left foot until the next heel strike of the left foot or between any two such consecutive events. The stance phase extends from heel strike until the next toe-off of the other foot. Single support exists when either leg is in the swing phase.

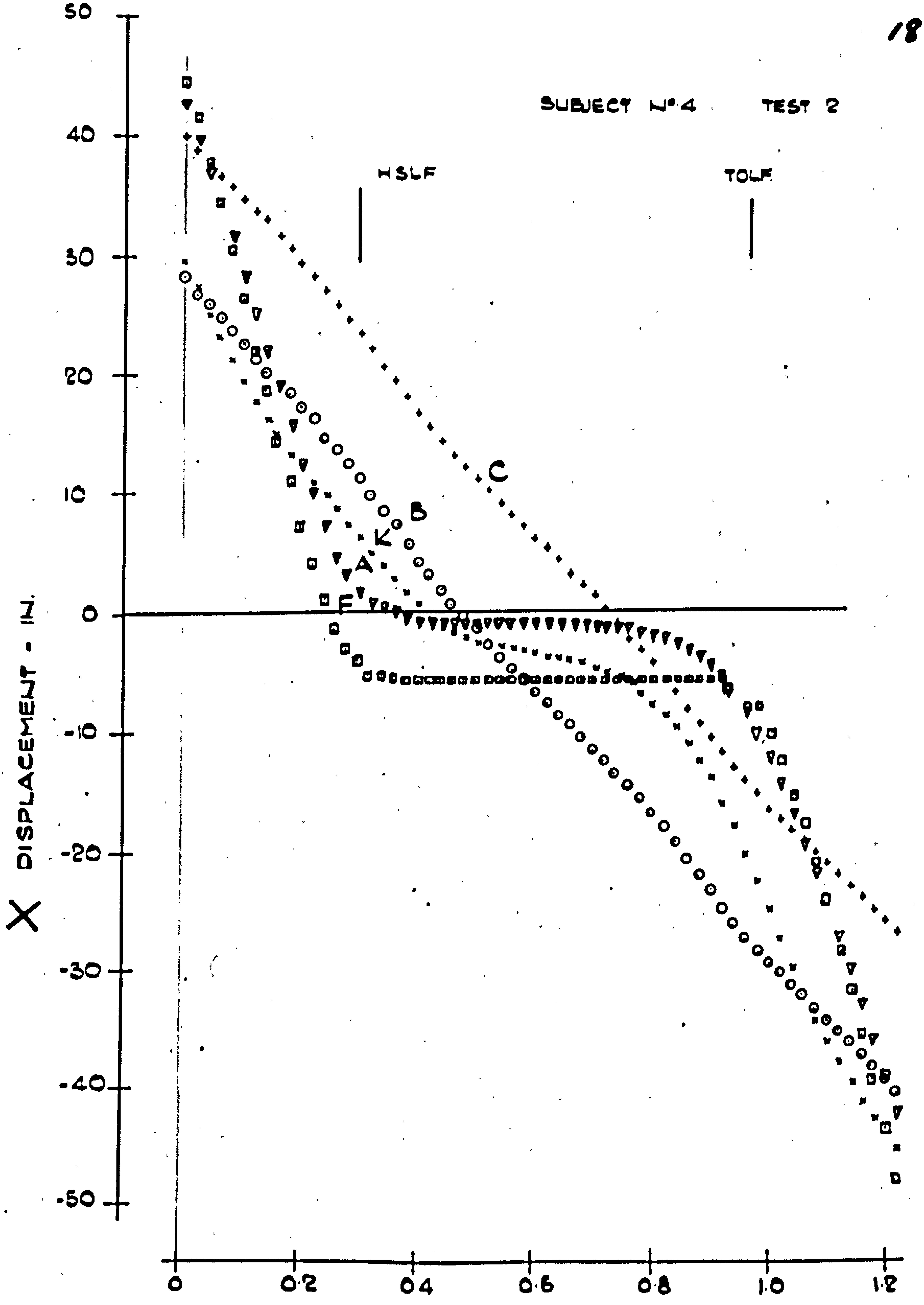
a) Displacements. In Fig. 58 the variation with time of the measured x - co - ordinates of the body markers is shown, x being measured from the centre of the force plate backwards along the line of progression. The markers B and C on the pelvis are seen to have relatively minor deviations from uniform velocity.



CONFIGURATION OF LEFT LEG AT SALIENT POINTS IN THE WALKING CYCLE.

TO BE SUPERIMPOSED ON FIGS. 58 - 72

INSERT FIG. 1



APPARENT X - VALUES OF MARKERS

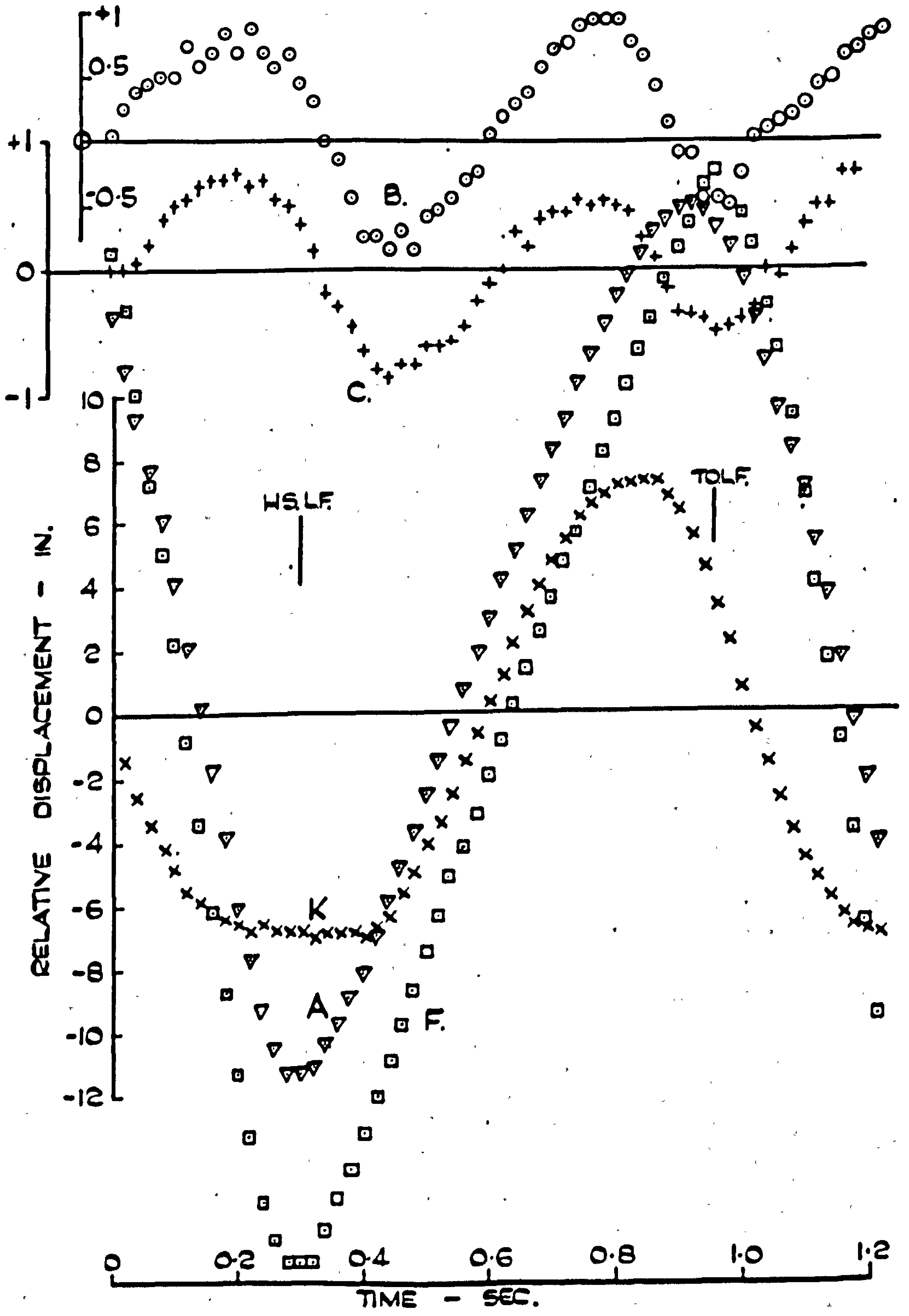
FIG 5 8

Markers further down the leg show considerably greater velocity variations. When plotted in this way small errors in readings were not obvious and a plot was made instead of the form shown in Fig. 59. In this the displacements of the markers are shown relative to a notional point moving at uniform velocity equal to the average speed of body movement. The C marker curve shows an undulation relative to the notional point of $\pm$ 0.6 in. Fischer (1895 etc) shows the corresponding curve for the centre of gravity for one subject as having amplitude of 0.43 in. but this is not, of course, directly comparable since the centre of gravity is not stationary relative to the body structure. For C the velocity of translation can be expressed approximately as:-

$$v_x = 57.4 + 7.3 \sin (11.2t - 3.6) \qquad (63)$$

where t is the time in seconds after heel strike, i.e. the undulations in velocity of C are only 13% of the mean whereas the maximum velocity of the foot is over three times the mean speed. The points in Fig. 58 and 59 are as read from the film, without connection of the parallax errors. The maximum parallax errors occur at each end of the cycle and are of the order $\pm$ 1.1, $\pm$ 0.2, $\pm$ 0.8, $\pm$ 1.0, $\pm$ 1.0 in for the C, B, knee, ankle and foot markers respectively.

SUBJECT No 4   TEST 2.

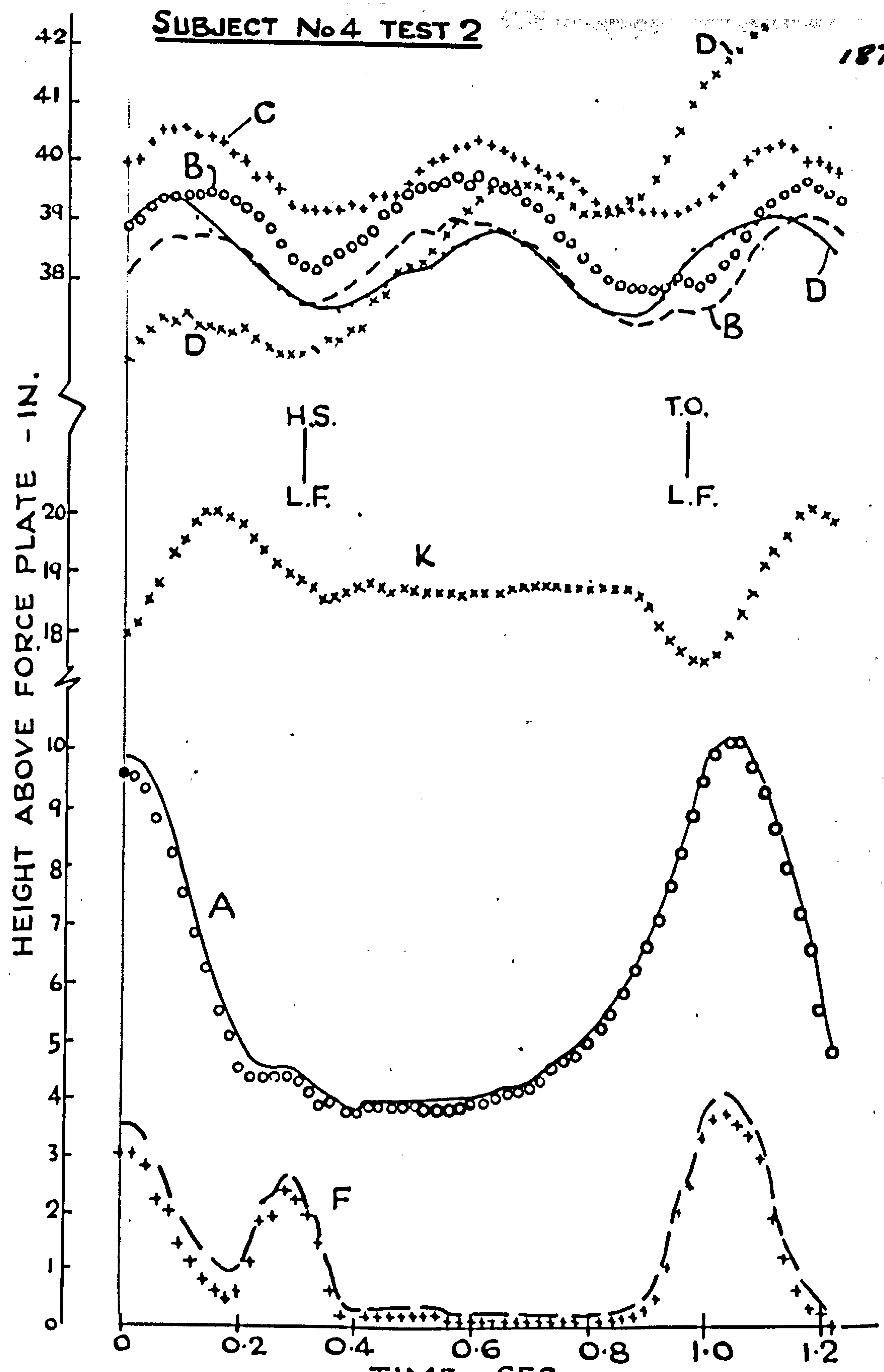


X-DISPLACEMENT OF MARKERS RELATIVE TO A NOTIONAL POINT MOVING WITH UNIFORM VELOCITY EQUAL TO THE AVERAGE SPEED.

FIG. 59

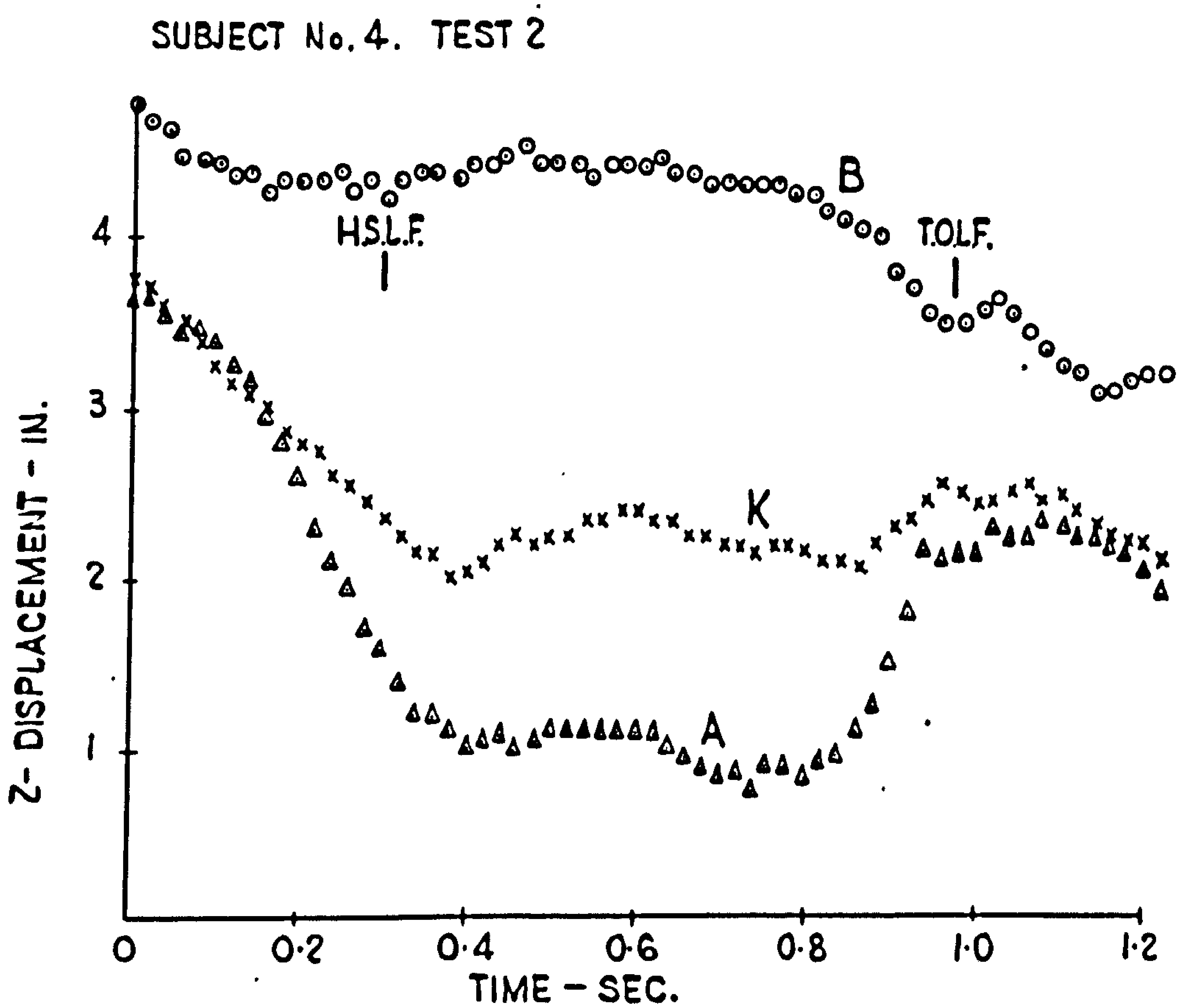
The variation of marker height, y, and the corresponding parallax errors are shown in Fig. 60. The large error for marker D arises since this is viewed through the x-axis camera and there is therefore considerable displacement of the marker relative to the measuring grid. The variation in height of the C marker which represents most closely the movements of the centre line of the trunk is $\pm$ 0.62 in. The ankle and foot markers show the largest undulation. The double undulation of the foot marker is found most useful in identifying phases of walking in the absence of other information. For the lateral or z displacement curves of Fig. 61 the maximum parallax correction was 0.7 in. but this was an appreciable fraction of the actual value of themeasurement. In this cycle the subject is seen to be moving from left to right by about 1.3. in. compared with the 57 inches of forward movement.

As shown in the theoretical analysis, the inclination of the limb segments relative to the reference axes is determined in the course of calculations and the variation with time of these angles is presented in Fig. 62. Features of interest here are the rapid angular acceleration $\ddot{\theta}_z$ of the foot and shank at heel strike and toe off. The values obtained for these acclerations, using equation (43), are - 74 and +89 rad/sec^2 for the foot and +59 and - 60 rad/sec^2 for



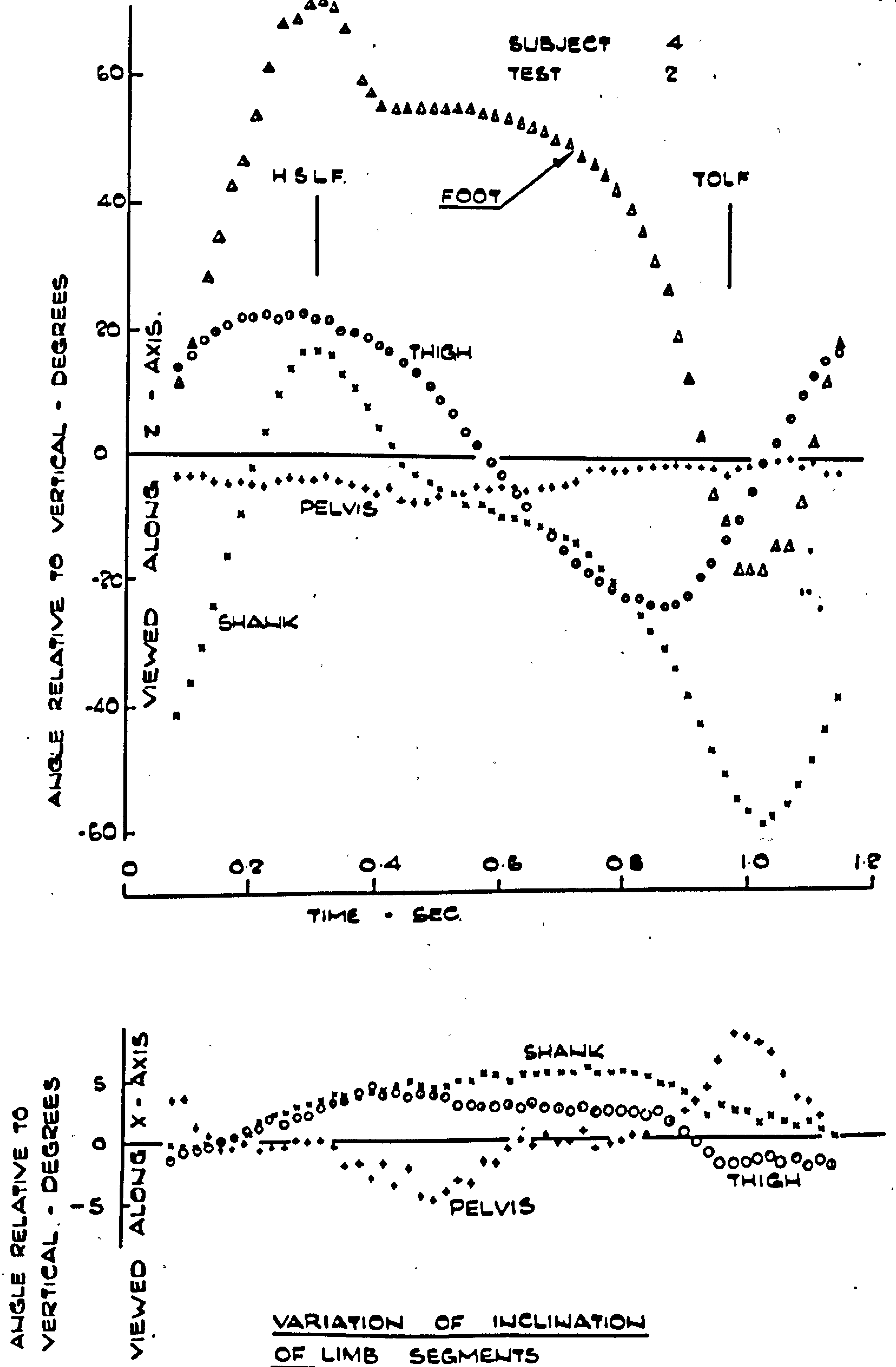
APPARENT AND CORRECTED Y VALUES OF MARKERS

FIG 60



CORRECTED Z-VALUES OF MARKERS

FIG. 61.



VARIATION OF INCLINATION OF LIMB SEGMENTS

FIG. 62

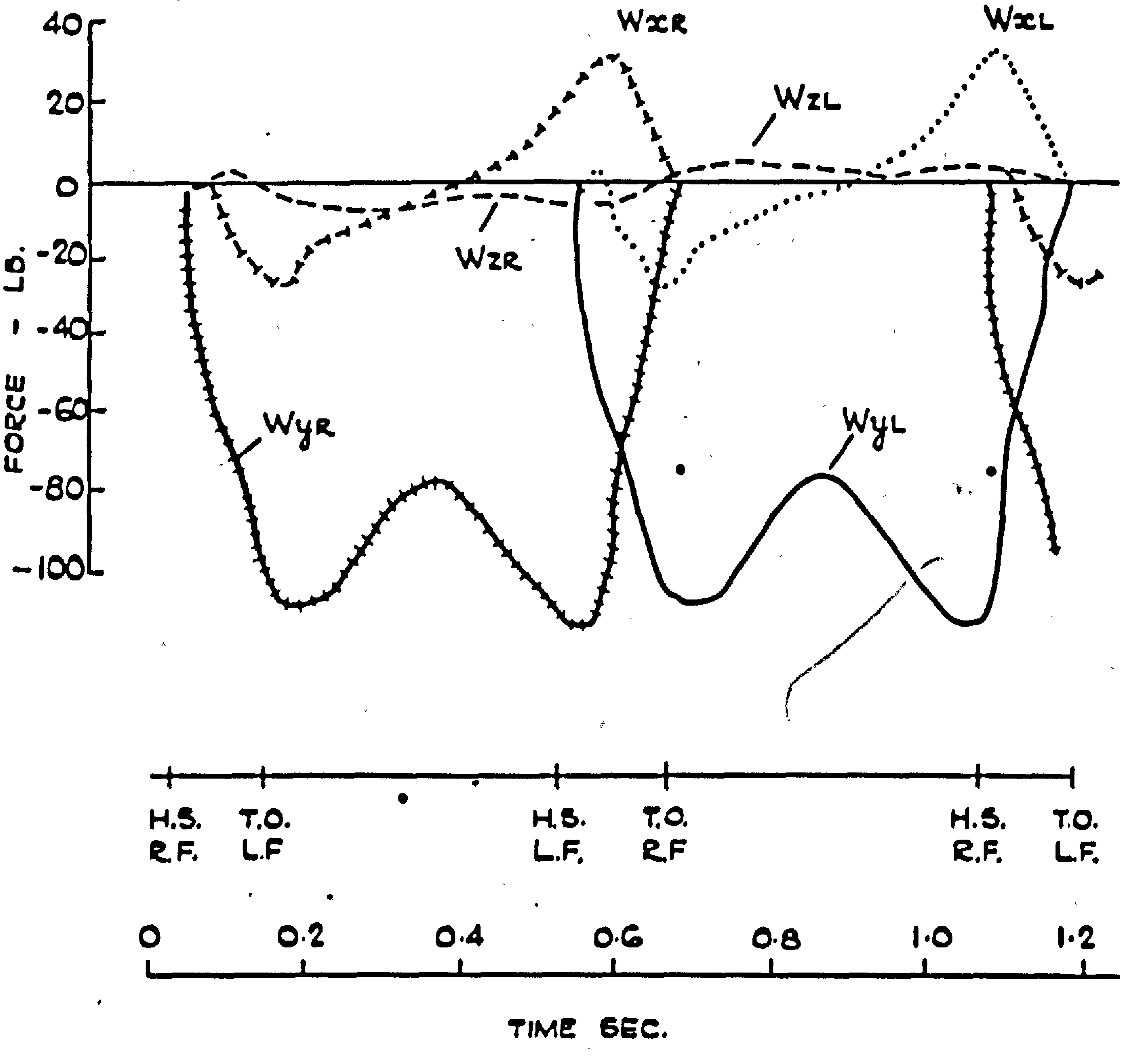
the shank respectively. When account is taken of the mass properties of the segments the maximum inertia torques are 22.5 and 22.0 lb. in. for the foot and the shank respectively. The maximum recorded moments at the hip are however over 300 lb. in.

b) External Force Actions.

The ground to foot force actions for this subject have already been shown in Fig. 43. The vertical component of force W_y shows the characteristic double peaked curve with the peaks occurring at the limits of the swing phase of the other foot. At these points the body is in its lowest positions and the ground force is therefore increased by the downward inertia force. The trough in the curve occurs at mid-stance when the body is at its highest position giving an upwards inertia force. This is counteracted to some extent by the centrifugal force of the swinging leg and the arms. Immediately after heel strike, the force in the direction of movement, W_x, has a characteristic short period in the positive sense, i.e. in the direction of movement. This feature, corresponds to a partial kick on the ground by the heel, varies in magnitude from subject to subject and is sometimes absent. Thereafter this force is negative until mid stance, and positive until toe-off. The greatest negative value of W_x here is 26 lb. which applied to the whole body mass of 127 lb. would cause a component acceleration of 79 in/sec^2. Equation (63) on differentiation gives a maximum acceleration of 82 in/sec^2 which is in surprisingly close correspondence considering that the movement of the marker does not necessarily correspond to that of the centre of gravity.

To get a complete picture of the external forces acting on the body Fig. 63 has been prepared to show together the forces on the left foot and what may be presumed to be the corresponding forces on the right foot. The latter have been obtained by redrawing the curves for the left foot in the appropriate time relationship. It is apparent for this subject that the maximum vertical component of force occurs at the middle of the double support phase and that the maximum forward force is exerted then. For the second half of the stance phase the forward and backward forces on the two feet are in opposition. The lateral force is alternately to right and left during left foot and right foot support respectively.

In the lower diagram of Fig. 43 the curves of moments of ground/foot forces about the force plate reference axes are shown. The



PRESUMED GROUND FORCE COMPONENTS ON RIGHT FOOT TOGETHER WITH VALUES FOR LEFT FOOT

FIG. 63

M_{wx} curve corresponds to the product of the vertical force component and the distance of its line of action laterally from the x axis, At the first instant of maximum force the offset is 172 lb. in./107 lb = 1.61 in. compared with z ankle of 1.1 in. It appears therefore that the subject is at this time bearing more on the outside of the shoe. At the next force peak the calculated offset is 1.58 in compared with 0.9 in from the film record. This is surprising since it would be expected that the greater load bearing would occur on the ball of the foot and the great toe. The M_{wz} curve corresponds to the product of the vertical force component and the distance of its line of action along the x axis. In this case the greater part of the curve is positive, indicating that soon after heel strike the line of the resultant lies forward of the force plate centre line. A curve of more familiar form from another test of the same subject is also shown. In it the M_{wz} term changes sign at mid-stance indicating more central placing of the foot on the force plate.

The form of the M_{wy} curve corresponds to two factors:-

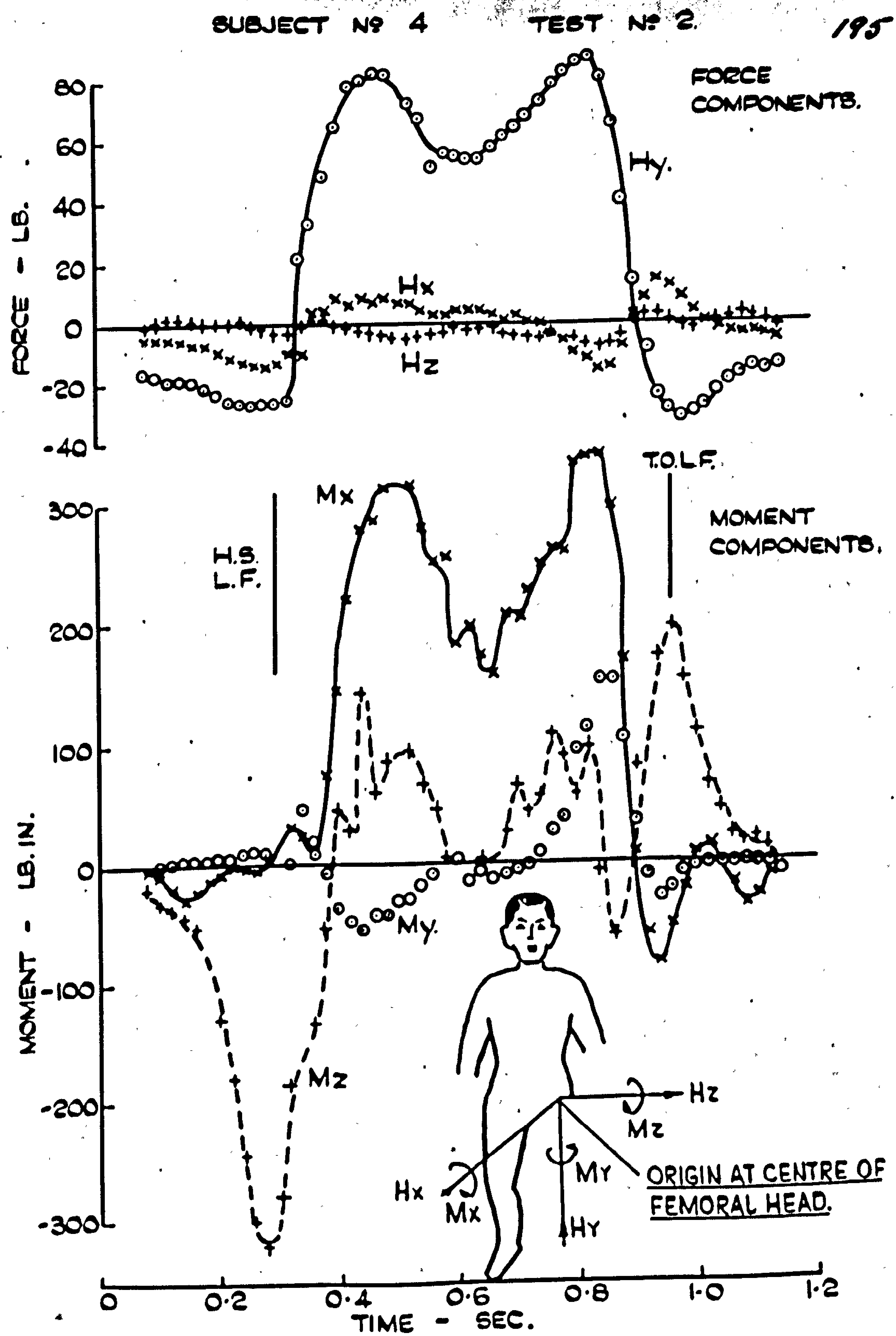
a) the moments about the centre of the force plate of the W_x and W_z forces.

b) the moment transmitted at the force plate/ foot interface by rotation or a tendency to rotate about an axis parallel to the y axis. The magnitude of themeasured moment is so small as to make it unreasonable to allocate its nature between a and b.

c) Leg/Trunk Force Actions.

