A BIOMECHANICAL ANALYSIS OF HUMAN STRENGTH

A Thesis presented to the University of London
for the degree of Doctor of Philosophy,

by

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ABSTRACT

1. Human strength, regarded as the maximum force that subjects can exert under given test conditions has been frequently observed and reported. The influences of biomechanical factors on the results of tests of strength are both complex and obscure. Hitherto, these influences have received very little attention and they are therefore the subject of this thesis. Particular emphasis has been given to the influence of posture because of its theoretical and practical importance.

2. Three contrasting tests of strength were experimentally investigated.

(i) Strength of plantarflexion was investigated as a function of posture of the ankle and knee. These observations were collated with quantitative cadaveric studies of the relevant muscles and joints.

(ii) Strengths of pronation and supination were measured as a function of the posture of the forearm. The ability to transmit supinator torques using ranges of uniform cylindrical handles and the handles of commercially available screwdrivers was investigated.

(iii) Strength of pulling in the sagittal plane in a variety of two handed tests was measured in 165 subjects. A multivariate statistical analysis established the extent to which performance was determined by body weight and stature. Results confirmed the predictions of a theoretical analysis of the task by means of free-body diagrams. They

showed that, in any given tests, the proportion of the variance in strength which could be attributed to weight and stature was closely related to the posture of the subject. A further experiment investigated the strength of extension of the trunk as a function of posture.

3. The interaction of the physiological properties of the musculature; the transmission of stresses through the interfaces between the body and its mechanical environment; and the weight and leverage of the body is discussed.

CONTENTS

FRONTISPIECE

from

"De Motu Animalium"

by

Giovanni Alfonso Borelli

1680

footnote:-

Each figure caption in this thesis includes the number of the section in which the contents of the figure are principally discussed.

CHAPTER 1 - INTRODUCTION

1,1 Man's ability to perform forceful physical actions is determined by his anatomy, his physiology and the nature of the task in which he is engaged. The measurements of this capacity, commonly referred to as strength, has been the subject of many studies. The object of the present work is to provide a satisfactory biomechanical framework within which our observations of strength may be interpreted.

1,1,1 DEFINITION

Because "strength" has many connotations it is necessary to rigorously define the term as used in the present work.

The Shorter Oxford English Dictionary (Third Edition) defines strength as "Power of action in body or limbs, ability to exert muscular force".

Kroemer (1970) in his review of the subject proposed that "strength is the maximal force muscles can exert isometrically in a single voluntary effort". Both the layman's and the physiologist's definitions are to some extent at variance with the usage of the term by a physicist or an engineer in which the strength of a material or a structure is the greatest load which may be imposed upon it without failure.

Early techniques for the measurement of strength involved a progressive loading until the subject released his grip on the measuring device (Hunsicker and Donelly, 1955). Contemporary practice is overwhelmingly in favour of the specification of an actively exerted rather than an actively resisted force. These may or may not be quantitatively equal, but the former is substantially easier to measure using one or other of a whole family of devices which sense the resulting mechanical distortion of an elastic element.

Kroemer's definition raises several points. Firstly it places paramount and perhaps undue emphasis on the action of muscles. Dempster (1958) identified a class of activity in which "limb and trunk muscles do not directly affect pull forces; instead they maintain joint postures which permit body weight to exert an effective moment." In Dempster's view there are many situations in which a subject's effective strength is dependent not upon the capacity of his muscles to develop forces, but upon the distribution of his body mass and his postural stability. Furthermore, the measurement of muscle force demands a quantitative knowledge of the internal geometry of the body which is not readily available. In many cases the quantity actually measured is a net torque about an articulation; the extent to which various muscles contribute to this torque is indeterminate.

In the light of these problems it is advantageous to exclude specific reference to muscles from our definitions, in which case, the term "isometric" becomes obscure and requires deletion. Kroemer suggests that it is necessary to restrict ourselves to the static steady state condition in which no work (force x displacement or change in kinetic or potential energy) is performed on the body or by the body. Tests of so called dynamic strength are often measurements of work, power or a combination of both. Although the present work deals principally with static steady state conditions a number of studies dealing with well defined dynamic conditions exist and will be discussed.

For present purposes it is useful to adopt a purely operational definition which makes no prior assumptions about the biological or mechanical factors which determine human capacities in tests of this kind. The following definition of strength is proposed.

THE STRENGTH OF AN ACTION IS THE MAXIMAL STEADY FORCE OR TORQUE WHICH AN INDIVIDUAL CAN VOLUNTARILY EXERT ON AN EXTERNAL TEST OBJECT UNDER GIVEN CONDITIONS.

1,1,2 Given this working definition of strength, it is appropriate to consider some typical strength tests. If the test is to measure strength as defined, the subject must maintain his maximum effort against a rigidly mounted measuring device for a period of several seconds.

Such a device is usually referred to as a dynamometer, and it may be solely mechanical in its operation or may include electronic transducer elements. The nature of the dynamometer mounting depends entirely on the action to be tested. In a test of the strength of flexion or extension of a limb articulation, the force sensing device may be attached to a sling which is placed around the distal segment. Alternately the sensor may be linked to an axle which is aligned with the axis of rotation of the joint so that torque is measured directly. In other tests the anatomical action involved is not specified but the subject is required to exert a force against a dynanometer mounted in such a way as to simulate a device of practical importance such as a vehicle control. Some factors influencing the results of such measurements are common to all tests; other factors are highly specific. These factors underlying the expression of human strength are numerous, and the available literature is vast, much of it being peripheral to the subject of this thesis. In the review of literature which follows topics, such as the consequences of specific pathological processes and the long term effects of training have been deliberately excluded.

1,2 REVIEW OF LITERATURE

1,2,1 HISTORICAL BACKGROUND

Speculation concerning the process of muscular contraction and the mechanism of human and animal locomotion extends into antiquity. Early ideas concerning these matters have been reviewed by Bastholm (1950). The notebooks of Leonardo Da Vinci (1452-1519) contain abundant references to questions we should now consider to be biomechanical in nature (see Sections 3,1; 5,1). The earliest quantitative studies of biomechanics were those of the seventeenth century "iatrophysicists" of whom the foremost was Giovanni Alfonso Borelli (1608-1679) a mathematician, physicist and anatomist who was a student of Galileo and a close colleague of Toricelli and Malphigi. Borelli's magnum opus, entitled "De Motu Animalium" was published posthumously in 1680/81. It contains studies of the mechanics of individual articulations, body equilibrium and locomotion. The frontispiece of the present work is taken from a copy of the 1685 edition of De Motu Animalium kindly loaned by the Library of Queen's University Belfast.

De la Hire (1699) compared the strengths of men and beasts of burden in lifting and carrying tasks and discussed the muscles involved in these activities.

The first device specifically constructed for the measurement of strength is generally attributed to Regnier, who in 1807 constructed a dynanometer based on a deformable ellipse of spring steel. Monod (1972) stated that this device was designed at the request of the naturalist Buffon, who wished to study the effects of aging on strength, but as the latter died in 1785 the connection seems tenuous.

Since then, many generations of devices have been based on spring, or more rarely, hydraulic linkages, but the post-war application of the electrical resistance strain gauge has almost rendered them obsolete. Hunsicker and Donelly (1955) have reviewed the history of strength measuring instruments in detail.

1,2,2 BODY SUPPORT AND STABILITY.

In accordance with Newton's Third Law of Motion, any force exerted in a strength test must be opposed by an equal and opposite reactive force. The nature of the interfaces between the test subject and his mechanical environment may limit the reactive forces which can exist, and many studies have demonstrated that performance may be limited in this way. Empirical studies of the effects of backrest position on the forces which could be exerted on pedals were conducted by Hugh-Jones (1947) and Rees and Graham (1952). Caldwell (1962) investigated the effect of backrest position on the performance of an horizontal pulling task. Kroemer (1961, 1969, 1974) has studied intensively the pulling forces which a man may exert on a variety of "working positions" and usefully tabulated the coefficients of friction between shoes and floors of various kinds (1964).

The works of the late W.T. Dempster (Dempster, 1955; Dempster, 1958; Gaughran and Dempster, 1956) are of considerable importance to the theoretical basis of biomechanics.In this series of publications, Dempster laid the foundations of a general approach to the study of body mechanics by means of free-body diagrams. The actions of pushing and pulling in the sagittal plane, from standing and sitting positions, were analysed both experimentally and by means of free-body diagrams in a graphical form. Dempster referred to these activities as ones in which the body formed a "closed chain of forces" and stated that performance in such strength tests is determined not by the capacity of any group of muscles, but by the dead-weight of the body and by the leverage at which this weight acts. The muscles act to lock the joints into appropriate postures - a function which Dempster likens to tightening the nuts on a drafting board mannikin. Many of Dempster's results have been subsequently confirmed by Ayoub and McDaniel (1974).

Whitney (1958) adopted a similar approach to the above authors in a study of the vertical lifting action, and concluded that "...... the magnitude of the lifting force was largely determined by the magnitude of the force moment which the body mass could exert to counterbalance the reaction of the lifting force on the body." Foot-reaction forces were measured using a force platform and the results were analysed by algebraic rather than graphical means.

1,2,3 THE MECHANICAL PROPERTIES OF SKELETAL MUSCLE

The mechanical properties of an isolated muscle or muscle fibre are frequently described or analysed in terms of a three component analogue of the type shown in fig. 1,1 (a), (Hill, 1922; Hill, 1938; Pringle, 1960; Grieve, 1975). The contractile component (c.c.) (i.e. the"motor" which in its active state, consumes chemical energy and actively exerts tension or does mechanical work) is coupled to its external attachments through an undamped elastic element known as the series compliance (s.c.). A further undamped elastic element, the parallel compliance (p.c.), also runs between these same points of attachment, but acts independently of the contractile component, the tension in it being determined solely by the length of the muscle.

These three elements are abstractions in the sense that they can be clearly defined physiologically but cannot be located in specific anatomical structures. Hence, the contractile component is located within the acto-myosin arrays, but the serial compliance is, by definition, the functional equivalent of all the elastic elements in series with the contractile sites. These elastic elements include inactive portions of the acto-myosin arrays, the tendon and its associated structures and, in most experiments, deformable elements in the

Figure 1,1

The Mechanical Properties of Muscle.

(a) Three - Component Model

c.c. is contractile component

s.c. is serial compliance

p.c. is parallel compliance

1,2,3

(b) Length - Tension Curve

Lo is "resting length"

PR is probable physiological range

1,2,3,1

(c) Force - Velocity Curve

Vo is unloaded velocity

Po is isometric tension

1,2,3,4 (a)

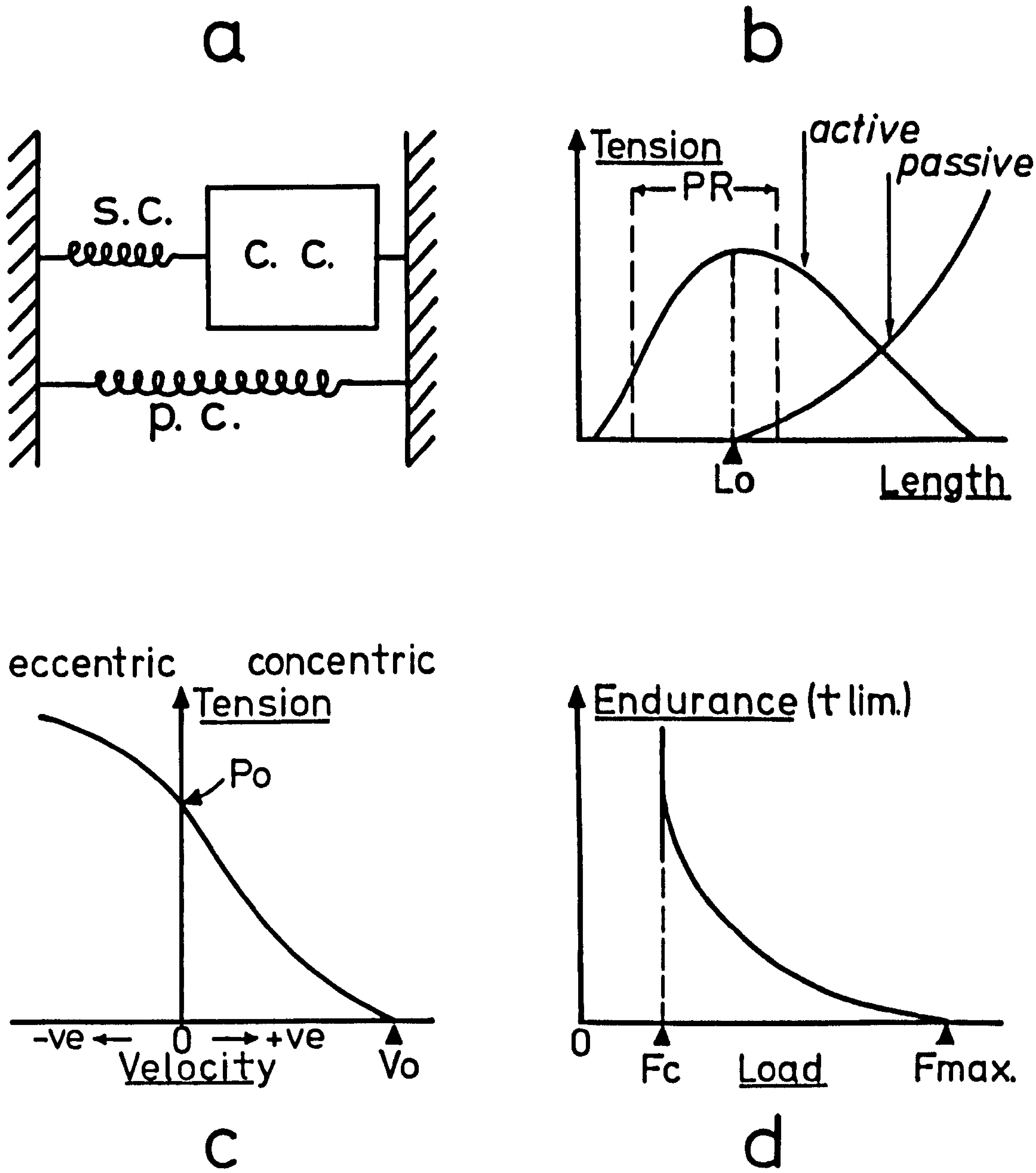
(d) Strength - Endurance Curve

Fmax. is maximal voluntary isometric strength

Fc is critical force

1,2,3,5

Figure 1,1


a
b
s.c.
c. c.
p. c.
Tension
PR
active
passive
Lo
Length
eccentric
concentric
Tension
Po
-ve
0
+ve
Velocity
Vo
c
Endurance (t lim.)
0
Fc
Load
Fmax.
d

linkage to the recording apparatus. Passive structures within the muscle, such as the sarcolemmal sheaths of the fibres, as well as the endomysium, epimysium and perimysium contribute to the parallel compliance.

The properties of these three elements determine the length-tension characteristic of the passive and fully active muscle. The remaining mechanical characteristics of the muscle are specifically associated with the contractile element.

1,2,3,1 THE LENGTH-TENSION CURVE

(a) Studies of Isolated Muscles and Muscle Fibres

The first description of the length-tension curve of skeletal muscle was that of Blix (1892) who measured the tension in frog muscles during isometric contraction elicited by tetanic stimulation at various muscle lengths. This relationship was further investigated by Ramsay and Street (1940) whose work on single frog fibres is usually considered definitive.

Fig. 1,1 (b) shows the length-tension relationship of an idealised muscle. All lengths are defined with respect to a standard condition usually referred to as the "Resting length" (Lo). At lengths greater than Lo there exists a passive tension due to the straining of the muscle compliances (fig. 1,1 (a)). The stiffness of these compliances increases dramatically with muscle length. When the muscle is in its fully active state, e.g. as the result of a maximal tetanic train of stimuli via the motor nerve, there exists an additional active tension which is greatest at or near the resting length and declines when the muscle is either longer or shorter than this. Certain small disparities exist in published accounts of this relationship. Ramsay and Street (1940) define Lo as "the length at which the muscle fibre developed maximum tension which corresponds to the length at which it was held just taut horizontally in Ringer's solution". Aubert _et al_ (1951) define their Lo (of the frog sartorius muscle) as the length measured in the body when the legs are in full extension and abduction. It must also be noted that some authorities consider the passive length-tension relationship to be a function of the resting condition of the contractile mechanism itself. (Buchthal and Kaiser, 1949; Cassella, 1950)

Gordon, Huxley and Julian (1966) measured the active tension of single fibres of frog muscle and measured muscle length in terms of sarcomere spacing. They interpreted their findings in terms of the now generally accepted sliding filament theory of muscular contraction. Rack and Westbury (1969) studied intact cat soleus muscle and found a relationship between sarcomere spacing and tension which differed from that of Gordon, Huxley and Julian (1966) in being substantially shifted to the right. It is not clear whether these disparities are attributable to difference in dimensions of the molecular structure of the actomyosin complex in mammalian and amphibian types or to different mechanical conditions in intact muscles and single fibres.

Whilst such questions are of considerable theoretical importance, knowledge of the length-tension relation of isolated fibres of muscles is of limited practical value in the absence of a clear indication of which portion of the curve represents the physiological range. Beck (1921) asserted that the gastrocnemius muscle of the frog operated always within the narrow portion of the curve ascending to the peak. Evidence in the papers cited above suggests that the "rest length" is close to the maximum length of the physiological range.

Hill (1970) defines the rest length of the sartorius of the toad or frog as "the distance between the points where the muscle joins its pelvic and tibial tendons when the legs are stretched out in line perpendicular to the body". Barfed (1973) published the length-tension curve of the triceps surae muscle of a rat and included the physiological limits of the muscle length.

Alder, Crawford and Edwards (1958) showed that the physiological range of the tibialis anterior muscle of the rabbit extended from a maximum length which was slightly greater than that at which greatest isometric tension was developed, to a minimum length which was approximately 70% of the maximum. In this study, one hind foot in each of a group of young rabbits was immobilised in a position of dorsiflexion for a period of 4 to 25 weeks. This procedure resulted in a reduction in the longitudinal growth of the tibialis anterior muscle and a length tension curve in which maximum active isometric tensions were exerted at lengths considerably less than those which were optimal for the control muscles of the opposite limbs. Passive tensions in the experimental muscles commenced at a length equivalent to that at which the muscle was immobilised. Tabary _et al_ (1972) immobilised the hind limbs of adult cats for a period of four weeks using plaster casts. Soleus muscles which had been immobilised in a lengthened position were found to have 20% more sarcomeres in each fibre than controls, whereas muscles immobilised in the shortened position had 40% less. Immobilisation in the lengthened position did not alter the passive length tension curve of the muscle whereas immobilisation in the shortened position reduced the extensibility of the muscle. Muscles which had been immobilised in their shortened position and then released for a further four weeks were found to have normal passive length-tension curves and numbers of sarcomeres. Goldspink _et al_ (1974) repeated the experiments of Tabary _et al_ (1972) and found that denervation of the test muscles had no effect on the observed results.