From the external and the body force actions the resultant force actions transmitted across theleg/trunk section are calculated and the curves for this test are shown in Fig. 64. The curves for force components H_x, H_y, and H_z correspond closely to the external force curves of Fig. 43, although the vertical component is reduced by the weight of the left leg and shoe. During the stance phase the M_x curve is similar to that for the vertical component of ground to foot force, since the greater part of the final value is accounted for by the product of vertical ground force times lateral offset. Since it acts inward the lateral ground force increases the value of this moment, i.e. in the rolling or "sailor's" gait greater M_x values may be expected.

Typically M_z has a negative value corresponding to the action of the hip extensors prior to heel strike as the forward swing of the leg is decelerated. As contact is made with the force plate, the line of action of the resultant ground force passes



VARIATION OF RESULTANT FORCE AND MOMENT TRANSMITTED FROM LEG TO TRUNK.

FIG. 64

in front of the hip joint and extensor action is again required. Depending on the subject's walking habits, these two peaks of M_z may appear separately or merge as shown for this subject. Generally thereafter M_z diminishes to zero at mid-stance and has a maximum positive value in the region of toe-off. This is due to the ground reaction whose line of action then passes behind the hip joint and also to the inertia force action of the left leg as it is accelerated into the swing phase. The graph for this subject is non-typical in respect of the swing to positive M_z immediately after the negative peak. It was noticed that this subject's foot had once slipped slightly on the force plate and this may be a nervous response to prevent its recurrence.

At the instant of toe-off of the right foot, time 0.44 sec, the contributions of the various external and body forces to the resultant hip force actions are shown in table 6 . At this phase of the cycle, acceleration forces are small but are still significant fractions of the external forces. For M_z the gravity and inertia actions are of the same order of magnitude as the net M_z due to external force actions. It should be noticed also that the M_x and M_z values involve subtraction of numbers of similar orders of magnitude, and that the accuracy of the results will be adversely affected by this.

HIP FORCE ACTION	FORCE ACTION FROM GROUND	FOOT	SHANK	THIGH	TOTAL
Hx - LB.	24·5	-0·5	-4·6	-9·6	9·8
Hy - LB	107	-2·5	-6·1	-13·6	84·8
Hz - LB	-3	0	0	0	-3
Mx - LB. IN					
FROM Y FORCES	337	-5	-10	-6	
" Z "	109	0	0	0	
" COUPLES	-160	0	0	0	
TOTAL	286	-5	-10	-6	265
My - LB. IN.					
FROM X FORCES	-77	1	7	20	
" Z "	11	0	0	0	
" COUPLES	-23	0	0	0	
TOTAL	-89	1	7	20	-61
Mz - LB. IN.					
FROM X FORCES	890	-17	-11	-74	
" Y "	-396	15	27	28	
" COUPLES.	-384	1	6	-8	
TOTAL	120	-1	+12	-54	77

FORCE ACTIONS AT HIP

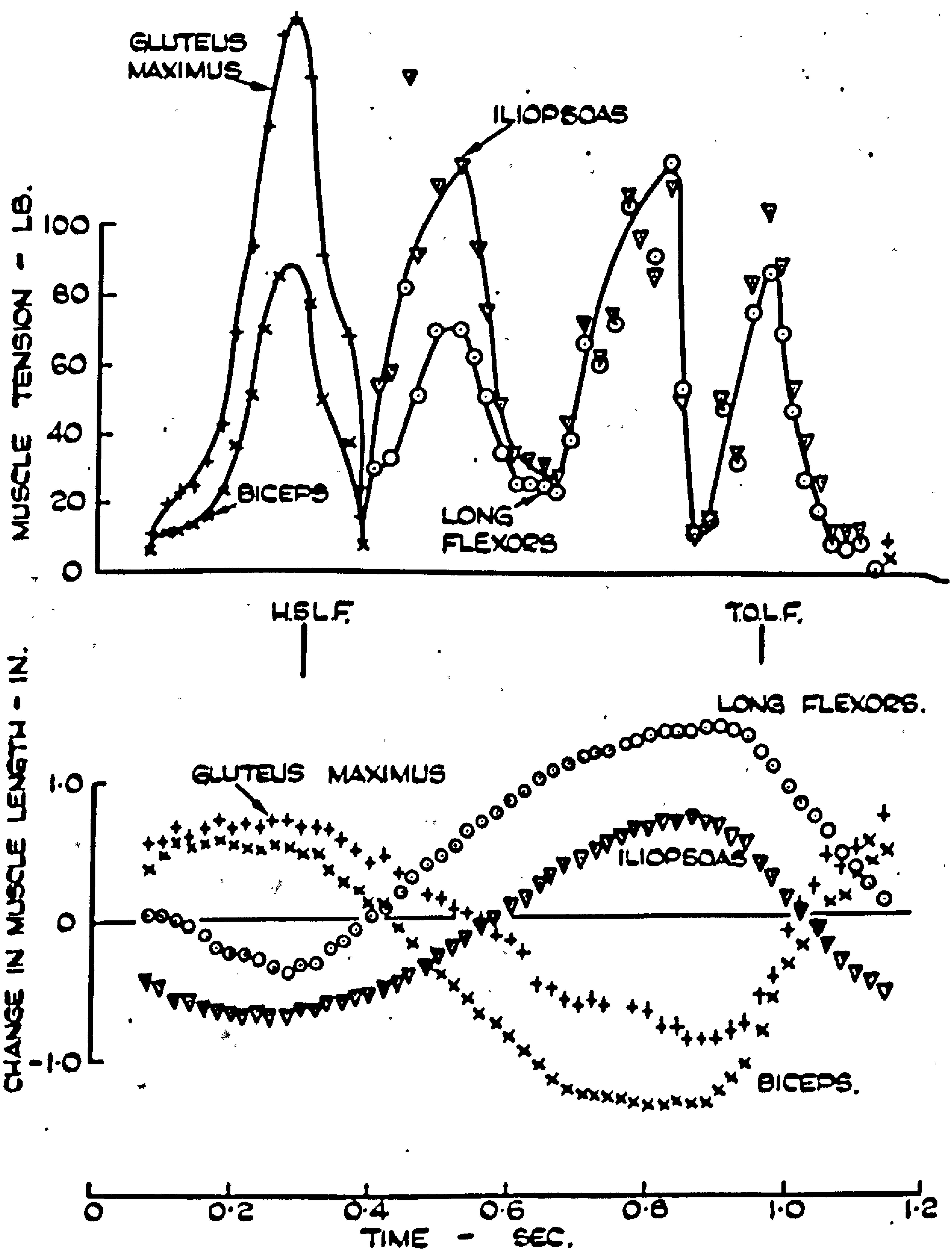
SUBJECT 4, TEST 2. TIME 0.44 SEC.

TABLE 6.

d) Muscle Forces.

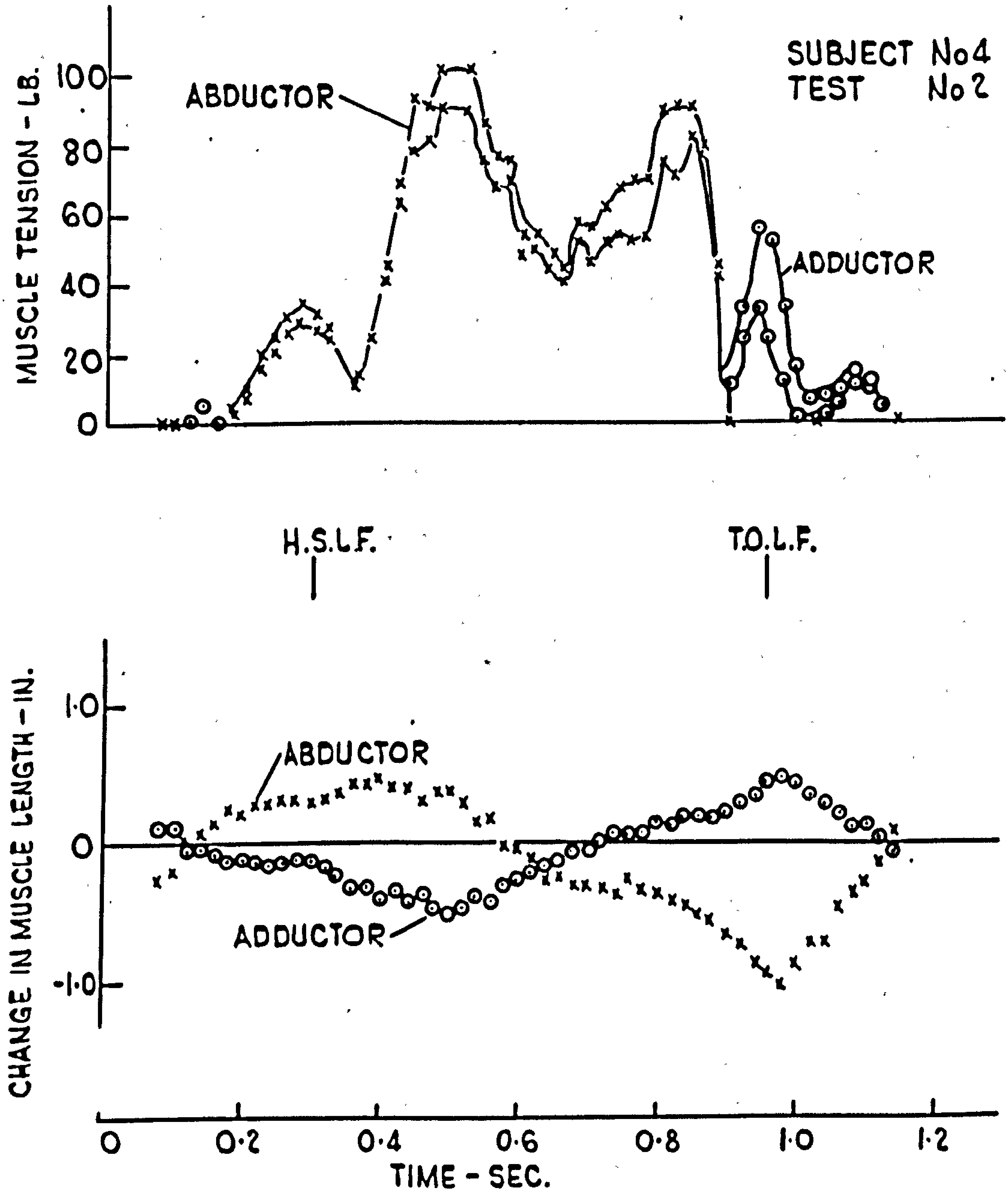
The muscle force curves shown in Figs. 65 and 66 follow in form the corresponding hip moment curves of Fig. 64. Where the thigh is flexed i.e. from mid-swing through heel strike to mid stance, there is a large difference between the forces in the long and short flexors and extensors due to the divergence of their lines of action. After mid-stance the thigh is extended and the long flexors and ilipsoas which are active then have lines of action which are adjacent. There is therefore little difference between the two possible solutions. It may also be contended that in this region, when the thigh is extended on the hip, ligamentous restraints will be sufficient to resist the moment tending to cause further extension. Since the ligaments are closer to the joint centre this would involve greater forces in the ligaments than in the muscles. At this phase of movement, also, the moment at the knee is tending to cause flexion there. This would be resisted by the quadriceps and the tension in rectus femoris is effective at the section through the hip also. If the hip moment is carried by the capsular ligaments, iliopsoas and rectus femoris with sartorius, the values of the forces shown in Fig. 65 will obviously be reduced, but the value of the resultant joint force will not be

SUBJECT № 4. TEST № 2.



VARIATION WITH TIME OF MUSCLE FORCE AND MUSCLE LENGTH.

FIG. 65



VARIATION WITH TIME OF MUSCLE FORCE AND MUSCLE LENGTH

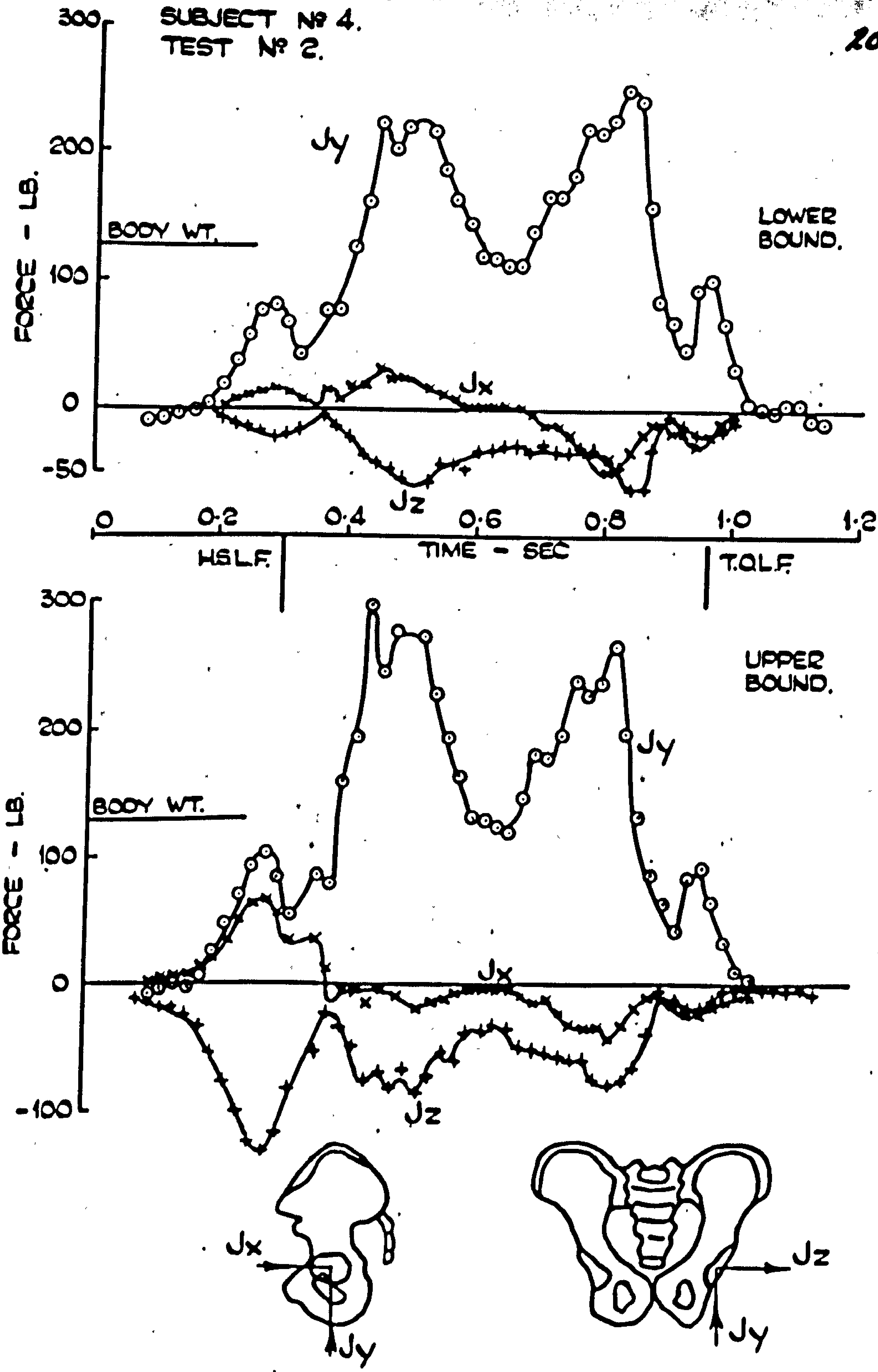
FIG. 66

greatly affected. The curve corresponding to the action of rectus femoris in carrying the moment alone will obviously give a lower bound to the region in which the correct solution lies.

Figs. 65 and 66 also show the change in length of the muscles relative to their length in the standard standing position. It is seen that these correspond approximately to the phenomenon described by Elftman (1967) namely that economy of use of muscles is obtained in normal activities by virtue of the muscles being extended from their resting position at times when they are required to develop most force.

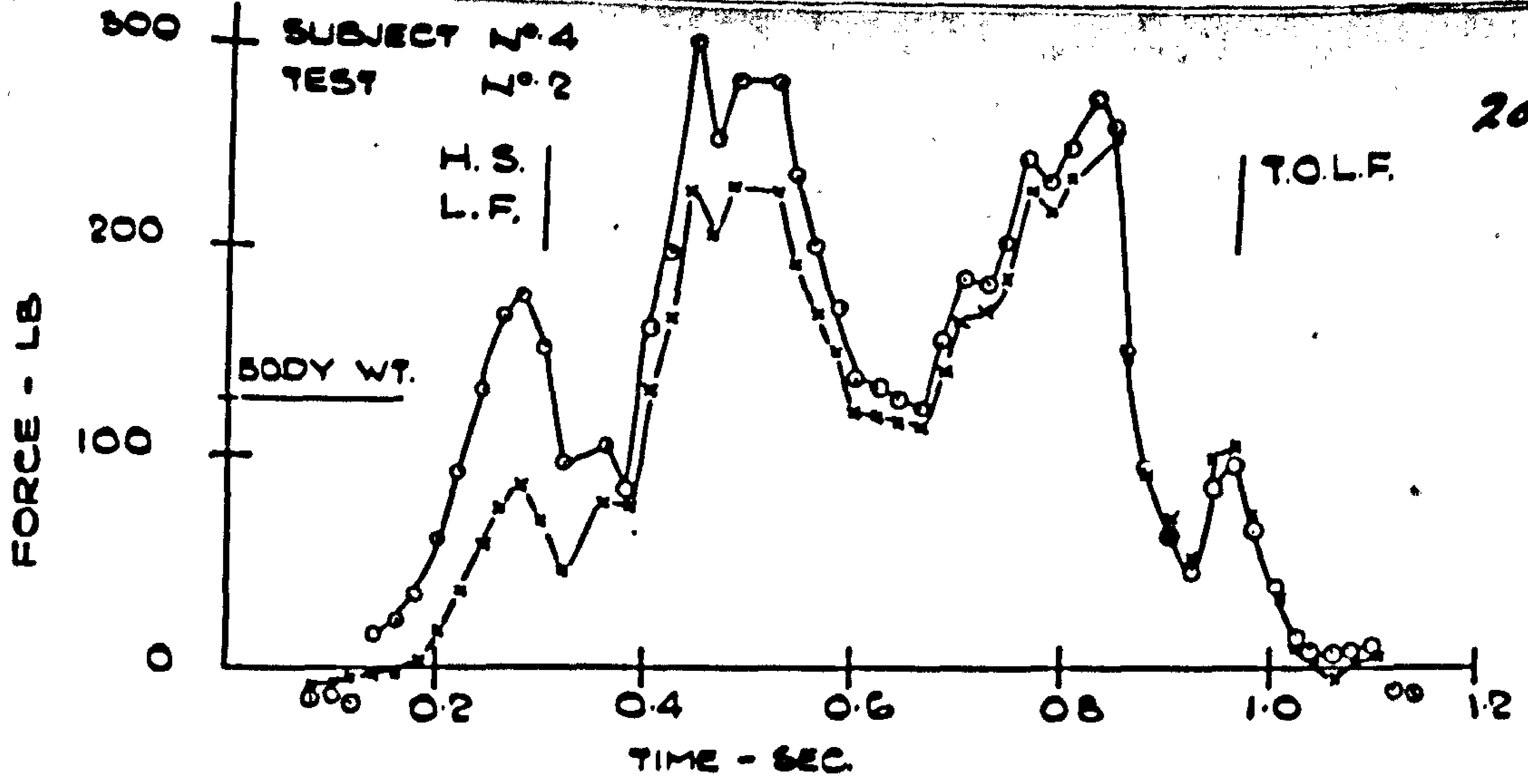
e) Joint Force.

As described in the theoretical analysis it is now possible to obtain the components of the hip joint force as shown in Fig. 67. Here separate graphs are drawn for the upper and lower bound values. The curves for the resultant forces and their inclinations to the reference axes are shown in Figs. 68, 69, and 70. It is seen that for this subject the maximum joint force has a value between 230 and 300 lb, occurs at toe-off of the opposite foot, and is directed upward, inward and very slightly backward on the acetabulum. The second highest force has a value between 260 and 280 lb, occurs at heel strike of the opposite foot and is directed upwards, inwards and slightly



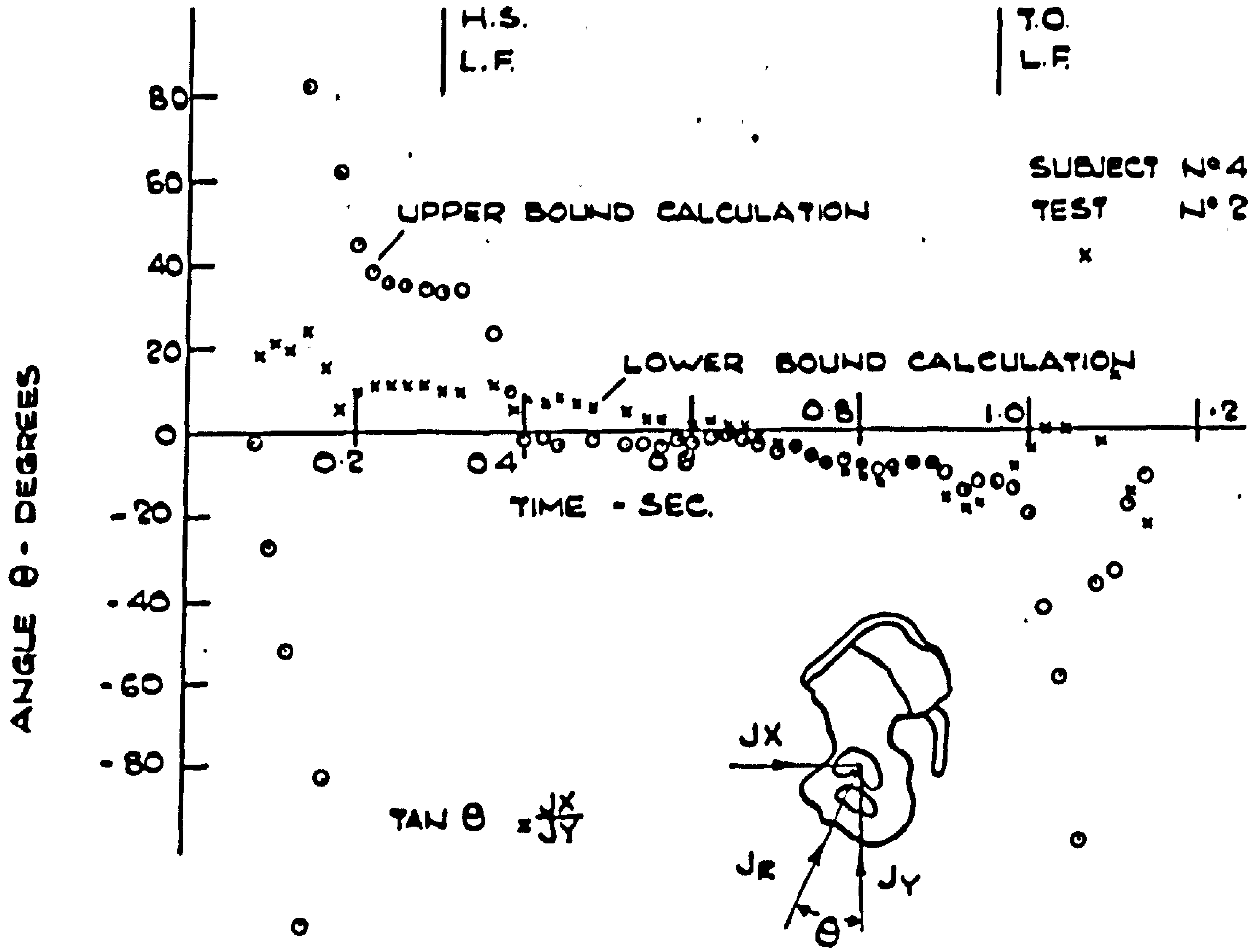
VARIATION WITH TIME OF COMPONENTS OF LOWER AND UPPER BOUNDS OF HIP JOINT FORCE.

FIG. 67



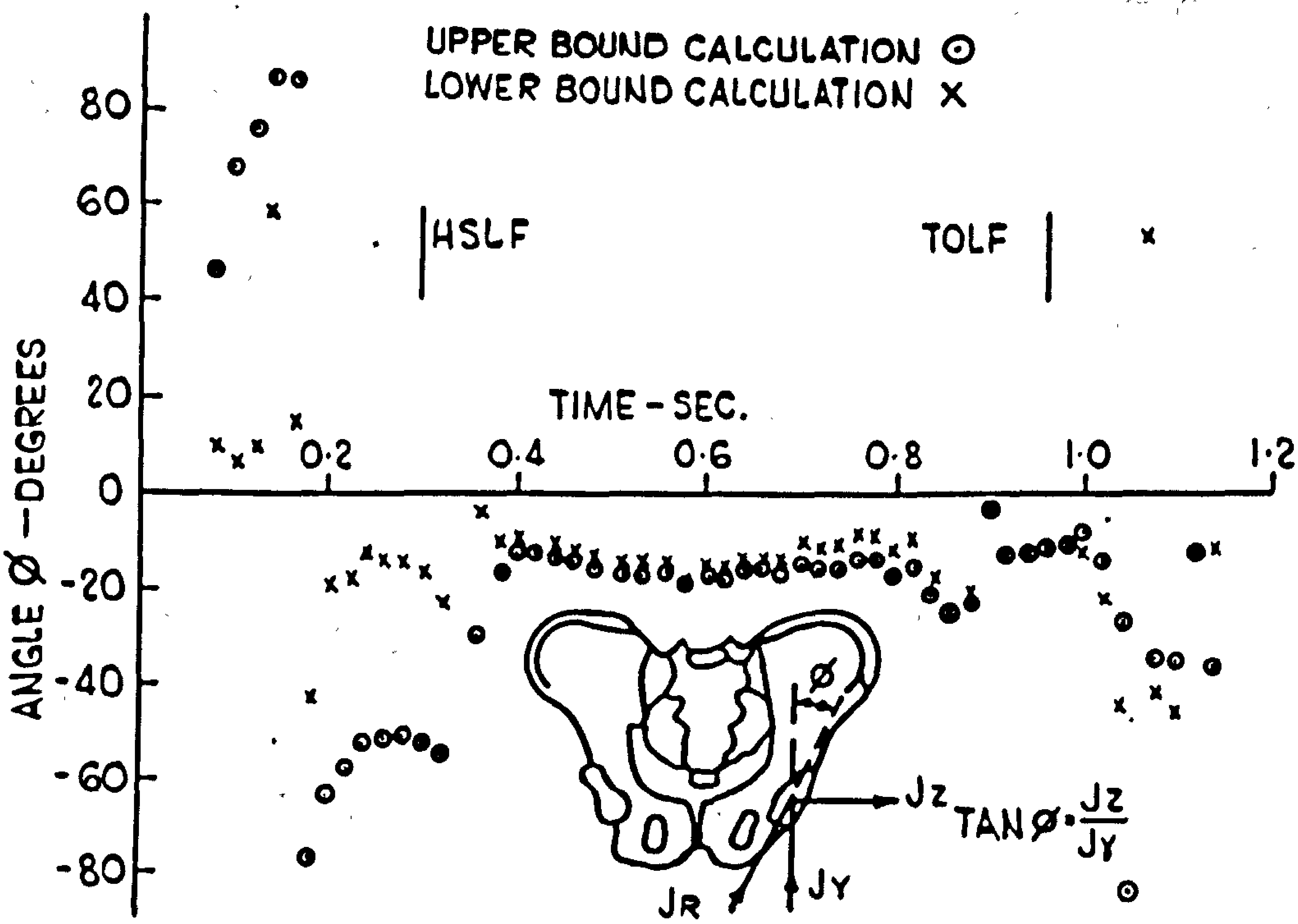
VARIATION WITH TIME OF RESULTANT JOINT FORCE. J_R.

FIG. 68.



θ · INCLINATION TO THE VERTICAL OF THE PROJECTION OF THE RESULTANT JOINT FORCE VECTOR ON THE YOX. PLANE.

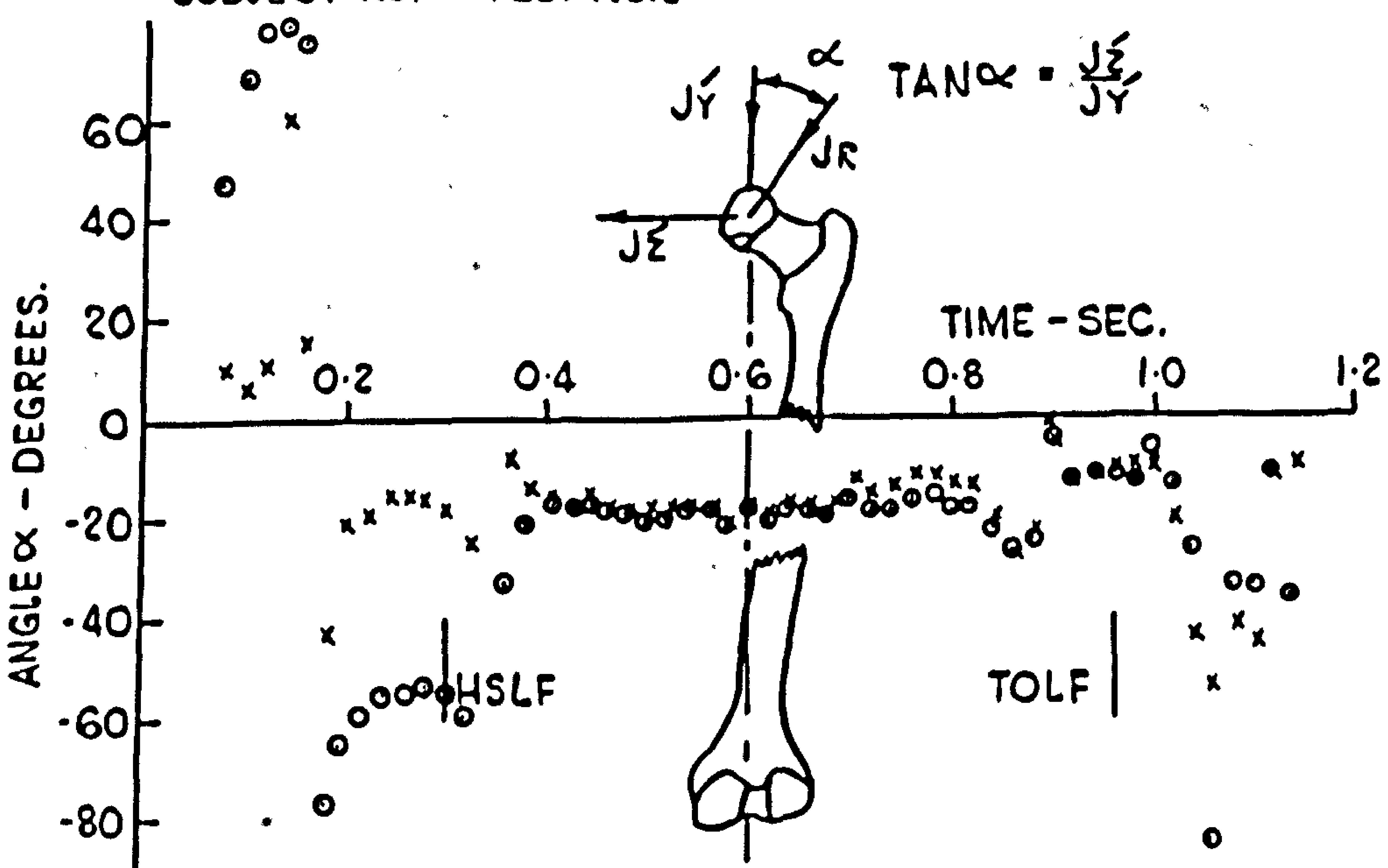
FIG. 69.



Ø-INCLINATION TO THE VERTICAL OF THE PROJECTION OF THE RESULTANT JOINT FORCE VECTOR ON THE YOZ PLANE.