Grieve (1975) has pointed out that the measured length-tension curve is modified in shape by the relationship between the magnitude of the series compliance and the isometric strength of the contractile apparatus. This phenomenon must be of considerable significance in those muscles in the body where relatively short fibres act on relatively long tendons. Grieve (1975) deduced that the effect of such a compliance would be to increase the muscle length at which greatest tension was generated.

(b) Cadaveric Studies of Muscle Excursion

Weber (1851) made a cadaveric study of the length changes of a large number of limb muscles as a function of joint posture. The functional excursion of these muscles (i.e. the total distance which a muscle may shorten, expressed as a percentage of its maximum physiological length) ranged from 34% for the innermost fibres of adductor longus to 89% for semimembranosus. It was subsequently hypothesised that these differences were due to post-mortem changes and that there was a constant excursion of 50% during life, which was the result of the orderly growth of muscle fibres to meet functional demands. Fick (1910) made the following statement:-

"Muscle fibres grow lengthwise until their length in extended state, (that is in the joint position opposite to that resulting from the appropriate muscle contraction) is twice as great as their capacity to shorten on the joint apparatus".

(Fick, R. 1910: p.300, translated from the original German).

This relationship became generally known as the Weber-Fick Law. It was further deduced that if the attachments of a muscle (typically one acting over two or more joints) were approximated to less than 50% of their maximum separation, then the muscle would enter a state of <u>active insufficiency</u> in which it was flaccid and no longer able to exert tension. (Bryce, 1923).

Wheeler-Haines (1934) stated that the fibres of most muscles in the human body are capable of shortening by 57% of their maximal length. These muscles he called "muscles of full action". Exceptions to this rule were certain limb muscles which he concluded must be slack in their shortest position. This latter class, which he named "muscles of short action" included the long head of biceps femoris, semimembranosus, rectus femoris, gastrocnemius, plantaris and soleus.

(c) <u>Studies of Cinetised Muscles</u>

The small but consistent interest in the surgical and prosthetic techniques of cineplasty which has existed from the time of Vaughetti (1912) to the present day (Anon, 1975) has shed some light on the human length-tension curve. Inman and Ralston (1954) present the length-tension curves of human forearm flexors and pectoralis muscles. These results are not dissimilar from those typically found in animal studies showing both ascending and descending portions. Blaschke <u>et al</u> (1952) in a very thorough paper show the results of measurements made on five biceps brachii muscles. "The average maximum isometric curve is seen to be virtually linear throughout the entire range from extreme shortening to the tolerable extent of stretching", i.e. the ascending portion only is present. The authors recorded the fact that the total muscle excursions of their experimental subjects were similar to the cadaveric excursions published by Weber for the same muscle, from which it can be argued that the whole physiological range of the muscle was tested.

1,2,3,2 Cross-Sectional Area and the Absolute Strength of Muscle.

It is often considered (and even taken as axiomatic) that the maximum contractile force of a muscle is proportional to its effective cross-sectional area. It is necessary to distinguish the anatomical cross-section, which is a geometric quantity of little functional significance from the physiological cross-sectional area (p.c.s.a.) which may be defined as the total area of a set of sections cut perpendicular to the direction of the muscle fibres such that no fibre is included in two sections and no fibre is excluded. It is a relatively simple piece of geometry to demonstrate that for any muscle this area is equal to the volume divided by the fibre length. (Weber, 18~~51~~49). (If we wish to be truly rigorous it is necessary to add a rider that all measurements are made when the muscle fibres are in a standard condition, e.g. at optimum length). Tables of p.c.s.a. of various human muscles have been given by many authors, including Weber (18~~49~~51), Fick (1911) and Schumacher and Wolff (1966).

If the proposition commencing the above paragraph is correct, it should be possible to determine a constant defined by the equation.

$$T = \rho \, . \, A$$

where T is the maximum muscle tension at the optimum length, A is the physiological cross-sectional area

and

ρ is the coefficient of absolute muscle strength.

Many authors have attempted to determine experimentally a value for this constant. This has either been done by direct measurement of isolated muscle preparations, or indirectly by comparing the voluntary isometric strengths of living subjects with the muscle cross-sections of cadavers. Table 1,1 shows a selection of values from the literature. In this table, S.I. units ($N\ m^{-2}$) have been placed alongside the more familiar $Kgf.\ cm.^{-2}$ The direct studies can be criticised on the grounds that the muscles used have been subjected to surgical insult, whereas the indirect measurements rely heavily on unproven assumptions. The available data presents a bewildering choice for the anatomist wishing to make functional predictions from morphological data. It is well known that muscles vary in their dynamic properties by virtue of their metabolic characteristics. This well researched field has been recently reviewed by Close (1972). The possibility that their static properties may vary in a similar way must be recognised. A further source of variation is differences in the proportions of a muscle cross-section which may be attributed to acto-myosin arrays, rather than connective tissue, vascular tissue or other passive components. Yamada (1970) presented the stress-strain curve, tensile strength and breaking strain for post-mortem specimens of several different human muscles. (The stress-strain curve of a dead muscle is not necessarily the same as the passive length-tension curve of a live one). The muscles studied show substantial differences in their mechanical properties, the most likely explanation for which is the varying amounts of connective tissue present in the bundles of muscle fibres which were tested. Differences of this nature may be responsible for some of the variation tabulated in table 1,1.

TABLE 1,1 Coefficients of Absolute Muscle Strength (ρ)

Source	Kgf. cm.$^{-2}$	N m^{-2} x 10^5	Comments
Haughton, 1873	7.2	7.06	Methods unknown
Fick, 1910	10.0	9.81	" "
Von Recklinhausen, 1920	3.6	3.53	" "
Fenn & Marsh, 1935	1.225	1.20	Cat quadriceps) calculated by Inman and Ralston (1954) on the basis of published data concerning length and weight of muscles
	6.356	6.24	Cat gastrocnemius)
Ramsay & Street, 1940	4.42 max.	4.34	Isolated muscle fibres "as a rule from the semi-tendinosus muscles of frogs".
	3.53 mean	3.46	
	2.49 min.	2.44	
Haxton, 1944	3.6	3.53	Human ankle flexors - comparison between cadaveric p.c.s.a. and voluntary strength.
Inman & Ralston, 1954	0.74 - 1.7	0.73 - 1.66	Gastrocnemius and anterior tibial muscles of the rat. pectoralis major) biceps brachii) triceps brachii) strengths of muscles of cineplastic amputees were measured directly and compared with the p.c.s.a. data of Weber (1849)

1,2,3,3, The Angle Torque Curve

Although the length-tension curves of isolated muscles are well documented, the application of the available data to the living case presents major problems. Consider two rigid limb segments acted upon by a single flexor muscle and articulating at a single hinge joint (fig. 1,2). Ignoring the effects of gravity, a simple equation may be written:

$$\tau_o = T_m . \quad lT_m \qquad (1)$$

where τ_o is the net torque about the joint,

T_m is the tension in the muscle,

and lT_m is the perpendicular distance between the line of action, the muscle and the centre of rotation of the joint.

It is τ_o that is the quantity which may be most readily measured or calculated in any strength test or action against a load. In a common kind of strength test in which a man exerts a force on a sling attached to the more distal of the two segments, then

$$\tau_o = F \quad . \quad lF \qquad (2)$$

where F is the measured force (load),

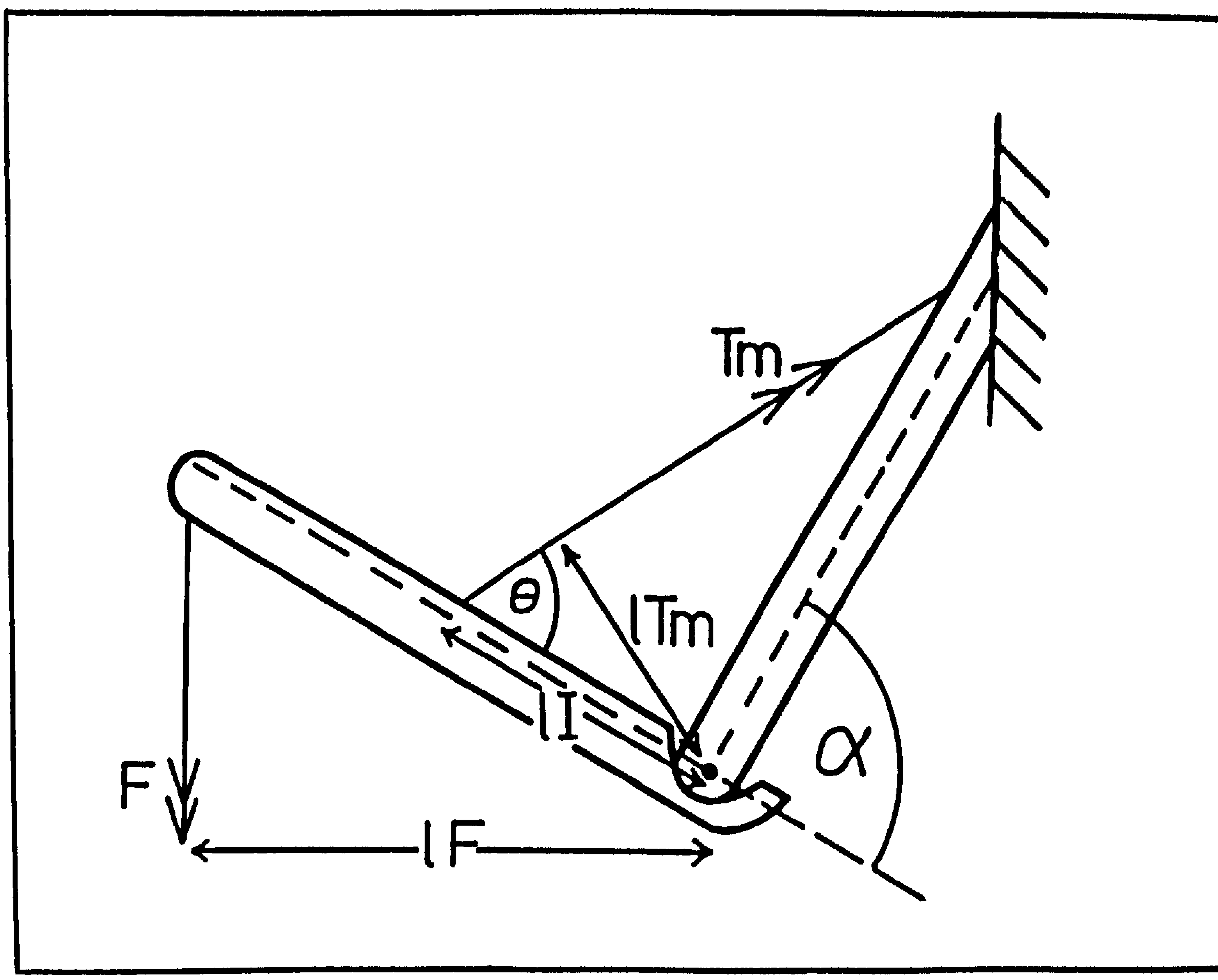
and lF is the perpendicular distance from the line of action of the load to the centre of rotation of the joint. Borelli (1680), as the frontispiece shows, was conversant with such a relationship. Basmajian ~~(1970)~~, in a widely used textbook draws the relationship incorrectly.

Figure 1,2

Mechanics of an Idealised Articulation
acted upon by a single flexor muscle.

Tm	is	tension in muscle
F	is	external load
lF	is	leverage of load
lTm	is	leverage of muscle
lI	is	distance from insertion of muscle to the centre of rotation of the joint.

Figure 1,2



An alternative expression pursued for example by Steindler (1955), and MacConaill and Basmajian (1969) is of the form.

$$\mathcal{T}_o = T_m .\ lI.\ \sin\theta \qquad (3)$$

where lI is the distance from the centre of rotation of the joint to the point of insertion of the muscle, and

θ is the angle at which the muscle inserts on to the bone of the segment. It does not seem likely that there are many situations in which θ is a measurable quantity and so such a relationship may be ignored for practical purposes.

If the posture of the limb is altered (i.e. angle α is changed) then both the length of the muscle and the leverage vary, and in a test of maximal strength

$$\mathcal{T}_o = \phi_1\ (\alpha) .\ \phi_2\ (\alpha) \qquad (4)$$

where $\mathcal{T}_o$ is the maximal torque, and

ϕ_1 and ϕ_2 are functions which cannot be defined by theory but require experimental determination and may well be unique for every muscle in the body. (As a later section 2,3,3) will show, ϕ_1 and ϕ_2 are themselves inter-related.)

Such speculations are of limited practical value for at least two reasons.

i) There are only a few unusual human actions which are performed by a single muscle. Several tensions each acting with a different leverage must therefore be summated, , and the difficulty of the analysis and prediction are multiplied.

ii) The functions and variables described are difficult if not impossible to measure either in the living or in the cadaver.

In contrast, it is a relatively simple matter to make measurements of maximal voluntary torques in living subjects, but many investigators have fallen into error in so doing.

Many authors have plotted "torque curves" in which the vertical axis is labelled in units of force. Brunnstrom (1966) quotes the work of Bethe and Franke in this way. Williams and Stutzman (1959) and Jensen _et al_ (1971) both correctly discuss the nature of torque measurement and then continue by plotting their extremely comprehensive data in pounds, without description of the leverages involved. Herman and Bragin (1967) on the other hand quote values for muscle tension in inch-pounds. The data of Clarke and Bailey (1950), Clarke _et al_ (1950), Clarke (1956) and Elkins _et al_ (1951) and Smith (1968) go as far as stating that measurements made with the transducer located at different positions along the axis of the limb show different readings. In the absence of any recorded information concerning the distance of the transducer from the fulcrum the application of the above data is severely limited. Campney and Wehr (1965) measured force at different joint angles in tests of shoulder extension and knee flexion. Data from 42 subjects was subjected to an analysis of variance to determine whether the strength variation throughout the range was significant. As no reference was made to the distance from the joint centre to the testing cable a major source of variation was omitted from the analysis.

There have subsequently been a number of studies in which the authors do clearly distinguish between force and torque. Darcus (1951) and Salter and Darcus (1952) studied the actions of pronation and supination. Provins and Salter (1957) studied elbow flexion and extension, Lindahl et al (1969) studied knee extension and Murray and Sepic (1968) studied abduction and adduction of the hip. Exceptionally thorough studies were carried out by Olson, Smidt and Johnson (1972) on the hip abductors and by Smidt (1973) on the knee flexors and extensors. In both investigations maximal voluntary torques were measured in a variety of postures throughout the range of the joints under isometric conditions and also during concentric and eccectric movements at a constant angular velocity of 13 degrees per second. The muscle moment arms were estimated from radiographs and the tensions in the muscles themselves were estimated.

The above studies have not considered in detail the role of passive elastic structures (whether or not these be located within the muscles) in the measured torques. Smith (1956, 1957) studied the passive torques which existed in the absence of muscular activity about the knee and ankle joints. In the first case his subjects were under general anaesthesia whereas in the second a large sphygmomanometer cuff was inflated around the thigh to 200 mm. of mercury until "sensory and motor paralysis below the cuff were complete".

Wright and others (Wright and Johns, 1960; Wright and Johns, 1961; Johns and Wright, 1962; Johns and Wright, 1964; Goddard et al, 1968; Wright et al, 1969; Wright et al, 1971; Wright, 1973) described the passive angle-torque characteristics of human metacarpo-phalangeal and knee joints during sinusoidal cycles of angular displacement. Parameters of the angle-torque hysteresis loop recorded for different amplitudes and frequencies of displacement allowed the authors to calculate elastic, viscous, plastic, frictional and inertial elements in the resistance to movement. Elastic and plastic elements contributed the greater part of the resistance under the conditions tested. Experiments on the wrist joints of cats (Johns and Wright, 1962) enabled the contributions of various tissues to be estimated.

Although a great deal of research has been done on the angle-torque relationship, some notable gaps exist. To the author's knowledge no study has been made in which active and passive torques are separately recorded and compared. Furthermore, with the exception of the work of the Oxford team of the fifties,(Darcus, 1951; Salter and Darcus, 1952; Provins and Salter, 1955), little attention has been paid to the possible effects of two-joint muscles.

1,2,3,4 DYNAMIC PROPERTIES OF MUSCLE

Although measurements made under dynamic conditions cannot be considered strength measurements within the limits of the narrow definition proposed above (1,1,1) it is of interest to devote some attention to the work on which contemporary concepts of dynamic strength are based.

(a) The Force-velocity Curve

The capacity of a muscle to exert tension while changing length, or alternately to change length against a load are described by the relationship generally known as the force-velocity curve which would perhaps be more accurately called the tension velocity curve (Fig. 1,1 c) Fenn and Marsh (1935) working with cats and frogs and Hill (1938) working with frogs dealt with conditions in which fully activated isolated muscles shortened under load. Katz (1939) extended their description to deal with the case in which an active muscle is loaded to a tension which is greater than that of a maximal isometric contraction (Po) and therefore lengthens. It is customary to refer to the shortening and lengthening portions of the curve as concentric and eccentric conditions respectively. Under concentric conditions tension is less than isometric (Po) and under eccentric conditions it is greater. The fully active unloaded muscle shortens at a maximal velocity (Vo).

Hill (1938) proposed that the force-velocity relationship be described by the equation

$$(P + a)\ (V + b) = (Po + a)b = a \text{ constant.}$$

where P is the tension in the muscle,

V is the velocity of shortening

Po is the isometric tension

and a and b are constants.

Since this equation can be written in the form

$$\left(\frac{P}{a} + 1\right) \cdot \left(\frac{V}{b} + 1\right) = \frac{Po}{a} + 1$$

Then the shape of the curve may be fully described by the ratio a/Po.

Fenn (1938) preferred to describe the relationship by the equation

$$P = Po.e^{-av} - KV$$

in which a and K are constants.

Both of these equations apply to concentric conditions alone, and are inaccurate in their prediction of eccentric conditions. In the words of Wilkie (1950) "It is the shape of the curve which is important, while the equation used to describe that shape is largely a matter of convenience."

The condition Po can be equivalent to any point on the length-tension curve, and in the practical measurement of the force-velocity curve all readings must be taken as the muscle passes through a constant portion of its range of lengths. Gordon, Huxley and Julian (1966) showed that for all lengths greater than Lo, Vo remained constant. At lengths less than Lo, Vo decreased with Po.