FIG. 70

SUBJECT No.4 TEST No.2



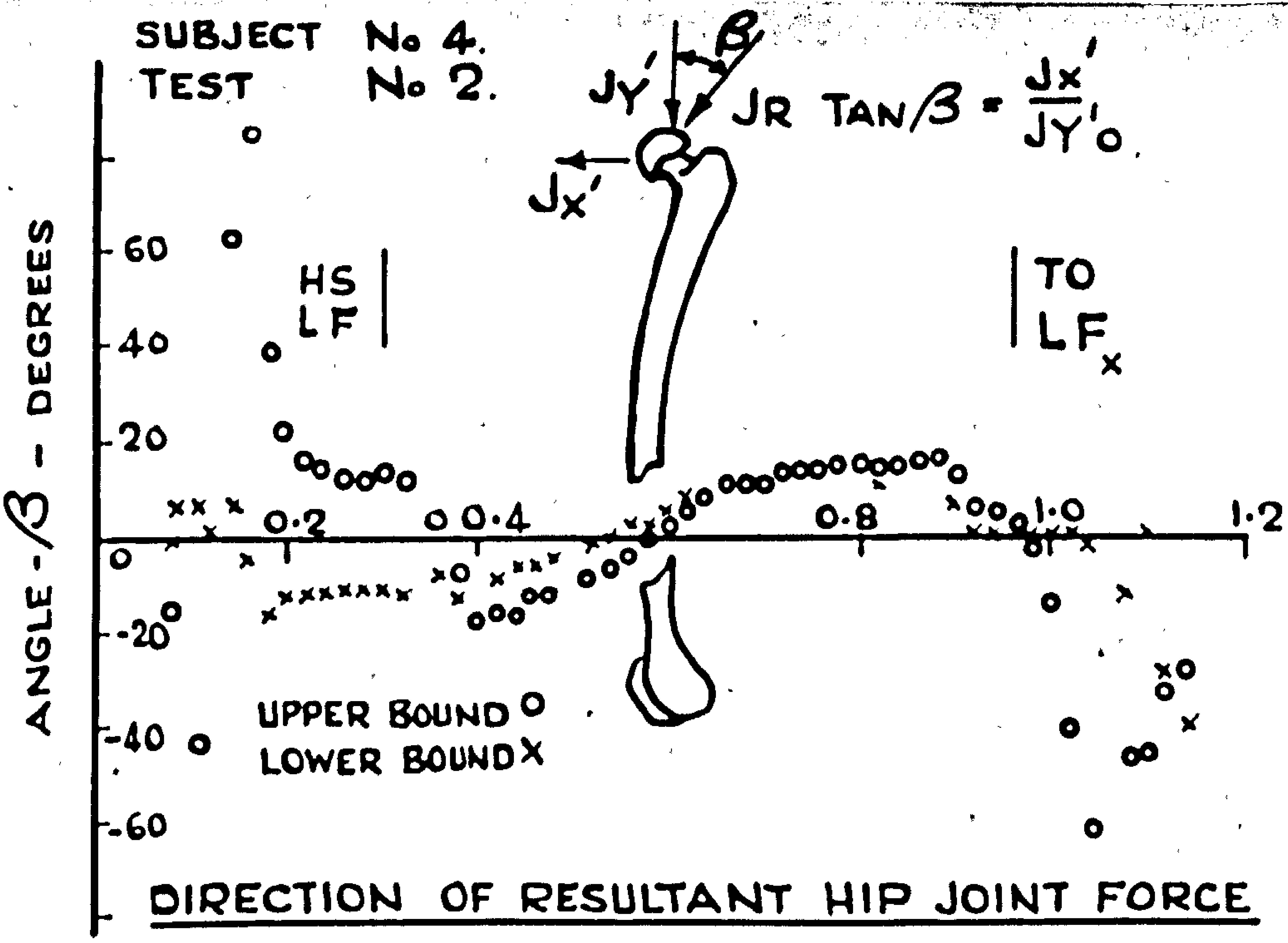
DIRECTION OF RESULTANT HIP JOINT FORCE RELATIVE TO FEMORAL HEAD – FRONTAL VIEW.

FIG. 71

forwards on the acetabulum. Relative to the femur the directions of the joint forces are shown in Figs. 71 and 72. Referring to the femur in a reference position with its line of joint centres vertical the force is vertically downward, outward and ± 15 degrees from the vertical in the anterior posterior plane.

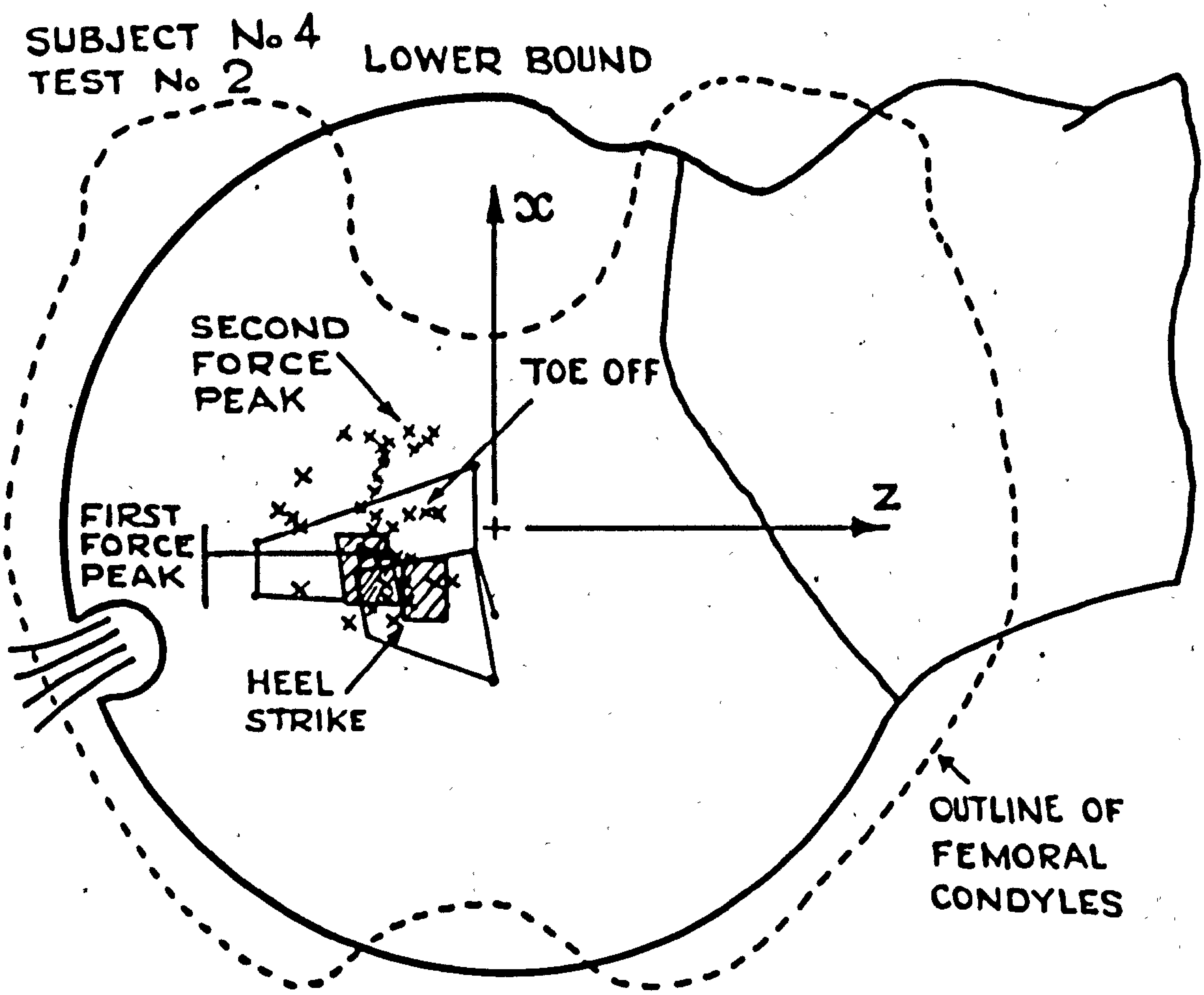
The major features of the lower bound curve for resultant hip joint force can be described by six parameters as shown in Fig. 75, and the ranges and means of these for 18 test runs are shown in the Figure. The first maximum value, Ra, is less than the second in all except four tests, and occurs on average 13 per cent of the cycle after heel strike. Toe-off of the opposite foot occurs generally 14 per cent after this heel strike. This turning value corresponds to the maximum values of M_x and M_z at the hip as already discussed. The minimum turning value Rb has magnitude of approximately 1.1/4 times body weight and occurs 32 per cent of the cycle after heel strike; mid-swing of the opposite leg occurs at the same time. This minimum value occurs since the fore and aft moment is zero and the abductor moment is at a minimum corresponding to the minimum value of the ground to foot vertical component of force.

The second maximum R_c is generally the greater and is almost co-incident in time with the heel strike of the opposite foot.



DIRECTION OF RESULTANT HIP JOINT FORCE RELATIVE TO FEMORAL HEAD - SIDE VIEW.

FIG. 72



INTERSECTION OF RESULTANT FORCE VECTOR WITH FEMORAL HEAD

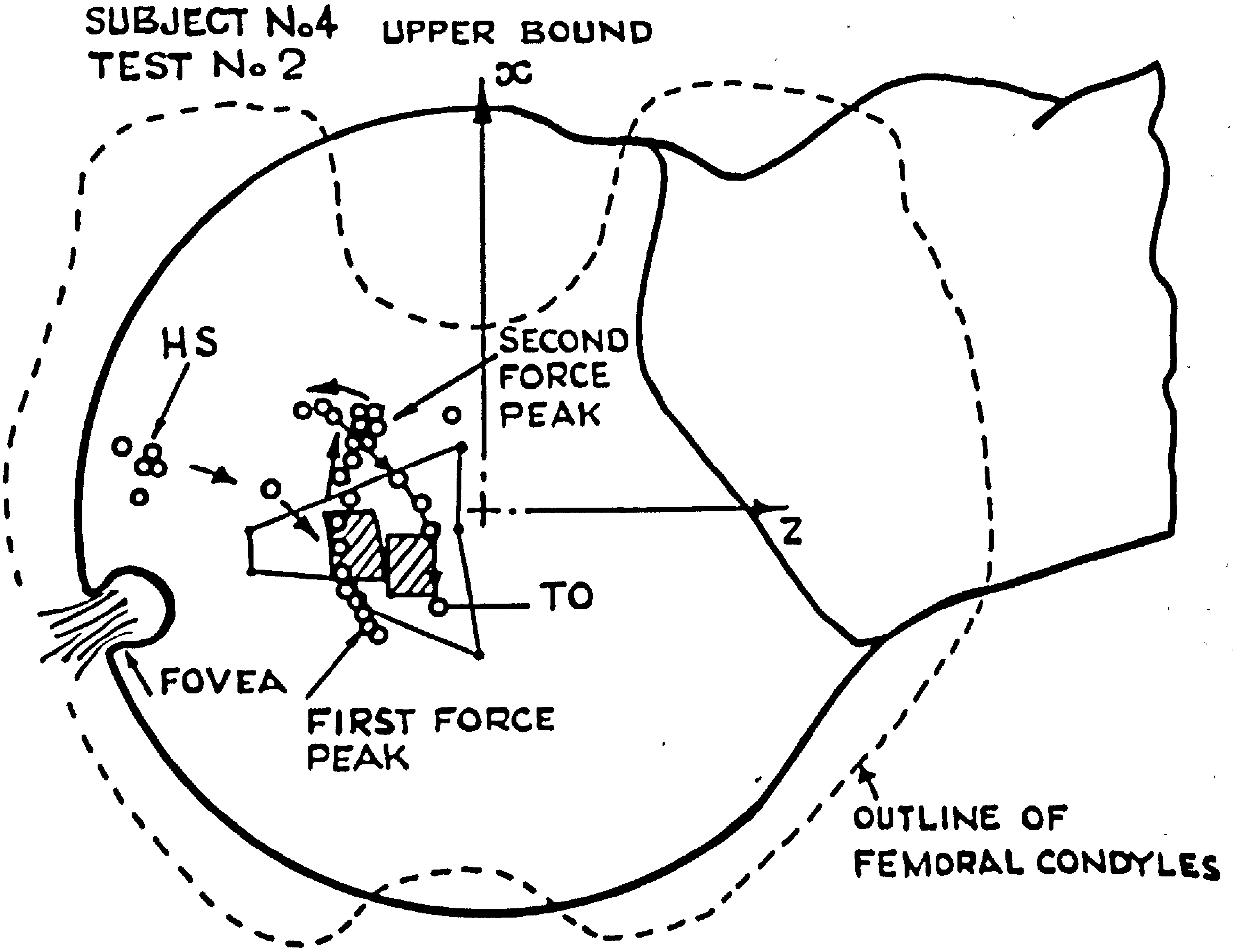
FIG. 73

These average figures presented correspond to the selected walking speed of the test subjects, and are therefore affected by the variables of their speed, stride length and stature. These factors are considered under statistical analysis.

The intersections of the line of action of the resultant force with the surface of the femur are shown in Figs. 73 and 74. In the swing phase, when the y component of joint force is small, these inclinations are of doubtful accuracy. The shaded and the outlined areas are those within which Rydell (1966) found the force to be in the stance and swing phases respectively.

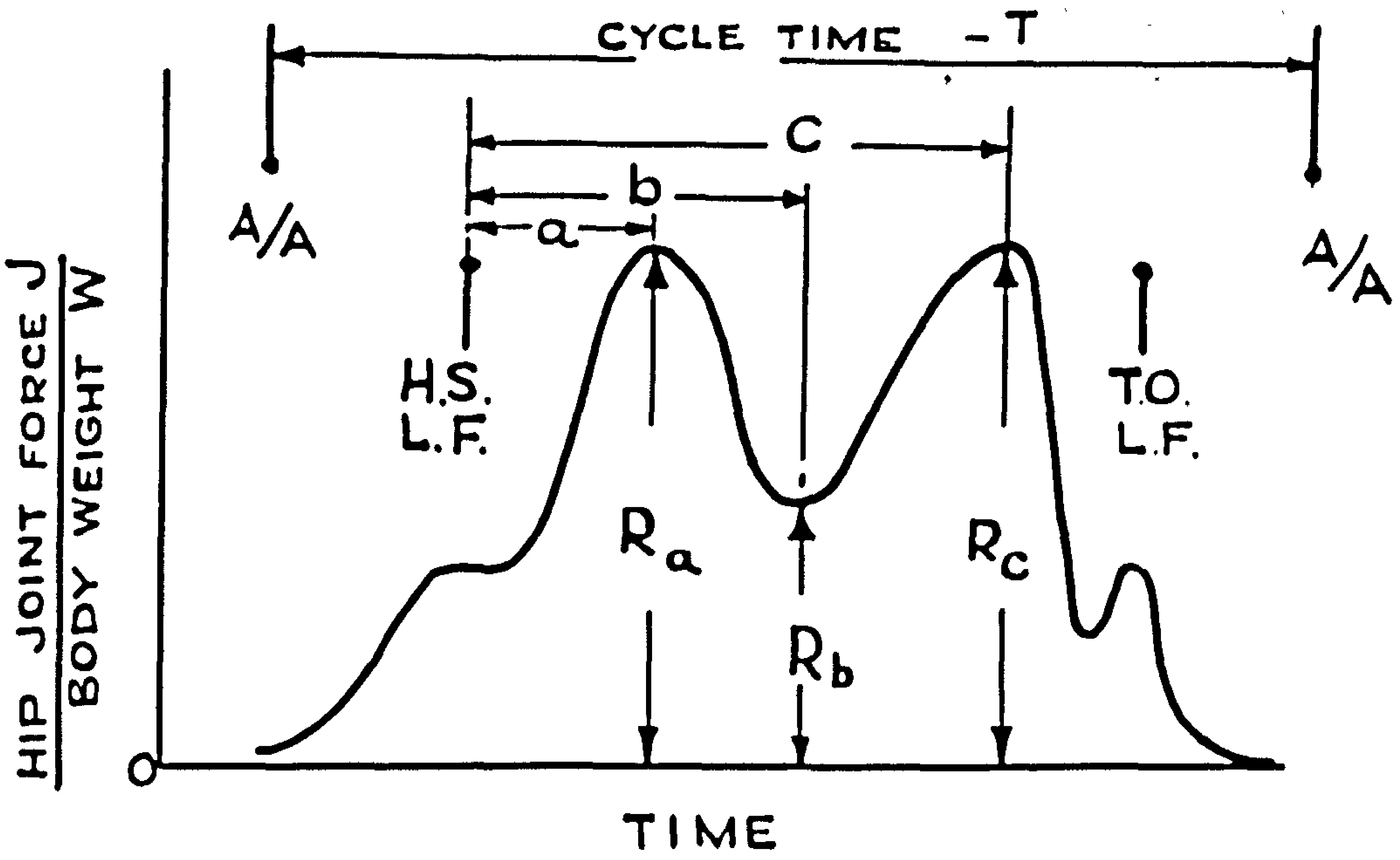
2. Comparison of Results with Published Work.

The only three dimensional analysis of leg/trunk forces and moments known is that of Bresler and Frankel (1950). Their curves for hip moments for four subjects are shown in Fig. 76 The curves for all the tests in the present series are shown in Figs. 77, 78 and 79. The envelopes of Bresler and Frankel's curves are drawn in Figs. 80 and 81 to the same scale as the summary graphs for the male subjects of the present series. The forms of the curves are closely related. Bresler and Frankel show higher values for M_x the adducting moment and lower values of M_z the flexing moment than found here. It is



INTERSECTION OF RESULTANT FORCE VECTOR WITH FEMORAL HEAD

FIG. 74

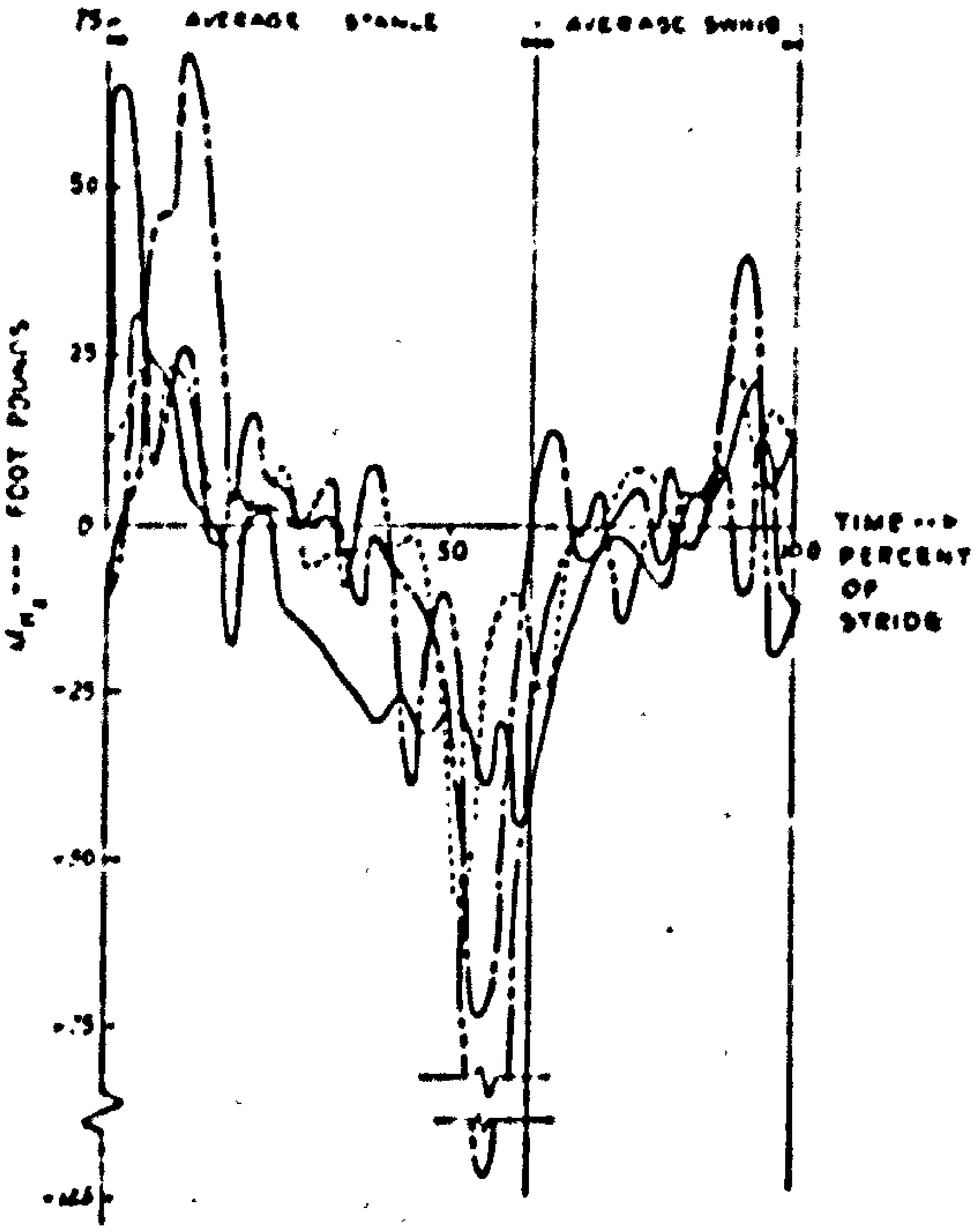


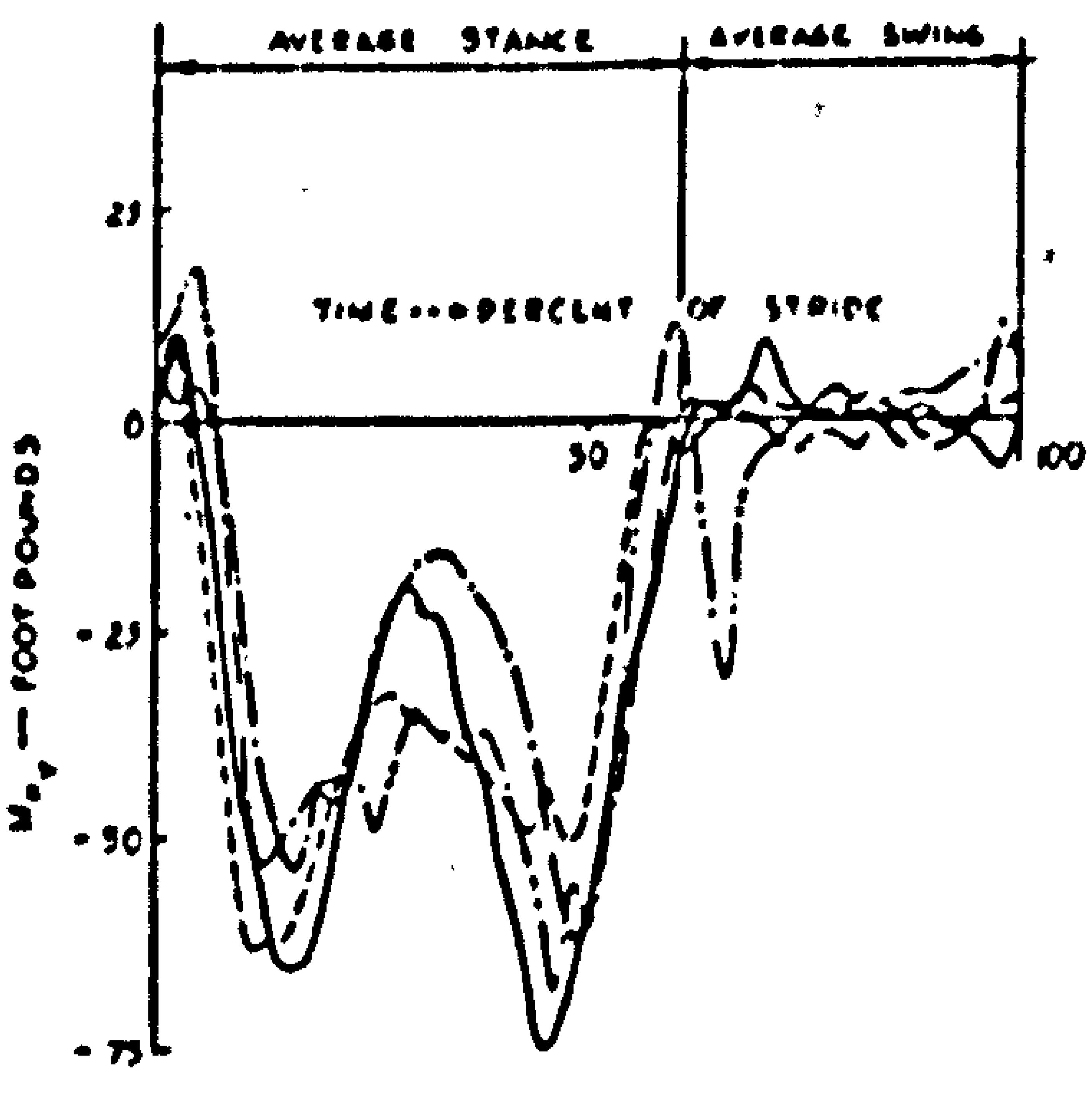
TYPICAL CURVE OF JOINT FORCE TO BODY WEIGHT RATIO WITH TIME

QUANTITY	MIN.	MAX.	MEAN
a/T	0·04	0·20	0·13
R_a	1.74	5·67	3·22
b/T	0·26	0·41	0·32
R_b	0·80	2·30	1·29
c/T	0·45	0·54	0·51
R_c	2·01	9·23	4·41
T sec.	1·02	1·24	1·13

GENERAL DESCRIPTION OF VARIATION WITH TIME OF LOWER BOUND OF RESULTANT HIP JOINT FORCE

FIG. 75.

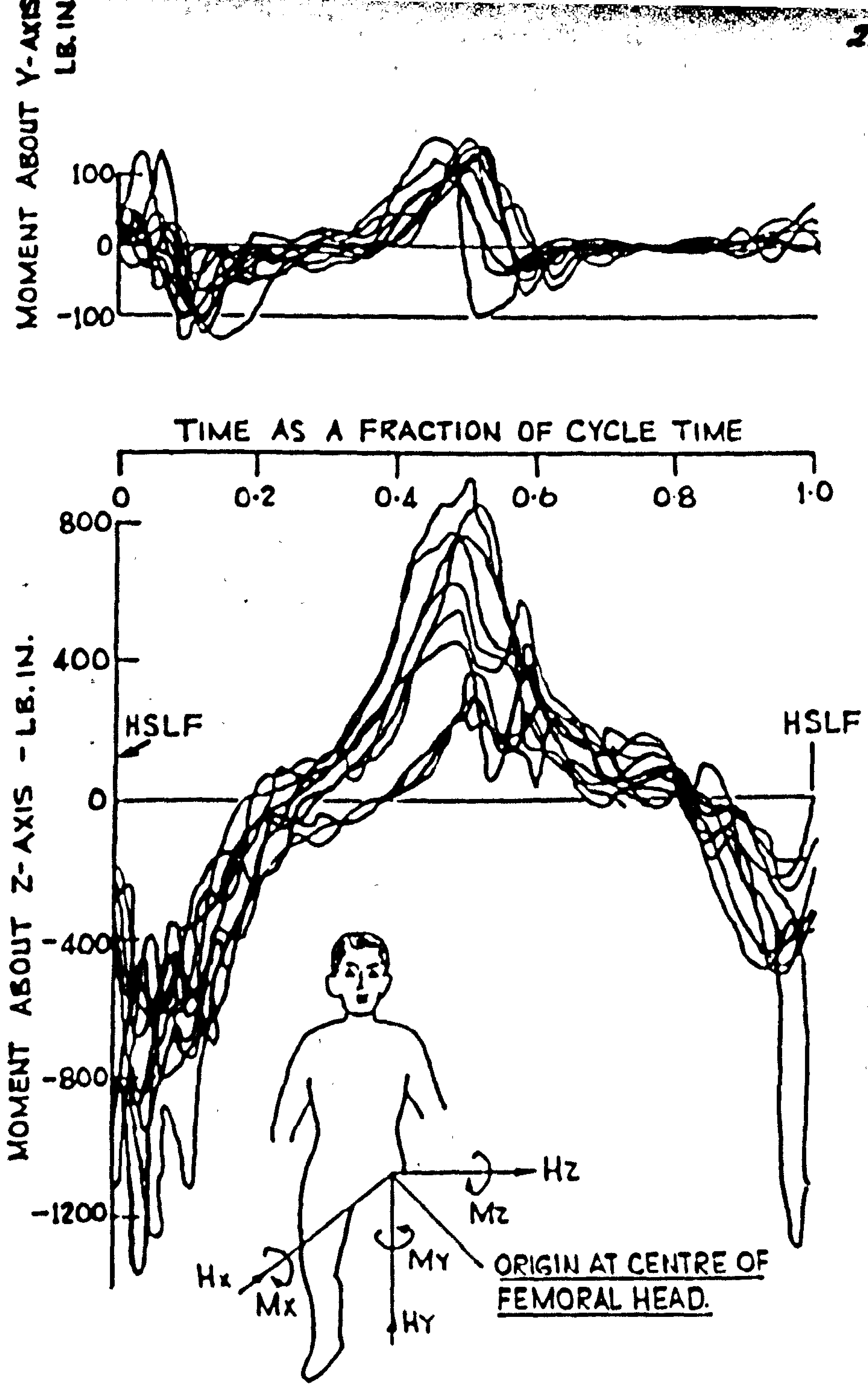




Leg to Trunk Moments from

Bresler and Frankel (1950)

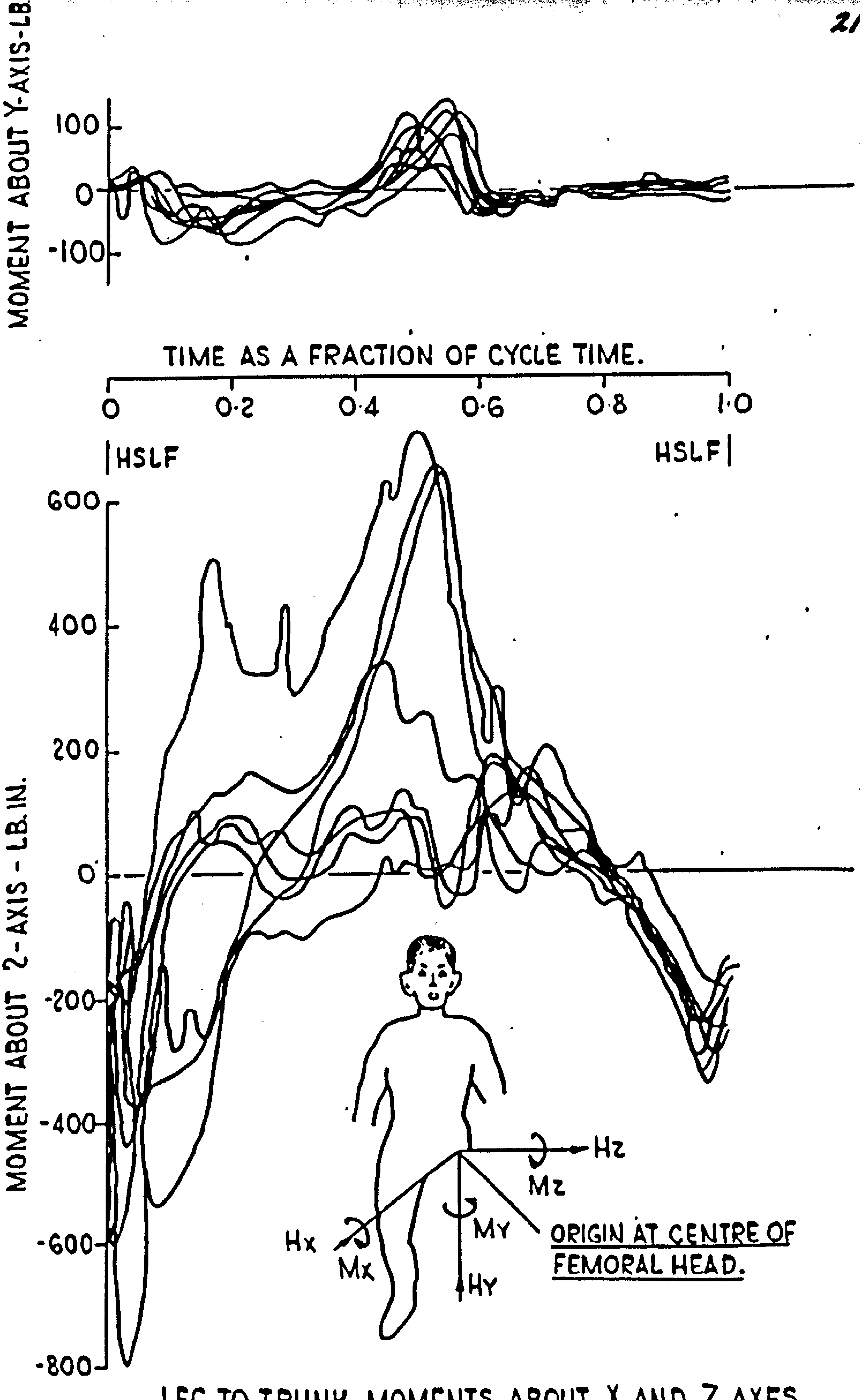
Fig.76.