The power output (rate of work) of the muscle in concentric activities is given by the product of the force and velocity (P.V). This is zero at both Po and Vo. The load at which maximum power may be developed as Hill (1938) has shown may be calculated by differentiating P.V with respect to P and setting d (P.V)/dP = o. This leaves the result that maximal power output may be achieved in the condition

$$\frac{P}{Po} = \frac{a}{Po}\left(\sqrt{1 + \frac{Po}{a}} - 1\right)$$

Inman and Ralston (1954) describe the force-velocity curve of maximal voluntary efforts of the pectoralis major muscle of a cineplastic amputee. Wilkie (1950) conducted a very detailed study of the force-velocity relationship of a maximal voluntary elbow flexion. It was found that an equation of Hill's type could be fitted to the experimental results provided that corrections were first made for the inertia of the forearm and hand. Values of a/Po ranged between 0.2 and 0.48 suggesting optimal loadings of 0.26 Po to 0.36 Po. Cavanagh and Grieve (1970) extended the work of Wilkie by analysing the partitioning of the mechanical energy released during a maximal effort of elbow flexion into elastic and kinetic energy stored and useful work done.

Although it is possible to represent the force-velocity characteristics of muscle by adding a non-Newtonian viscous element ("dashpot") to the muscle model (see Fig. 1,1), the work of Fenn (1924) showed that this is inappropriate. It is a general property of any purely visco-elastic system that the energy liberated on shortening a fixed amount is always the same. Fenn showed that the total energy released by a muscle (i.e. the sum of the mechanical work done and the heat dissipated) is not constant for different rates of contraction. It is commonly concluded (Carlson and Wilkie, 1974) that the force-velocity curve is a result of the kinetics of the energy yielding processes of the contractile mechanism. Close (1972) has reviewed substantial evidence showing that the a/Po parameter of animal muscles is closely related to the histochemical properties of the fibres or motor units in the muscle, especially the level of myosin ATPase activity. Thorstensson et al (1976) investigated the force-velocity relationship of knee extension in human subjects. A correlation coefficient of 0.5 ($p < 0.05$) was found between the maximum angular velocity of knee extension and the percentage of fibres in biopsy specimens of the vastus lateralis muscle which revealed ATPase activity by histochemical methods.

Knowledge of the force-velocity curve has been used in recent years to considerably refine our concepts of dynamic strength. Doss and Karpovitch (1965) studied the angle-torque relationship of isometric, concentric and eccentric efforts of the flexor muscles. At all joint angles eccentric strength was greater than isometric which was greater than concentric, but the angular velocity was not closely controlled. Olson, Smidt and Johnson (1972) and Smidt (1973) plotted angle-torque relations for the abductors of the hip and for the flexors and extensors of the knee respectively. Use of an electric motor resistance allowed the measurement of torque at a controlled angular velocity of 13 degrees per second in both concentric and eccentric conditions. In these studies the eccentric torques occasionally fell below isometric level and the concentric torques were consistently lower.

Asmussen et al (1965) performed a detailed study of a horizontal pulling task. Subjects were restrained by a sling around the trunk and pulled with one hand at shoulder height against a variable hydraulic or electric motor resistance. The results of this study gave a force velocity curve of exactly the type anticipated from the animal studies. Consistent relations also existed in any arm position between the force and the rate of lengthening or shortening. Eccentric efforts at a velocity of 60% arm length per second registered force as high as 140% of the isometric maximum.

Komi (1973) has subsequently conducted similar studies for the elbow flexor and recorded forces in the order of 120% of the isometric maximum during eccentric activities.

Grieve and Arnott (1969) studied axial rotation of the trunk and by indirect means calculated an equivalent torque- angular velocity curve for this complex multi-link activity. Their results were consistent with the existence of torques of over 200% of isometric maximum during the eccentric phase of this action.

"Break tests" of muscle strength, in which the subject resists an externally applied force have been found to give higher readings than conventional isometric tests. Bethe (1925) found that subjects in a test of grip strength could resist an externally applied force 25% greater than they could actively exert. Rasch and Pierson (1960) in a study of elbow flexion failed to find a statistically significant difference between isometric strength and "break strength". The action of resisting an external force may allow muscles to operate under eccentric conditions and this possibly accounts for such differences as have been observed.

(b) History Dependence

It is a matter of common experience that most kinds of impulsive or forceful muscular activities are performed most effectively when the phase of muscle shortening is preceded by a "back-swing" or "wind-up" movement in which the active muscle is stretched by its antagonist or by an external force.

Abbott and Aubert (1952), using isolated toad, frog and dogfish muscles, demonstrated that if a muscle was stretched during a tetanic stimulus its subsequent tension whilst at constant length was greater than the isometric tension for that same length. (The reverse occurred if it was allowed to shorten). Edman et al (1976) subjected single fibres of the frog semitendinous muscle to quick stretches whilst tetanically stimulated. The tension following this procedure was greater than isometric and the difference was most pronounced when the muscle fibre was at a length greater than the optimum.

Cavagna et al (1965, 1968) studied isolated toad sartorius and frog gastrocnemius muscles as well as voluntary contractions of the elbow flexors in man. The work done by a muscle which shortened after previously being stretched whilst tetanically active (W') was greater than the work done by the same muscle when shortening through the same range of length subsequent to an isometric contraction (W). The ratio W′/W increased with both speed of stretch and muscle length to a maximum measured value of 2.5. Tensions in the shortening phase remained higher than isometric during the early part of the movement.

Cavagna et al (1971) recorded vertical force at the feet during a vertical jumping task. The work done in the jump was greater when the jump was preceded by a rapid full flexion, than when the jump commenced from a static position with the knees flexed. Asmussen and Sørensen (1971), using the apparatus described by Asmussen et al (1965) showed that the work done in the early stages of a concentric contraction is greater if that contraction is preceded by an eccentric exertion than if it is preceded by an isometric exertion which in turn is greater than if the concentric contraction is initiated from rest.

Cavagna et al (1968) calculated that the quantity of work gained by pre-stretching a muscle (i.e. W′ - W) was greater than the quantity of energy which could be stored by the deformation of the series compliance. It was therefore concluded that the act of stretching must in some way enable the contractile component itself to achieve a higher power output during the shortening phase. The suggestion has been made, (Hill and Howarth, 1969; Hill, 1960) that the work absorbed in stretching an active muscle can reverse the chemical processes by which potential energy is liberated when a muscle becomes active. Cavagna et al (1968) concluded that the kinetics of the mechano-chemical coupling reactions are such that the reversal of chemical processes cannot account for the observed phenomena of their experiments.

The evidence of the experiments cited in this section shows that the action of muscle is history dependent in a way that cannot be predicted from a consideration of the length-tension and force velocity relations or from the simple models of muscle currently in use.

1,2,3,5 THE STRENGTH-ENDURANCE RELATIONSHIP

The literature concerning muscular fatigue and its associated phenomena is too extensive to review comprehensively in the present context. Quantitative studies date back at least to the time of Mosso (1915)

The most concise descriptions of decrement of performance in a sustained effort are in the form of strength-endurance curves of the type shown in fig.1,1(d) in which the limit time (t lim), or maximum voluntary duration, of a continuous static exertion is plotted as a function of the strength of that exertion (F) expressed as a percentage of the isometric maximum(F max) as measured in the unfatigued state. This curve has been plotted by Rohmert (1960) for "arm, leg and trunk muscles", by Caldwell (1963) for an upper limb pulling action and by Monod and Scherrer (1965) for biceps brachii, triceps brachii, the middle finger flexor and quadriceps femoris. The curves plotted by these three authorities are almost identical. The limit time tends to infinity at a critical force (Fc) of 15-20% F max and tends to Zero at F max.

Monod and Scherrer (1965) describe the strength-endurance relationship by the equation:

$$t\ lim = \frac{K}{\left(\frac{F}{F\ max} - Fc\right)^{n}}$$

where K is a constant in the order of 2.5, n is a constant in the order of 2.4 and the remaining quantities are defined above.

Pottier _et al_ (1969) extended this relationship to deal with limit times for intermittent static efforts. Wilkie (1960) discussed the maximum power output of individuals engaged in extended periods of muscular work of a variety of tyres and Monod and Sherrer (1965) and Monod (1972) described similar relationships for the power output of single muscle groups.

It has been shown (Barcroft and Millen, 1939; Barcroft _et al_, 1963; Humphreys and Lind, 1963) that during isometric efforts, occlusion of the muscle blood flow commences at 20% of the maximum force and is complete by 70%. The lower figure is strikingly similar to that for the critical strength (Fc) and it seems probable that vascular factors are important but by no means exclusive determinants of the strength-endurance relationship.

The closeness of agreement in this relationship for different muscle groups (Rohmert, 1960; Caldwell, 1963; Monod and Sherrer, 1965) strongly suggests that it might describe a general property of all muscle groups in the body. Subsequent investigations have demonstrated the existence of exceptions. Jorgensen (1970) plotted a strength-endurance curve for trunk extensors and found that the endurance level at any proportional load was around 80% greater than would be anticipated from the results of the authors cited above. Fc was found to be

0.25 F max. Molbech and Johansen (1973) tested the endurance of the muscle groups responsible for dorsiflexion and plantarflexion at a force of 0.5 F max and found that the endurance of the latter was greater than that of the former which in turn was greater than would be predicted from the standard curve. It is possible that such differences as exist between muscle groups may be attributed to the differing histochemical properties of their motor unit populations, or to differences in vascularity. Hulten et al (1975) measured isometric endurance at a level of 0.5 F max in a test of lower limb strength involving "mainly the thigh muscles". A correlation coefficient of 0.7 ($p < 0.01$) was found between endurance time and the percentage of fibres in biopsy specimens of the vastus lateralis muscle which did not show ATPase activity histochemically. Thorstensson and Karlsson (1976) obtained comparable results in a test of endurance which required repeated fast extensions of the knee.

Although all workers in the field are in agreement that stronger subjects are capable of maintaining a given absolute load for longer than weaker subjects (a finding which is implicit in the strength-duration curve) there is some controversy concerning limit times for proportional loads (i.e. loads expressed as a percentage of F max) in subjects with different absolute strength. Tuttle, Janey and Salzano (1955) claim that "stronger individuals maintain a smaller proportion of their maximum back and leg strength than individuals with less initial strength". Caldwell (1963) tested a group of males and a group of females in an upper limb pulling task. He found that although the maximum strength of the males was much greater than that of the females, their strength-endurance curves were identical. Carlson (1969) tested elbow flexion at 100%, 80%, 70%, 60% and 50% maximum load and found that subjects who were stronger in the maximal test had greater endurance in the sub-maximal tests.

It would be of interest to elucidate the way in which the strength-endurance curve is modified by changes in the posture of the subject that are known to affect absolute strength (1,3,3). Caldwell (1964) studied the strength of a pulling action in 20 different postures together with the endurance of a load set to 80% of the strength of the weakest posture. A correlation of 0.76 ($p < 0.01$) existed between the mean strength and the mean endurance for each of the 20 postures. The present author has not located any studies which deal with this particular question in more detail.

Caution is called for in the interpretation of tests of the type described above, as several authors (Schwab, 1953, Ikai and Steinhaus, 1961) have demonstrated the importance of motivation in prolonged muscular activity.

1,2,4 ANTHROPOMETRIC AND DEMOGRAPHIC VARIABLES RELATED TO HUMAN STRENGTH

When strength measurements are made on any given subject population, it is common to find a variability in performance which is greater than the variability in other anthropometric measurements made on the same population. Taking for example the first thirty sets of strength data tabulated by Damon, Stoudt and MacFarland (1971) we find coefficients of variation ranging from 12.9% for the left handed grip strength of U.S. aircrew students measured by Clarke (1945), up to 51.5% for the force exerted on aircraft foot pedals in a specific pedal seat configuration by U.S. airforce males (Hertzberg, 1954). By contrast, U.S. airforce flying personnel show coefficients of variation of 3.5% and 12.9% for height and weight respectively (Hertzberg, Daniels and Churchill, 1950).

It is of considerable interest to ask ourselves what biological variables might relate to such differences in a given population.

1,2,4,1 - AGE

The effects of age on muscle strength have been widely investigated starting with Galton *et al* (1881) (1883), who found that the strength of the arms increased rapidly to 19 years, more slowly to 30 years and thereafter commenced to decline. These findings have been substantially confirmed by subsequent investigators. (Cathcart *et al*, 1935; Hettinger, 1961). Asmussen, Heebøll-Neilsen and Molbech (1959) studied increases in strength of children between the ages of 7 and 15 years and gave tables of standards for strength of various muscle groups as related to the child's height. Asmussen and Heebøll-Nielsen (1961,1963) measured the strength of 25 different actions in male subjects ranging from 15 to 65 years of age and female subjects ranging from 15 to 55 years. It was found that the strength of the hands and upper limbs reaches a plateau at 20 years and does not decline until 40 years, whereas the trunk and lower limb muscles have a peak of shorter duration at around 30 years. It was also shown that the ages of peak and decline were earlier in women than in men.

The majority of studies of this nature have been cross-sectional - i.e. cohorts of individuals of various ages were sampled at a given time. The changes therefore cannot be attributed to the ageing process *per se*, but are confounded with the effects of overall changes in the population (such as the well known world wide increase in stature) and the effect of differential survival. Longitudinal studies have been understandably scarce, but Damon and Stoudt (1963) citing unpublished data from a study of the alumni of Columbia College claimed that no major decrements in an individual's strength occurred between the ages of 19 and 56 years.

1,2,4,2 SEX

It has been widely stated that women are two-thirds as strong as men in any given task. Hettinger (1961) lists sources showing that this ratio ranged from 55% for the elbow flexors and extensors to as much as 80% for the hip flexors and extensors, the knee flexors and the muscles of mastication. The conclusion may be very reasonably drawn that these differences reflect the extent to which the various muscle groups are employed in everyday life, the groups with the biggest differences being employed in predominantly male activities. Hettinger (1961) claims that the strength differences between men and women may be accounted for by differences in muscular cross-sectional area. Morris (1948) however had presented data leading to the contrary conclusion. It is not easy to decide this question with any certainty as the measurement of p.c.s.a. in living subjects cannot be performed with any accuracy. It is also necessary to consider the subjects' motivation to do well in the test situation. As Ikai and Steinhaus (1961) point out, it is likely that, for social reasons, a woman will be more inhibited in her performance, and this confounding effect is necessarily present in any investigation of sex differences.

1,2,4,3 BODY BUILD

Studies which attempt to correlate strength with more easily measured anthropometric measures have been widespread, but at first sight inconclusive, since strength has been reported to be modestly correlated with many other variables but highly correlated with none.

Roberts et al (1959) using Royal Navy personnel measured grip strength and the strength of elbow flexion and extension together with stature, weight and the lengths and girths of the limbs. Most pairs of variables yielded significant correlation coefficents at the 1% or 5% levels. Partial correlation showed that with the elimination of stature and weight there remained a significant correlation between strength and limb circumference. Factor analysis by Hotelling's method indicated the presence of a "common size factor" which saturated all the variables measured to some extent.

Rasch and Pierson (1963) measured strength of elbow flexors and extensors, stature, weight, arm girth and arm volume. It was found that the correlation between strength and body weight was greater in trained than in untrained men, and that the girth and volume measures were too poorly correlated with performance to be of predictive value.

Laubach (1969) performed what is probably the most detailed study of this kind. Using 45 U.S.A.F.

personnel as subjects he measured the strength and range of flexion of both the hip and the trunk, together with stature, weight, body composition measures based on skinfolds and somatotypes on the Sheldon system. It was possible to generate predictive multiple regression equations which in the best case accounted for 56% of the variance in performance, but in the worst case only 26%.

Several authors (Laubach, 1969; Rasch _et al_ 1960; Everett and Sills, 1954) have confirmed the anticipated correlation between strength and the mesomorph component of the Sheldon system.

Some interest has centred around the relationship between weight, lifting performance and body weight (O'Carrol, 1968; Keeney, 1955; Lietzke, 1956). (Weight lifting cannot strictly speaking be considered a test of strength within the limits of the present working definition, but strength _per se_ must be an important pre-requisite for the weight lifting task). Examination of world or Olympic records in different weight classes reveals that performance/weight ratio decreases with increasing body weight. Lietzke (1956) fitting a least squares regression line to world records found that the best fitting equation was

$$P = 1.458 \; W^{0.6748}$$

where P is weight lifted and

W is body weight.

This equation is based on a highly atypical subject group which in no way reflects the population at large. Poulsen (1970) in untrained subjects found the very low correlations of 0.06 (males) and 0.28 (females) between the maximal weight which could be lifted in a box and body weight, whereas correlations of 0.72 (males) and 0.78 (females) were found between maximal weight which could be lifted and the isometric strength of trunk extension.

1,2,4,4 OCCUPATION

Cathcart _et al_ (1935) conducted a survey under the aegis of the Industrial Health Research Board which included the measurement of grip strength and vertical pulling strength. All subjects were male, and were divided according to age, geographical locality and occupation. "Black coated workers" (clerks, typists, commerce, finance and insurance) although taller and heavier than manual workers were markedly inferior in pulling strength. Differences in grip strength were small, but as the authors did not quote variances the statistical significance of these differences cannot be ascertained.

Nemethi (1952) tested the grip strength of 1,000 job applicants and found the greatest strength in semi-skilled workers, followed by skilled men, labourers, office workers and women in that order. Unfortunately, it is impossible to compare measures of strength made on different populations by different investigators. The most widely measured variable over the years has been "grip strength" and this is so sensitive to the precise nature of the measuring device as to cast doubt on the validity of any comparisons which might be made between investigators.

Considering the obvious theoretical and applied interest of such data, it is surprising that more studies have not been undertaken. A number of questions arise. Do physically strong individuals gravitate towards certain work in their youth, or is their strength acquired along with their skill? Is the supposed "lightening" of industry together with the more uniform environmental circumstances of modern society extinguishing differences which existed forty or twenty-five years ago?

1,2,5 PHYSIOLOGICAL, PSYCHOLOGICAL AND ENVIRONMENTAL FACTORS MODIFYING HUMAN STRENGTH.

1,2,5,1 VOLUNTARY AND STIMULATED ACTIONS

Whether a muscle can be voluntarily activated to create as much tension as it would produce during a fully fused supramaximal tetanic stimulation, is a question of considerable theoretical interest.

Merton (1954) investigated the human adductor pollicis muscle and found that maximal stimulation was achieved by a 50 Hz train to the ulnar nerve, and that this produced a mechanical response equal to that of a maximal voluntary effort. Furthermore, he found that a stimulus supplied during a maximal voluntary effort did not produce a measurable additional mechanical response. It was therefore concluded that the tetanic and voluntary conditions were entirely equivalent. Ikai *et al* (1967) repeated these experiments using the same muscle and the same stimulus conditions but found that electrical stimulation produced a stronger response. The conflict remains unresolved, but it is more a theoretical question than a practical one as maximal tetanic stimulation via the motor nerve is not possible for any but the smallest muscles. Grieve and Pheasant (1976) attempted to elicit a maximal contraction from triceps surae by stimulating the tibial nerve in the popliteal fossa, but abandoned the procedure as unacceptably painful.