LEG TO TRUNK MOMENTS ABOUT Y AND Z AXES

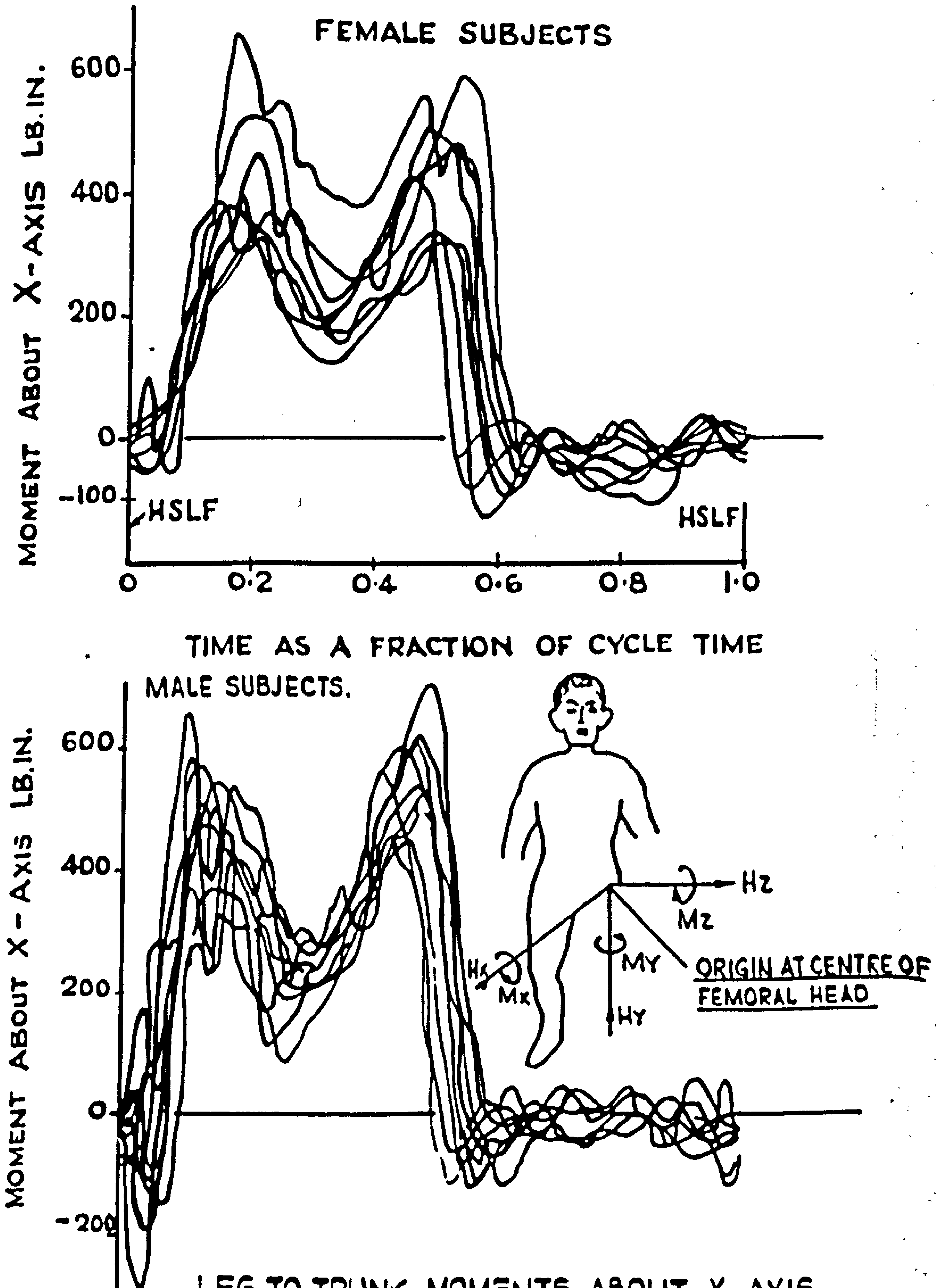
SUMMARY CURVES FOR MALE SUBJECTS

FIG. 77



LEG TO TRUNK MOMENTS ABOUT Y AND Z AXES
SUMMARY CURVES FOR FEMALE SUBJECTS.

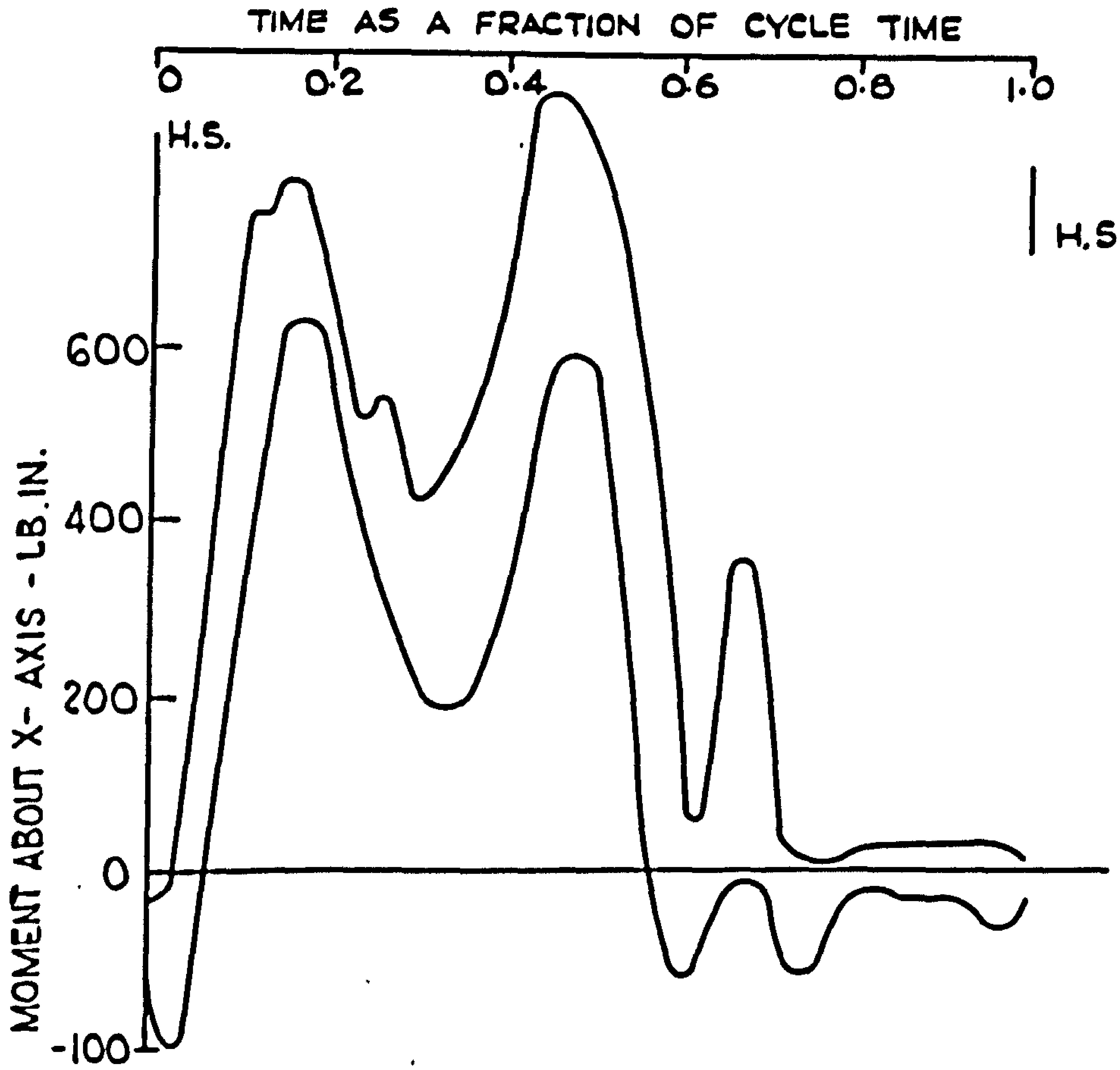
FIG. 78.



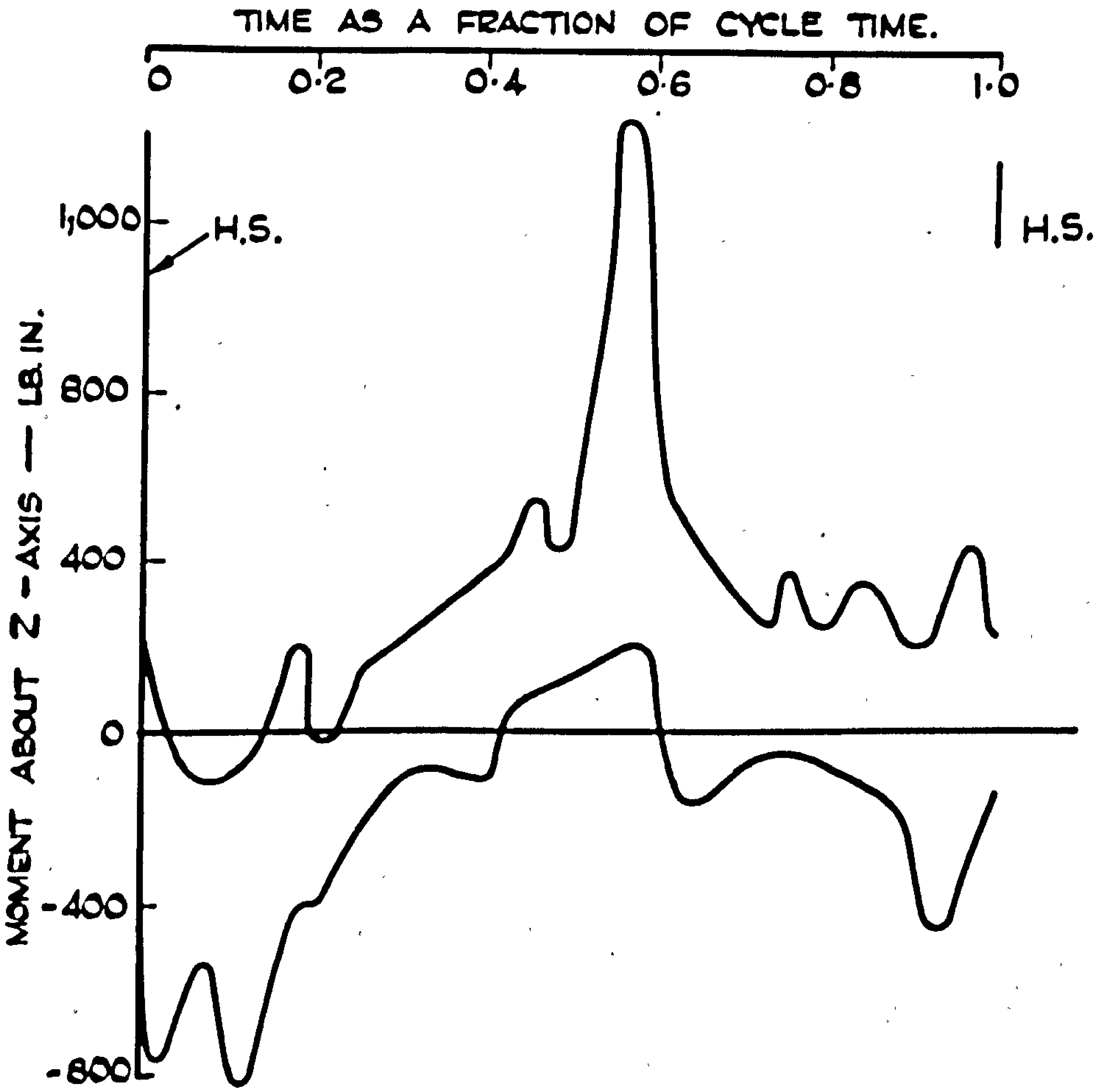
LEG TO TRUNK MOMENTS ABOUT X-AXIS.

SUMMARY CURVES FOR MALE AND FEMALE SUBJECTS.

FIG. 79



LEG TO TRUNK MOMENTS ABOUT X- AXIS
ENVELOPE OF CURVES FOR FOUR MALE SUBJECTS
FROM BRESLER AND FRANKEL (1950)
FIG. 80



? Z

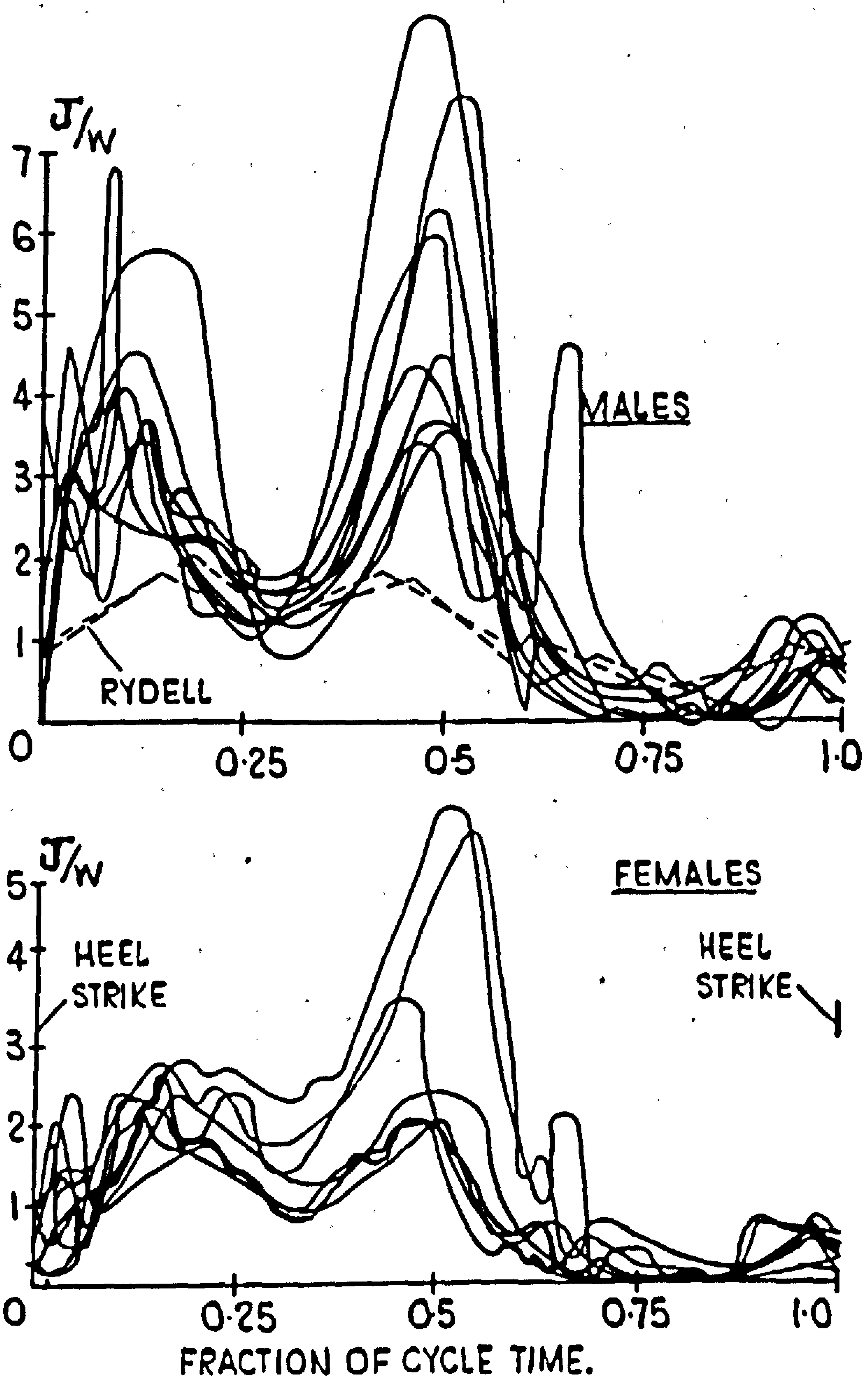
LEG TO TRUNK MOMENTS ABOUT X - AXIS
ENVELOPE OF CURVES FOR FOUR MALE SUBJECTS.
FROM BRESLER AND FRANKEL (1950)

FIG. 81

particularly interesting to see the narrow band within which the curves of the moments for the male subjects lie. With reference to the M_z curves for the female subjects in Fig. 78, it should be noted that this is plotted at twice the scale for the male subjects and this accounts for the apparently greater spread of these test results. Generally it can be considered that the validity of the authors curves for leg trunk moment is confirmed by the results of Bresler and Frankel.

The only experimental determination of hip joint force is that of Rydell (1966). Summary curves for the variation with time of the resultant hip joint force of the present test series are shown in Fig. 82. The joint force J is shown as a multiple J/W of the subject's weight. There is seen to be considerable variation in the value of the ratio at the second peak value which occurs at or near heel strike of the opposite foot. In the graph for male subjects, the two walking tests reported by Rydell are shown by dotted lines and lie below the other values. In the graph for female subjects, Rydell's curves could not be distinguished from the four lower curves of this figure.

The reasons for the difference in one case and the agreement in the other can be found in the walking speeds and stride lengths of the respective tests. Rydell's male subject performed tests



SUPERIMPOSED CURVES OF JOINT FORCE VARIATION WITH TIME FOR ALL TESTS.

FIG. 82

walks at speeds of 35 and 51 in/sec and stride lengths of 30 and 47.5 in respectively. The values for the author's tests are for walking speeds between 54 and 82 in/sec and for stride lengths between 62 and 88 in. Rydell's female subject performed test walks at speeds of 35 and 53 in/sec with a stride length in the former case of 41 in. The author's female subjects had speeds between 41 and 57 in/sec and stride lengths between 50 and 60 in. Thus Rydell's results would appear to substantiate the results of Fig. 82.

Referring to the locus of the resultant joint force on the femoral head in Figs. 72 and 73, the shaded and outlined areas within which Rydell's results lie are transposed from Rydell's reference axes relative to his prosthesis to the present axes by the procedure shown in Appendix VII. Rydell shows only these areas, the locus of the resultant force is not presented. The results are seen to be in agreement in respect ofmedial offset from the centre of the head. The difference in the anterior/posterior offset corresponds to two factors.

1. The angles of flexion and extension of the thigh for the subjects. differ as follows:-

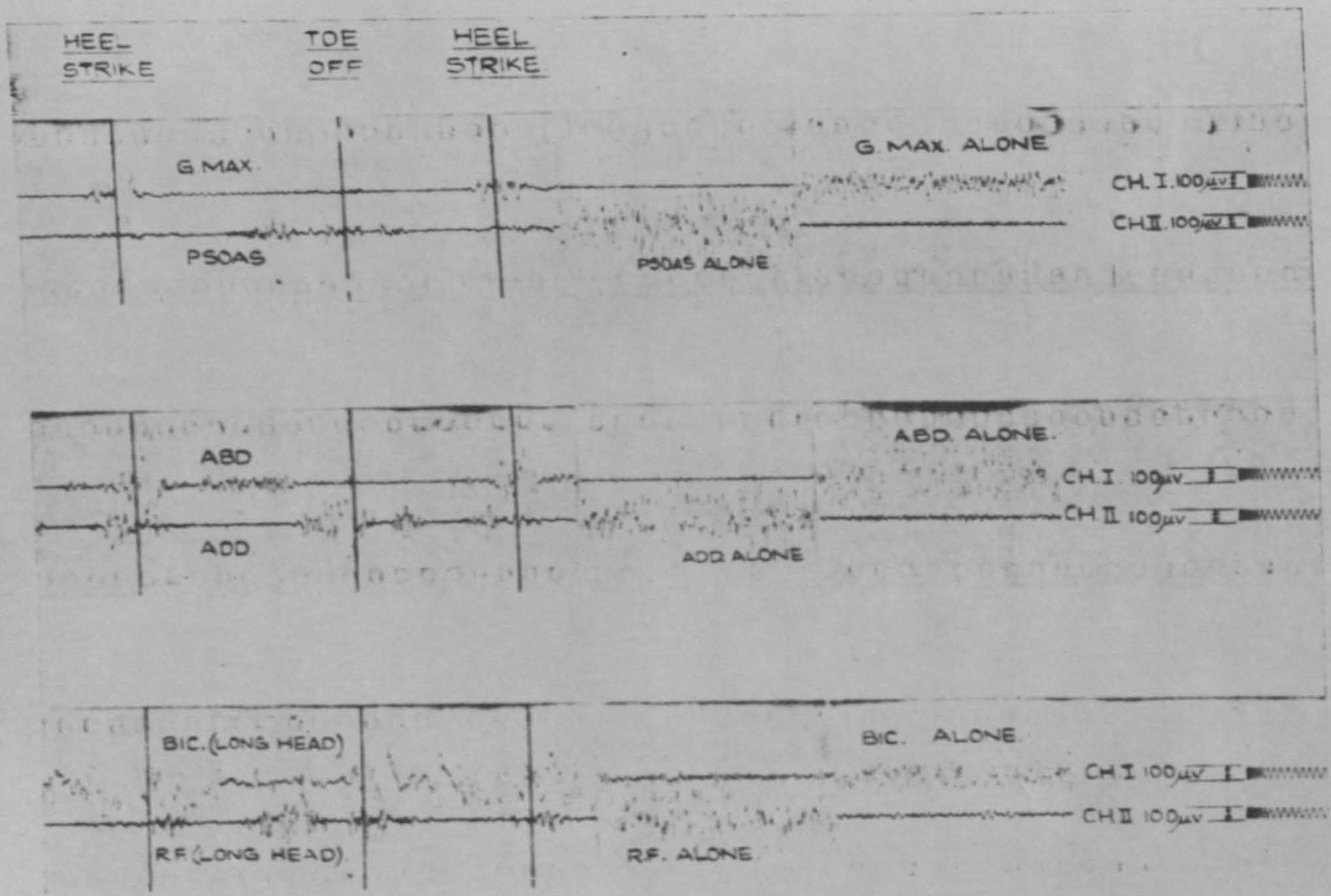
Subject	Rydell 1	Rydell 2	4
Thigh Flexion	23°	22°	23°
Thigh Extension	12°	19°	28° .

This large difference in extension corresponds to the spread of this subject's points posterior to Rydell's.

2. In the present analysis the effect of rotation of the thigh about its long axis is neglected. At heel strike the thigh is rotated outwards, and rotates inwards relative to the pelvis during the immediately following period of double support (Univ. California 1947). This would imply that the points corresponding to heel strike should be moved posteriorly in Figs. 73 and 74. There is only a small rotation of the femur in the region of single support and the point between the two force peaks should therefore be moved only slightly. In the following period of double support the femur rotates outwards and the points between second force peak and toe-off should therefore move in the anterior direction. For 10° rotation which is a typical amount of femur rotation relative to the pelvis, the A/P movement of these points is however represented by only .09 in. displacement on the diagram.

3. Electromyographic Results.

A typical edited record taken during the course of these investigations is shown in Fig. 83 which is taken from Sorbie and Zalter (1965). At the extreme right the 100 μV calibrating signal is shown. To the left of this is the record



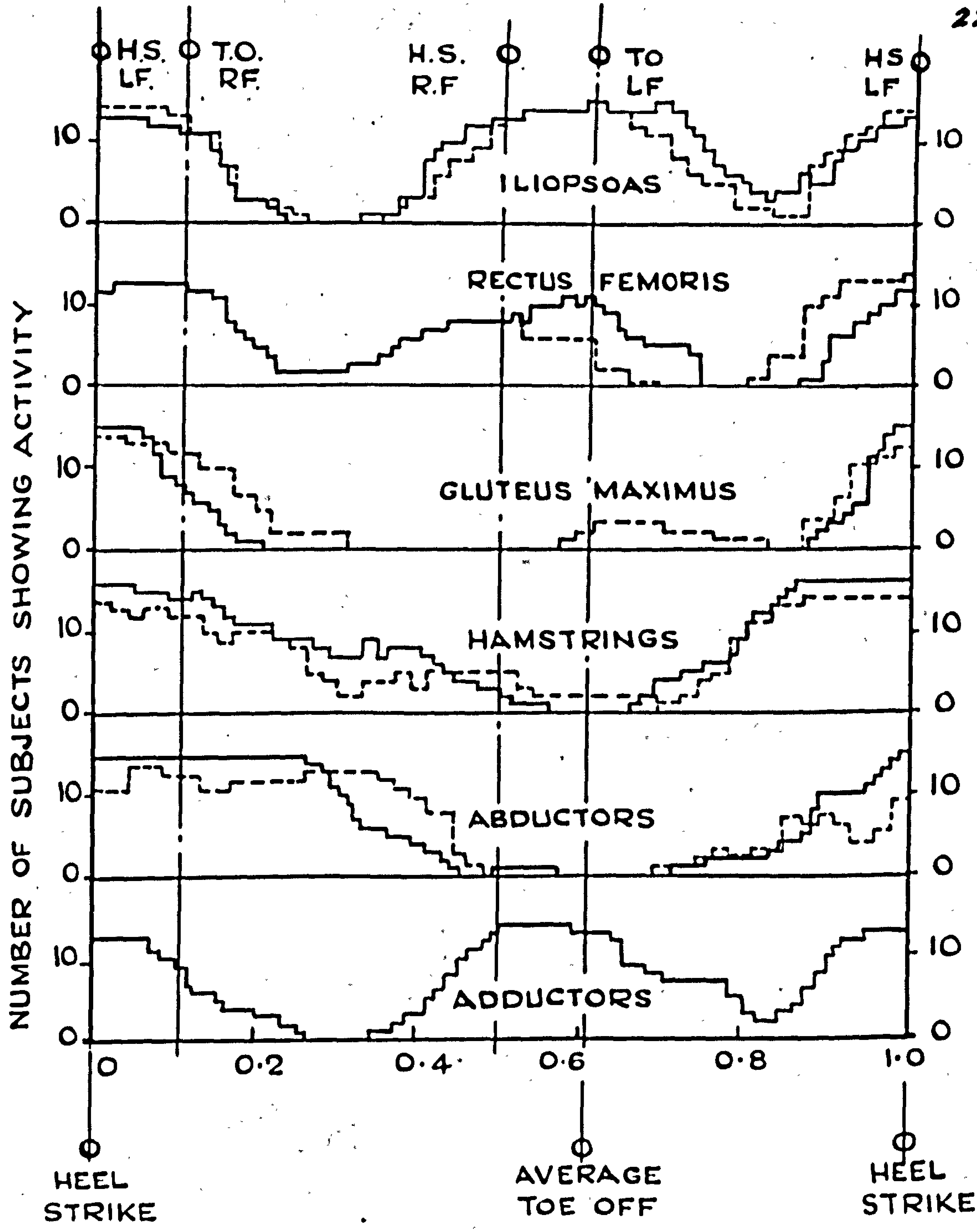
TYPICAL SET OF E.M.G. RECORDS

FROM SORBIE AND ZALTER (1965)

FIG. 83.

taken as the subject was exerting a force action which loaded only the relevant group of muscles, e.g. for Psoas the thigh was flexed with the shank hanging freely from the knee : for the abductors the subject, standing on the right leg, pushed the left foot outwards against resistance, etc. This portion of the record shows also the amount of background "noise" on the record. The left hand portion of the diagram shows the records taken during a walking test. It is obvious that although the signals can be taken to indicate muscular activity, they are not precisely the same from cycle to cycle. It is difficult to decide the exact time at which activity commences and ceases and the intensity of force cannot be determined. By a comparison with the record obtained during the gauge localisation tests it can be seen that for this subject the walking effort is much less than that exerted during calibration.

The records from 17 subjects are summarised in Fig. 84 which is presented in the manner adopted by Joseph and Battye (1966) and their results are illustrated also. The first comment to be made is that this diagram does not indicate the intensity of recorded electrical activity and cannot therefore be compared quantitively with the muscle force graphs of Figs. 65 and 66.



SUMMARY DIAGRAMS FOR 17 SUBJECTS SHOWING PHASES OF E.M.G. ACTIVITY

JOSEPH & BATTYE (1966) ------

PAUL ———

FIG. 84.

The differencesbetween the two sets of results which appear significant are:-

1. Joseph and Battye's results show a narrower band of activity of Rectus Femoris at toe-off and a wider band at heel strike.

2. Joseph and Battye's results show an earlier cessation of abductor activity.

It is submitted that no great signifcance can be attached to these minor differences and that they may be accounted for as follows:-

a) by the use of different criteria for the threshold at commencement and termination.

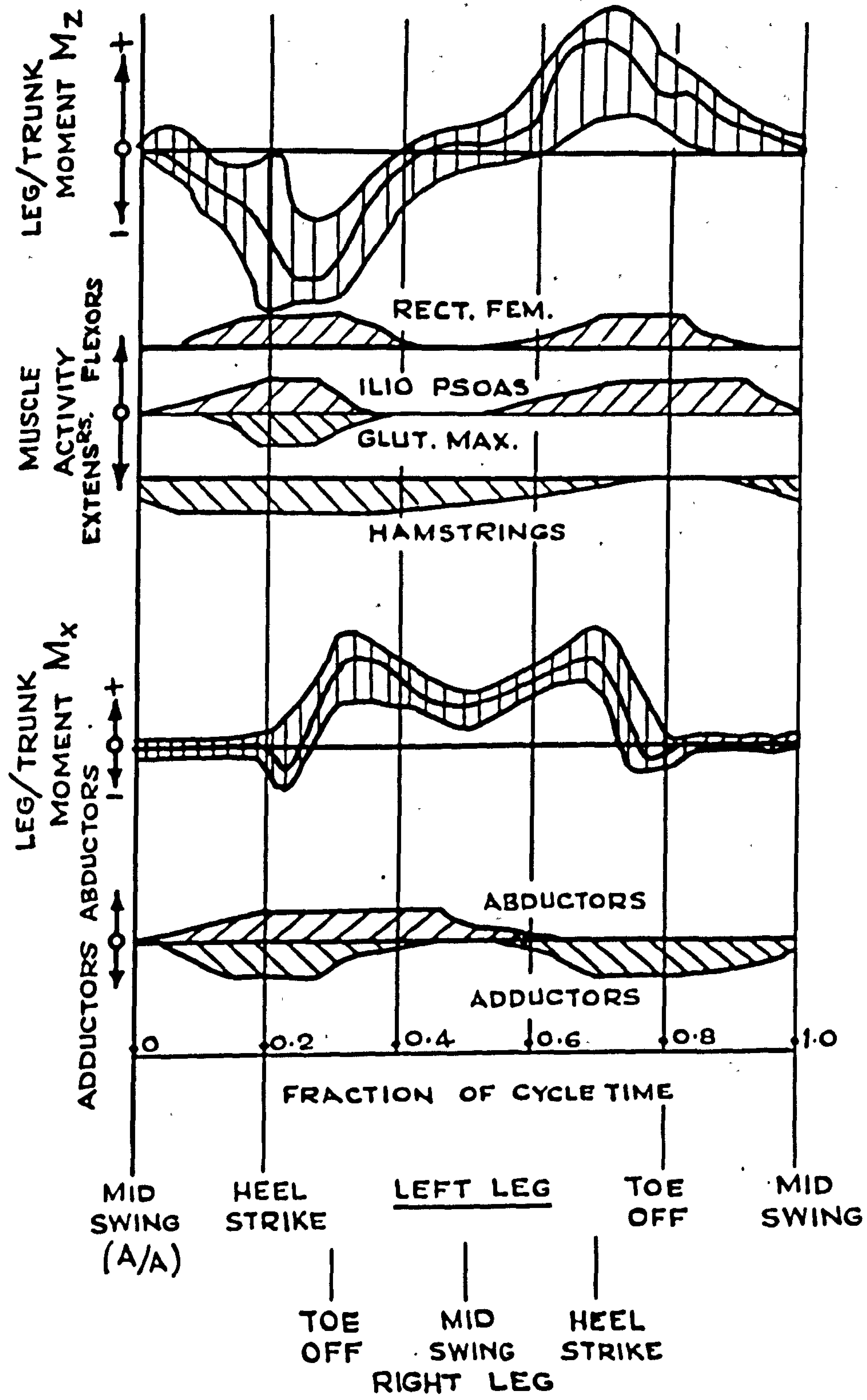
b) by errors in the phasing.

With regard to b) Joseph and Battye's method has been criticised in the review on physiology and the writer's procedure is open to error in relating the galvanometer record to the film of the E.M.G. record.

The University of California (1947) results, reproduced in Fig. 53, show no action of iliopsoas at heel strike. It may be that the signals which Joseph and Battye and the writer obtain are "noise" from Rectus Femoris. However in view of the identification procedure adopted this seems unlikely. Sartorius which might cause "noise" is also reported by the California study to be quiet at heel strike. Pectineus is approximately in the area of the iliopsoas electrodes but there is no information on its activity in walking in the studies cited.

It must be considered therefore that a) the signals shown by the studies using surface electrodes include a signal from Pectineus or b) the needle electrodes of the California studies are picking up local signals in the muscle and that action is occurring elsewhere which is not being recorded or c) the surface electrodes are picking up signals from the abdominal muscles.

To compare the summary diagram of leg to trunk moment with the diagram of E.M.G. activity Fig. 85 has been drawn from the information presented in Figs. 77 - 79 and 84. In the region of heel strike there is a large negative M_z value and corresponding activity in the extensor muscles, Gluteus Maximus and the Hamstrings, but there is significant activity also in the flexor muscles Iliopsoas and Rectus Femoris. The action of the latter may be interpreted as stabilisation of the leg until firm contact of the foot on the ground has been established. The effect of this is to increase the value of joint force above that shown in the graphs of Figs. 67, 68, and 82. Also since Gluteus Maximus is showing activity, the lower bound curve must lie between that for Gluteus Maximus and the Hamstrings. Similarly at heel strike the value



RELATIONSHIP OF MOMENT AND E.M.G.

FIG. 85.

of M_x is low but there is antagonistic activity between the abductors and adductors. This disappears as the M_x curve reaches its first peak and for the remainder of the cycle there is little antagonistic activity. At heel strike of the right leg both Rectus Femoris and Iliopsoas show activity and the true curve of joint force must lie between the curves corresponding to their separate action, there being no indication of load transmission by the capsular ligaments. In this region the curves lie close together and this load sharing therefore does not affect the joint force significantly.