1,2,5,2 PSYCHOLOGICAL INFLUENCES

Anecdotal evidence for the importance of motivation in the expression of strength is widespread in the often bizarre accounts of feats of strength performed under conditions of acute stress. For example, in the Guinness Book of Records (1974) we read that "It was reported that an hysterical 55 kg. woman, Mrs Maxwell Rogers, lifted one end of a 1632 kg. station wagon, which, after the collapsing of a jack, had fallen on top of her son at Tampa, Florida, USA, on 24th April 1960. She cracked several vertebrae".

The effects of motivation on skill (Fitts and Posner, 1967) and endurance (Schwab, 1953) have been widely studied, but the testing of strength has received little attention in this respect. Pierson and Rasch (1964) showed that knowledge of results gave a modest but statistically significant improvement in a test of elbow flexion when compared with the blind condition. Johnson and Nelson (1967) report a study in which subjects were trained in an isometric task over an eight week period. At the end of the training programme, such motivators as band music, enthusiastic spectators and a supposed contest with a rival p.e. college produced a very significant increase in performance.

Whitley and Elliot (1968) examine the proposition that the exertion of a maximal isometric effort is a skill which must in part be learned. The action studied was inward rotation of the leg - a deliberately novel task. A learning curve was demonstrated over time intervals in which muscular training in the physiological sense could not have occurred.

Roush (1951) and Ikai and Steinhaus (1961) both found that hypnosis improved performance in a test of elbow flexion strength. The results plotted by the latter are very striking. Ikai and Steinhaus (1961) ~~and Ikai (1962)~~ also achieved an increase in performance by firing a gun behind the subject's back a few seconds before he was due to perform, or asking him to shout loudly as he made his effort. Although these stimuli were given during the course of a repetitive exercise which was presumably resulting in progressive fatigue, the shot or shout elicited a better performance than the original unfatigued condition. The authors discussed these results in Pavlovian terms.

Moral fibre could be said to interact strongly with muscle fibre.

1,2,5,3 DRUGS

Prompted by the real and imaginary effects of drugs on athletic performance, several investigations have been conducted into their effects on muscle strength. The amphetamine group of C.N.S. stimulants have been consistently found to increase strength whereas alcohol, adrenaline and chlordiazepoxide have shown no statistically significant effects. (Ikai and Steinhaus, 1961; Adamson and Findlay, 1965; Hurst *et al* 1968).

1,2,5,4 CIRCADIAN RHYTHMS

Wright (1959) produced evidence for the existence of a circadian cycle in strength of grip and found that this was closely linked to the body temperature cycle. It was also shown that hot baths had an effect on both temperature and strength, but rightly stated that this did not necessarily imply a causal relation for the circadian phenomenon. Wright's graphs show that grip is in one case less than half as strong in the troughs of the cycle as it is in the peaks; whereas raising body temperature as high as 101°F. in a hot bath produces strength changes in the order of 10 percent. As the next section (1,2,5,5) will show, the effects of raised temperature on muscle strength are not entirely predictable. On balance it seems more likely that cyclical changes of both temperature and strength reflect the function of a common clock in the central nervous system.

1,2,5,5 TEMPERATURE

Every athlete knows that in order to achieve peak performance in an event he must first "warm up" his muscles. Asmussen and Bøje (1945) demonstrated that this is a phenomenon linked specifically to local muscle temperature. Included in their battery of tests was one of the strength of plantarflexion. This showed a marked increase when the calf muscles were raised two degrees C. above their resting temperature. These results are comparable to those of Wright (1959) cited above. Sedgewick and Whelan (1964) reviewed the sometimes conflicting reports in the physical education literature and in their own elbow flexion experiments reported a modest decrease in strength ($p < 0.05$) after muscle warming by diathermy.

Craik and MacPherson (1943) studied the effect of submerging the hand as far as the wrist in an ice/water mixture for 15 minutes. Hand-grip strength was decreased by 21% whereas the strength of opposing the thumb to the fingers was reduced by 44% Clarke, Hellon and Lind (1958) found a 60% reduction in grip strength when the forearm was immersed in water at $2^{o}C$. By cooling either the intrinsic muscles of the hand or the intrinsic and the extrinsic muscles these experiments show that a reduction in local muscle temperature causes a profound reduction in muscle strength.

Cullingham, Lind and Morton (1960) investigated the effects of reduced temperature on the maximum isometric tetanic tensions of the tibialis anterior muscle of the cat. The tensions fell to 70% of the value at body temperature when the muscle temperature was reduced to 20°C. The shape of the length-tension curve remained unchanged. The stimulus rate required to achieve maximal response was greater at low temperatures. Comparisons of the findings of this study with those of Clarke, Hellon and Lind (1958) lead to the conclusion that reduction in strength of voluntary contractions at low temperatures was due in equal parts to neuromuscular block and to changes in the contractile mechanism.

Discussion of the many factors which affect human strength would not be complete without passing reference to the work of Fischer (1947) who measured grip strength daily over a period of several months and claimed to detect a relation between grip strength and the weather.

1,3 INTRODUCTION TO EXPERIMENTAL WORK

The above review of the literature illustrates the multiplicity of factors which together determined the results of a test of strength. In an ideal experimental study of any of these factors, the remainder should be held constant, or varied in a known and systematic fashion. In practice, the uncontrolled factors which exist in any experimental design introduce sources of variance which tend to obscure the relationships under investigation. In the worst instances observations became irrelevant and spurious conclusions are drawn. It is of importance to discover which factors impose the ultimate limitation of performance in any given test of strength. In so doing, it is instructive to attempt to divide the total range of possible limitations of strength into classes, such that the qualitative differences between the classes are as large as possible, whereas the differences within the classes are as small as possible.

1,3,1

The factors which interact to determine strength operate at levels which range from the mechanical through the anatomico-physiological to the behavioural. The present work deals only with the former levels of analysis. For present purposes, strength is defined as the maximal steady force or torque which an individual can voluntarily exert on external test objects under given conditions (1,1,1).

The human body may, for the purposes of mechanical analysis, be considered as a kinetic chain with a finite number of articulations. The limit of the strength of any given action is determined by the weakest link in that portion of the total chain which participates in the action concerned. It is possible, in a biomechanical analysis of strength, to recognise <u>at least</u> three ways in which such limitations may occur. Each of these three possibilities will now be considered in isolation. In real tests of strength, two or more of them may interact.

a) Firstly there are those situations, initially described by Dempster (1958, etc.) and by Whitney (1958) in which the force output is limited by the weight of the body and by the distribution of body mass. Dempster (1958) cited the example of two-handed pulling actions in the sagittal plane. Taking moments about some pivot point, the torque exerted by the applied force can never exceed the counter-torque due to gravity acting at the centre of mass of the body. An analysis of the statics of pulling and lifting actions will be presented at the beginning of Chapter 4. Dempster characterised such tasks as "closed chain" activities. This term is not entirely satisfactory. A closed kinetic chain is one which has two or more points of contact with the outside world which are rigidly linked in the plane of analysis. A man standing in the "at ease" posture forms a closed chain in the coronal and transverse planes by virtue of his two feet being on the ground. He is an open chain in the sagittal plane and in all planes with regard to actions in which his lower limbs and pelvis play no direct part. When he exerts a force on any object rigidly linked to the floor, he closes the chain. Hence, the majority of strength tests performed on terra firma are closed chain actions and Dempster's terminology should therefore be abandoned. It is proposed therefore that the limitation of strength by the weight and mass distribution of the body be termed a GRAVITATIONAL (Type I) limitation.

b) A second type of limitation is one in which an element within the body is tested to its limit and therefore determines the measured output. It is this situation which pertains, or at least is assumed to pertain in the vast majority of formal measurements of "muscle strength". Because the local mechanics of muscles are inextricably linked with the mechanics of the articulations across which they function, this will be referred to as a MUSCULOSKELETAL (Type II) limitation.

c) There is a third type of limitation which has received little formal study in spite of its practical importance. A limitation of this kind occurs, for example, if a man attempts to exert a force on a slippery handle or while standing on a slippery floor. In these cases the man's capacity to exert a measurable force is limited by the nature of the interfaces between the outside world and himself. This situation will therefore be named an INTERFACE (Type III) limitation.

d) Mechanically, all possible tests of strength fall into one of the above three categories. It is readily acknowledged however that strength is frequently reduced to levels below the mechanical limits described above by the influence of a variety of other factors. Examples of such factors are motivation, pain, discomfort and varieties of psychological and physiological fatigue. These factors may override any of the three biomechanical limitations and such an occurrence will be named a MISCELLANEOUS (Type IV) limitation of strength. In practical tests of strength, Type IV factors are always present to some extent and act as a source of variance in the results. It can be assumed that, given good experimental design, the effects of Type IV factors will be random ones, and no systematic bias will be introduced. Hence, when the strength of a group of subjects is tested under different conditions, the relative results for these conditions are not affected by type IV factors.

The experiments which will be described constitute an exploratory study of some properties of gravitational, musculoskeltal and interface limitations of human strength and the ways in which they interact.

1,3,2 The experimental work to be presented is divided into three sections. Chapter 2 deals with the action of plantarflexion; Chapter 3 deals with the action of supination, and Chapter 4 deals with a variety of pulling tasks. Each section includes accounts of several separate, but related experiments and these will be referred to by the number of the subsection in which their methods and results are described. Each section includes one or more sub-sections in which the results of these experiments are discussed and conclusions are drawn. Certain aspects of experimental procedure are common to many of the experiments which have been performed. To avoid repetition these have been described in an appendix (1). All statistical methods used in the analysis of the data have been described in a second appendix (2).

CHAPTER 2

A STUDY OF THE ANGLE-TORQUE CURVE OF PLANTARFLEXION

"Write how each muscle can become extended or contracted or made thinner or thicker, and which is more or less powerful".

Leonardo Da Vinci

Fog. A 14 J Institut de France.

after MacCurdy, 1938

Introduction

It has been noted above (1,2,3,3) that certain aspects of the human angle-torque curve have not yet been discussed. The present study aims to investigate the ways in which two joint muscles influence the angle-torque curve and the relative contributions of active and passive torques. The action of plantarflexion at the ankle (talo-crural) joint was chosen for investigation because of the following considerations. All joints in the body are complex in both their muscular and articular mechanics but the ankle presents certain specific advantages. Torques about the ankle may be directly measured using a suitably designed pedal, and the foot is sufficiently light in weight that errors in corrections for gravity will be relatively insignificant. No major studies of the angle torque curve of the ankle have appeared in the literature (in contrast for example to the knee and elbow joints which have been widely investigated) and data on the strength and posture of the ankle would be of relevance to locomotor and other research in progress in the author's laboratory.

Herman and Bragin (1967) present some torque data in an electromyographic study of gastrocnemius and soleus muscles and Hertzberg and Burke (1971) tabulated maximal ankle torques in efforts against an aircraft brake pedal in various locations and orientations. The articular geometry of the talo-crural joint and its associated articulations, the sub-talar and transverse tarsal joints have been described by several authors. Barnett & Napier (1952); Isman and Inman (1968).

The present study is in two parts. The first is an experimental determination of the active and passive angle-torque relationships of a small group of normal subjects. In this experiment efforts were made to conduct an investigation which was parallel to one which might be made in plotting the length-tension curve of an isolated muscle. The second part consists of measurements made on cadaveric limbs in order to acquire more information concerning the anatomical basis of the strength measurements.

2,1 MEASUREMENT OF PLANTARFLEXOR TORQUE IN LIVING SUBJECTS.

2,1,1 METHODS

(a) Subjects.

Nine fit young adults acted as subjects in this study. Seven were male and two female, but as sex differences have no obvious relevance to this particular study, this population is treated as being homogeneous. Several anthropometric measures were taken, the means and standard deviations of which are tabulated in Table 2,1.

(b) Sign Conventions.

The sign conventions shown in figure 2,1 are used throughout this study. All measurements are made in the plane common to the palpable eminences of the lateral malleolus of the fibula, the head of the fibula and the greater trochanter of the femur. The knee angle is the angle subtended through the line joining the axes of rotation of the knee and hip joints and the line joining the axes of rotation of the knee and ankle joints.

(The axis of rotation of a joint may for present purposes be located upon the skin by a black tape marker. The marker is placed in such a position that it suffers minimum displacement when either of the body members articulating at the joint moves relative to the other (Dempster, 1955).)

The ankle angle is the angle subtended by the line joining the axes of rotation of the knee and ankle joints, and a plane tangent to the sole of the foot.

The knee angle is zero when the axes of the hip, knee and ankle joints are colinear, and increases with knee flexion. The ankle angle is ninety degrees when the line joining the axes of the knee and ankle is perpendicular to the tangent to the sole; dorsiflexion increases the ankle angle, and plantarflexion decreases the ankle angle.

Torques exerted about the axis of the ankle which tend to cause plantarflexion are positive.

(c) Torque Measurement.

The measuring apparatus is shown in figure 2,2. The subject was prone and kneeling on a padded table top. The foot to be tested (right) was free from the edge of the table and was securely strapped to the torque measuring device which was adjustable for height along two substantial piers. The device consisted of a footplate which pivoted on an axle which could be aligned with the axis of rotation of the ankle joint of the experimental subject, hence ankle joint torques could be measured directly about the axle without the necessity for any corrections other than a simple calibration. The shaft was mounted in low friction needle-roller bearings and was connected by a rigid lever to a load cell. (Pye Ether type U.F.2). This transducer was equally sensitive to both tension and compression, and could be calibrated

Figure 2,1

Definition of Anthropometric Quantities measured and Descriptive Terms used.

is knee angle

is ankle angle

ve is plantarflexor torque

2,1,1 (b)

Table 2,1 <u>Anthropometric Data for Subjects of Experiment 2,1</u>.

	Mean	S.D.
Weight	676 N	88 N
Stature	1759 mm	75 mm
Shank length (D1)	414 mm	32 mm
Thigh length (D2)	430 mm	24 mm
Foot length (D3)	259 mm	18 mm
Height of centre of rotation of ankle joint from floor (D4)	85 mm	8 mm
Distance of posterior limit of heel to centre of rotation of ankle joint (D5)	58 mm	9 mm

2,1,1 (a)

Figure 2,1

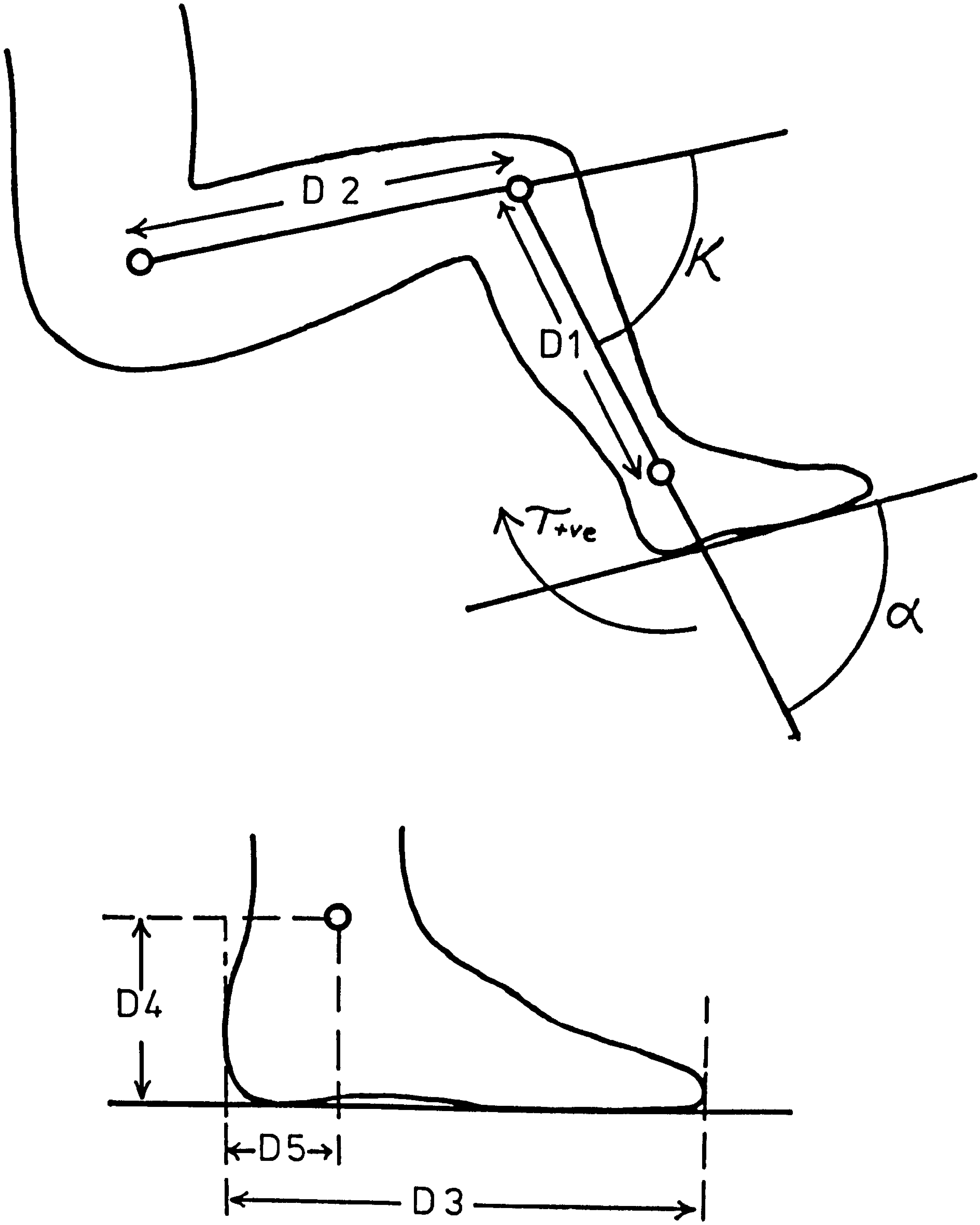
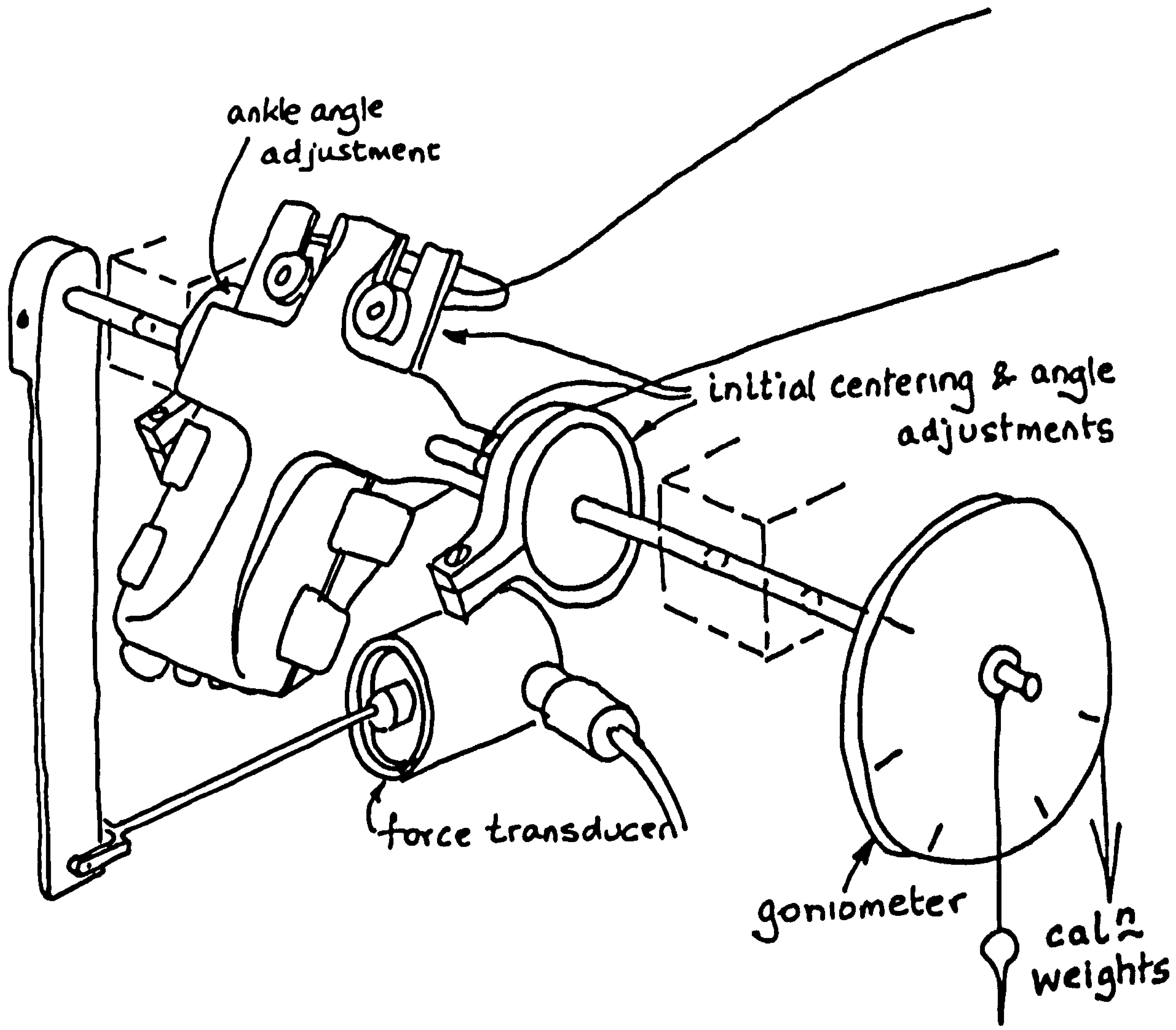