The present studies and the other E.M.G. studies cited all show cessation of abductor muscle action at 0.45 of the cycle from heel strike, i.e. 0.65 of the cycle from mid-swing in Fig. 85. Yet this is prior to the second peak of M_x shown by the present studies and those of Bresler and Frankel. The time delay of 0.08 ± 0.02 sec. between E.M.G. activity and muscle force quoted by the University of California studies amounts to between 0.053 and 0.089 of the mean cycle time and is insufficient to account completely for this discrepancy. It appears that Inman's (1947) contention that there is "passive tension" in the ilio-tibial tract might account for this loading in the absence of E.M.G. activity but there is no significant change in the femur/pelvis angles 0_x at this phase of the movement shown by either Fig. 62 or the California results. Activity is reported in Tensor Fascia Lata but only starting some 4% of the cycle before toe-off, which is not sufficiently early to correspond. The present evidence appears insufficient to form a final conclusion.

Generally, however, the E.M.G. studies indicate that in the graphs of joint force the first peak indicated in this study is less than the actual value due to antagonistic activity of the muscles. Thereafter the appropriate value of joint force lies between the upper and lower bound curves presented.

4. Statistical Analysis of Joint Force Results.

The details of the test subjects and the results obtained are shown in Table 7. The parameters expected to be of significance in the results obtained are shown in Table 8 together with the results of a series of simple linear bivariate statistical analyses of 4 variable dependent quantities pertaining to the curve of resultant hip joint force with time, namely:-

J_1 the height of the first peak. J_2 the height of the second peak.

J_{max} the greater of J_1 and J_2. $J_{mean} = (J_1 + J_2)/2$.

SUBJECT Nº	SEX	AGE	HEIGHT	WEIGHT	THIGH LENGTH	SHANK LENGTH	SOMATO TYPE			TEST Nº	STRIDE LENGTH	STRIDE TIME	J_1	J_2
							ENDO	MESO	ECTO					
		YEARS	IN.	LB.	IN.	IN.					IN.	SEC.	LB.	LB.
1	F	20	62·5	131	15·7	13·4	5	2	4	5	50·7	1·24	334	330
										6	50·1	1·20	380	327
3	F	20	64·5	127·5	16·8	14·8	4	5	3	1	55	1·16	366	699
										8	51·7	1·06	361	714
4	F	18·6	67	129	16·8	14·8	4	4	3	2	59·5	1·04	232	263
										3	59·2	1·04	260	259
										4	56·2	1·04	224	265
										6	57·5	1·08	274	456
7	M	19·3	69	140	17·4	14·6	4	3	4	1	65·2	1·20	581	1091
11	M	19·7	70	166	17·4	16·3	4	4	3	1	71·2	1·10	685	582
14	M	19·0	71·5	133	19·2	15·3	3	1	6	3	78·5	1·04	526	800
15	M	20·2	66	140	14·2	14·6	4	5	3	1	71·7	1·16	641	613
16	M	24	69·5	129	16·8	15·8	2	4	5	1	66·7	1·02	354	823
17	M	19·4	66·5	140	15·8	15·1	4	4	3	1	64·7	1·16	454	522
										3	64·5	1·16	427	534
18	M	22	67	143	17·5	15·5	5	3	3	3	62·5	1·10	436	583
22	M	20·8	72	130	18·8	16·4	1	3	6	2	88·3	1·08	601	1201
23	M	36·1	72·5	180	16·5	15·5	3	4	4	1	69·2	1·12	1040	855
R1	M	51	APPROX 67	165						1	30·2	0·86	260	255
										2	47·5	0·92	354	311
R2	F	56	APPROX 61	98·5						1	41·3	0·96	291	220
										2			322	251

DETAILS OF SUBJECTS AND TEST RESULTS

R - DENOTES RESULTS OF RYDELL (1966)

TABLE Nº 7

PARAMETER		J MAX.		J MEAN		J_1		J_2	
		r	t	r	t	r	t	r	t
AGE	α	-0·29	1·37	-0·28	1·31	-0·08	0·34	-0·38	1·85
HEIGHT.	H	0·68	3·69	0·70	3·87	0·61	3·05	0·63	3·24
WEIGHT	W	0·28	1·32	0·40	1·95	0·57	3·13	0·21	0·97
PONDERAL INDEX.	$H/\sqrt[3]{W}$	0·55	2·64	0·45	2·02	0·11	0·46	0·62	3·20
THIGH LENGTH	LT	0·43	1·94	0·33	1·42	0·06	0·23	0·48	2·20
SHANK LENGTH	LS	0·55	2·68	0·54	2·60	0·40	1·76	0·54	2·59
LT + LS		0·53	2·50	0·47	2·10	0·22	0·91	0·56	2·71
STRIDE LENGTH	L	0·73	4·68	0·76	5·03	0·60	3·28	0·74	4·81
STRIDE TIME	T	0·32	1·48	0·37	1·72	0·34	1·60	0·33	1·51
L/T.		0·64	3·71	0·64	3·74	0·49	2·50	0·64	3·76
L/H.		0·64	3·30	0·68	3·68	0·54	2·55	0·65	3·45
WL		0·74	4·73	0·82	6·34	0·84	6·67	0·68	4·02
WL/H.		0·65	3·49	0·78	5·02	0·87	7·08	0·56	2·72
WL/HT		0·61	3·70	0·70	3·92	0·76	4·61	0·47	2·12

STATISTICAL ANALYSIS OF HIP JOINT FORCE RESULTS

TABLE 8

The coefficient of correlation 'r' and the 'Student' factor 't' defined by Gosset (1908) corresponding to the linear regression of J on the other parameters were obtained using a digital computer analysis system "SCAN" (MacGregor (1965). The joint force values of Rydell were included where possible but the absence of information on height in both cases and stride length in one case restricted their inclusion. The number of variables was between 18 and 22 depending on this. For a linear regression of 18 variables a 't' value exceeding 2.92 is required to demonstrate significance at the 0.01 level. From table 8 it can be concluded therefore that the following factors may be excluded from the analysis:-

Age, Weight, Ponderal Index, Thigh and Shank Length, and Stride Time. It is seen that correlation is not so good for J_{max} and J_2 as for J_{mean} and J_1. Since the intention was to obtain a representative expression describing both maxima of the joint force pattern, attention was given to J_{mean} subsequently.

The hip joint force depends on the ground to foot reaction W, and its moments M_x, M_z about the hip axes. For a simplified analysis the vertical component W_y only may be considered and its value taken as $W_y = W + IF$

where the Inertia Force $IF = W \ddot{y}/g$

and the vertical acceleration of the body C.G. $= \ddot{y} = \omega^2 y_o \sin \omega t$

y_o = maximum amplitude of vertical displacements

ω = frequency of oscillation.

$= 2\pi \times 2/T = 4\pi/T$

T = cycle time for one stride.

Grieve and Gear (1966) quote $f = 64.8\, V'^{0.57}$ and Dean (1965) quotes $f = 63\, V'^{0.5}$ for the equations describing the relationship between step frequency f and relative walking speed V', for adult subjects.

Taking Dean's result

$$1/T = \text{const} \times (L/HT)^{0.5}$$

$$\text{i.e. } T = \text{const} \times H/L.$$

Assuming the legs to be straight and length L_L at heel strike and mid-stance and neglecting pelvic rotation, y_o, the amplitude of vertical displacement can be expressed as:-

$$y_o' = L_L - \sqrt{L^2_L - (L/4)^2}$$

$$= L_L (1 - 1 + L^2/8L^2_L + \text{other terms})$$

$$= L^2/8L_L = \text{const} \times L^2/H.$$

assumingleg length L_L to be a constant proportion of height H .

Hence $W_y \max = W(1 + AL^4/H^3)$

where A is a constant.

The hip joint force $J = W_y + M_z/r_z + M_x/r_x$

M_z can be taken as:- M_z = constant x W_y x L

M_x can be taken as:- M_x = constant x W_y.

Thus $J_{max} = W(1 + \lambda L^4/H_3)(1 + B_o + B_1 L)$

$$= A_1 W + A_2 WL + A_3 WL^4/H^3 + A_4 WL^5/H^3$$

where A_1, A_2 A_3 and A_4 are constants.

Graphs of the mean joint force to various functions of W, L and H are shown in Figs. 86 - 93 and it will be seen that the points indicate generally an increase in joint force with weight and stride length, and also that Rydell's results fit well into the established pattern. If a statistical analysis is performed for the relationship between mean joint force J and certain other simple functions of WL and H the following results are obtained:-

function	$\frac{WL^5}{H^3}$	$\frac{WL^4}{H^3}$	$\frac{WL^3}{H}$	$\frac{WL^2}{H}$	WL^2	L^2	L^2/H^2	WL^2/H^2
r	0.68	0.70	0.70	0.79	0.78	0.77	0.56	0.71
t	3.76	3.93	3.97	5.16	5.46	5.27	2.69	5.76

Since none of the above functions gives as good a correlation coefficient as does WL in table 8 it appears that the best representation for the present values of hip joint force may be given by the following equation:-

$$J_{mean} = 0.0865WL - 225$$

where W is in lb, L in inches and J_{mean} in lb. The standard error is 127. The t value of 6.34 implies correlation to

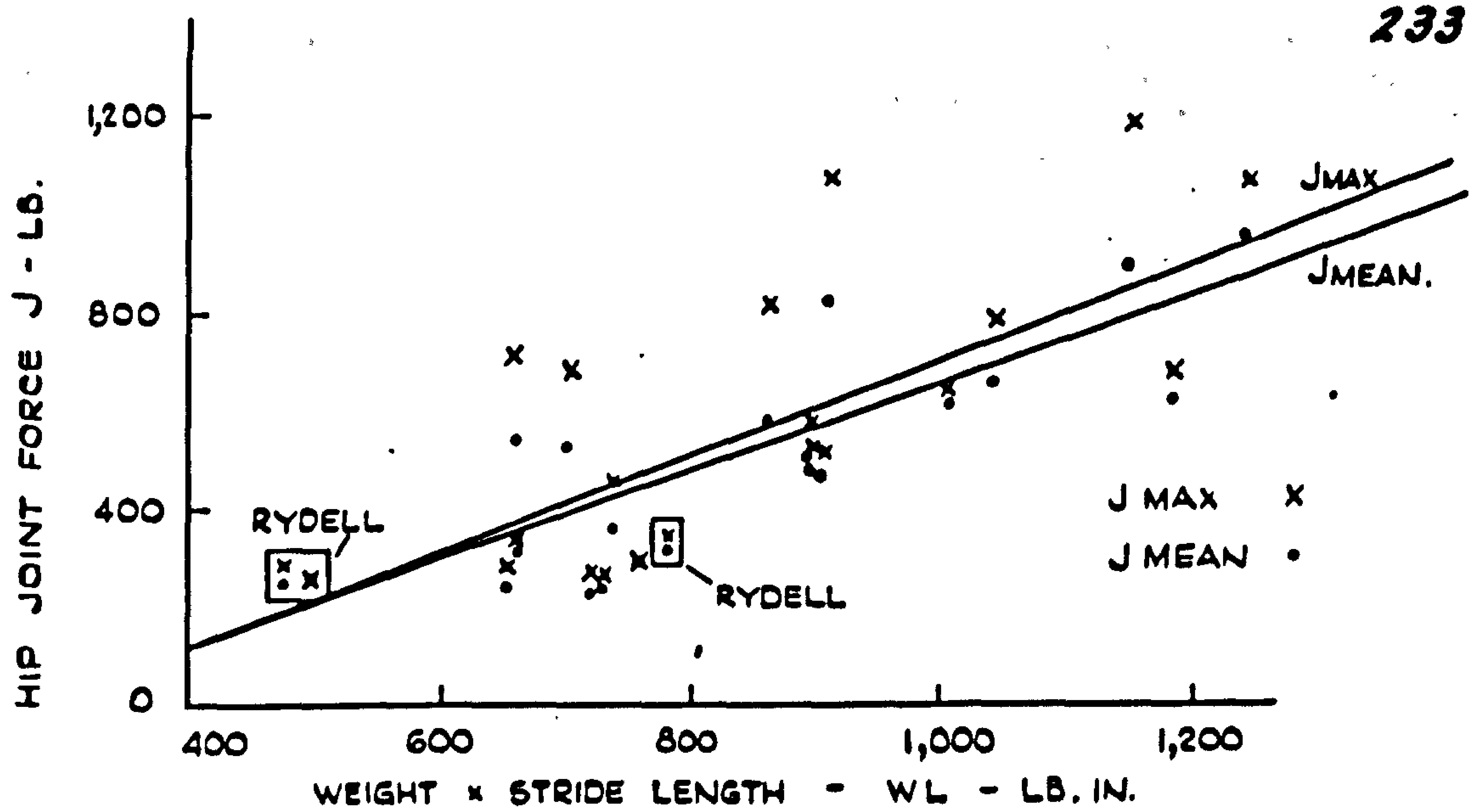

HIP JOINT FORCE J - LB.
1,200
800
400
0
400
600
800
1,000
1,200
WEIGHT × STRIDE LENGTH - WL - LB. IN.
RYDELL
RYDELL
JMAX
JMEAN.
J MAX X
J MEAN •

FIG. 86

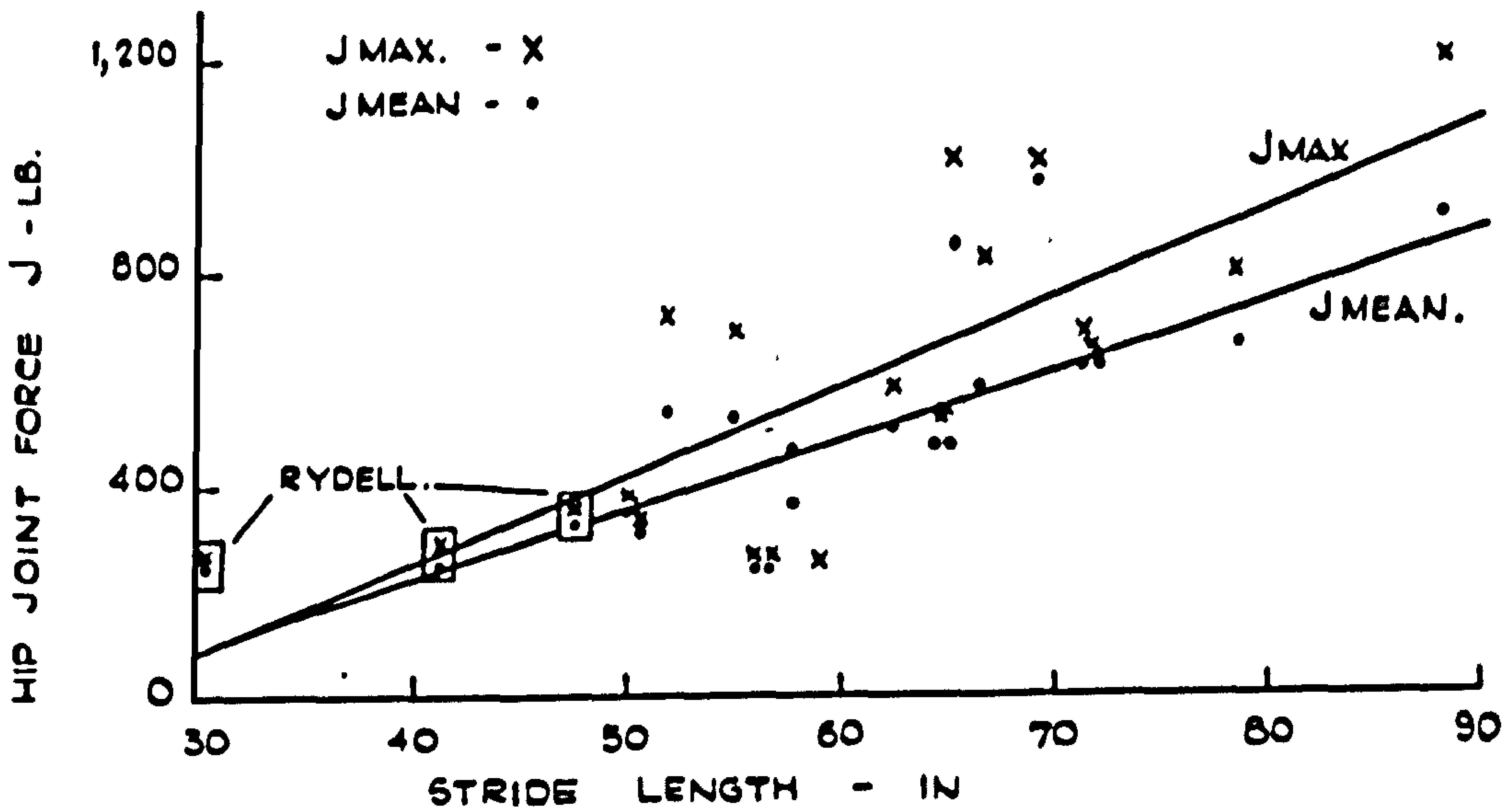

HIP JOINT FORCE J - LB.
1,200
800
400
0
30
40
50
60
70
80
90
STRIDE LENGTH - IN
J MAX. - X
J MEAN - •
RYDELL.
JMAX
JMEAN.

FIG. 87.

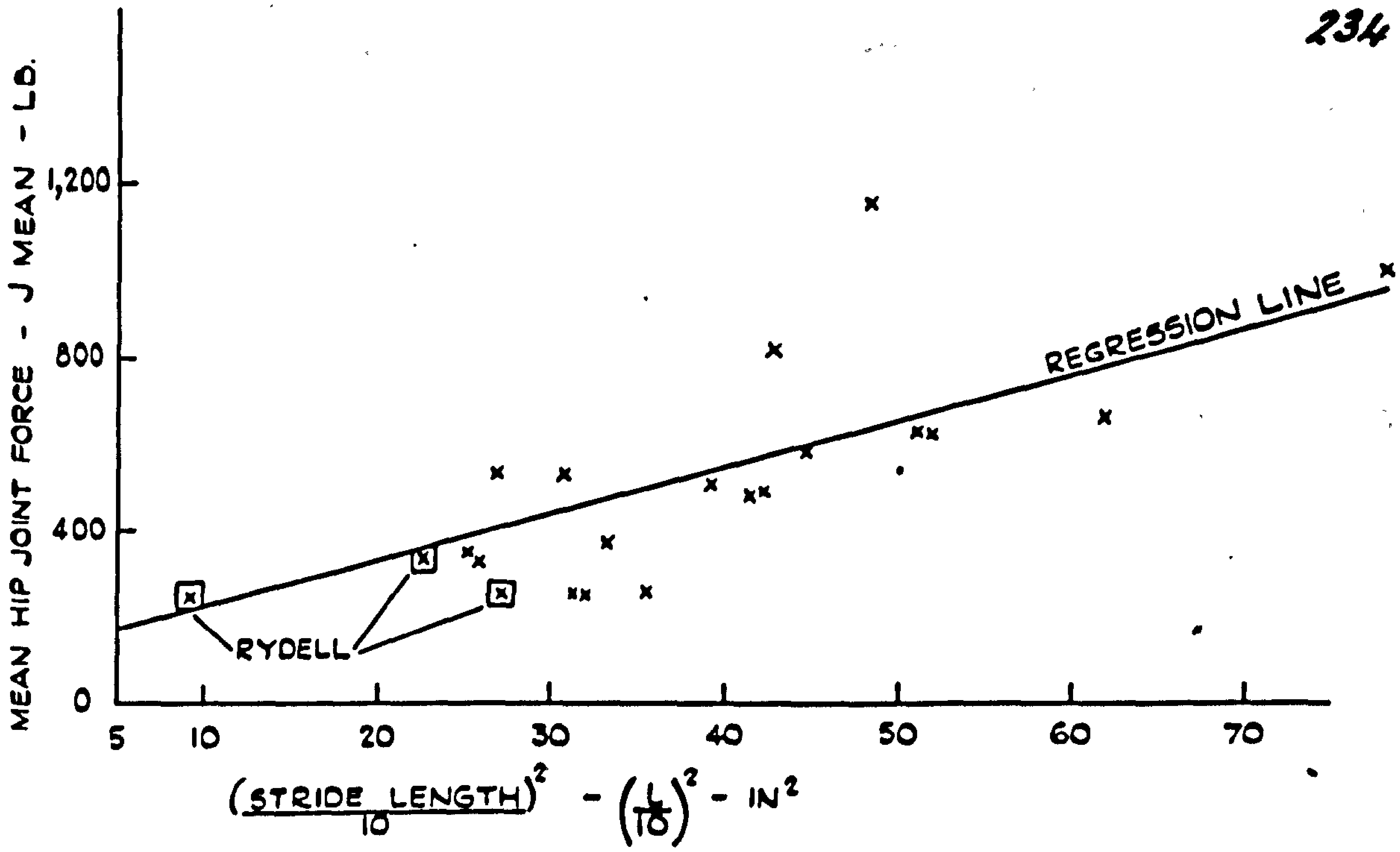

MEAN HIP JOINT FORCE - J MEAN - LB.
1,200
800
400
0
REGRESSION LINE
RYDELL
5
10
20
30
40
50
60
70
(STRIDE LENGTH / 10)² - (L/10)² - IN²

FIG. 88

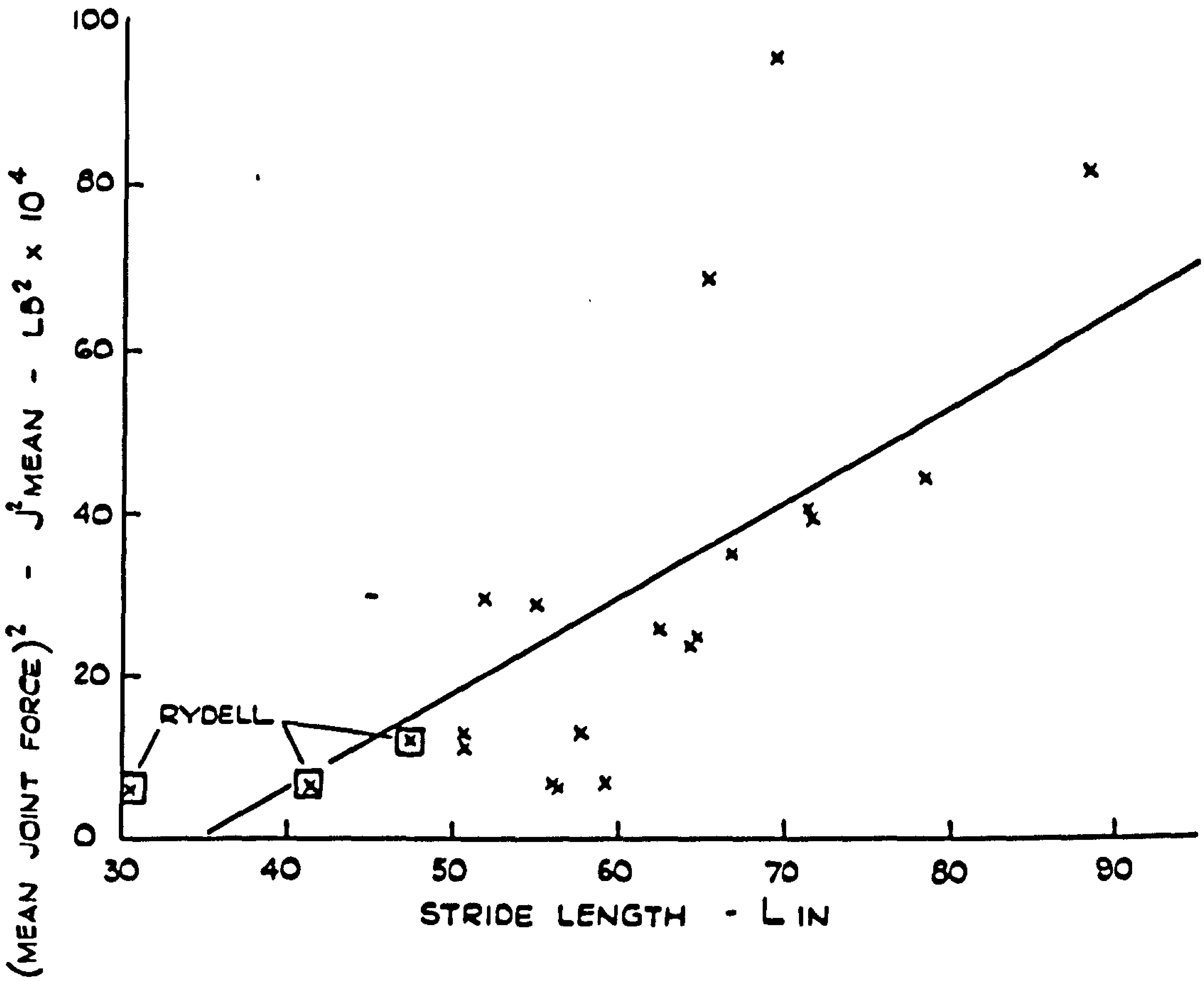
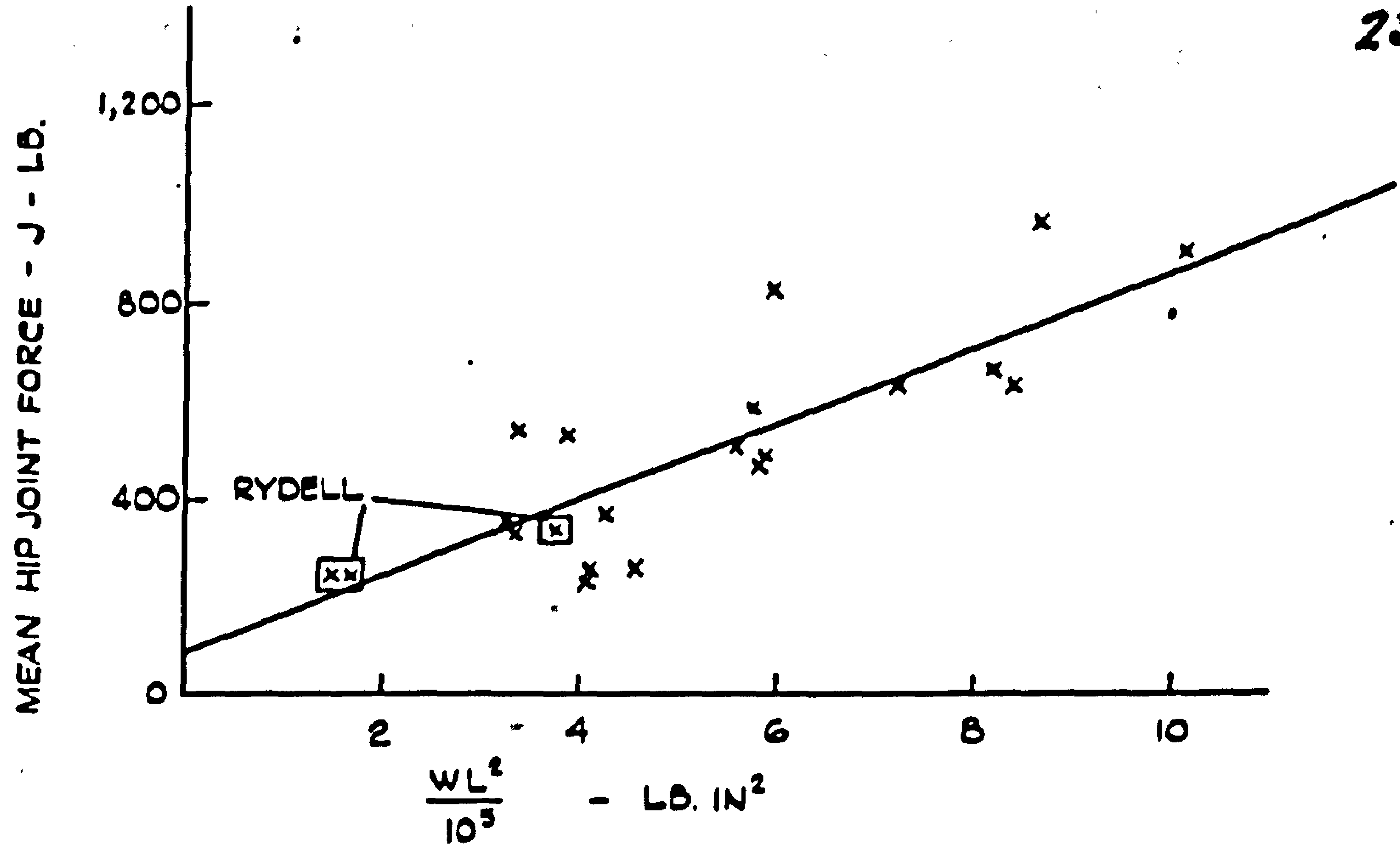
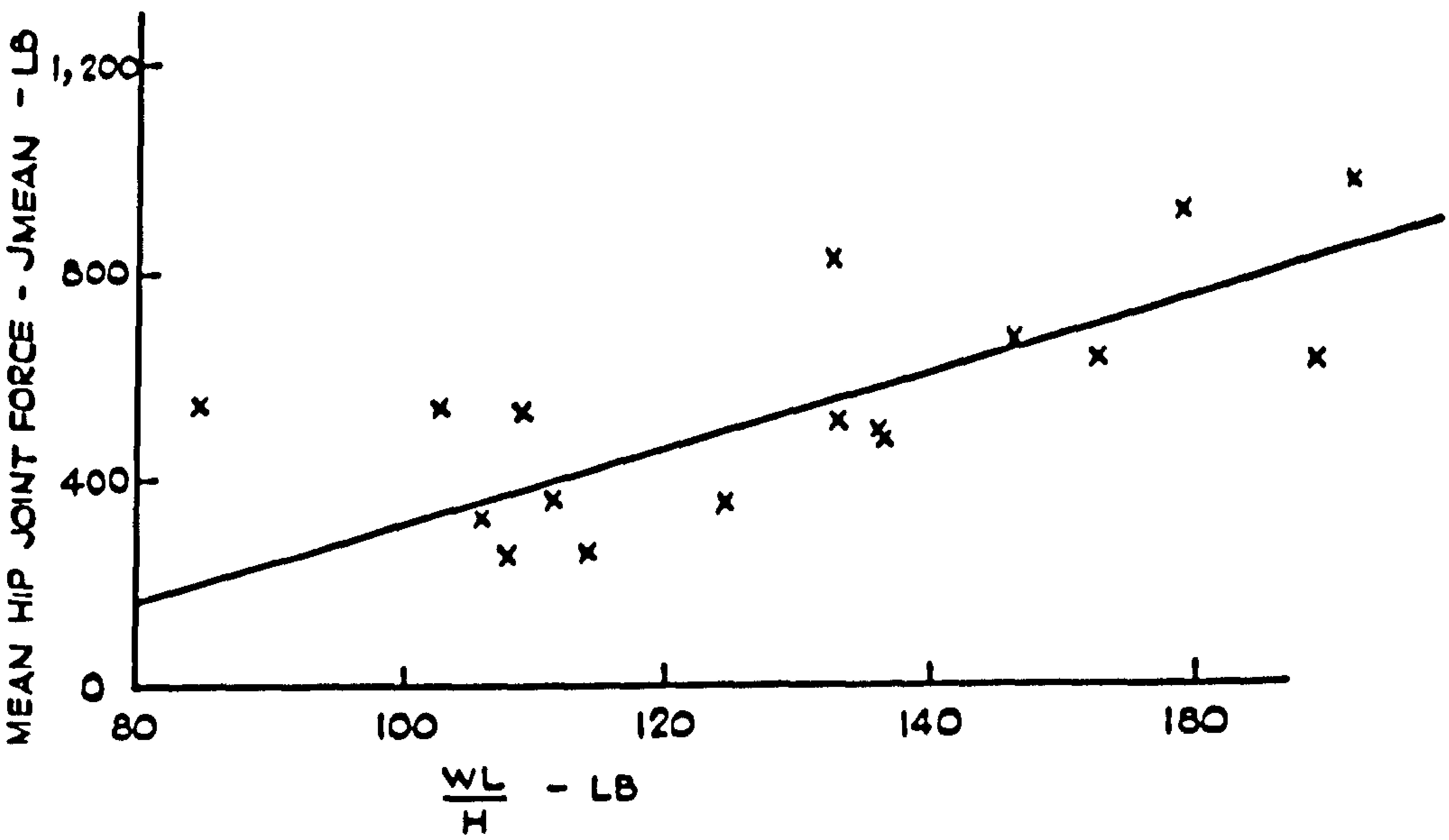

(MEAN JOINT FORCE)² - J²MEAN - LB² x 10⁴
100
80
60
40
20
0
RYDELL
30
40
50
60
70
80
90
STRIDE LENGTH - L IN

FIG. 89



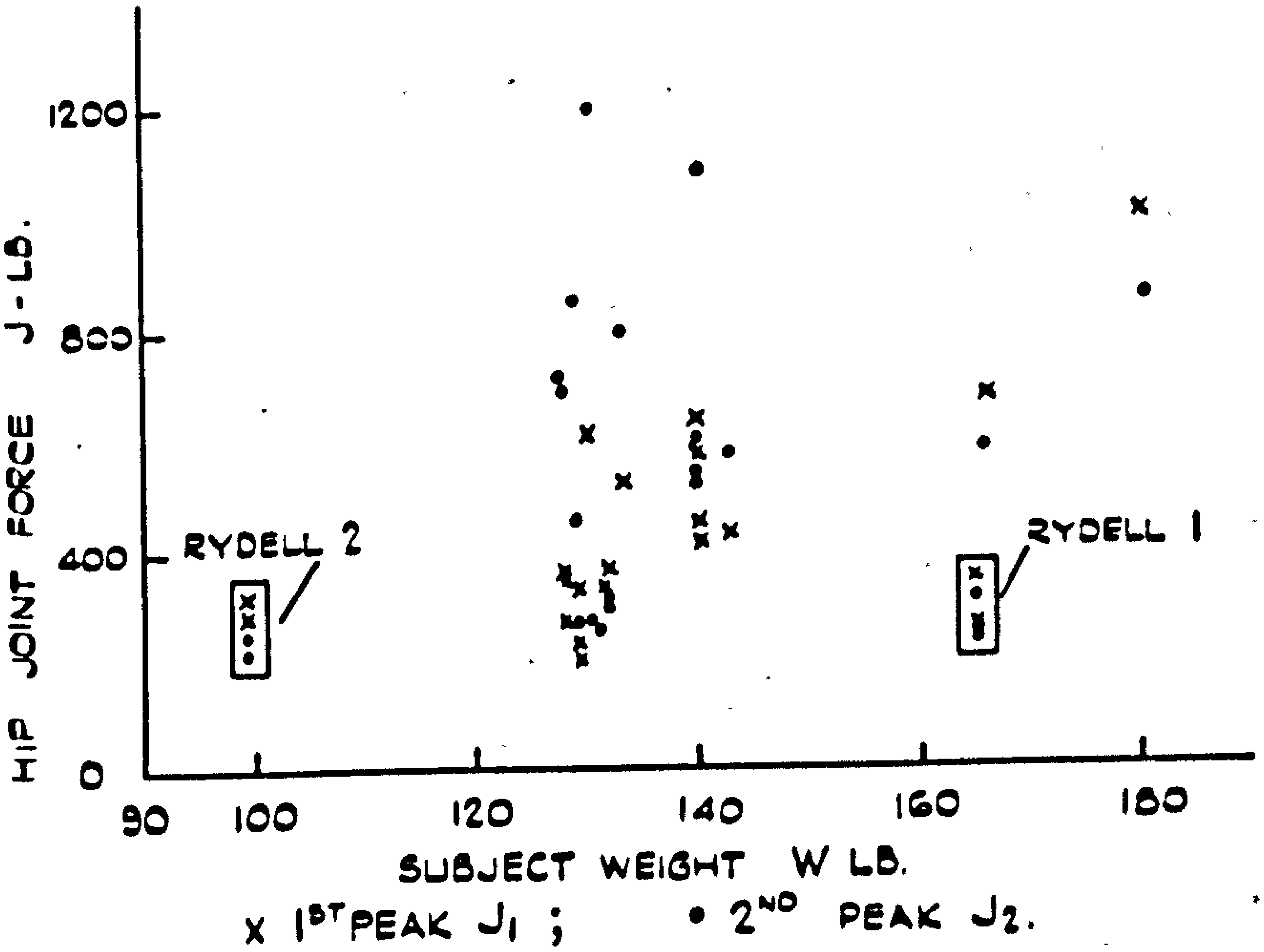
RELATIONSHIP BETWEEN MEAN JOINT FORCE 'J', SUBJECT WEIGHT 'W' AND STRIDE LENGTH 'L'.

FIG. 90



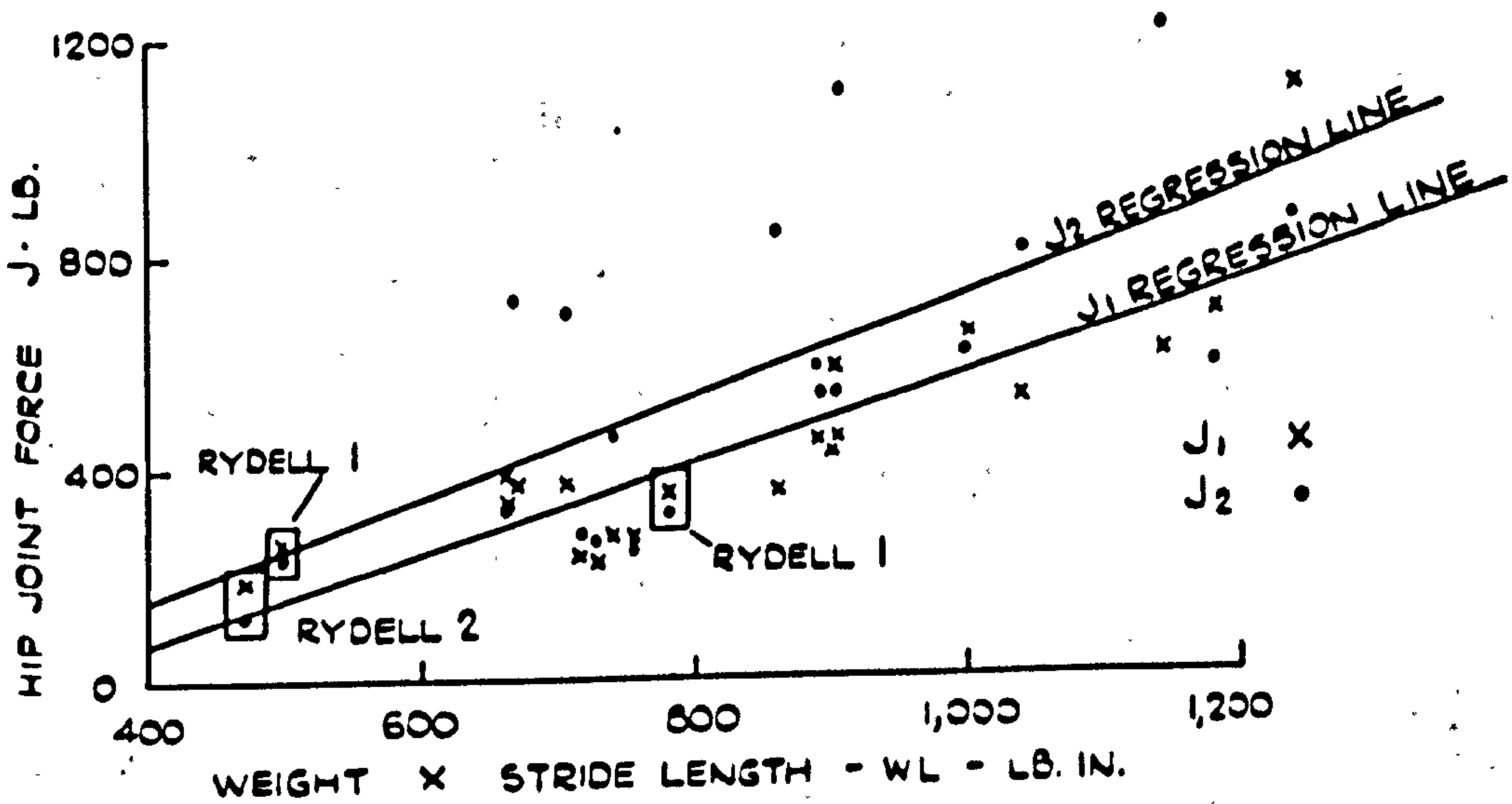
RELATIONSHIP BETWEEN MEAN JOINT FORCE 'JMEAN' AND WL/H WHERE W = SUBJECT WEIGHT, 'L' = STRIDE LENGTH AND 'H' = HEIGHT

FIG. 91



HIP JOINT FORCE J - LB.
1200
800
400
0
RYDELL 2
RYDELL 1
90
100
120
140
160
180
SUBJECT WEIGHT W LB.
X 1ST PEAK J1 ; • 2ND PEAK J2.

FIG. 92



HIP JOINT FORCE J. LB.
1200
800
400
0
J2 REGRESSION LINE
J1 REGRESSION LINE
RYDELL 1
RYDELL 1
RYDELL 2
J1 X
J2 •
400
600
800
1,000
1,200
WEIGHT X STRIDE LENGTH - WL - LB. IN.

FIG. 93

better than the 0.001 level. The better correlation with WL implies that the inertia term in the analysis is of smaller significance than the gravitational term. It should be noted also that Dean's equation implies that LT/H is constant for all subjects. In fact the subjects in this test series demonstrated LT/H values in the range 0.85 - 1.33 sec. (mean 1.027 S.D. 0.11) compared with Dean's mean of 0.908 sec. and the reliability of this approximation in the above analysis is correspondingly reduced.

5. Analysis of Experimental Errors.

The calculation procedure involves complicated mathematical expressions in which it is not possible to apply the normal methods of estimating the amount of variation in the end result due to errors in reading. The order of magnitude of these variations was therefore estimated by systematically varying the experimental quantities for selected frames in the analysis for one subject. This procedure was applied to frames 27, 30, 45 and 55 of test No. 3 on subject No. 18 in which heel strike occurred at frame 28 and toe-off at frame 63. The analysis is conveniently considered in separate parts corresponding to the first and second programmes of the computer analysis.

Leg/Trunk Force Actions.

Table (9) shows the results obtained from perturbations of

PARAMETER	ESTIMATED EXPERIMENTAL ERROR - %	PERCENTAGE CHANGE IN JOINT FORCE FOR 1 PER CENT. CHANGE IN PARAMETER.			
		FRAME 27	FRAME 30	FRAME 45	FRAME 55
$W + W_S$	4	1·0	0·01	0·07	0·01
W_S	10	0·54	0·04	0·08	0·06
L_T	1	0·08	0·01	0·16	0·11
L_S	1	0·40	0·01	0·34	0·21
L_F	1	0·22	0·02	0·35	0·24
a	1	0·57	2·75	2·42	1·31
b	1	2·33	2·10	2·0	1·8
$x_H - x_B$	10	0·72	0·22	1·24	1·41
$y_H - y_B$	10	0·14	0·03	0·01	0·06
$z_H - z_B$	10	0·42	0·05	1·58	1·16
G1	1	0	0	0·26	0·30
G2	1	0	0·13	0·17	0·83
G3	1	0	0	0	0
G4	1	0	0·40	1·0	1·13
G5	1	0	0·06	0·07	0·39
G6	1	0	0·84	1·0	2·4
G15	1	0	0	0·01	0·02
G24	1	0	0·01	0·01	0·06

EFFECT ON CALCULATED JOINT FORCE OF SYSTEMATIC PERTURBATIONS OF BASIC DATA

TABLE 9

the basic data for the first computer programme. For convenience in comparisons these are presented as variations in the lower bound calculation of the resultant hip joint force. As expected the terms have a significance which depends on the phase of the walking cycle.

In frame 27, which is at the end of the swing phase the mass properties of the body exert significant effects on the end result, whereas the characteristics of the force plate are obviously irrelevant. In the other frames which correspond to the peaks and trough of the joint force curve, the converse holds. It is seen that the analysis is sensitive to the dimensions a and b which define the distances between the pelvic markers and to $x_H - x_B$ and $x_H - z_B$ which define the position of the hip joint centre relative to the B marker. Considering the indicated estimates of experimental error and the most pessimistic view, namely that all errors are present and that their effects are additive, it is found that their total for frames 27, 30, 45 and 55 are 26%, 10%, 37%, 35% in total joint forces of 117, 388, 199 and 583 lb. respectively.

The possibility of errors in phasing between film and force records was considered in the theoretical analysis. At the three points in the cycle defined by a, b and c in Fig. 75, the vertical components of ground to foot force generally exhibit turning values with respect to time. This is also the case for the moments about the force plate axes at a and c. The leg-trunk moment actions, which depend principally on external force actions as shown in table 7, will not vary significantly with phase errors at these points. The moments due to the horizontal force components correspond to the y component of joint position which varies by only small amounts from frame to frame. Therefore the effect of the changes in these components at points, a, b and c due to phasing errors will be small.

If perturbations of film co-ordinate measurements are undertaken, for frame 65 it is found that 0.1 in. variation in the three hip joint co-ordinate measurements gives errors of 5.3 % in M_{HZ} and 2.6% in M_{HX} which implies a total joint force error of 3%. If a corresponding perturbation is considered for one term of equation 43 for the linear acceleration of a marker, the worst errors in joint force occur if the marker is on the foot, and amount to 0.9% for M_{HZ} and 0.3% for M_{HX}. These correspond to a joint force error of 0.6%.

These assumed errors would correspond to inaccuracies in measurement, movement of the skin marker relative to the skeleton and local distortion within a five inch grid square due to film processing or camera or projector lens. Errors in body mass properties arise because of the known inaccuracies of Fischer's coefficients and movement of soft tissue between adjacent segments and within each segment due to joint rotation and muscular action.

Muscle and Joint Force Actions.

Table 10 shows the effects of perturbations on the measured co-ordinates which define the lines of action of muscle groups and on the scaling factor used to relate subject measurements to those of the standard skeleton. Again the importance of the measurements depends on the phase of the walking cycle since different muscle groups act with varying force at the selected points in the cycle. The sum of the numerical values of the errors due to 1% errors in all quantities in no case exceeds 2.1%. The possible error varies with the muscle group and depends particularly on the assumption that the line of action of the muscle group force extends between two points taken to be at the centres of areas of the corresponding origin and insertion. Where this area is large the approximation is greater, particularly with the Gluteal muscles which are claimed to exhibit "fan action", in that muscular activity appears to commence at one border of the muscle and spread successively across its breadth. Another error in this scheme is due to the bulging of underlying muscle groups which will tend to force the muscle to act along a different line of action at the hip/trunk section. The amount of this will be small in comparison with that due to the grouping of muscles.

The use of the scaling factors to relate the dimensions of the test subjects to those of the standard skeleton is open to question since it does not take into account the possibility of differences in the shapes of the pelvis and femur from subject to subject. On medical advice no attempt was made to take x-ray measurements of the appropriate regions and in any case the accuracy of such x-ray measurements is open to question. It is contended that the use of separate scaling factors for the three co-ordinate directions of measurement for the two segments is the best compromise in the prevailing circumstances. The total errors in joint force results due to perturbations of the scale factors shown in Table 10 in no case exceed 2% for simultaneous and additive perturbations of 1% in each factor. If the possibility of 10% errors in co-ordinates due to the scaling procedure is conceded then the most pessimistic error in joint force is 20%. It is submitted however that this would include the errors involved in the first 12 rows of Table 10 and would not be additive to them.

EXPERIMENTAL MEASUREMENT.			UPPER BOUND CALCLN.				LOWER BOUND CALCLN.			
			FILM FRAME No				FILM FRAME No			
GROUP	SITE	DIRECTN	27	30	45	55	27	30	45	55
FLEXOR OR EXTENSOR MUSCLES	ORIGIN	x	0.94	0.76	0.17	0.44	0.63	0.75	0.15	0.62
		y	0.08	0.07	0.01	0.02	0.16	0.20	0.01	0.16
		z	0.07	0.06	0	0	0.07	0	0	0.05
	INSERTION	x	0.23	0.09	0.14	0.47	0.04	0.04	0.02	0.07
		y	0.49	0.33	0	0	0.02	0.02	0.01	0.01
		z	0.21	0.22	0.01	0.04	0.04	0.03	0.02	0.08
ABDUCTOR OR ADDUCTOR MUSCLES	ORIGIN	x	0	0.01	0.02	0.01	0	0	0.01	0.02
		y	0.02	0.05	0.22	0.11	0.03	0.05	0.20	0.07
		z	0	0	0.02	0.01	0	0.01	0.02	0.01
	INSERTION	x	0.03	0.08	0.31	0.18	0.05	0.07	0.19	0.15
		y	0.01	0.04	0.14	0.18	0.04	0.07	0.13	0.15
		z	0	0.02	0.80	0.56	0.09	0.10	0.68	0.49
SCALING FACTOR	PELVIS	x	0.84	0.68	0.18	0.48	1.23	0.46	0.09	0.48
		y	0.10	0.13	0.19	0.15	0.34	0.26	0.16	0.15
		z	0.08	0.06	0.02	0.11	0.27	0.18	0.06	0.11
	FEMUR	x	0.15	0.02	0.35	0.09	0	0.04	0.20	0.09
		y	0.40	0.28	0.01	0.02	0.02	0.01	0	0.02
		z	0.34	0.20	0.71	0.44	0.10	0.15	0.60	0.44

PERCENTAGE CHANGE IN CALCULATED VALUE OF JOINT FORCE FOR ONE PER CENT CHANGE IN EXPERIMENTAL MEASUREMENT.

TABLE 10

Summary.

If the errors due to film readings, leg/trunk force actions ,basic data and muscle and joint force calculations are added ,the maximum possible error in calculation is 58 per cent. This is a very high figure and corresponds to:-

1) the large number of experimental measurements.
2) the approximations necessary in skeleton measurements.
3) the approximations in the form of analysis for muscle group forces.

The most pessimistic view has of course been taken in this analysis namely that every measurement is simultaneously in error to its maximum amount and that each error is additive. The statistical analysis of the results indicates a standard error of 127 on a mean joint force of 531 lb. i.e. 24%. This of course applies to random variation in errors and gives no indication of possible systematic errors in the analytical procedure. The amount of systematic error should however be small in view of the close correspondence of Rydell's results with those presented here. The other point to be made is that the assumed muscle groups are taken to be in the positions best adapted to transmitting the leg/trunk moments. If other groups of muscles or ligaments transmit these load actions they will inevitably involve higher values of joint force due to their smaller moment arm, and the values presented may be taken to represent a lower bound to the values existing in the body.

6. Proposed Extension of the Project.

The present procedure for determining joint force has been shown to be complicated and have significant possibilities of error. Any extension of this work should aim at simplification of the procedure and reduction of experimental error. Investigations are currently proceeding into the possibility of automatic registration of displacement of body markers using television cameras and recording the information directly on punch type or magnetic tape for input to a digital computer. It is also possible that a simplified analysis might be performed for the periods of maximum joint force considering only external force actions, and using the present measurement techniques.

The analytical procedure for joint force estimation should be improved in accuracy. The accuracy of film measurement might be improved by the use of 35 mm. cameras but this would involve initial and recurrent expenditure on a large scale. The procedure for determination of joint centre should be improved, either by the use of image intensifier X-ray techniques or by an adaptation of ultrasonic mapping techniques.

There is scope for the investigation of the possibility of using some function of E.M.G. signals to estimate muscle force. This would require simultaneous measurement of relative movement of limb segments in order to assess changes and rates of change of muscle lengths.

The tests should be extended to obtain values of force at the hip knee and ankle joint during walking at different speeds and on stepped and inclined surfaces.

In view of the dispersion of the values obtained in the tests on normal subjects it is suggested that the technique is not yet sufficiently refined for clinical use where it may be required to assess changes in condition due to therapy or surgical intervention. The technique can however be usefully applied to obtain information on the gait of amputees using prostheses. It is suggested also that the technique may be adapted to allow the assessment of energy expenditure and energy interchange between limb segments in normal and prosthetic gait.

SUMMARY.

A procedure has been devised to calculate from experimental measurements of walking subjects the pattern of variation of hip joint force. 18 tests have been performed on 3 female and 9 male subjects.

These tests show maximum values of hip joint force between 2.02 and 9.23 times body weight (mean 4.53 times). The analysis of the results indicate that joint force may be expressed by the equation:-

$$J_{mean} = 0.0865 \ WL - 225.$$

The experimental results of Rydell are in close agreement with this equation.

The experimental results of Bresler and Frankel are in close agreement with values calculated as an intermediate stage in the general calculation.

The phases of muscle activity assessed by E.M.G. signals are in general agreement with previously published work and are compatible with the measured force actions.

The amount of experimental error possible could at worst introduce variations in the results by 58%. The statistical analysis suggests a possible error of 25% in practice.

Suggestions are made for the improvement and simplification of the technique and for possible extensions of the investigations.

APPENDIX 1.

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APPENDIX II.

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APPENDIX III

Analysis of Signals of Strain Bridges.

The general Wheatstone bridge circuit as shown on Fig. AIII.1 can be represented by four arms each of nominal resistance R to which is applied a bridge voltage V. The measuring instrument has a resistance kR and it is required to find the current flow in the instrument, i_3, when resistances $R_1 - R_4$ change by fractions $m_1 - m_4$ of their original values.

Assuming current flows i_1 and i_2 as shown and zero source resistance

From ABC : $i_1 (2 + m_1 + m_4) + i_3(1 + m_4) = V/R$ AIII.1

From ADC: $i_2 (2 + m_2 + m_3) + i_3(1 + m_3) = V/R$ AIII.2

From ABD: $i_1 (1 + m_1) - i_2(1 + m_2) - ki_3 = 0$ AIII.3

hence

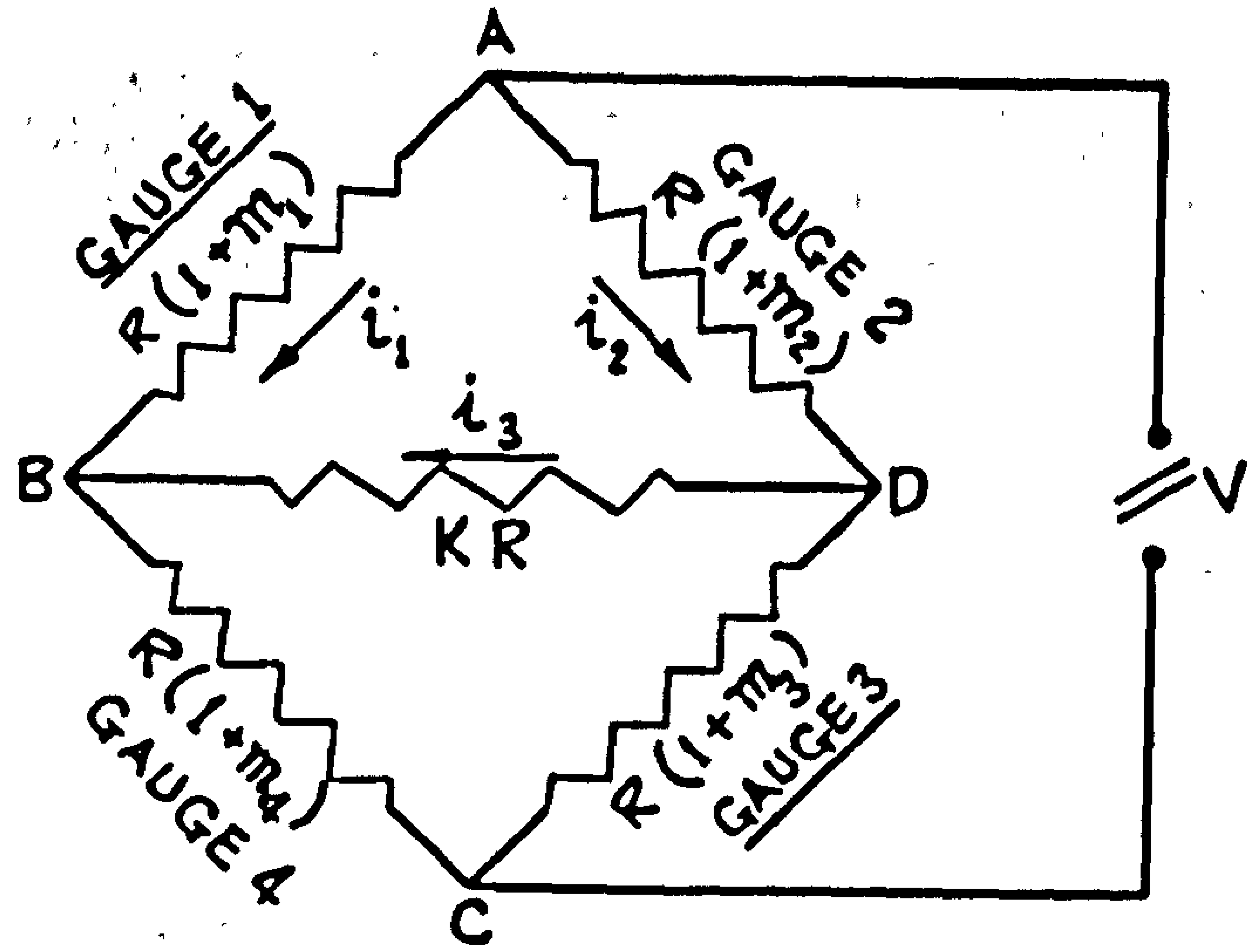
$$i_3 = \frac{(V/R)\left[m_1 - m_2 + m_3 - m_4 + m_1 m_3 - m_2 m_4\right]}{4(1 + k) + D_1 + D_2 + D_3 + D_4}$$

where

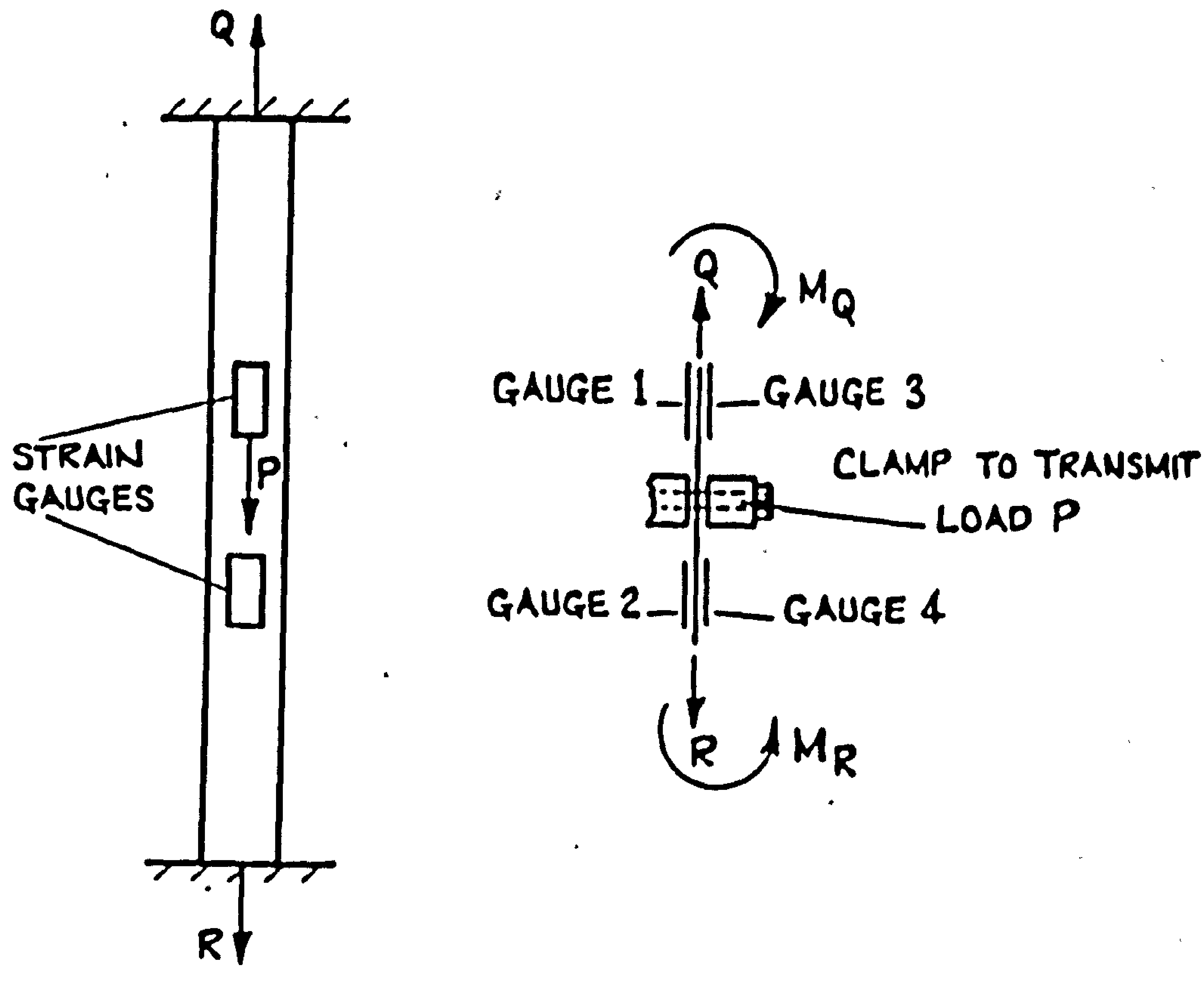
$$\begin{aligned} D_1 &= (3 + 2k)\times(m_1 + m_2 + m_3 + m_4) \\ D_2 &= (2 + k)\times(m_1 + m_4)(m_2 + m_3) \\ D_3 &= 2(m_1 m_4 + m_2 m_3) \\ D_4 &= m_1 m_3 (m_2 + m_4) + m_2 m_4 (m_1 + m_3) \end{aligned} \qquad \text{A.III.4}$$

if $m_1 - m_4$ are small numbers

AIII.4 becomes $i_3 = \dfrac{V}{4R} \cdot \dfrac{m_1 - m_2 + m_3 - m_4}{(1 + k)}$



BASIC WHEATSTONE BRIDGE CIRCUIT



GAUGE LAYOUT ON DYNAMOMETER STRIPS

FIG. AIII 1.

and the power input to the instrument = (current)^2.resistance

$$= i_3^2 \cdot kR$$

$$= \text{const} \times \frac{k}{(1+k)^2}$$

giving after differentiation the standard result that maximum instrument power input obtains when $k = 1$.

i.e. instrument resistance = bridge arm resistance.

the fractional resistance change $m = a\sigma + B$

where $a = S/E$ = gauge factor/Youngs Modulus for the underlying material - assumed constant.

An analysis of errors due to differing S values is given in Paul (1949).

σ = linear stress in the line of the gauge.

The stress is assumed uniform and the material linear elastic.

$b = C \times T$ = temperature coefficient of resistance x temperature change.

Equations A.III.4 become

$$i_3 = \frac{\left[a\,(\sigma_1 - \sigma_2 + \sigma_3 - \sigma_4)\,(1+b) + a^2(\sigma_1\sigma_3 - \sigma_2\sigma_4) + 2b^2\right] V/R}{4(1+k) + E_1 + E_2 + E_3 + E_4}$$

where :-

$$E_1 = (3+2k)\,(a\sum\sigma_m + 4b)$$

$$E_2 = (2+k)\left[a^2(\sigma_1 + \sigma_4)(\sigma_2 + \sigma_3) + 2ab\sum\sigma_m + 4b^2\right]$$

$$E_3 = 2\left[a^2(\sigma_1\sigma_4 + \sigma_2\sigma_3) + ab\sum\sigma_m + 2b^2\right]$$

$$E_4 = a^3\left[\sigma_1\sigma_3(\sigma_2 + \sigma_4) + \sigma_2\sigma_4(\sigma_1 + \sigma_3)\right] + 2a^2b\sum\sigma_m\sigma_n + 3ab^2\sum\sigma_m + 4b^3$$

A.III.5.

In the spring steel member of the initial force plate, the applied load P is in equilibrium with the changes in tension Q and R in the two sides of the strip as shown in Fig AIII.1. If the gauges are mounted axially on the centre line of the strip no signals arise due to offset load transmission in the plane of the strip. If the load actions at section KK cause direct stress σ_Q and bending stress σ_{M_Q}, and at LL σ_R and σ_{M_R} where

$$\sigma_1 = \sigma_Q + \sigma_{M_Q} \qquad \sigma_Q = Q/bt \;:\; \sigma_{M_Q} = \frac{6M_Q}{bt^2}$$

$$\sigma_2 = \sigma_R + \sigma_{M_R} \qquad \sigma_R = R/bt \;:\; \sigma_{M_R} = \frac{6M_R}{bt^2}$$

$$\sigma_3 = \sigma_Q - \sigma_{M_Q}$$

$$\sigma_4 = \sigma_R - \sigma_{M_R}$$

The relevant equation of force equilibrium is

$$P = Q - R$$

If the arrangement is symmetrical and the pretension is not reduced to zero in either part then $Q = 0.5\,P$.

If this is not fully realised, let $Q = gP$.

Then substituting in equation A.1.5.

$$i_3 = \frac{\left\{2aP(1+b)/A + a^2\left[P^2(2g-1)/A^2 - \sigma_{M_Q}^{\,2} + \sigma_{M_R}^{\,2}\right] + 2b^2\right\}V/R}{4(1+k) + F_1 + F_2 + F_3 + F_4}$$

where $F_1 = 2(3+2k)\left[aP(2g-1)/A + 2b\right]$ AIII.6

$$F_2 = (2+k)\left[a^2P^2(2g-1)^2/A^2 - a^2(\sigma_{M_Q} - \sigma_{M_R})^2 + 4abP.(2g-1)/A + 4b^2\right]$$

$$F_3 = 4\left[-a^2gP^2(1-g)/A^2 + abP(2g-1)/A + b^2\right]$$

$$F_4 = 2a^3\left[- g(1 - g)(2g - 1)\frac{P^3}{A^3} + (1 - g)P\,\sigma M_Q^2\,/A - nP\,\sigma M_R{}^2/A\right]$$
$$+ 2a^2 b\; P^2 (6g^2 - 6g + 1)/A^2 - 2a^2 b\,(\sigma M_Q{}^2 + \sigma M_R{}^2) + 6ab^2 P(2g - 1)/A + 4b^3$$

The following numerical values can be assumed to determine the magnitude of error terms for normal electrical resistance gauges.