Figure 2,2



Apparatus for measurement of plantarflexor torque.

2,1,1 (c)

by hanging weights on a pulley wheel mounted on the end of the shaft. A clutch was incorporated into the design allowing rapid alteration of the ankle-angle together with rigid locking in any position, the ankle angle being measured by a protractor and plumb line mounted on the end of the shaft. The transducer signal was amplified using the AD6 amplifier of the Medelec MS6 modular recording system.

(d) Electromyography.

Surface electrodes (I.M.I. Ag/AgCl) were mounted over the surface features of muscles tibialis anterior and triceps surae. The former were placed approximately 30 mm above and below the centre of the palpable tibialis anterior muscle, the latter were placed with a 60 mm separation spanning the centres of soleus and the lateral head of gostrocnemius. An indifferent electrode was mounted just above the ankle. The subject's skin was prepared by shaving (where necessary), washing with alcohol, and abrading lightly using fine emery paper until a slight erythema was produced. The electrodes were mounted and remounted until a low noise condition was achieved. The myoelectric signals were amplified by two AA6 amplifiers of the Medelec MS6 system. The signals were then displayed on the C.R.T. screen of the system and recorded along with the torque measurements. The minimal detectable signal of the overall system was approximately 10 microvolts.

(e) Procedure

Torque measurements were made at ankle angles of 50, 60, 70, 80, 90, 100 and 110 degrees, as measured using the protractor incorporated in the testing. (The protractor had originally been adjusted so that it read correctly for each individual subject). At each ankle angle measurements were made when the knee was at 0, 45, 90 and 135 degrees, as measured using a simple goniometer of local construction. The subjects were given precise instructions of the commencement of the experiment concerning the nature of the exertions which were required and were allowed several practice efforts. Each subject commenced in the posture of 90 degrees at the ankle and 0 degrees at the knee, but the postures were randomised thereafter. Constraints imposed by shortage of time unfortunately made it necessary for the 60, 80 and 100 degrees positions of the ankle to be omitted for two subjects.

In each posture the subject was given visual feedback of his e.m.g. signal until he achieved myoelectric silence in both muscle groups monitored. The torque existing in this state was recorded, and is referred to henceforth as the passive torque for the posture concerned.

The subject was then required to exert a maximal voluntary plantarflexor torque. These recordings were made according to standard procedure (Appendix 1) and the results measured from the paper trace are henceforth referred to as the total torques. The difference between the total torque for a given posture and the passive torque for the same posture will be described as the active torque.

2,1,2 RESULTS.

All the results of these experiments are to be found in Table 2,2 at the end of this section.

(a) Passive Torques

Figure 2,3 shows the dependence of passive torque upon joint angles. The means of the subjects are plotted together with vertical band representing ± 1 standard error of the mean. A net plantarflexor torque exists in positions of dorsiflexion. This torque is strongly dependent upon knee angle. Passive torque is always zero at 90 degrees ankle angle. A net dorsiflexor torque was observed in positions of plantarflexion, but differences at the different knee angles were in this case negligible and the data has been pooled.

It was noted that many subjects experienced difficulty in relaxing tibialis anterior when in positions of dorsiflexion, although all subjects succeeded eventually.

(b) Active Torques

Figure 2,4 shows the absolute values of both active and passive torques (mean of all subjects). The precise optimal position of active plantarflexor torque varied from subject to subject. In all cases it was at an ankle angle greater than or equal to 90 degrees with either 0 or 45 degrees at the knee. The best performances of the subjects taken as a group had a mean value of 86.8 Nm with a standard deviation of 40.7 Nm.

Data was normalised with respect to the mean active torque and the mean normalised torque is plotted

Figure 2,3

Passive plantarflexor torques.

(Mean $\pm$ 1 s.e.m., plotted against ankle angle)

(a)	Knee	=	0 deg.
(b)	Knee	=	45 deg.
(c)	Knee	=	90 deg.
(d)	Knee	=	135 deg.

At ankle angles $\leqslant$ 90 deg., data for the knee angles has been pooled.

2,1,2 (a)

Figure 2,3

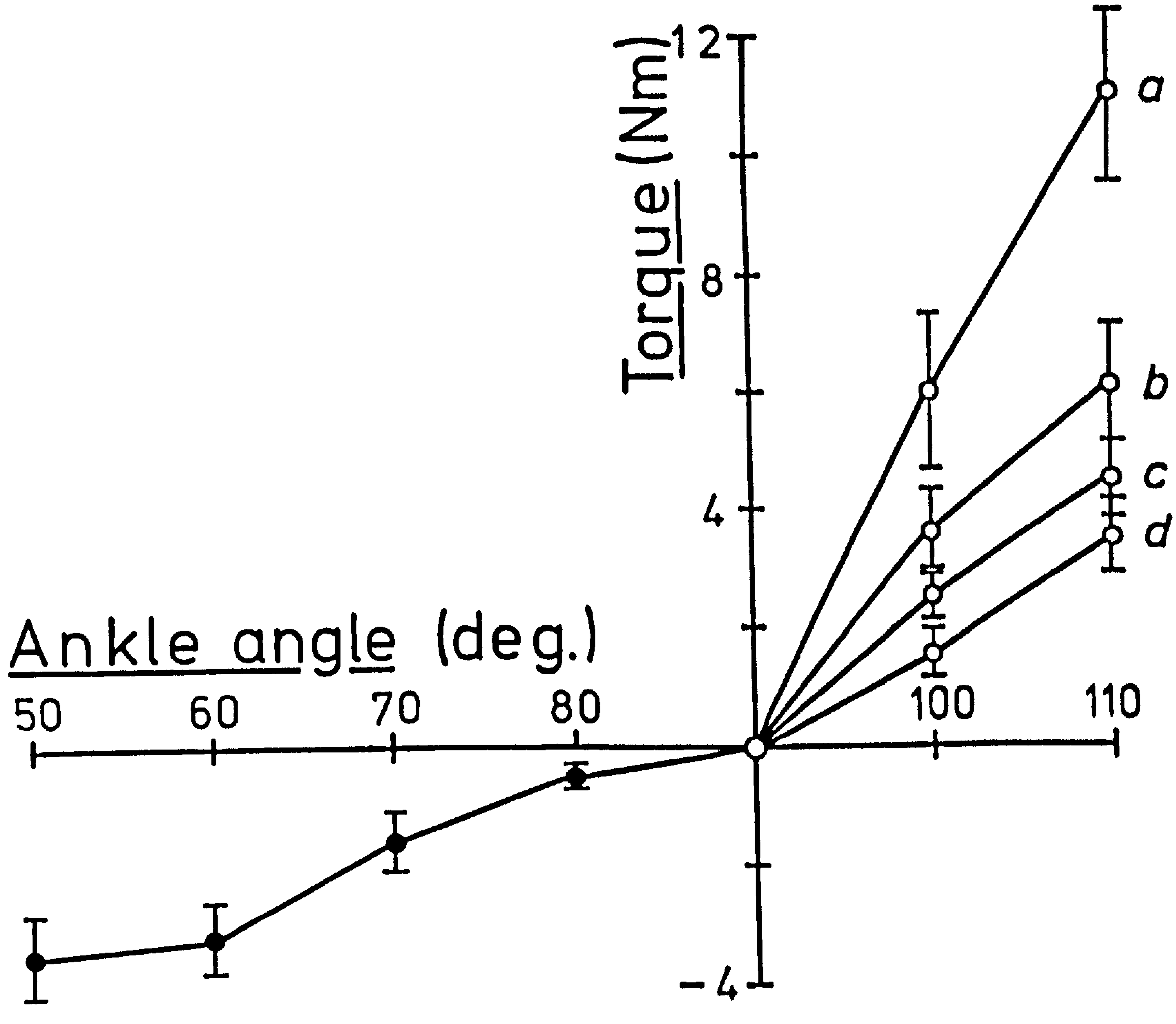


Figure 2,4

Active and Passive Plantarflexor Torques.

(Mean plotted against knee angle)

——o——	Knee = 0 deg.
——●——	Knee = 45 deg.
——X——	Knee = 90 deg.
——+——	Knee = 135 deg.

2,1,2

Figure 2,4

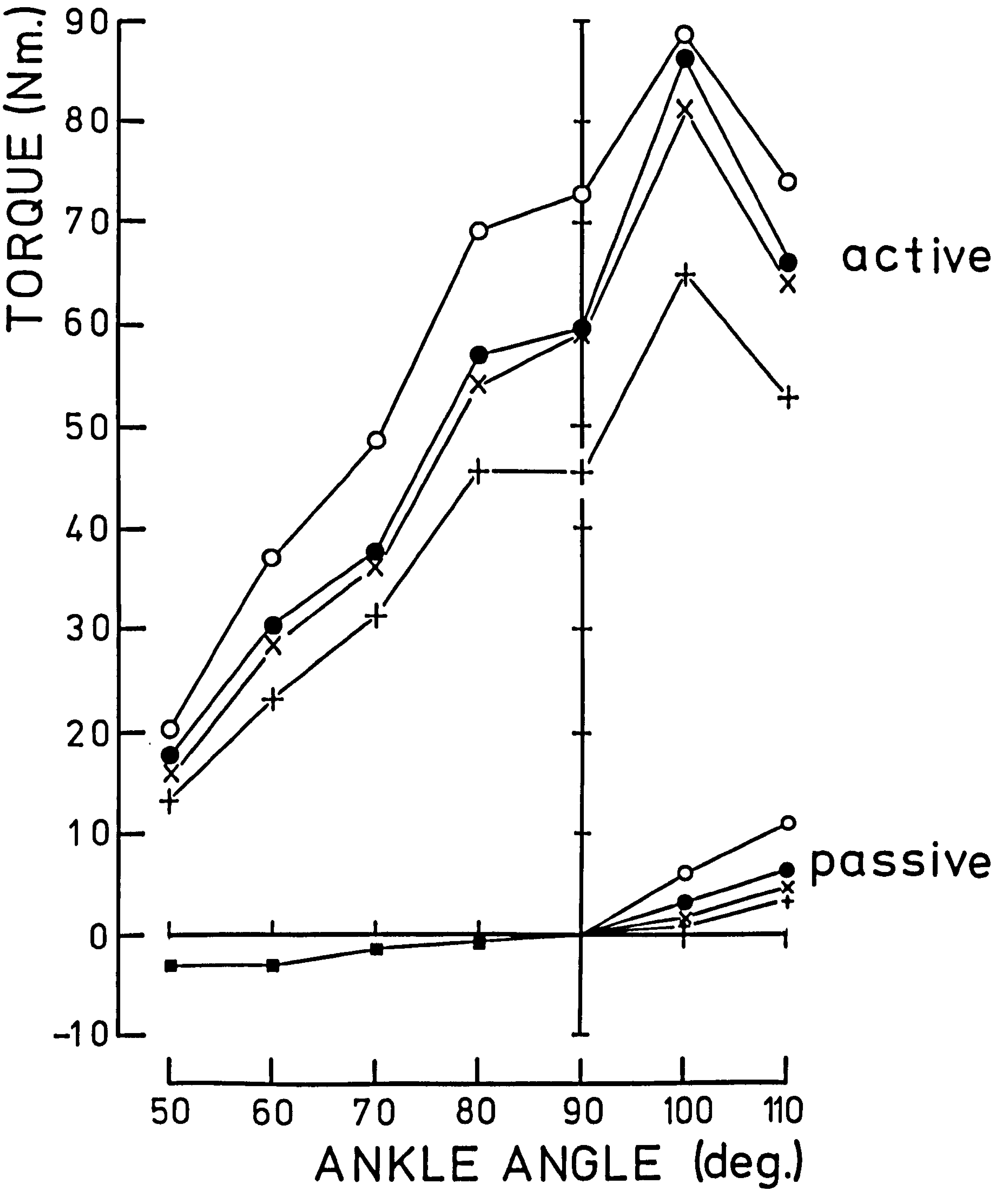

TORQUE (Nm.)
90
80
70
60
50
40
30
20
10
0
-10
active
passive
50
60
70
80
90
100
110
ANKLE ANGLE (deg.)

Figure 2,5

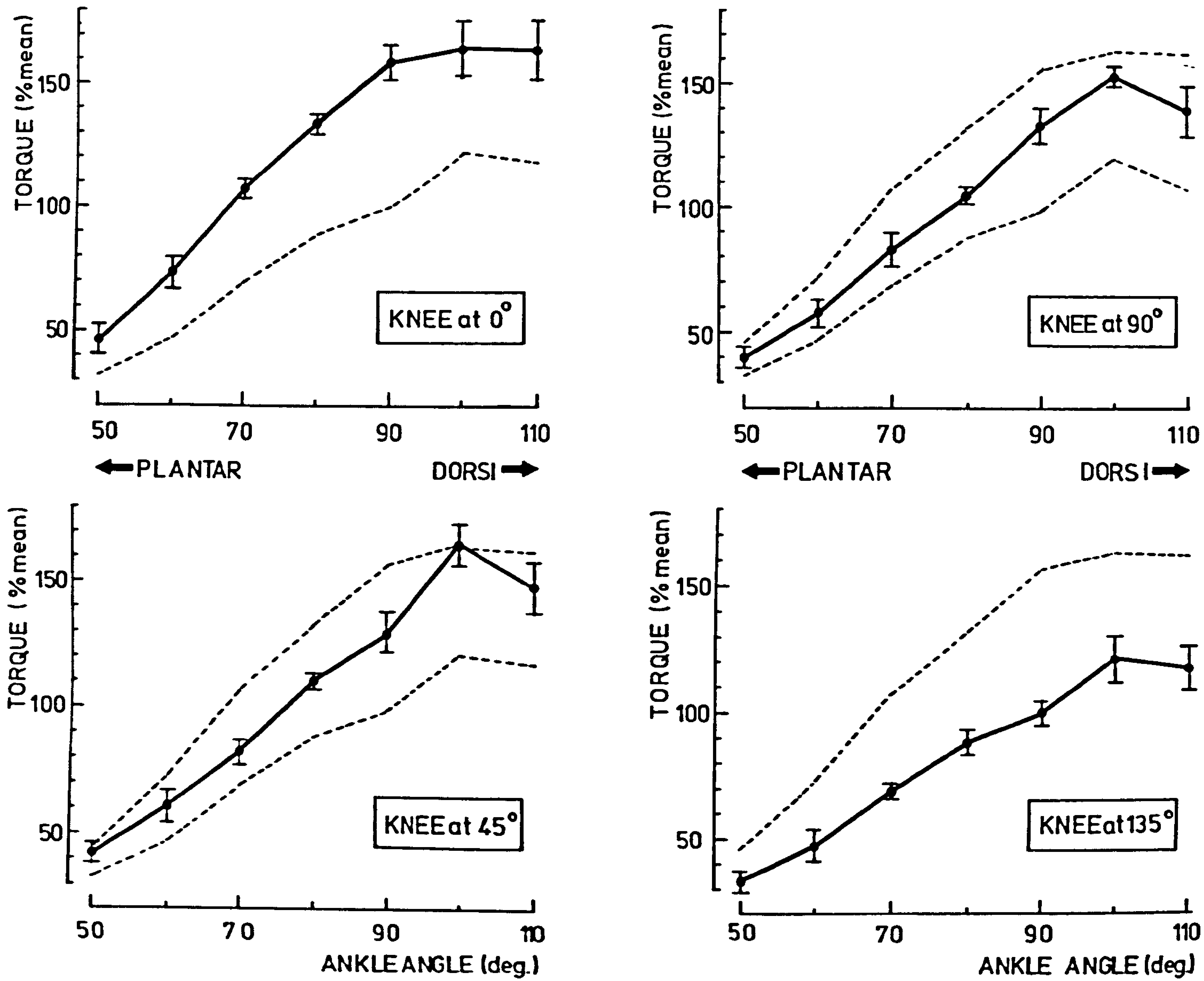
Normalised Plantarflexor Torques

plotted against ankle-angle.