$S = 2.0$: $E = 13{,}400$ ton/in^2 : $b' = 0.25$ in : $t = 0.02$ in.

$b = 10^{-4}$: $g = 0.55$: $k = 1.$

$Q = 0.2$ ton $\sigma M_Q = \sigma M_R = 4$ ton/in^2.

For these values the instrument current i_3 is given to first order of accuracy by

$$i'_3 = \frac{a P V}{4A R}$$

Taking account of the error terms in the numerator and in term F1:-

$$i_3 = i'_3 (1 - 1.1 \times 10^{-3} P + 7.5 \times 10^{-4}) = i'_3 (1 - 3.7 \times 10^{-4}) \quad \text{A.III.7}$$

The non-linearity of the bridge and the effect of other error terms in this expression is obviously negligible. The effect of error terms $F_2 - F_4$ in A.III.6 is even smaller.

If semi-conductor gauges having gauge factors of approximately 100 and a thermal sensitivity .12% per C° are used, equation A.III.7 becomes:-

$$i_3 = i_3' (1 - 0.0036 - 0.03P)$$

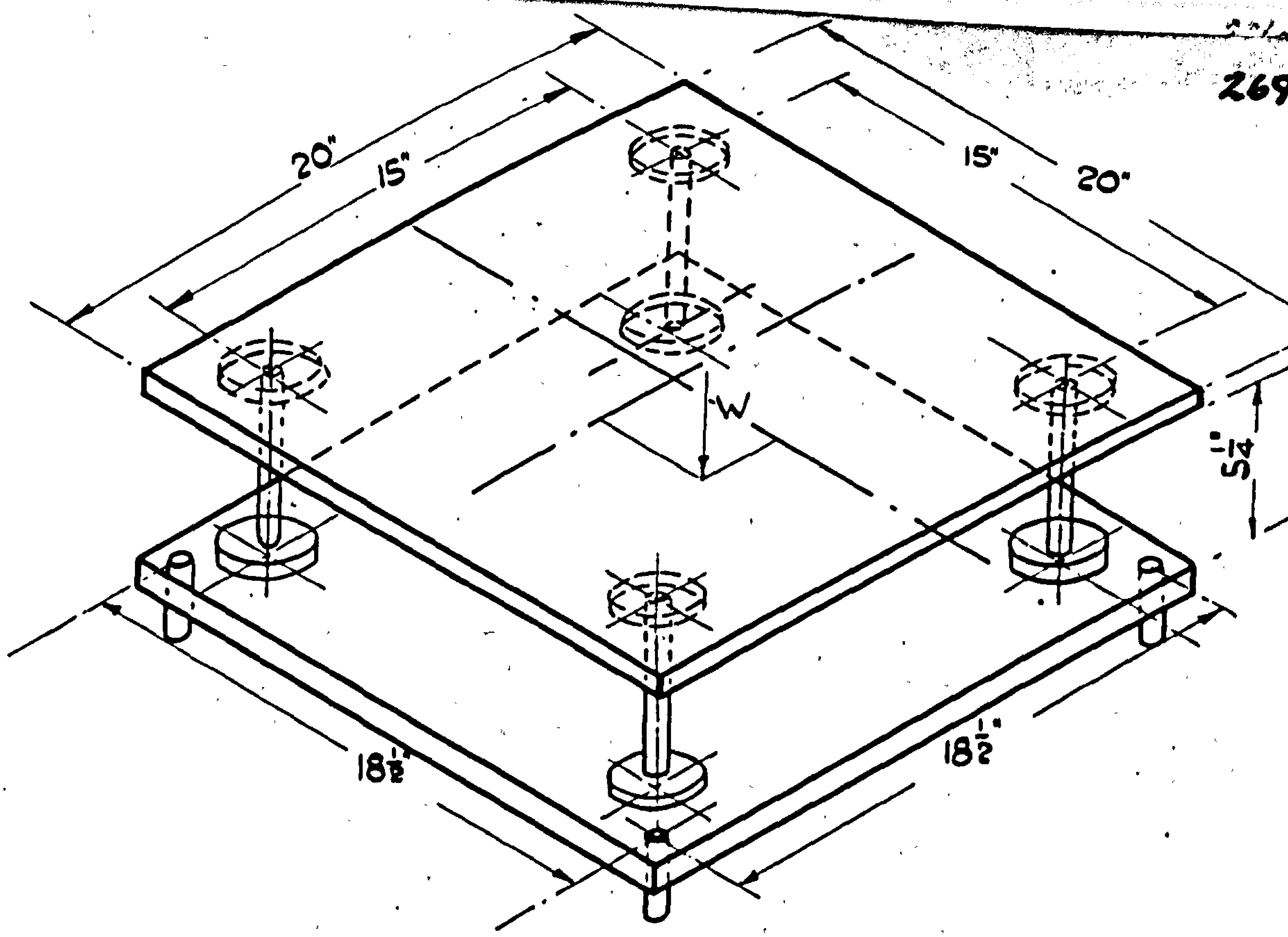
This indicates the change in sensitivity due to temperature and the possible non linearity of the bridge at large P values resultant on the use of semi-conductor gauges. The apparent error due to temperature is slightly reduced by a further factor not so far considered, namely temperature coefficient of gauge factor, but this is typically of the value 0.1 per cent per C° and does not eliminate the error due to change in bridge resistance. This can only be eliminated by using a constant current power supply. The non linearity of the bridge could be kept to a small value by using a more rigid dynamometer and working to a lower value of strain.

APPENDIX IV.

Structural Analysis of Four Column Force Plate.

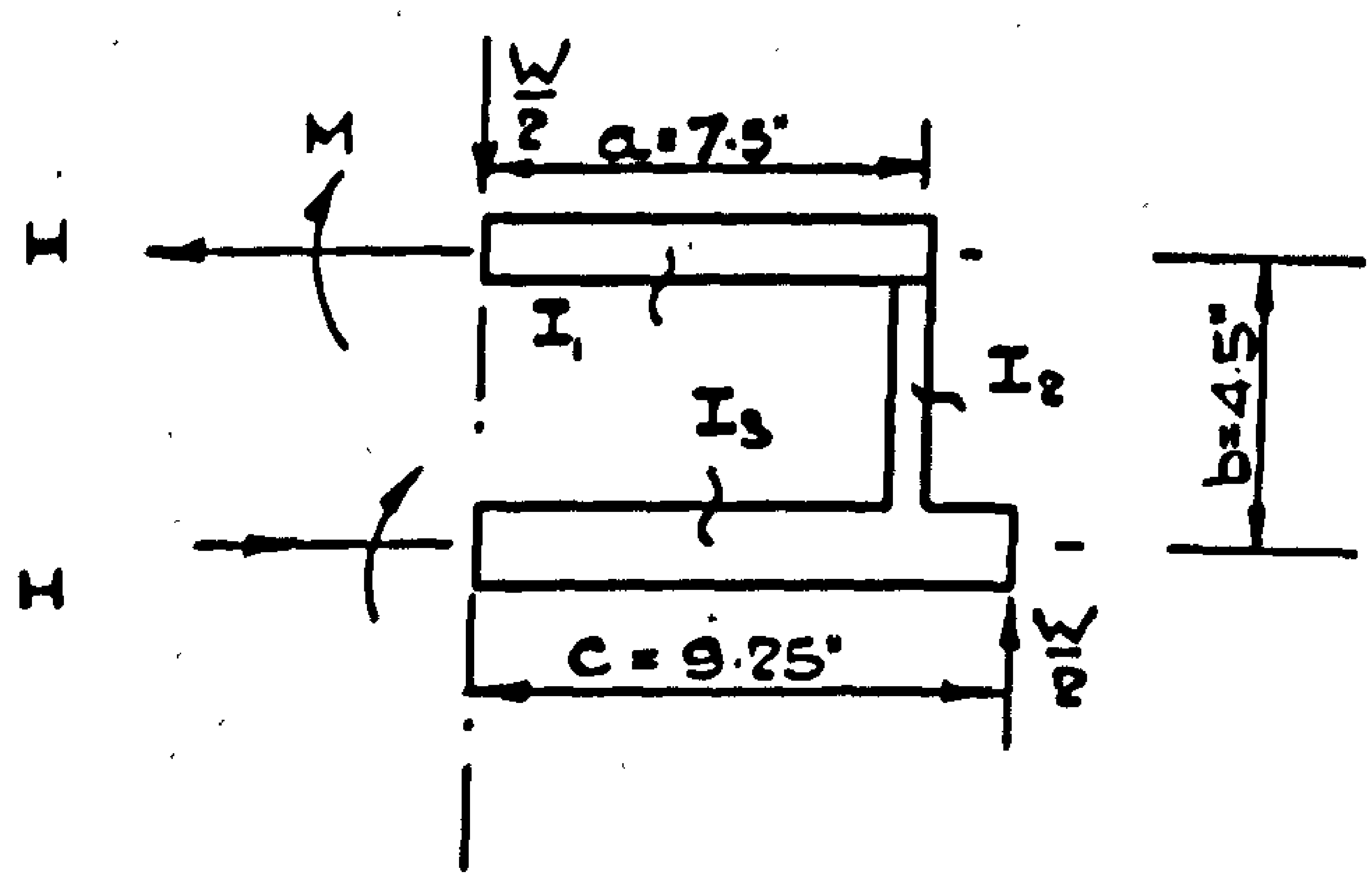
The force plate comprises an upper plate rigidly fixed to four symmetrically situated tubular columns fixed to a lower plate supported by levelling screws as shown in Fig.A.IV.1 The layout and circuits for the strain bridges are shown in Figs. 41 and 42. No exact analysis of the deformation of a flat plate loaded in the manner of the upper and lower members was found and the lengthy procedure of numerical analysis was not considered to be justified. As a design approximation the bending of the plate about each horizontal axis of symmetry was treated separately, and the curvature of each plate was calculated from the expression for a flat bar. As an initial approximation it was thought that the columns could be considered to carry axial force only since the second moments of area for the top and bottom plates, I, and I_3 had values of 0.407 in^4 and 1.068 in^4 respectively compared to the value for two columns of 0.007 in^4.

To check this the simple analysis for a central point load was performed. Fig. A.IV.2 shows the dimensions a, b and c defining the configuration of the symmetrical half of the structure with the statically indeterminate force actions H and M at the



PRINCIPAL DIMENSIONS OF FORCE PLATE.

FIG. A IV, 1.



SYMMETRICAL HALF OF EQUIVALENT STRUCTURE OF FORCE PLATE.

FIG. A IV. 2

load point. To reduce the size of the expressions, constants k_1, k_2, k_3 are defined as a/I_1, b/I_2 and a/I_3 respectively. The total strain energy U due to longitudinal stress due to bending is given by:-

$$U = \int_0^a (M - \tfrac{1}{2}Wx)^2\, dx/2EI_1 + \int_0^b (M - \tfrac{1}{2}Wa - Hx)^2\, dx/2EI_2$$

$$+ \int_0^a (M - \tfrac{1}{2}WC - Hb)^2\, dx/2EI_3 + \int_0^{c-a} (\tfrac{1}{2}Wx)^2\, dx/2EI_3 \qquad \text{AIV.1}$$

Applying Castigliano's theorem and the condition of symmetry requiring zero change of slope $\emptyset_A$ at A

$$E\emptyset_A = E\frac{\partial U}{\partial M} = 0 = M(k_1 + k_2 + k_3) - 1/4W(ak_1 + 2ak_2 + 2ck_3) - \tfrac{1}{2}Hb(k_2 + k_3) \qquad \text{A.IV.2}$$

Similarly the component displacement at A in the direction of H, δH is zero.

$$E\delta H = E\frac{\partial U}{\partial H} = 0 = \tfrac{1}{2}Mb(k_2 + 2k_3) + (1/4)Wb(ak_2 + 2ck_3) - (1/3)Hb^2(k_2 + 3k_3) \qquad \text{A.IV.3.}$$

Equations AIV.2 and AIV.3 can be solved simultaneously to get:-

$$H = \frac{3W\left[(ak_1 + 2ak_2 + 2ck_3)(k_2 + 2k_3) - 2(ak_2 + 2ck_3)(k_1 + k_2 + k_3)\right]}{2b\left[4(k_2 + 3k_3)(k_1 + k_2 + k_3) - 3(k_2 + 2k_3)^2\right]}$$

$$M = \tfrac{1}{2}W\frac{\left[2(k_2 + 3k_3)(ak_1 + 2ak_2 + 2ck_3) - 3(k_2 + 2k_3)(ak_2 + 2ck_3)\right]}{\left[4(k_2 + 3k_3)(k_1 + k_2 + k_3) - 3(k_2 + 2k_3)^2\right]} \qquad \text{A.IV.4.}$$

If the stiffnesses of the top and bottom plates are indefinitely large k_1 and k_3 tend to zero,

$H \rightarrow 0$ and $M \rightarrow \tfrac{1}{2}Wa$ and the bending moment in the column is zero.

If k_3 alone tends to zero

$$H \rightarrow -1.5 \; \frac{Wa}{b} \cdot \frac{k_1}{4k_1 + k_2}$$

$$M \rightarrow \tfrac{1}{2}Wa \; \frac{2k_1 + k_2}{4k_1 + k_2}$$

Similarly if k_1 alone tends to zero

$$H \rightarrow \frac{3W(a - c)}{b} \times \frac{k_3}{k_2 + 4k_3}$$

$$M \rightarrow \tfrac{1}{2}W \times \frac{ak_2 + 2k_3(3a - c)}{k_2 + 4k_3}$$

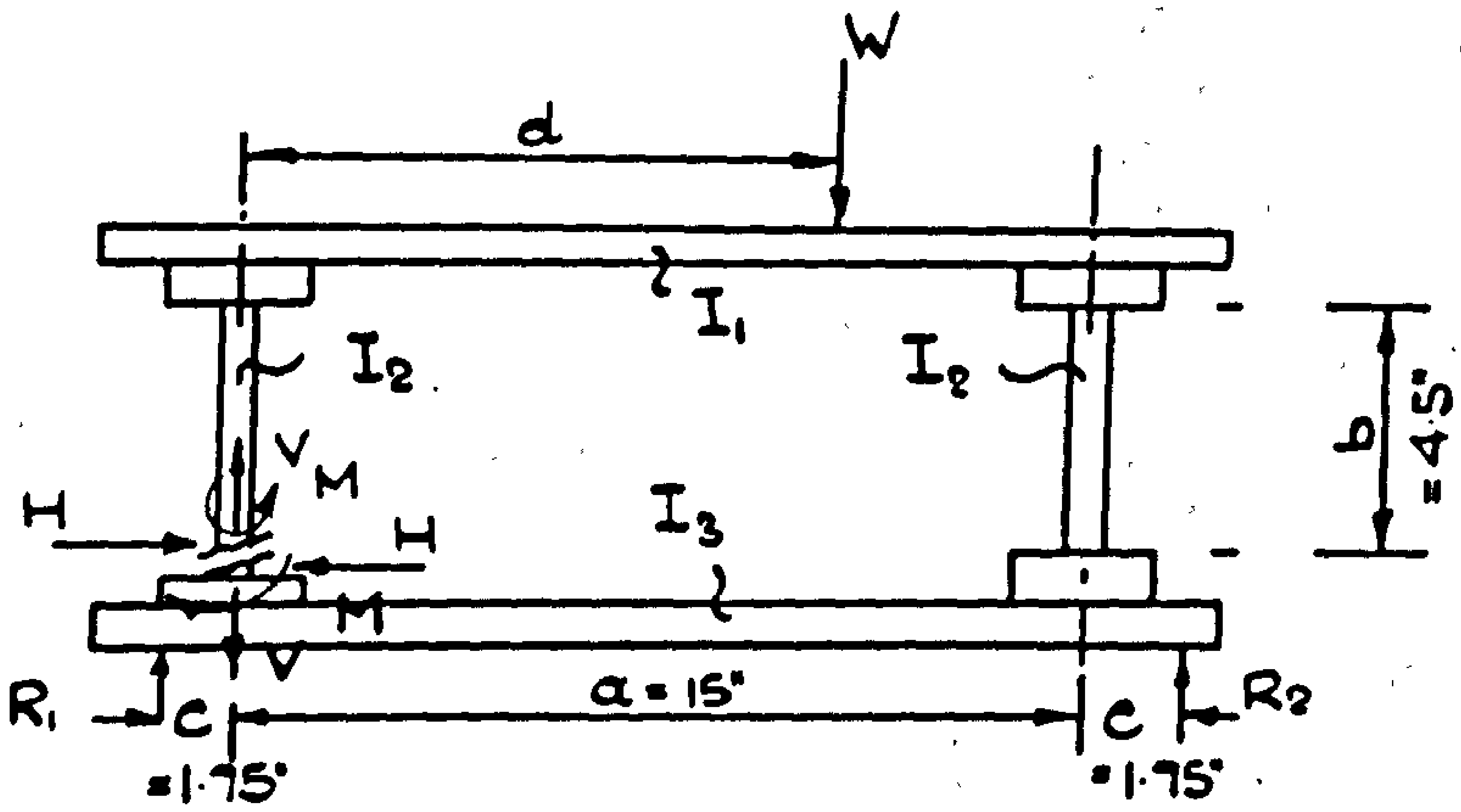
Tables A.IV.1 shows a comparison of the stresses in the tubular columns due to direct and bending actions for the above extreme conditions.

It is seen that the bending stresses in the columns have significant values compared with the longitudinal forces for the condition of central loading.

A further analysis was therefore performed using the actual values of the relevant moments of area and obtaining values of the load actions when the load is applied eccentrically. The relevant dimensions and load actions are shown in Fig. A.IV.3.

The reactions R_1 and R_2 are statically determinate and are given by

$$R_1 = W(a + c - d)/(a + 2c) \qquad R_2 = W(d + c)/(a + 2c) \qquad \text{A.IV.5.}$$



EQUIVALENT STRUCTURE OF FORCE PLATE

FIG. A IV 3

ASSUMED CONDITION FOR ANALYSIS	H TON	M. TON IN.	BENDING MOMENT AT COLUMN CENTRE TON IN.	σ_B BENDING STRESS AT COLUMN CENTRE TON/IN^2	σ_D DIRECT STRESS P/A TON/IN^2
K1 = K3 = 0	0	3.75W	0	0	3.86W
K3 = 0	-0.064W	3.56W	-0.05W	2.41W	3.86W
K1 = 0	-0.012W	3.73 W	+0.007W	0.34 W	3.86W
EXACT SOLUTION	-0.0745W	3.545W	-0.037W	1.82W	3.86W

ANALYSIS OF STRUCTURE OF FIG. A IV 2.

TABLE A IV 1.

Three indeterminate force actions exist in the closed frame and may be represented by the actions H, V and M acting at the cut section in the left hand pillars. Considering bending actions only the total strain energy, U in the structure, is

$$U = \int_0^b \frac{(M + H_x)^2\,dx}{2E\,I_2} + \int_0^d \frac{(M + Hb - Vx)^2 dx}{2EI_1}$$

$$+ \int_d^a \frac{(M + Hb - Vx + Wx - Wd)^2 dx}{2EI_1}$$

$$+ \int_0^b \frac{(M + Hx - Va + Wa - Wd)^2 dx}{2EI_2} + \int_0^a \frac{(R_1x + R_1c + M - Vx)^2 dx}{2EI_3}$$

$$+ \int_0^c \frac{(R_1x)^2\,dx}{2EI_3} + \int_0^c \frac{(R_2x)^2 dx}{2EI_3} \qquad \text{A.IV.6.}$$

The deformation conditions are as follows:-

$$\frac{\partial U}{\partial H} = \frac{\partial U}{\partial V} = \frac{\partial U}{\partial M} = 0 \qquad \text{A.IV.7}$$

Substituting the numerical values corresponding to the size of the force plate and solving the three simultaneous equations gives:-

$$H = W\,(0.0116249 + 0.0168412d - 0.0011227d^2)$$

$$V = W\,(0.996591 - 0.065909d - 0.00012605d^2 + 0.00000560d^3)$$

$$M = W(0.0387930 - 0.01965209d + 0.0007025d^2 + 0.0000420d^3) \qquad \text{A.IV.8}$$

The sum of the vertical loads in the two columns is equal to the applied load regardless of the value of d so that strain measurements of the axial strains in these columns will give a sum proportional to applied load for all load positions. The results of the plane analysis can obviously he used to obtain the same results in the general case. The axial strain in a column is obtained fromthe sum of the strains of linear gauges on opposite sides of the column which will automatically balance out bending strains. If one pair of gauges is mismatched the automatic balancing will not be perfect and the output voltage can be obtained from

$$V_S = \frac{V\sigma_D}{2} \quad \frac{S}{E}\left[1 + \frac{n}{8}\frac{\sigma_B}{\sigma_D}\right] \qquad \text{A.IV.9}$$

where V is the bridge voltage, S the gauge factor, n the fractional error in S, E, Young's Modulus and σ_B and σ_D the bending and direct stresses respectively.

From A.IV.8 themaximum bending moment occurs where $\frac{\partial}{\partial d}(M + H\frac{b}{2}) = 0$, i.e. at d = 6.51 in, and has the value 0.118W.

The error term $\frac{n}{8}\frac{\sigma_B}{\sigma_D}$ in A.IV.9 then becomes 0.19 n : or a 2 % error in gauge factor gives a bridge error of about 0.4%. This effect may therefore readilv give variations in bridge signal per unit vertical load, depending on the position of the load on the force plate.

The moment of W about the central position, M_z = W(d - 7.5), is signalled by linear gauges at the centre of the columns which effectively measure the difference between the direct loads on the two pairs of columns. In fact the moment is given by

$$M_z = W(d - 7.5) = (V_L - V_R)\, {}^a/2 + M_L - M_R .$$

where V and M are the direct load and bending moment in the columns respectively and L and R refer to the left and right column. This can be written

$$M_z = (V_L - V_r)\, {}^a/2 + V_L a - W(a - d)$$

Expressing V_L in terms of W and d from equation A IV.8.

$$M_z = (V_L - V_r)\, {}^a/2 + \text{error terms}.$$

The error terms have a maximum value of $\pm$ 0.0108 W when d = a/2 $\pm$ 4.01 in, the actual moment is then W x 4.01 and the error due to this effect is 0.25% only.

The moment term $M_L - M_R$ affects the bridge for measurement of shear force, and the vertical forces in the columns affect the bridges for moment about horizontal axes. When loaded by a horizontal shear force P as shown in Fig. A.IV.4 equation, AIV.6 becomes

$$U = \int_o^b \frac{(M + H_x)^2}{2\ EI_2}\, dx + \int_o^a \frac{(M + Hb - Vx)^2 dx}{EI_1}$$
$$+ \int_o^b \frac{(M + H_x - V_a + P_b - P_x)^2\, dx}{2EI_2} + \int_o^a \frac{\left[\frac{Pb(x + c)}{a + 2c} + M - V_x\right]^2 dx}{2EI_3}$$

A.IV.10

The deformation conditions remain as in A.IV.7. Using the same numerical values as before, the following results are obtained:-

$$H = -0.5000\,P$$
$$V = -0.1478\,P \qquad \text{A.IV.11}$$
$$M = 1.1411P$$

The bending moment at the same level on the right hand column is 1.1411 P also.

In the shear force bridge the signal derives from the sum of the bending moments on the columns, i.e. 2.2822 P. The error term due to offset vertical loading $M_L - M_R$, has a maximum value of 0.0108 W. Since W may be typically 140 lb. and P 20 lb the fractional error in P in these circumstances becomes .033 or 3.3 %.

Conversely due to shear force P the difference in vertical load in the columns is:-

$$V_L - V_R = 0.296\,P$$

Thus a maximum shear force of P_x = 20 lb would give an indication of an apparent M_Z = 45.2 lb. in. This is a gross cross-sensitivity which most users take account of.

Rotation of the top of the force plate about a vertical axis through its centre under the action of a couple C will result in rotation by angle θ corresponding to rotations of each column

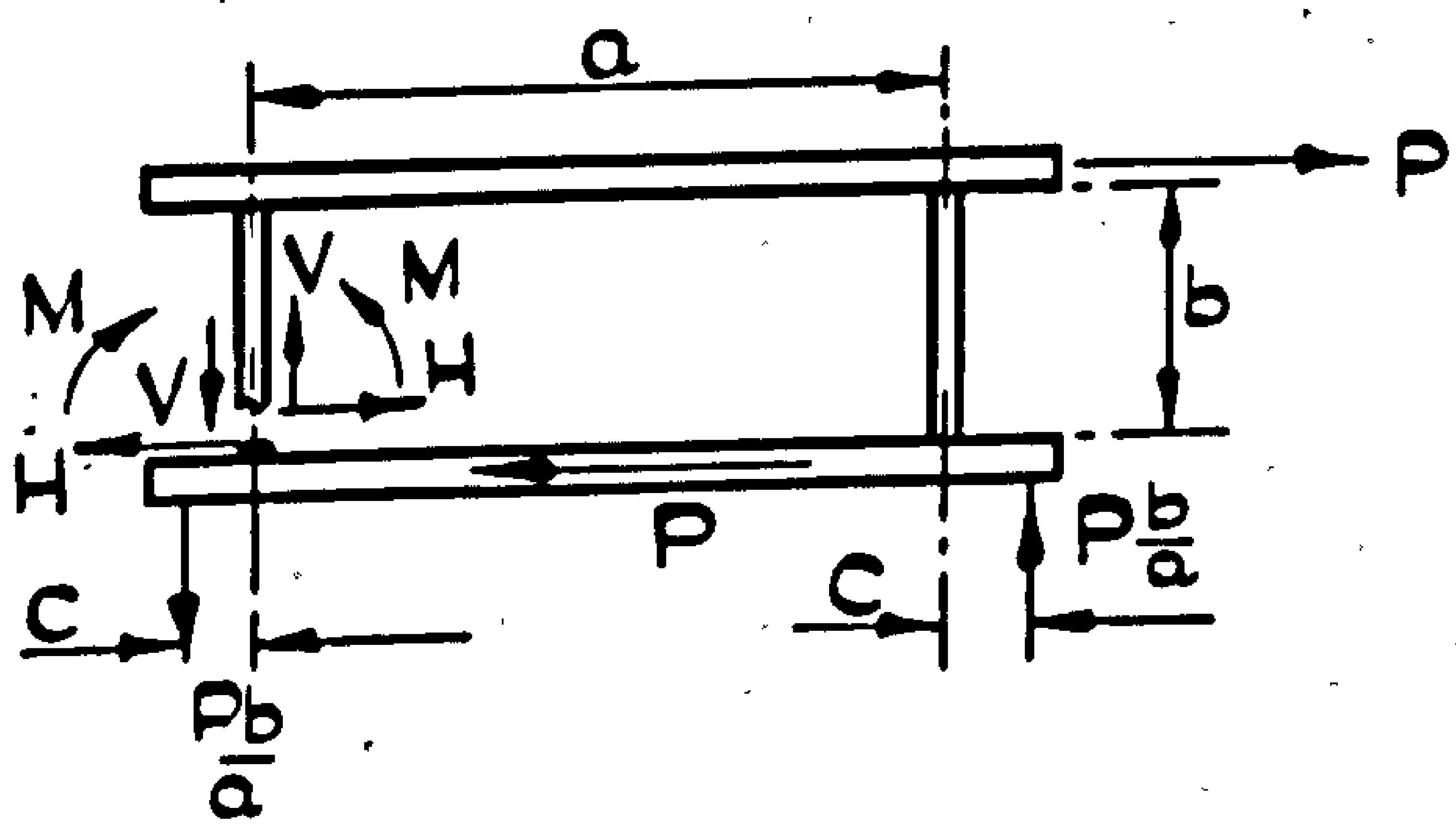


FIG. AIV4.

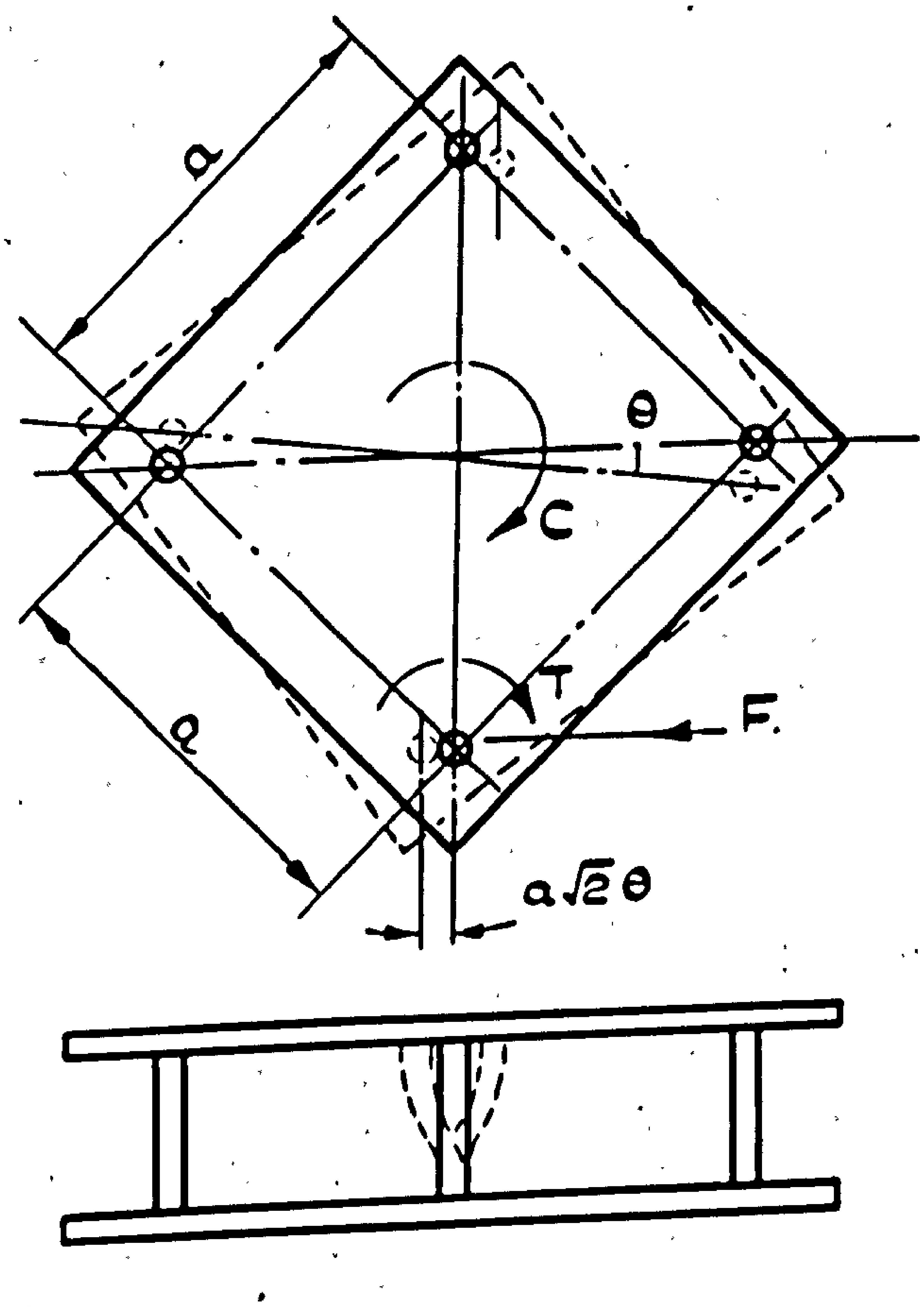


FIG. AIV5.

through θ and their translation along an arc by a distance $a\sqrt{2}\,\theta$ as shown in Fig. A.IV.5. The displacement corresponds to the end deflection of a fixed end beam under a transverse force F. Under this loading action it was considered that the deflections of the top and bottom plates would be negligibly small.

The overall equilibrium of the top plate in respect of moments about the vertical axis is given by:-

$$C - 4T - 4Fa\sqrt{2} = 0 \qquad \text{A.IV.12}$$

where T is the torque necessary to twist each tube through an angle θ.

The load deformation equations are:-

for column bending $a\sqrt{2}\,\theta = Fb^3/12EI$ A.IV.13

and $\theta = Tb/(0.4E) x(2I)$ A.IV14.

Hence $F = 0.35\ Ca/(a^2 + b^2/30)$

and $T = 0.25\ C/(1 + 15\ a^2/b^2)$.

On the force plate columns all gauges measure extensional strain and the shear stress due to T will not produce a signal.

Substituting the actual values of a and b the bending and shear stresses due to C amount to 4.57×10^{-3} C ton/in² and 3.25×10^{-4} C ton/in² respectively. The bending stress is zero at the centre and maximum at the ends of the column. The longitudinal gauges connected to measure vertical force and moments about horizontal axes will not therefore be affected by the force actions due to a moment about a vertical axis.

Referring to Fig. A.IV.5. it is apparent that the component displacements of the four columns in the X or Z direction sum to zero: i.e. the bending moments on the column in planes parallel to YOZ are equal and opposite in columns 12 and 34, and presuming satisfactory matching of gauges there should be no cross-sensitivities between the Torsion and shear force channels.