(Mean $\pm$ 1 s.e.m.)

Dotted lines plotted on each chart represent the results for the extreme knee angles.

2,1,2 (b)



TORQUE (%mean)
150
100
50
KNEE at 0°
50
70
90
110
PLANTAR
DORSI
TORQUE (%mean)
150
100
50
KNEE at 90°
50
70
90
110
PLANTAR
DORSI
TORQUE (% mean)
150
100
50
KNEE at 45°
50
70
90
110
ANKLE ANGLE (deg.)
TORQUE (% mean)
150
100
50
KNEE at 135°
50
70
90
110
ANKLE ANGLE (deg.)

in figure 2,5 $\pm$ 1 S.E.M. Results for active torques, both before and after normalisation were tested against the results for every other posture using the Student's t-test for independent samples (see Appendix 2). The outcome of these tests are shown in figure 2,6. In the case of the normalised data, at a given knee angle, differences were highly significant between adjacent ankle angles up to 90 degrees, but beyond 90 degrees such differences were not found. For a given ankle angle, the significance of differences between knee angles increased with the difference between the angles.

(c) <u>Calculation of the Gravitational Contribution to the passive torques</u>.

The dead weight of the foot exerts a torque about the shaft of the apparatus which is included in the passive torque as measured. This contribution may be calculated by combining the anthropometric data of the subjects in the experiments with the segmental data of Drillis and Contini (1966). The location of the centre of gravity of an average foot is shown in Fig. 2,7 (a). Combining this with present mean values, the data of Fig. 2,7 (b) is generated (dimensions in mm) and the angle θ is 48.3 degrees. The moment of the foot about the ankle joint in the experiment is given by the equation:

$$M = - \left[74 \times \cos (\theta + \quad - 90) \right]$$

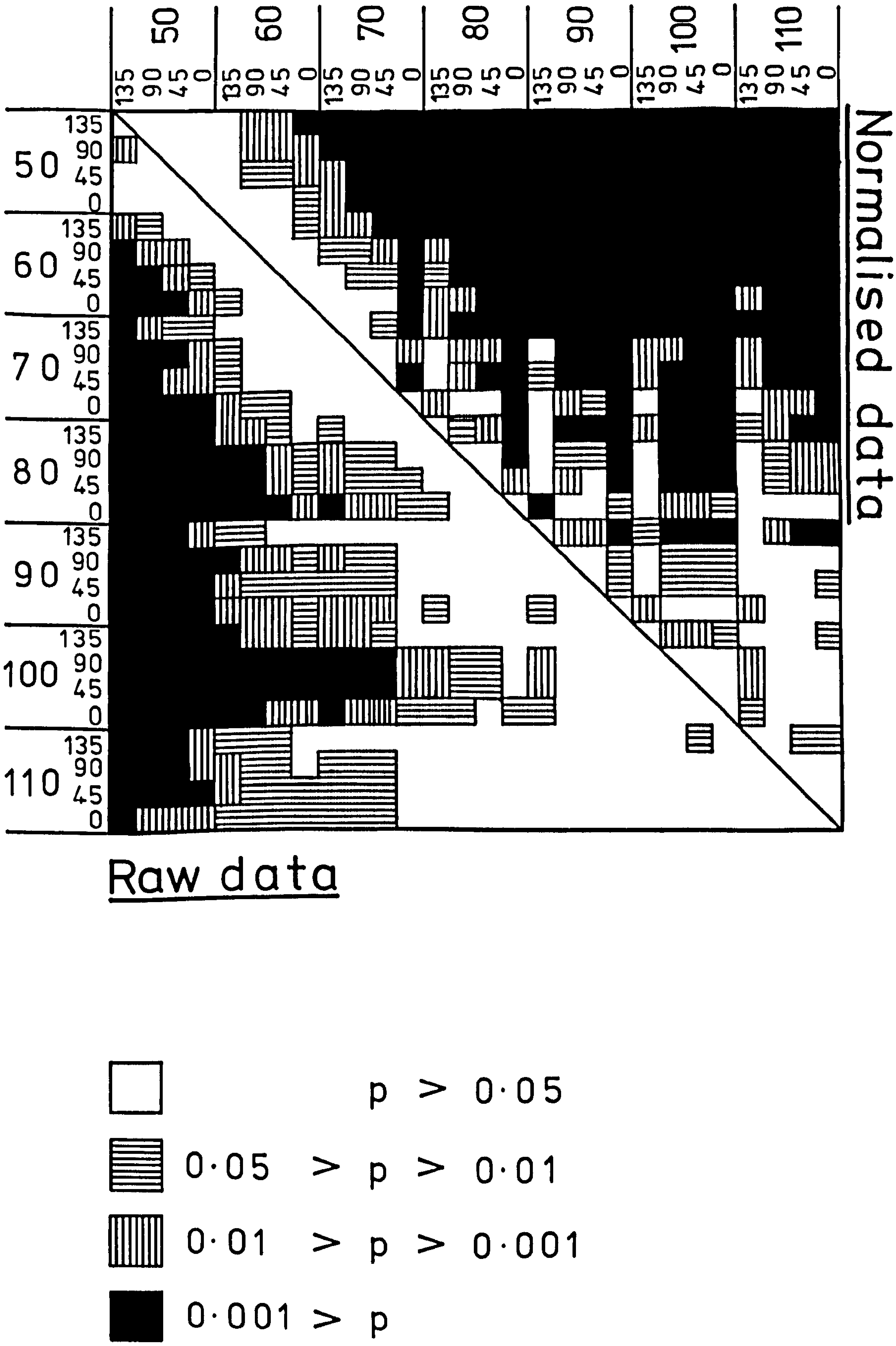
Figure 2,6

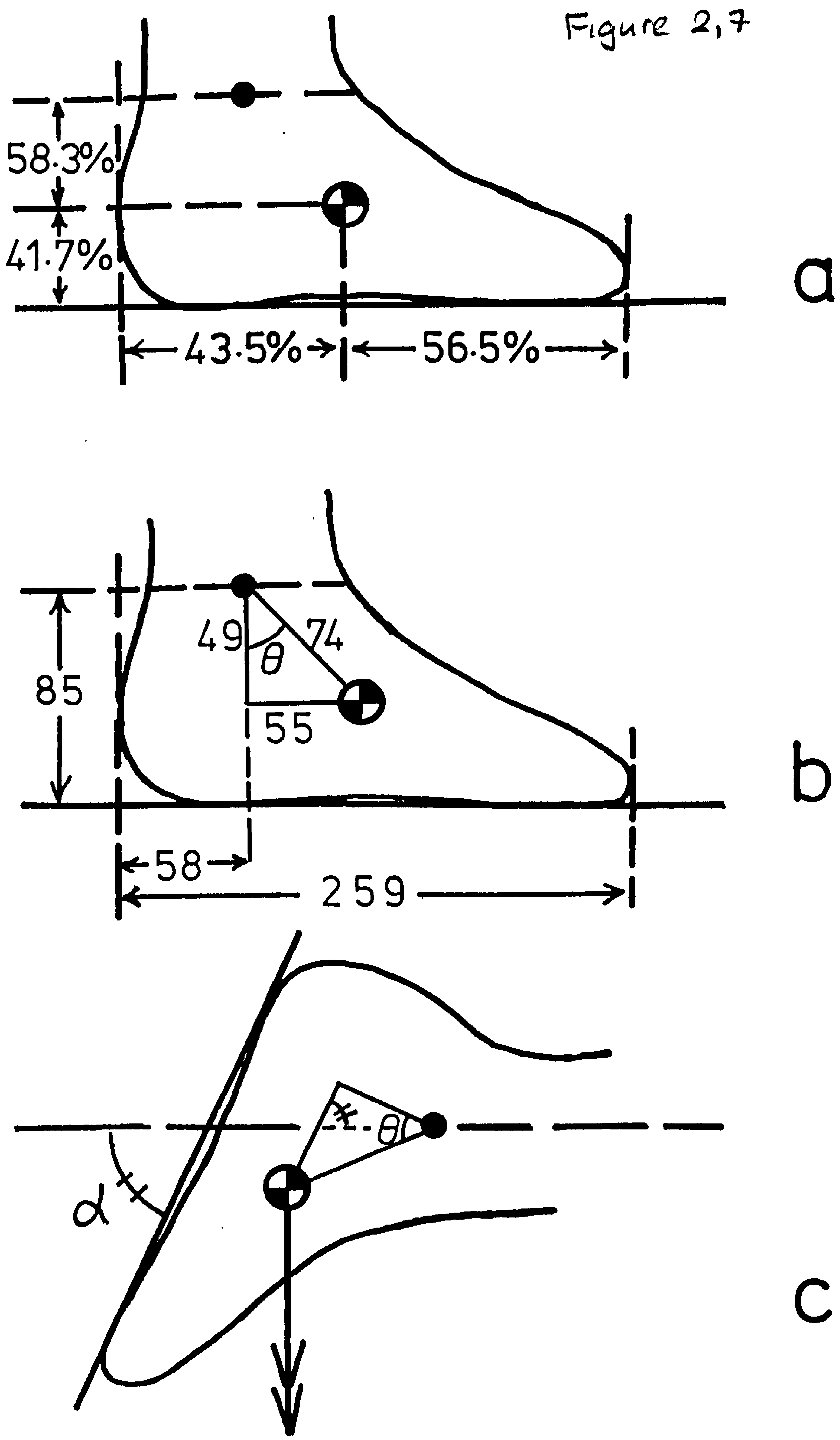
Significance levels for students' t test performed on active plantarflexor torque data in all pairs of conditions. Top right of diagram, normalised data; bottom left, absolute data.

2,1,2 (b)

A2

Figure 2,6


50
60
70
80
90
100
110
135
90
45
0
Normalised data
Raw data
p > 0·05
0·05 > p > 0·01
0·01 > p > 0·001
0·001 > p



Calculation of gravitational contribution to passive plantarflexor torques. Dimensions in fig. b are in mm.

From the data of Drillis and Contini (1966) the mean weight of a subject's foot is 9.5 Newtons. Hence the dead weight contributions of the passive torque may be estimated (Table 3,2)

Ankle Angle (deg.)	Calculated Dead-weight Torque (Nm)
50	- 0.69
60	- 0.67
70	- 0.62
80	- 0.55
90	- 0.47
100	- 0.37
110	- 0.26

Hence, although the errors in these calculations are potentially large it may be seen that the dead-weight contribution to the passive torque is at least an order of magnitude less than the elastic contribution.

Table 2,2 PLANTARFLEXOR TORQUES

POSTURE ANKLE	KNEE	ACTIVE TORQUE (Nm) $\bar{x}$	(Nm) s	(%) $\bar{x}$	(%) s	PASSIVE TORQUE (Nm) $\bar{x}$	(Nm) s		GRAVITY TORQUE (Nm)	N
50	0	20.2	9.2	47	18	- 3.5	2.2	-3.1 ± 1.9	- 0.69	9
	45	17.8	6.1	42	13	- 3.6	2.4			
	90	16.3	5.4	39	14	- 2.7	1.1			
	135	13.5	4.6	33	13	- 2.4	1.6			
60	0	37.0	10.9	73	19	- 3.6	1.6	-3.2 ± 1.5	- 0.67	7
	45	30.5	8.1	61	18	- 3.1	1.3			
	90	28.6	7.3	57	15	- 3.0	1.2			
	135	23.2	7.6	47	17	- 3.2	1.6			
70	0	48.7	17.7	107	9	- 1.7	1.6	-1.5 ± 1.6	- 0.62	9
	45	37.7	13.8	83	14	- 1.7	1.6			
	90	36.3	11.4	84	20	- 1.4	1.5			
	135	31.6	12.6	69	9	- 1.4	1.5			
80	0	69.3	22.1	132	12	- 0.2	0.5	-0.3 ± 0.4	- 0.55	7
	45	57.1	14.9	111	10	- 0.3	0.6			
	90	54.2	13.9	105	9	- 0.2	0.5			
	135	45.5	12.7	88	4	- 0.4	0.7			
90	0	72.7	31.5	157	22	0	0		- 0.47	9
	45	59.8	27.4	130	24	0	0			
	90	60.0	21.3	133	21	0	0			
	135	45.6	18.9	99	16	0	0			
100	0	88.7	38.5	164	33	6.0	3.3		- 0.37	7
	45	86.3	28.6	166	25	3.6	1.8			
	90	81.2	29.4	154	12	2.5	1.0			
	135	64.5	24.6	121	24	1.6	1.0			
110	0	73.9	42.9	162	37	11.0	4.1		- 0.26	9
	45	66.4	33.0	149	31	6.1	3.1			
	90	63.7	31.1	139	32	4.6	1.9			
	135	53.8	29.1	118	27	3.5	2.0			

2,2 CADAVERIC STUDIES OF THE EXCURSIONS OF MUSCLES ACTING ABOUT THE ANKLE JOINT.

An extensive study of cadaveric material was originally conceived with the intention of predicting the human angle-torque curve from a knowledge of musculo-skeletal morphology and of published data concerning the length-tension relationship of mammalian muscle. Whilst such a fusion of the reductionist and holistic approaches remains a laudable aim, the author decided to abandon the project, due to the fact that the number of variables involved in such a model is so large that it seems unlikely that a model based upon them could ever be truly tested against reality. Without recourse to such complexities it is nonetheless possible to use certain simple morphological data to inform our understanding of the observed angle-torque curve. The present section deals with excursions of the muscles spanning the ankle joint measured on eight cadaveric limbs.

2,2,1 METHODS

(a) Materials

The present author investigated five cadaveric limbs. Three subjects were female, two were male and all had died above the age of sixty years from myocardial infarction or cerebro-vascular accident. The cadavers were embalmed and used for undergraduate teaching purposes; undissected limbs were available for research at the end of the academic session. It is acknowledged that all specimens showed the degree of muscular wastage typical of elderly subjects, but in no case was this sufficiently pronounced to be pathological. The limbs were removed by disarticulation at the hip or by hemisection of the pelvis and lumbo-sacral disarticulation. Muscles investigated were gastrocnemius, soleus, peroneus longus, peroneus brevis, flexor hallucis longus, flexor digitorum longus, tibialis posterior, tibialis anterior, extensor hallucis longus, extensor digitorum longus, and peroneus tertius. M. peroneus tertius was present in all specimens, m.plantaris was present in three specimens but was ignored. The results tabulated have been supplemented by the inclusion of measurements of gastrocnemius made by P. Cavanagh. (Cavanagh, 1972) on three further cadaveric limbs.

(b) Technique

The skin, superficial fascia and deep fascia (fascia lata, fascia cruris) from the entire limb as far distal as the mid-tarsal region. The muscles of the calf were cleaned and their tendons were severed at a level approximately three centimeters proximal to the talo-crural articulation. (In the case of m.flexor hallucis longus this demanded that a few muscle fibres were also severed). Fat and connective tissue was then removed from around the talo-crural joint until a full range of movement at the joint was achieved. This involved a variable degree of destruction of the extensor, flexor and peroneal retinacula together with the making of small insertions in the joint capsule and its associated ligaments. All muscles spanning the knee, with the exception of gastrocnemius, were severed, and the joint was freed in a similar fashion.

A wooden board was nailed to the sole of the foot to define its tangent plane and pins were driven into the estimated centres of rotation of the ankle, knee and hip joints so that angular measurements could be made as described for the living subjects.

The ankle joint was then manipulated through its range of motion and at ten degree intervals the gap between the cut ends of the tendons was measured with a pair of spring calipers, readings being recorded to the nearest milimeter. Efforts were made to sustain the tendons in a lifelike condition; a little tension was applied to take up the slack, and "bow stringing" due to the removal of restraining bands of fibrous tissue was avoided as far as possible. It is impossible to gauge to what extent these ends were achieved. The aponeurosis by which gastrocnemius joins the tendo calcaneus was then severed from the soleus aponeurosis and the excursions of gastrocnemius with flexion of the knee were measured at ten degree intervals. For each muscle in each limb four sets of measurements were made (twice in ascending and descending order of joint angle, and means were calculated for each position).

2,2,2 RESULTS

All excursion measurements were standardised by converting each reading to a value for the difference of the length in which the reading was taken from the length in a standard position when the knee or ankle joint was at 90 degrees. These results were then normalised by expressing each individual reading as a percentage of the shank length (the distance between centres of the knee and ankle joints). The normalised data was then treated by least squares polynomial regression (see Appendix 2) and significant terms were determined by the analysis of variance. A scatter plot of the excursion data for m. gastrocnemius is given in fig. 2,8. Results of the statistical procedure are shown in table 2,3 and the regression lines are plotted in fig. 2,9. It may be seen that only in the case of the triceps surae data does the quadratic term reach statistical significance.

Figure 2,8

Excursions of gastrocnemius muscle ΔL expressed as a percentage of the shank length, plotted against ankle and knee angles.