APPENDIX V.

Dynamic Performance of Force Plate.

The force plate may be considered to be a single mass, comprising the top plate and attachments, supported by an elastic system allowing six modes of simple vibration, 3 linear and 3 angular. In fact the system is more complicated than this since flexural vibration of the top plate is possible in addition. In this analysis the plate vibration is neglected and the effect of coupling of different modes of vibration is considered as a factor of secondary importance. The most important modes of vibration are those involving movement of the top plate in the horizontal plane, and Cunningham and Brown (1952) found that viscous dampers were necessary to reduce oscillations in these modes. As previously described similar damping units were fitted to the force plate used in the present investigation.

The analysis in Appendix IV can be extended to give the sideways deflection under horizontal load and this was found to be 2.01×10^{-5} in/lb. An experimental investigation was attempted using dead weights and a dial gauge reading to 10^{-4} in but in the range of loads applied this system was unable to give reliable results. The natural frequency of transverse vibrations was measured to be 56 cycles/sec. Taking a top plate weight of 70lb and the calculated stiffness, gives a theoretical natural frequency of 83 cycles/sec. Since the calculation assumes rigid fixing at the joints and neglects the depth of the plate and flanges this figure may not be unreasonable.

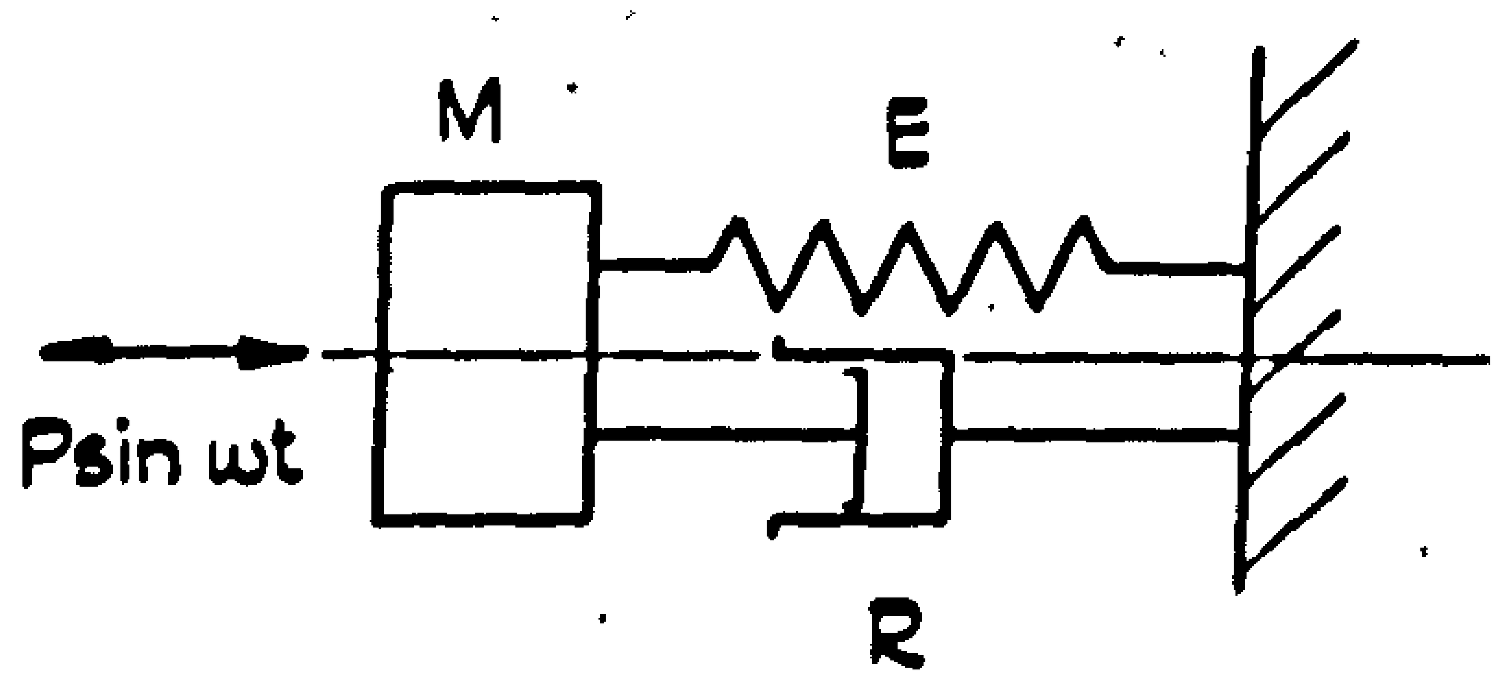
The natural frequency was measured by striking the top plate at the centre of one side with a weight moving horizontally and noting the response of the galvanometer which indicated shear force. This also allowed the measurement of the damping. After 5 cycles the amplitude of this vibration had diminished from 1.00 to 0.18 experimental units. When in use the top plate of the force plate is an elastically restrained, damped mass to which a disturbing force is applied. As the test subject steps on the plate additional mass is added to top plate but the connection involves the joints of the body whose elastic stiffness depends on muscular tension and can therefore be taken to be negligibly small. The effect of the added body mass is therefore neglected.

For the system sketched in Fig. A.V.1 the controlling differential equation is

$$\ddot{x} + \frac{E}{M}x + \frac{K}{M}\dot{x} = \frac{P}{M}\sin\omega t \qquad \text{----- A.V.1.}$$

This has the solution

$$x = A^{-\Delta t}\sin mt + Be^{-\Delta t}\cos mt - C\frac{P}{M}\cos\omega t - D\frac{P}{M}\sin\omega t$$



SIMPLIFIED SYSTEM FOR
FORCE PLATE.

FIG. A.V. I.

where A and B are constants of integration:-

$$D = {}^{K}/_{2M} \;:\; n^2 = {}^{E}/_{M} \;:\; C = \frac{2\Delta\omega}{(n^2-\omega^2)+4\Delta^2\omega^2}$$

$$m^2 = n^2 - \Delta^2$$

$$D = \frac{(n^2 - \omega^2)}{(n^2 - \omega^2)^2 + 4\Delta^2\omega^2}$$

Initial conditions are $t = 0 : x = \dot{x} = 0$

$$\therefore B = C \; {}^{P}/_{M} \quad \text{and} \quad A = \frac{\omega}{n} D \; {}^{P}/_{M}.$$

$$\text{i.e. } x = \frac{P}{M} D\left[({}^{\omega}/_{n})e^{-\Delta t} \sin mt - \sin \omega t\right]$$

$$+ \frac{P}{M} C(e^{-\Delta t} \cos mt - \cos wt)$$

The measuring circuits are arranged so that the signal is proportional to the force action in the columns, i.e. the signal is proportional to -Ex.

The quantity to be measured is P sin ωt.

The error in the reading η is given by

$$\eta = P \sin \omega t - (-Ex)$$

$$\eta/P = \sin \omega t - n^2 D \sin \omega t - n^2 C \cos \omega t$$

$$- e^{-\Delta t}(n\,\omega D \sin nt + n^2 C \cos nt)$$

The steady static error $\eta_S = \sin\omega t - \dfrac{n^2(n^2 - \omega^2) \sin \omega t}{(n^2 - \omega^2)^2 + 4\Delta^2\omega^2}$

$$- \frac{n^2\, 2\Delta\omega \cos \omega t}{(n^2 - \omega^2) + 4\Delta^2\omega^2} = \sin \omega t - \frac{n^2}{\sqrt{(n^2 - \omega^2) + 4\Delta^2\omega^2}} \sin(\omega t + \alpha)$$

where $\tan \alpha = \dfrac{2\Delta\omega}{n^2 - \omega^2}$

i.e. an amplitude error of $1 - \dfrac{n^2}{(n^2 - \omega^2) + 4\Delta^2\omega^2}$

and a phase error, α, of $\tan^{-1} \dfrac{2\Delta\omega}{n^2 - \omega^2}$

The transient error $\eta_T = -e^{-\Delta t} (n\omega D \sin m t + n^2 C \cos mt)$

$$= -e^{-\Delta t} F \sin (mt + \beta)$$

where $F = n^2 \omega^2 D^2 + n^4 C^2 = \dfrac{n \;\omega^2(\omega^2 - n^2)^2 + n^2 4\Delta^2 \omega^2}{(n^2 - \omega^2)^2 + 4\Delta^2\omega^2}$

$$= n\omega \frac{(n^2 - \omega^2)^2 \; 4n^2\Delta^2}{(n^2 - \omega^2)^2 + 4\Delta^2\omega^2}$$

$$\tan \beta = \frac{n^2 \; 2\Delta\omega}{n\omega \; (n^2 - \omega^2)} = \frac{2\Delta n}{n^2 - \omega^2}$$

From the experimental observations of free vibrations

$m = 2\pi \times 56 = 352$ rad/sec

$$\Delta = \frac{m}{2\pi n} \log_e \frac{A_1}{A_{1+n}} = \frac{56}{2\pi \times 5} \log_e {}^1/0.18$$

$= 19.2.$ rad/sec.

$n^2 = m^2 + \Delta^2 = 124,00 \; {}^1/sec^2$: $n = 352$ rad/sec.

Steady state error in amplitude $= 1 - {}^1/\sqrt{(1 - \dfrac{\omega^2}{n^2}) + 4\Delta^2 \; {}^{\omega^2}/n^4}$

for a maximum error of 1%, ${}^{\omega}/n = 0.071$: $\Omega = 4.0$ cycles/sec.

Taking the stance phase as one half of the cycle time of the fundamental term of a Fourier series, the fundamental frequency is 0.8 cycles/sec i.e. the steady state error is less than 1% up to the fifth harmonic. The maximum steady state time lag is $/\omega = \dfrac{1}{\omega} \tan^{-1} 2\Delta\omega/ (n^2-\omega^2) = 2\Delta/n^2 = 0.0003$ sec. the time interval between successive measurements is 0.02 sec. and the steady state time lag is therefore negligible. The expression for the amplitude of the transient error η_T is given approximately by $\eta_T \doteq (\omega/n)\, e^{-\Delta t} \sin(nt + \beta)$.

For the fifth harmonic of the walking cycle, $\omega/n = 0.071$ and β the phase angle is given by

$\tan \beta = 2\Delta/n = 0.109$ hence $\beta = 0.11$ rad $= 6.3°$.

The first peak of the transient error term will occur when

$nt + \beta = \pi/2$ i.e. $t = .0043$ sec.

i.e. $\eta_T = 0.071/109 = 0.07$

This is a significant error at the level of the fifth harmonic. The value at the fundamental frequency of the walking cycle is 0.014 or 1.4%. This transient error however has a frequency of 56 cycles per second and readings are taken at 0.02 sec intervals. Where higher frequency undulations were obviously present in the records the observations taken were on a mean line selected by eye and this error was not therefore likely to be significant. A Fourier analysis of the force plate record from a typical test subject is shown in table 4 page 128. It is seen that except in the case of channel 1 the coefficient of any term after the seventh harmonic is less than one twentieth of the major harmonic. It appears therefore that the force plate dynamic response is adequate to record the force actions transmitted to it during walking. The possibility remains that its performance might be inadequate for activities developing higher frequency spectra such as running, jumping or even for an amputee using a prosthesis.

The other lateral mode of vibration is identical to that considered. Linear vibrations in the direction of the y axis are found to occur at 260 cycles/sec and all response errors will thérefore be negligibly small. Rotational vibration about the reference axes could not be identified experimentally but simple theoretical analysis indicated that they should in each case have a higher natural frequency than the corresponding linear vibration.

APPENDIX VI.

Analysis of Angular Acceleration Force Actions.

Each body segment may be assumed to be a rigid body moving in space. If Fischers' results are to be relied on the principal moments of inertia, I_1, I_2, I_3 are given for the thigh, say,by:-

$$
\begin{aligned}
I_1 &= C_{1T}\, W\, (C_{3T}\, L_T)^2 \;/g \\
I_2 &= C_{1T}\, W\, (C_{4T}\, D_T)^2 \;/\, g \qquad\qquad \text{A VI.1} \\
I_3 &= C_{1T}\, w\, (C_{3T}\, L_T)^2 \;/g
\end{aligned}
$$

where the axes 1, 2 and 3 coincide with front to back,vertical and lateral lines of the limb segment located for reference with its lines of joint centres vertical. The moments about these axes due to angular displacements, θ_1, θ_2, θ_3 are given by

$$
\begin{aligned}
M_1 &= I_1\ddot{\theta}_1 - (I_2 - I_3)\,\dot{\theta}_2\,\dot{\theta}_3 \\
M_2 &= I_2\ddot{\theta}_2 - (I_3 - I_1)\,\dot{\theta}_3\,\dot{\theta}_1 \qquad\qquad \text{A.VI.2.} \\
M_3 &= I_3\ddot{\theta}_3 - (I_1 - I_2)\,\dot{\theta}_1\dot{\theta}_2
\end{aligned}
$$

As noted in the experimental procedure it was not found possible to measure component rotations of limb segments about their long axes and these are in any case small. If θ_2 and its derivatives are neglected, and $I_1 = I_3 = I_{max}$ equations A. VI. 2 become

$$
\begin{aligned}
M_1 &= I_{max}\,\ddot{\theta}_1 \\
M_2 &= 0 \qquad\qquad \text{A.VI.3} \\
M_3 &= I\;\ddot{\theta}
\end{aligned}
$$

If direction 1 is inclined in space to the reference x, y, z directions by angles defined by direction cosines t_{1x}, t_{1y} and t_{1z} and directions 2 and 3 similarly by t_{2x}, t_{2y}, t_{2z}: t_{3x}, t_{3y}, t_{3z}, the corresponding moments about the x, y, z axes are

$$\begin{aligned} M_x &= M_1 t_{1x} + M_3 t_{3x} = I_{max}\left[\ddot{\theta}_1 t_{1x} + \ddot{\theta}_3 t_{3x}\right] \\ M_z &= M_1 t_{1z} + M_3 t_{3z} = I_{max}\left[\ddot{\theta}_1 t_{1z} + \ddot{\theta}_3 t_{3z}\right] \end{aligned} \qquad \text{A.VI.4.}$$

If the angular accelerations measured with respect to the x and z axes are respectively $\ddot{\theta}_x$ and $\ddot{\theta}_z$ then

$$\begin{aligned} \ddot{\theta}_1 &= \ddot{\theta}_x t_{1x} + \ddot{\theta}_z t_{1z} \\ \theta_3 &= \ddot{\theta}_x \; {}_{3x} + \ddot{\theta}_z t_{3z} \end{aligned} \qquad \text{A. V1.5.}$$

Equations A.V1.4. then become

$$\begin{aligned} M_x &= I_{max}\left[\ddot{\theta}_x (t_{1x}^2 + t_{3x}^2) + \ddot{\theta}_z (t_{1z} t_{1x} + t_{3z} t_{3x})\right] \\ M_z &= I_{max}\left[\ddot{\theta}_x (t_{1z} t_{1x} + t_{3z} t_{3x}) + \ddot{\theta}_z (t_{1z}^2 + t_{3z}^2)\right] \end{aligned} \qquad \text{A.V1.5.}$$

From equation (58) developed earlier

$$t_{1x} = 1 - \tan^2 \theta_z (1 - \frac{1}{R})/S$$

$$t_{1z} = t_{3x} = \tan \theta_z \tan \theta_x (1 - \frac{1}{R})/S$$

$$t_{3z} = 1 - \tan^2 \theta_x (1 - \frac{1}{R})/S$$

where $R = \sqrt{1 + \tan^2 \theta_x + \tan^2 \theta_z}$ and $S = \tan^2 \theta_x + \tan^2 \theta_z$

$$M_x = I_{max}(\ddot{\theta}_x \sec^2\theta_x + \ddot{\theta}_z \tan\theta_z \tan\theta_x)$$

$$M_z = I_{max}(\ddot{\theta}_x \tan\theta_z \tan\theta_x + \ddot{\theta}_z \sec^2\theta_z)/R^2 \qquad \text{A.VI.7.}$$

For the test subject No. 4 test 2 the graphs of angular inclination θ_x, θ_z are shown in Fig. 62 . At the time of maximum θ_z the values of θ_z, θ_x, $\ddot{\theta}_z$ and $\ddot{\theta}_x$ are respectively 24.8°, 4.3°, 38.7 rad/sec^2 and 4.3 rad/sec^2. Taking subject weight of 127 lb and thigh length 16 in. equation A. VI.7 gives the following values:-

$$M_x = 4.14 \text{ lb in.} \qquad M_z = 36.3 \text{ lb in.}$$

The corresponding values calculated from equations 46 are

$$M_x = 3.11 \text{ lb. in} \quad \text{and } M_z = 36.2 \text{ lb in.}$$

It is apparent that the contribution of $\ddot{\theta}_z$ apparent to M_x is a considerable fraction of the result, but the amount of the term is small in any case.

The accuracy of the larger inertia term is seen to be satisfactory.

APPENDIX VII.

Correlation with the Results of Rydell (1966).

Rydell quotes the force actions on the head of the femur relative to axes HX, HY, HZ through H the centre of the femoral head where HX coincides with line OHC in Fig. A.VII.1. Rydell defines the angles t and u for his test subjects and Backman (1957) quotes angle p as 5 - 6 degrees for a wide range of femora. To relate the present author's results to those of Rydell it is necessary to define $\emptyset$ the apparent angle of ante-torsion when the femur is viewed along the axis HK and it can be seen from Fig. A.VII.2 that:-

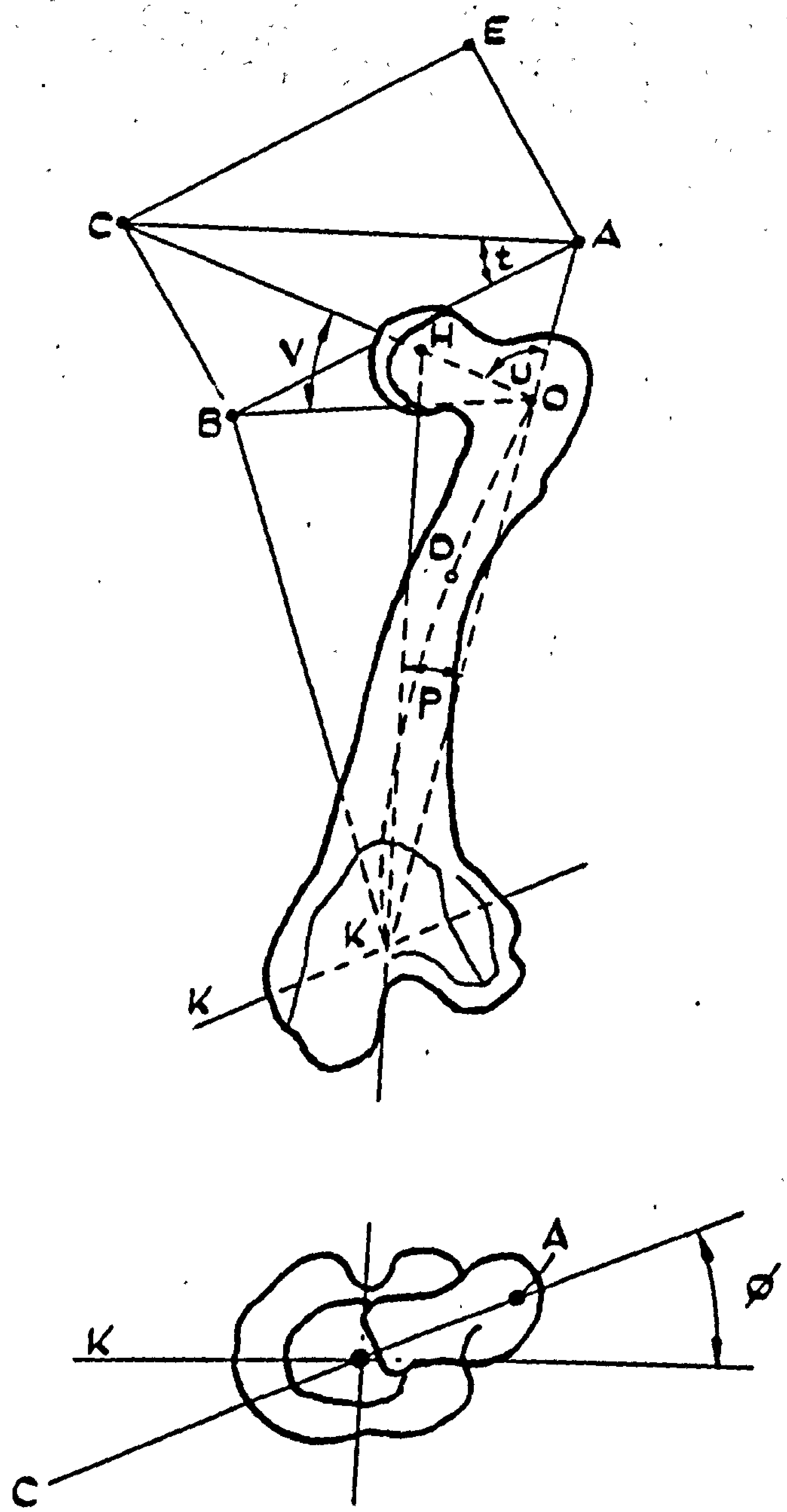
$$\tan\emptyset = B_3P_3/A_3P_3 = B_1P_1/A_2B_2 \cos p.$$
$$= A_1B_1 \sin t / A_1B_1 \cos t \cos p$$
$$= \tan t / \cos p \qquad \text{A.VII. 1}$$

Rydell defines a point $P(X_R, Y_R, Z_R)$ on the surface of the femoral head by angles α and γ measured from the axes X_R and Y_R of the prosthesis Fig. A.VII.3 If the femoral head is of radius R,

$$X_R^2 + Y_R^2 + Z_R^2 = R^2$$

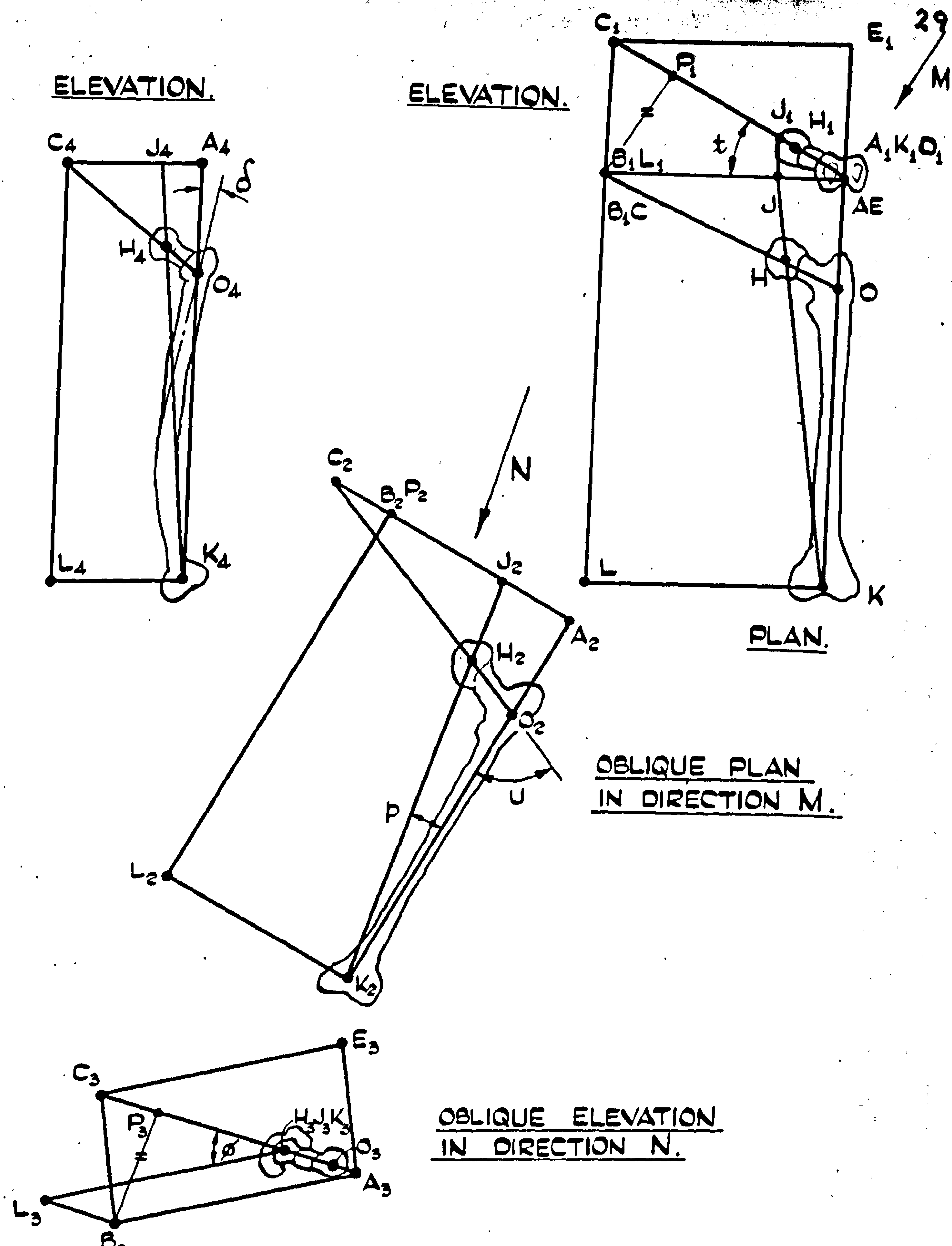
and $\quad Y_R/X_R = \tan\alpha \qquad Z_R/Y_R = \tan\gamma.$

hence $\quad X_R = R/\sqrt{1 + \tan^2\alpha + \tan^2\alpha\tan^2\gamma} = \dfrac{R\cos\alpha}{\sqrt{1 + \sin^2\alpha\tan^2\gamma}}$



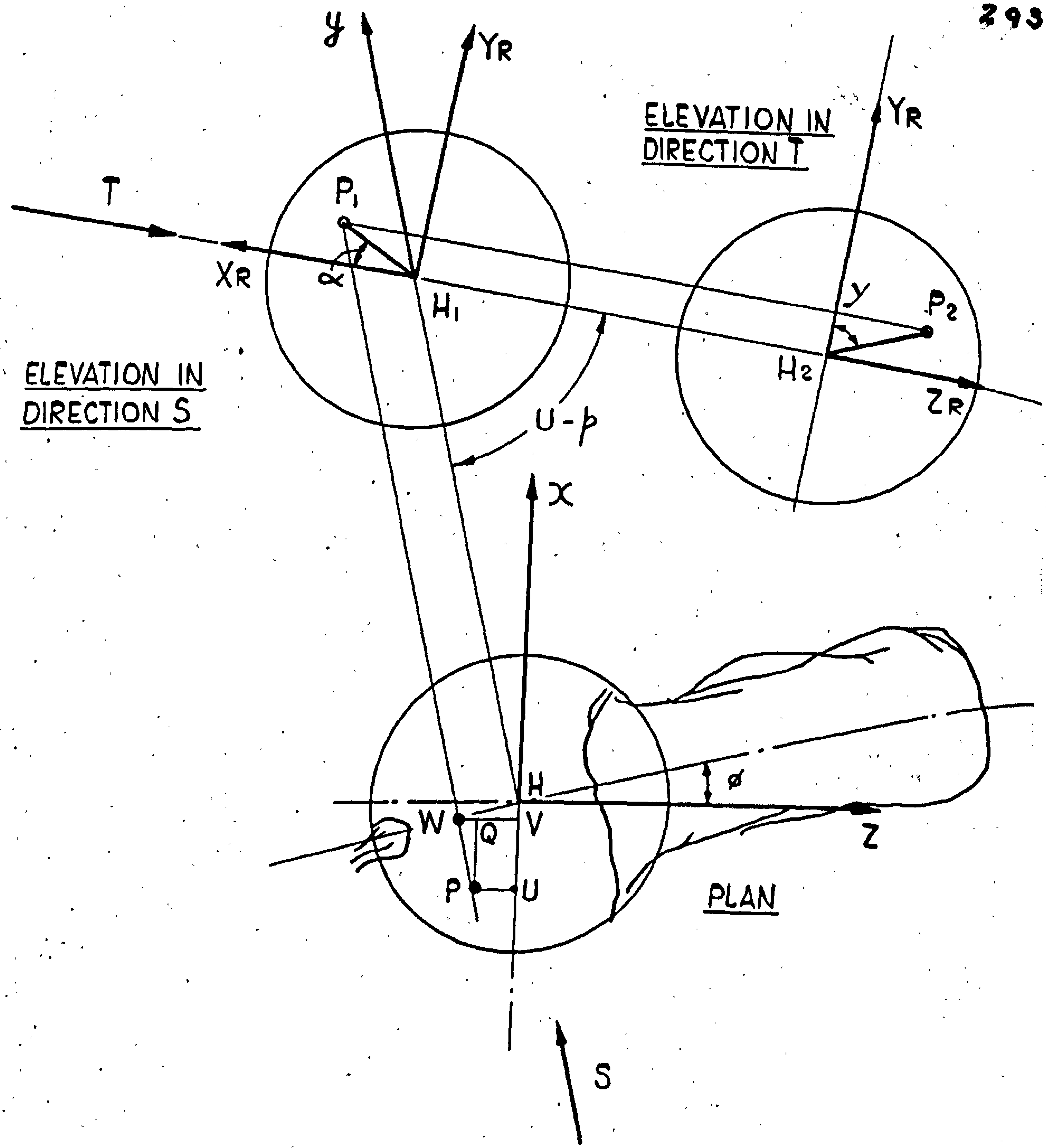
REFERENCE ANGLES OF THE FEMUR ADAPTED FROM BACKMAN (1957)

FIG. A VII 1.



GEOMETRIC PROJECTIONS OF FEMUR

FIG. A VII 2.



GEOMETRIC PROJECTIONS OF HEAD OF FEMUR.

RYDELL'S AXES : X_R, Y_R, Z_R.

PAUL'S AXES : x, y, z.

FIG. A VII 3

$$H_1P_1 = \sqrt{X_{R^2} + Y_{R^2}} = R./\sqrt{1 + \sin^2\alpha \tan^2\gamma}$$

$$Z_R = R \tan\alpha \tan\gamma/\sqrt{1 + \sin^2\alpha \tan^2\gamma} = WP$$

$$HW = H_1P_1 \sin(U - p - \alpha)$$

$$\therefore x = -(HV + VU) = -(HW \sin\emptyset + WP \cos\emptyset) \quad \text{(A.VII.2)}$$

$$= \frac{-R\left[\sin(U - p - \alpha) \sin\emptyset + \tan\alpha \tan\gamma \cos\emptyset\right]}{\sqrt{1 + \sin^2\alpha \tan^2\gamma}}$$

$$z = -(WV - WQ) = -HW \cos\emptyset + WP \sin\emptyset \quad \text{(A.VII.3)}$$

$$= \frac{-R\left[\sin(U - p - \alpha) \cos\emptyset - \tan\alpha \tan\gamma \sin\emptyset\right]}{\sqrt{1 + \sin^2\alpha \tan^2\gamma}}$$

Referring to Fig. A.VII.2 it is seen in the side elevation that the axis of the stem of the prosthesis is not coincident with the "ideal" axis OK. This may be treated approximately by subtracting from Rydell's values of γ the value of the angle δ between these axes. The following values of angles are common to both test subjects : $\delta = 8°$: $p = 5°$: $u = 60°$

For case 1 :- $t = 8°$ and therefore $\emptyset = 8.1°$

For case 2 :- $t = 35°$ and therefore $\emptyset = 35.1°$

The position of the point of intersection of the resultant joint force vector with the surface of the joint is defined by angles α and γ and the corresponding x and z values can therefore be calculated using equations A.VII. 2 and 3.

THESIS BY J.P. PAUL

"FORCES AT THE HUMAN HIP JOINT"

Amendments

p.ii	Add Appendix VI p.287 Appendix VII p.290.
p.iii	C_o Table 3 Add p.69.
p.vi	After See fig. 47, p.152, add and fig. 52 p.166.
p.4	Line 16. The combination of the two hip bones with the sacrum .....
p.75	3rd para. line 2 for $\emptyset_2$ read $\emptyset_1$.
p.118	Line 5 dynamometer 1.
p.131	After 1/100 delete the.
p.147	Line 2 for "from" read "frame".
p.150	Line 3 forces.
p.159	After $W_T = C_{1T}W$ add (41).
p.173	4th last line for "volume" read "value".
p.174	After eqn. 56 "plan view of fig. 54" substitute "fig. 55".
p.176	Last set of equations number (59).
p.177	First line $P_z = T_z p$.
p.215	Title "moments about X-AXIS" substitute "Z-AXIS".
p.239	Line 13 for "$X_H - z_B$", read "$z_H - z_B$".
p.259	Add PAUL, J.P. (1967) "Forces transmitted by joints in the human body" Proc. I. Mech. Eng. 181, 3J; 8.....
p.281	Last line $x = Ae^{-At} \sin mt + ...$
p.284	Last line "is α/ω = ..."

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p.281	Last line $x = Ae^{-\Delta t} \sin mt + ...$
p.284	Last line "is α/ω = ..."