2,2,2

Figure 2,8

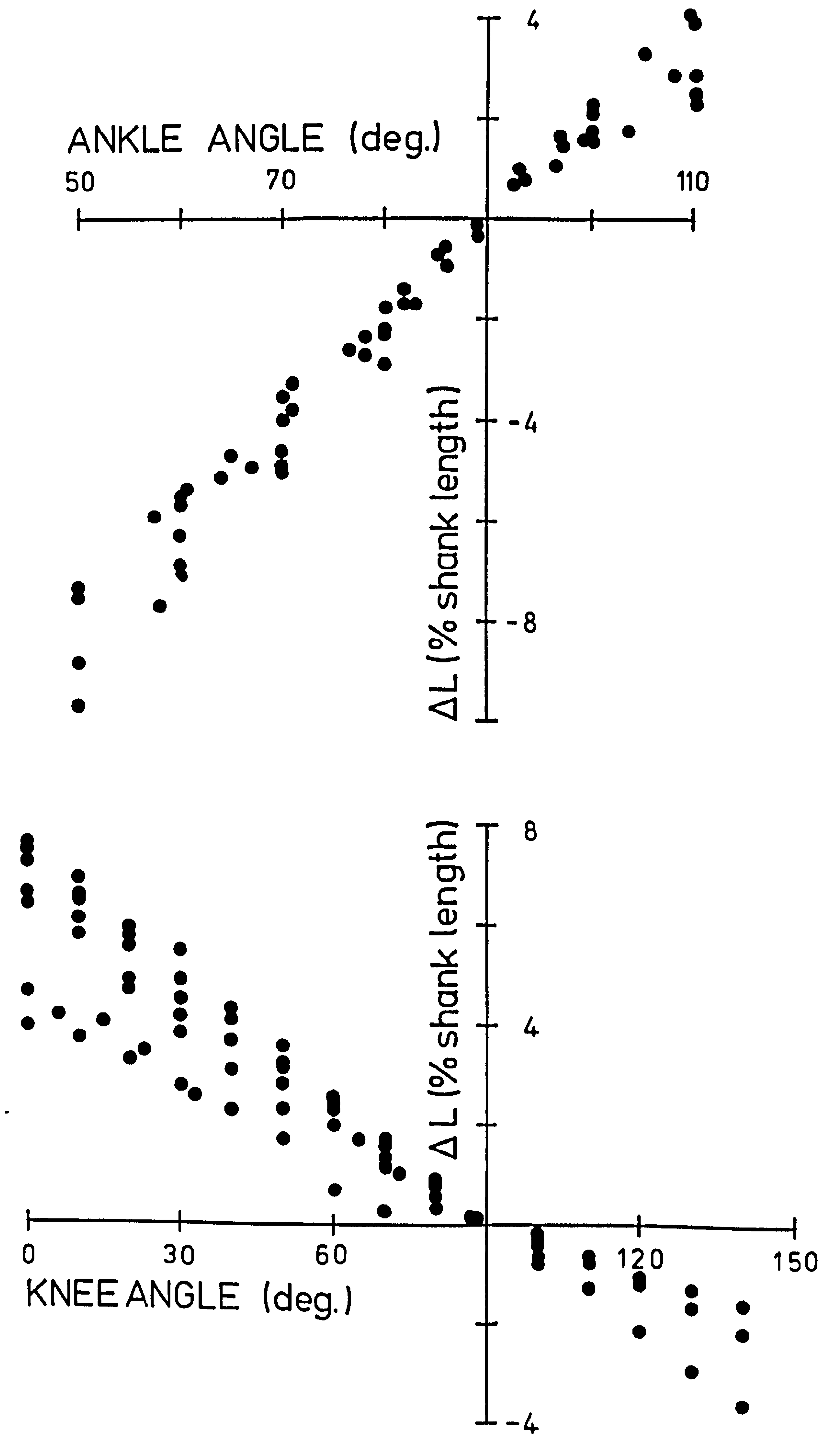

ANKLE ANGLE (deg.)
50
70
110
4
-4
-8
ΔL (% shank length)
8
4
-4
ΔL (% shank length)
0
30
60
120
150
KNEE ANGLE (deg.)

Figure 2,9

Regression lines of muscle excursion (ΔL) plotted against

TA	-	tibialis anterior
EDL	-	extensor digitorum longus
EHL	-	extensor hallucis longus
PT	-	peroneus tertius
TP	-	tibialis posterior
PB	-	peroneus brevis
PL	-	peroneus longus
FDL	-	flexor digitorum longus
FHL	-	flexor hallucis longus
TS	-	triceps surae

2,2,2

Figure 2,9

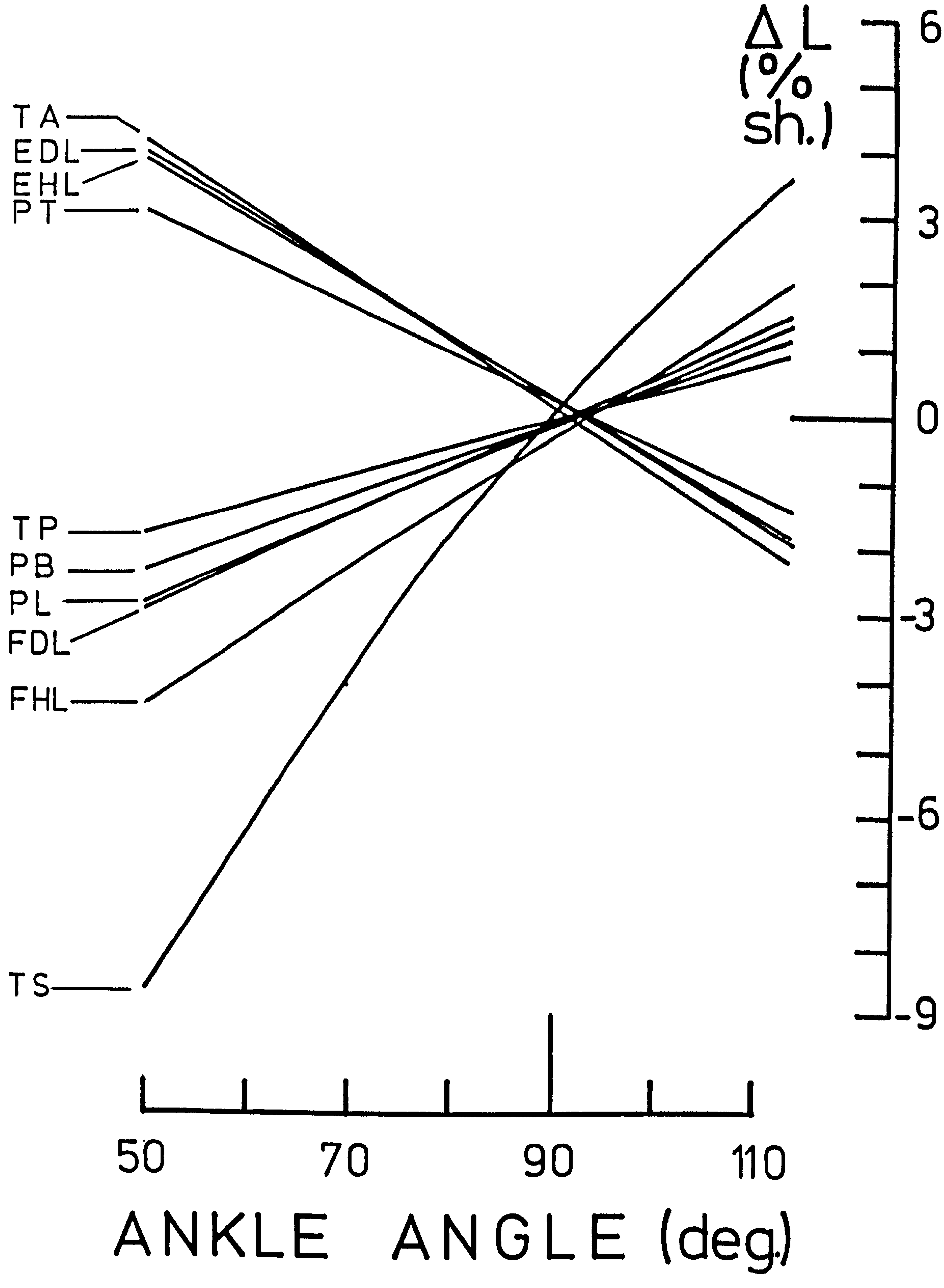

ΔL
(%
sh.)
6
3
0
-3
-6
-9
TA
EDL
EHL
PT
TP
PB
PL
FDL
FHL
TS
50
70
90
110
ANKLE ANGLE (deg.)

Table 2,3

REGRESSION EQUATIONS ($\pm$ 1 R. M.S. ERROR) FOR MUSCLE EXCURSION (ΔL) AS A PERCENTAGE OF LINK LENGTH ON ANKLE ANGLE (α) IN DEGREES.

Tibialis Posterior

$$\Delta L = -3.89118 + 0.04347\,\alpha \pm 0.312$$

Flexor Digitorum Longus

$$\Delta L = -6.44065 + 0.07151\,\alpha \pm 0.302$$

Flexor Hallucis Longus

$$\Delta L = -9.54597 + 0.10427\,\alpha \pm 0.333$$

Peroneus Brevis

$$\Delta L = -5.2531 + 0.05831\,\alpha \pm 0.263$$

Peroneus Longus

$$\Delta L = -6.3186 + 0.06969\,\alpha \pm 0.369$$

Tibialis Anterior

$$\Delta L = 9.23164 - 0.10172\,\alpha \pm 0.314$$

Extensor Digitorum Longus

$$\Delta L = 8.84326 - 0.096997\,\alpha \pm 0.317$$

Extensor Hallucis Longus

$$\Delta L = 8.59070 - 0.09514\,\alpha \pm 0.587$$

Peroneus Tertius

$$\Delta L = 6.95258 - 0.07604\,\alpha \pm 0.364$$

Gastrocnemius + Soleus

$$\Delta L = -22.18468 + 0.30141\alpha - 0.00061\,\alpha^2 \pm 0.492$$

Regression equation of Gastrocnemius Excursion on Knee angle (K).

$$L = 6.46251 - 0.07987K + 0.00011\,K^2 \pm 0.806$$

2,3 DISCUSSION

2,3,1 THE ANGLE-TORQUE CURVE

The angle-torque curve of plantarflexion may be interpreted by a simple consideration of the muscles of the region. Let us ignore for the moment any question of changing leverages or torque contribution from the deep flexor muscles or the peronei: These questions will be raised again at a later stage.

In positions of dorsiflexion, in the living subject, the plantarflexor muscles are under strain; hence the passive torques are positive according to our sign convention and dependent upon the position of the knee. The knee dependence reflects the contribution of the gastrocnemius muscle which spans the knee joint. In positions of plantarflexion the dorsiflexor muscles are under strain, the passive torque is negative and it is not dependent on knee posture. No dependence on the knee is to be anticipated as no dorsiflexor muscles span the knee joint. It is recognised that structures other than muscle contribute to these passive torques.

The active torque curves are those which would be expected if both the one-joint soleus muscle and the two-joint gastrocnemius muscle operated on a portion of their length-tension curve ranging from the middle part of the rising limb up to and including the plateau. This is the range which has been shown to be physiological for certain isolated muscles.

We may conclude therefore that under the conditions of the experiment the isometric behaviour of a muscle during maximal voluntary effort has the same dependence on length as an isolated muscle during tetanic stimulation.

The present results for passive torque are not in agreement with the findings of Smith (1957) who reported the presence of a net plantarflexor torque at ankle angles substantially less than 90 degrees. It is difficult to imagine what structures posterior to the ankle joint would be under strain in such a position and hence exerting torque. The discrepancy in findings is possibly due to the different criteria of muscular rest. Smith (1957) presumed that there was no muscular activity below the knee due to the fact that his readings were taken after a period of total ischaemia. In the present study the validity of the measurement depends upon the sensitivity of the e.m.g. system. Electromyographic criteria of muscle silence are least reliable in positions where muscles are at their greatest length, for it is in these positions that the greatest mechanical output is associated with the least myoelectric activity (Grieve and Pheasant, 1976). Nonetheless, it is difficult to conceive how an activity of the anterior tibial muscles sufficient to account for the discrepancy could have been missed.

It is of some interest to compare the absolute torque values for the present study with those recorded in the nearest equivalent survey conducted on a statistically more significant sample. Hertzberg and Burke (1971) made measurements in a "neutral leg" position in which the knee was at approximately 45 degrees according to present notation. In the optimum ankle/pedal configuration the following plantarflexor torques were recorded:

Mean = 99 Nm., S.D = 47 Nm., N = 99.

In the present study, in the optimum ankle postiion (100 degrees) with the knee at 45 degrees, the results were:

Mean = 86 Nm., S.D. = 29 Nm., N = 7.

Applying Student's t-test for independent samples it is found that there is no significant difference between these sets of measurements. ($t = 0.72$; $p > 0.05$)

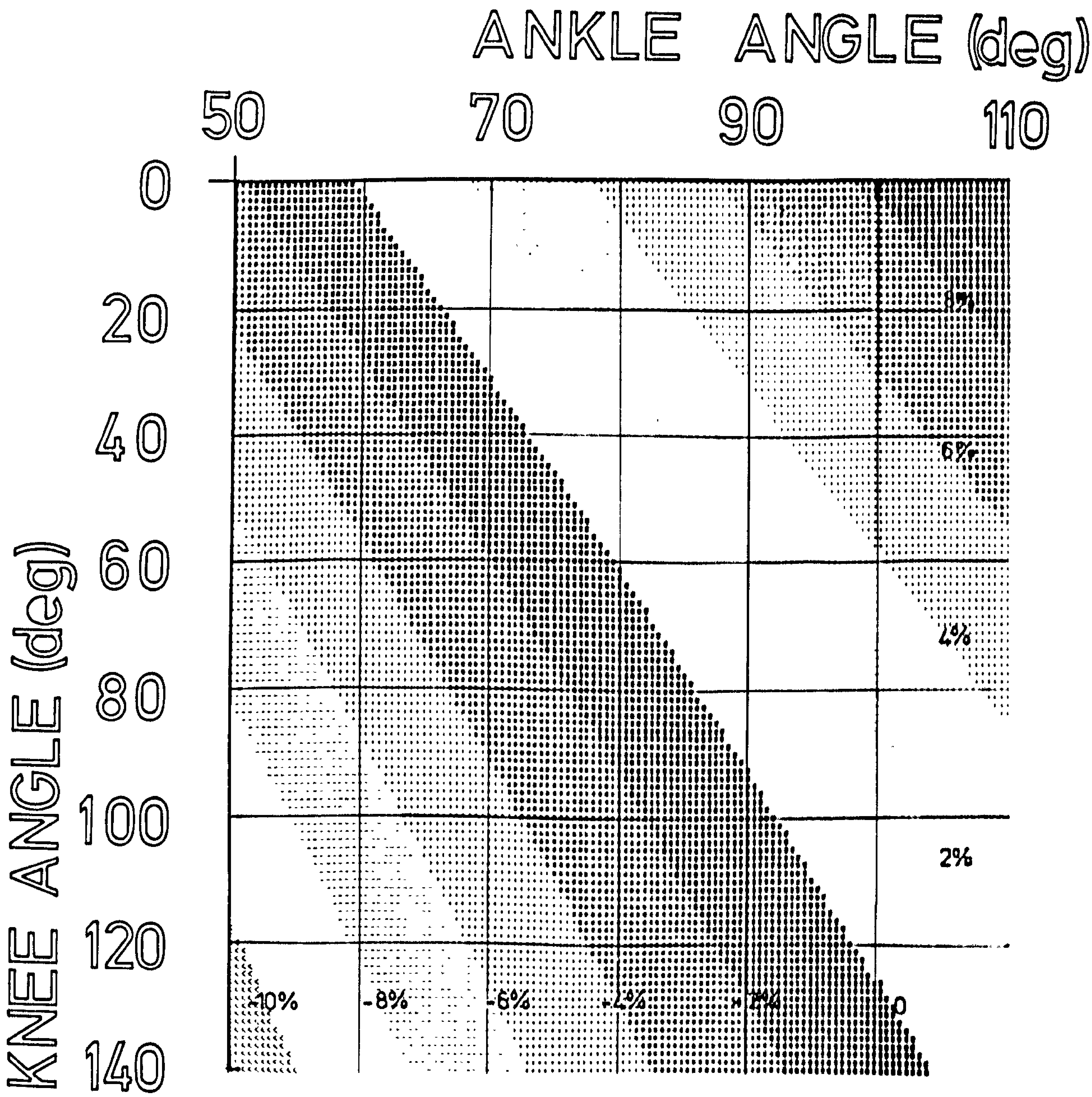
2,3,2 LINES OF CONSTANT LENGTH AND STRENGTH.

Grieve (1969) has used "angle-angle" diagrams to describe movements of the lower limb. An example of such a diagram is the ankle-knee diagram which is a graph in which knee angles are plotted against ankle angles. Using present muscle excursion data it is possible to plot on such a diagram a set of lines of constant length for the muscle gastrocnemius. Figure 2,10 shows such a set of lines which was plotted using the library programme UCLMAP (U.C.L. Computer Centre, Gordon Square, London WC1). The equation used to generate the lines was:

$$\Delta L = -15.72217 + 0.30141\alpha + 0.00061\alpha^2 - 0.07987K + 0.00011K^2$$

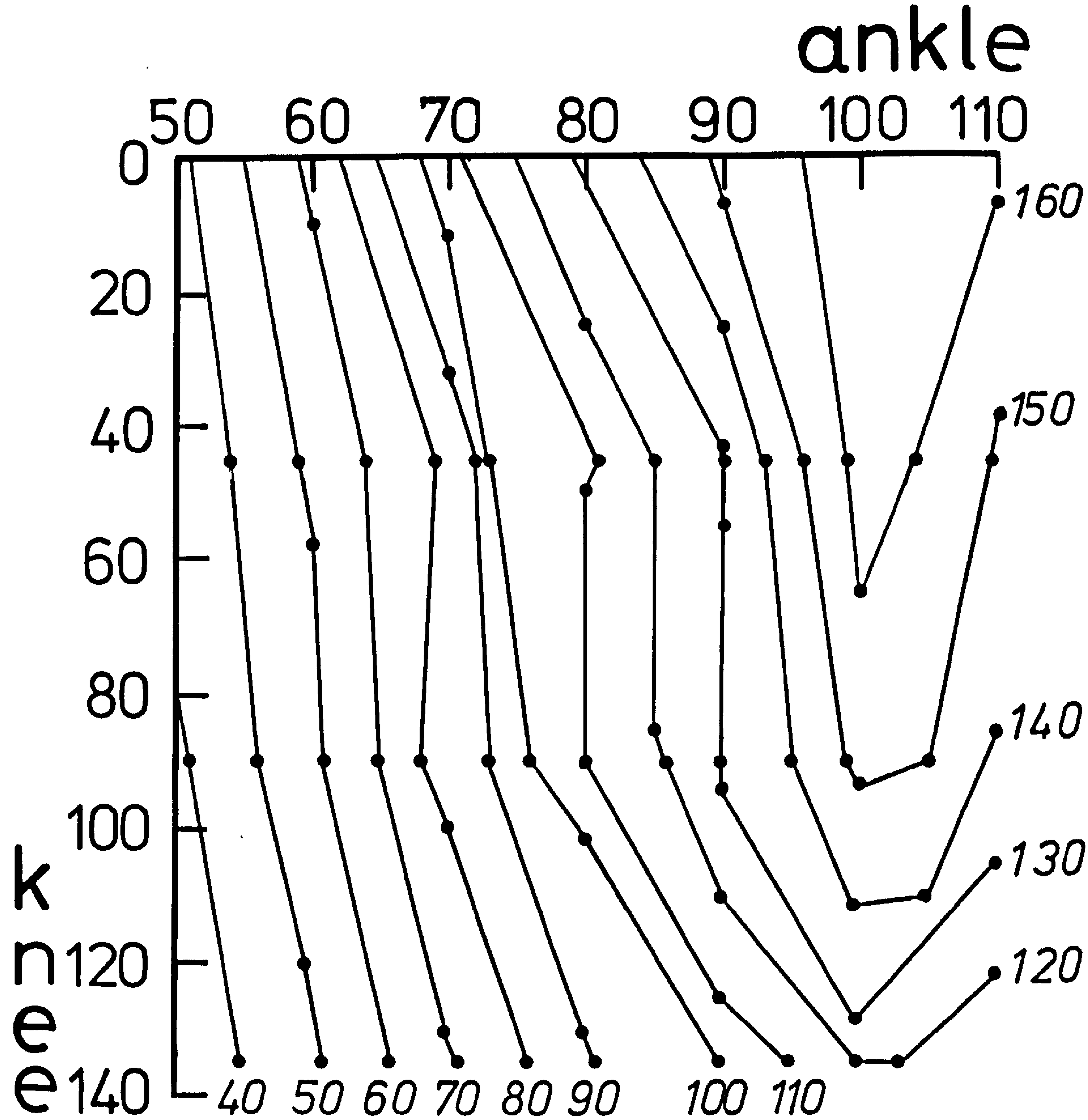
which is a combination of the two gastrocnemius regression equations in table 2,3.

Figure 2,11 is an ankle-knee diagram on which lines of constant strength are plotted. These were derived by taking the data from table 2,2 and plotting the normalised active torque in percentage as a function first of ankle angle and then of knee angle. Angular values corresponding to 10% intervals of torque were then read off from the chart, and the results were replotted in figure 2,11.



Lines of constant muscle length for the gastrocnemius plotted on an ankle-knee diagram.

Figure 2,11



Lines of constant plantarflexor torque (active normalised) plotted on an ankle-knee diagram.

2,3,3 THE LEVERAGE OF MUSCLES ACTING ABOUT THE ANKLE JOINT.

It is of considerable interest to discover the relative or absolute leverages of the various muscles acting about the ankle joint in the production of plantar-flexor torque and furthermore to determine whether these are constant or variable with ankle angle.

It is not possible to measure these quantities directly by taking the perpendicular distance from the centre of rotation of the joint to the line of pull of the muscles for the following reasons.

Barnett and Napier (1952) in an analysis of 152 specimens found that the true axis of the talo-crural joint shows a systematic shift in position in different parts of the joint's range. Isman and Inman (1968) however, concluded that shifts of axis if they existed were too small to be of practical importance. Dempster (1955) claimed that the axis of the joint shifted in an unpredictable manner which was idiosyncratic for each individual specimen he tested. Differences in conclusion probably reflect the fact that the three authors attempted to locate the axis by different techniques, but such differences do not bode well for the possibility of numerical studies. The precise lines of pull of the muscles are similarly elusive as most of the muscles concerned turn around "bony pulleys" the geometries of which are difficult to define. An indirect method for the calculation of these leverages will be demonstrated.

The torque ($\mathcal{T}$) that is exerted about a joint is equal to the product of the tension in the tendon (T) and the effective lever arm about that joint (R). The work done in moving that joint through an angular increment $\Delta\,\theta$ is given by $\mathcal{T}.\,\Delta\theta$, which (from the above) is equal to T. R. $\Delta\,\theta$. The work done may also be expressed as the product of the tension in the tendon (T) and the distance it moves (Δ L).

$$\therefore\ T.\ R.\ \theta = T.\,\Delta\,L.$$

$$\therefore\ R = \frac{\Delta L}{\Delta\theta}$$

Hence, at any given joint angle the lever arm of a muscle about a joint is defined as the rate of change of muscle length with joint angle

$$\therefore\ R = \frac{d\,L}{d\,\theta}$$

and this quantity may be readily evaluated as the first derivative of the equation which expresses muscle excursion as a function of joint angle. This quantity is not necessarily equivalent to any specific anatomical dimension and describes a simple functional property of the musculoskeletal system which is ultimately dependent upon many complicated structural properties.

For triceps surae acting about the ankle joint

$$\Delta L = -\,22.18468 + 0.30142\,\alpha - 0.00061\,\alpha^2$$

where ΔL is a percentage of shank length and

α is the ankle angle in degrees;

$$\frac{d\,L}{d\,\alpha} = 0.30142 - 0.00122\,\alpha$$

and making the necessary conversion to radians

$$\frac{d\,L}{d\,\alpha} = R = 17.27 - 0.07\,\alpha$$

where R is the effective leverage of triceps surae expressed as a percentage of shank length.

It must be noted that although the derived leverages are in all cases a simple function of ankle angle the true changes in leverage are possibly complex. This simplification is inherent in the statistical procedure used as higher orders of complexity are masked by the scatter in the data. The expressions for muscle leverage as a function of joint angle are the best that can be derived from the available data (see Appendix 2)

2,3,4 SOME SPECULATIONS CONCERNING MUSCLES OF PLANTARFLEXION.

Several authors have quoted values for the physiological cross sectional area of the muscles capable of producing plantarflexion of the ankle joint. Ignoring the implicit pitfalls of measurements of this nature (see section 2,3,2), it is possible to use such data to make certain deductions from the present measurements. Table 2,4 shows the data of three such sources together with an unweighted mean of all of them.

Table 2,4 PHYSIOLOGICAL CROSS SECTIONS AREAS (cm^2) OF PLANTARFLEXOR MUSCLES.

Muscle	Weber (1851)	Fick (1911)	Schumacher & Wolff (1966)	Unweighted mean
Gastrocnemius	57.3	23.0	16.9	32.4
Soleus	84.1	20.0	23.8	42.6
Tibialis Post.	16.9	5.8	5.8	9.5
Flexor Digitorum Longus	5.1	2.8	2.1	3.3
Flexor Hallucis Longus	13.4	4.5	4.1	7.3
Peroneus Longus	13.9	4.3	4.6	7.6
Peroneus Brevis	6.7	3.8	2.0	4.2

Although the absolute values of these sources vary widely we have no clear reason for rejecting any of them and the unweighted mean should give a reliable estimate. When the ankle angle is 90 degrees, then the lever arm is 11% of shank length. The mean shank length of the subjects in the angle-torque study was 414 mm which predicts a lever arm of 45 mm. This figure relates well to the most similar anthropometric measure taken; the horizontal distance from the back of the foot to the estimated centre of rotation of the ankle which had a mean value of 58 mm. (It is clear from the anatomy of the region that this latter dimension must exceed the effective lever arm).

Table 2,5 shows the calculated value of the effective leverage of all muscles acting about the talo-crural joint. Muscles of plantarflexion have positive values and muscles of dorsiflexion have negative values. The values are expressed as percentages of shank length, and in mm. calculated for a 414 mm shank.

Table 2,5 EFFECTIVE LEVER ARMS OF MUSCLES ACTING ABOUT THE ANKLE JOINT.

Muscle	%	m.m.
Triceps Surae		
(Max; $\alpha = 50$)	13.8	57
(Min; $\alpha = 110$)	9.6	40
Tibialis Posterior	2.5	10
Fl. Digitorum Longus	4.1	17
Flexor Hallucis Longus	6.0	25
Peroneus Longus	4.0	17
Peroneus Brevis	3.3	14
Tibialis Anterior	- 5.8	- 24
Extensor Digitorum Longus	- 5.7	- 24
Extensor Hallucis Longus	- 5.5	- 23
Peroneus Tertius	- 4.4	- 18

Multiplying the cross-sectional area by the moment arm gives a figure which may be considered as an estimate of the potential importance of a muscle in the generation of plantarflexor torques. Table 2,6 shows these muscles expressed as a percentage weighting.

Table 2,6 PERCENTAGE WEIGHTINGS OF MUSCLES OF PLANTARFLEXION.

Gastrocnemius	38.5	89.1%
Soleus	50.6	
Tibialis Posterior	2.0	10.9%
Flexor Digitorum Longus	1.2	
Flexor Hallucis Longus	3.8	
Peroneus Longus	2.7	
Peroneus Brevis	1.2	

The above figures were calculated for an ankle angle of 50 degrees. The deductions which can be made from them are limited as there is no available information concerning the length-tension curves of individual muscles. Consider the widely quoted hypothesis of Wheeler-Haines (1934) which was referred to in 1,2,3,1 (b). Wheeler-Haines considered that both gastrocnemius and soleus were "muscles of short action" and were incapable of exerting tension in their fully shortened position. The most extreme position measured in the angle-torque study was ankle 50°/Knee 135°, which is close to the limit of the range of both the ankle and knee joints, Dempster,(1955). Expressing the torque that each subject could exert in this postion as a percentage of the torque he could exert in his own strongest position,the mean torque in this position is 18.5% of maximum with a standard deviation of 8.0%

We are forced to conclude that either gastrocnemius or soleus or both are capable of exerting a modest tension in the fully plantarflexed position, since the deep flexors and peronei would be insufficinet to account for the measured torque: even on the unlikely assumption that the latter were at their optimal length in this position. There are, of course, considerable possibilities for error in these calculations, but on the basis of available data the hypothesis of Wheeler-Haines (1934) can be provisionally rejected, at least for triceps surae.

The greatest plantarflexor torque was measured when the ankle angle was 100 degrees. If we assume that all the muscles have a co-incident optimum of their length tension curve in this position, it is possible to derive a value for the coefficient of absolute of muscle strength (ρ) as defined in section 1,3,2. This is given by the equation:

$$\rho = \mathcal{T} / \Sigma\,(Ai.\ Ri)$$

where $\mathcal{T}$ is the voluntary torque measured,

Ai is the physiological c.s.a. of a given muscle,

and Ri is the effective lever arm of that muscle.

This gives a value of 2.41 $kgfcm^{-2}$ or in S.I. units 2.37 x 10^5 Nm^{-2}. This value is well within the range tabulated for ρ in table 1,1. It does not take into account the fact that the muscle fibres are pulling obliquely on the tendons, and hence a cosine effect should be introduced. Inspection of cadaveric material suggests that the mean angle between the fibres and tendon is unlikely to exceed 20 degrees and this would yield a value of

ρ = 2.56 $Kgf.cm^{-2}$ (2.52 x 10^5 Nm^{-2}). Such a value is of course dependent upon the choice of p.c.s.a. data used. If we choose to rely upon that of Schumacher and Wolff (1966) then we arrive at a figure of

ρ = 4.40 $Kgf.cm^{-2}$ (4.32 x 10^5 Nm^{-2}) which after correction for cosine effects gives a value of

ρ = 4.68 $Kgf.cm^{-2}$ (4.59 x 10^5 Nm^{-2}).

It is of some interest that the alternate calculations bracket those of Haxton (1944) who quoted a value of 3.9 Kgf. cm^{-2} for human calf muscles.

CHAPTER 3 - GRIPPING AND TURNING

3,1 INTRODUCTION

There are a large number of everyday activities in which an object must be gripped and turned against resistance. The actions of using a screwdriver, opening a screwtop jar and turning a door handle all fall within this category, as do a range of industrial assembly and control tasks. Considering the practical importance of the above activities, it is surprising that so few quantitative studies have been conducted into the forces which can be brought to bear in turning actions.

The task under consideration has at least two functional elements. A rotational torque is generated as a result of the activity of muscles acting about an axis running between the radio-ulnar articulations of the forearm, but this torque can only operate on the outside world through a complex intermediate link which requires stabilisation of the wrist joint, and of all the joints of the hand. It is therefore highly probable *a priori* that in many circumstances the latter intermediate link, henceforth referred to as 'the grip', is the weakest in the chain and prevents the full expression of the strength of the muscles which pronate or supinate the forearm. It is also highly probable that the strength of the grip may be modified by the nature of the object gripped. *Hand/handle interaction* will be discussed in some detail in the present chapter.

The strength of pronation and supination was studied in a number of postures of the elbow and shoulder by Darcus (1951) and Salter and Darcus (1952). These studies are certainly the most widely quoted angle torque curves in the literature. Darcus, however, used only three subjects in his experiments, and apparently did not consider the possibility that the handle used (in his case one shaped like a stirrup) might possibly modify the results in the ways suggested in the previous paragraph. The first part of this chapter deals with experiments which firstly repeat and verify Darcus' results on a larger number of subjects and secondly investigate the possibility of an interaction between forearm posture and handle dimensions.

Many investigators have studied the strength of grip, using a range of measuring devices which respond to the forces or their components with which an object may be squeezed between fingers and palm. A useful table of these results is to be found in Damon, Stoudt and McFarland (1966). Of particular interest in this respect are the studies of Hertzberg (1955), Ayoub and Lo Presti (1971) and O'Neill (1974) which quote data for the relationship between strength of squeeze and the separation of the two active elements of a force measuring device (i.e. that engaged by the fingers and that engaged by the palm). The results of these three studies have been compiled, and are plotted on a percentage basis in Fig. 3,1.

Figure 3,1

Results from the published literature relating grip strength to handle size.

Published accounts of the measuring equipment show them to have been similar (although Ayoub and Lo Presti's device is poorly described, the strength measurements being only a small part of an electromyographic study) and so it is difficult to account for the substantial discrepancies found. Bechtol (1954) measured grip strength using an adjustable hand dynomometer and found that a modest difference existed between the handle sizes which allowed the greatest forces to be exerted by male and female subjects respectively. The males showed a slight tendency to be stronger at two inches than at 1.5 and the females vice versa. This difference was attributed to hand size being greater in the males.

The type of squeezing action measured in such tests is artificial and only reflects one facet of the infinite range of activities of which the human hand is capable.

Various attempts have been made to subdivide or classify the prehensile activities of the human hand. Schlesinger (1919) identified six major classes of grasp:- cylindrical, spherical, hook, fingertip, "palmar" (between the pads of the thumb and finger) and "lateral" (between the pad of the thumb and the lateral surface of the index finger). Taylor (1954) quotes figures for the strength of the last three of these actions in addition to a grip strength of the more common kind mentioned above.

A more effective solution to the problem of the classification of prehensile activities is that proposed by Napier (1956) and elaborated by Landsmeer (1962). Napier pointed out that it is the "nature of the intended activity that finally influences the pattern of the grip" and that prehensile activities could be divided into two main categories.

"In *precision grip* the object is held as in a clamp between the flexor aspects of the fingers and that of the opposing thumb.

In *power grip* the object is held as in a clamp between the flexed fingers and the palm, counter pressure being applied by the thumb lying more or less in the plane of the palm".

Long *et al* (1970) reported a detailed electromyographic study of the function of various intrinsic and extrinsic muscles of the hand in power gripping and precision handling. Napier's study was entirely qualitative. Sharpe (1962) measured the efficacy of the precision grip during the exertion of supinator torques on knobs of various diameters. The grip strength studies cited above test one aspect of the power grip. Napier used cylinders of varying diameters to illustrate the power grip; it is this type of grip which is applied when using a screwdriver.

The second part of this chapter deals with the exertions of supinator torque on cylindrical handles and in a final experiment, these are compared with the exertions of torque on a range of commercially available screwdrivers.

3.2 EXPERIMENT TO INVESTIGATE THE EFFECTS OF FOREARM POSTURE ON THE STRENGTH OF PRONATION AND SUPINATION.

3,2,1 METHODS.

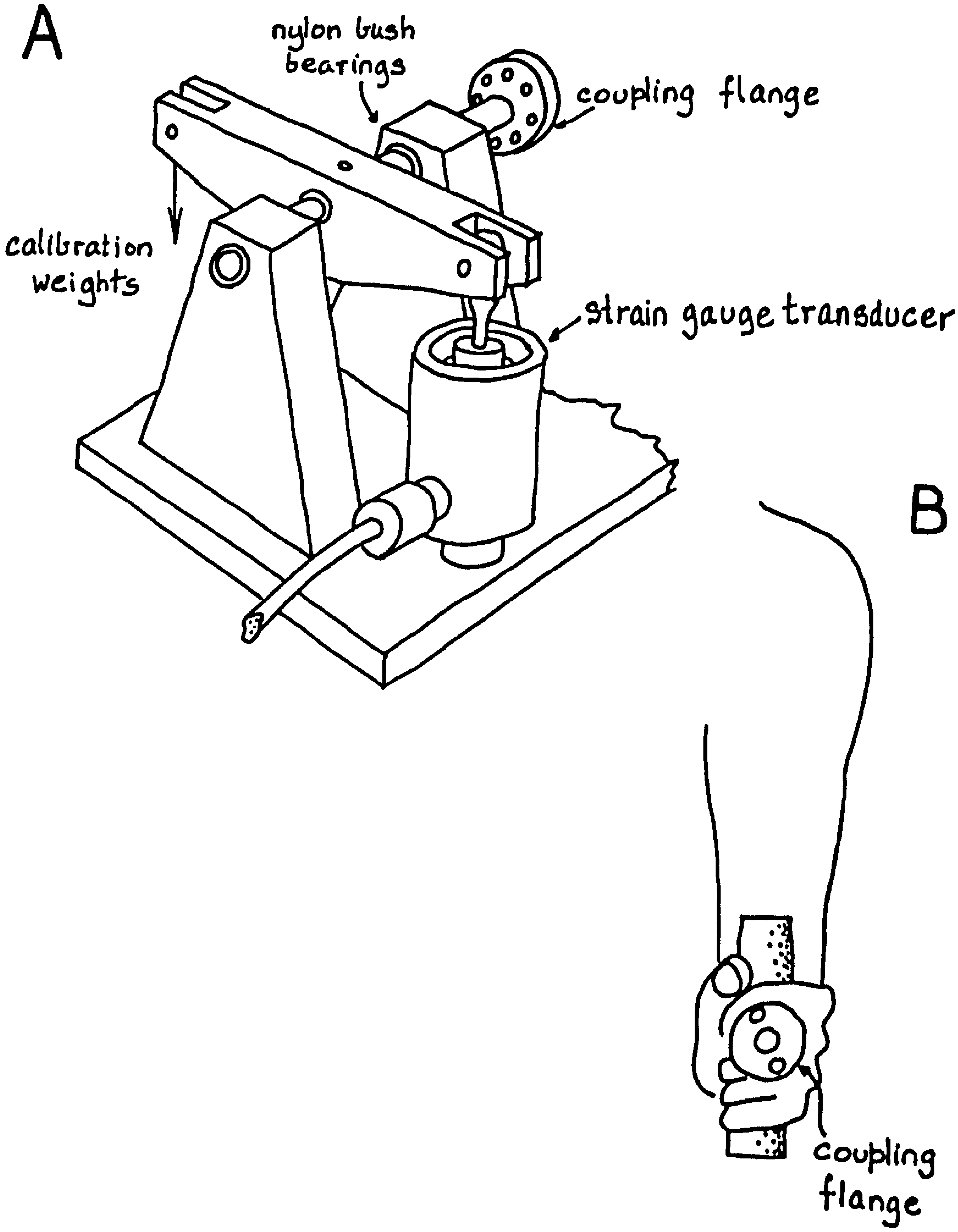
a) Apparatus.

A test rig was constructed for the measurements of isometric torques of pronation and supination. The design permitted rapid changes in handle type and orientation. The handles investigated were stirrup type (3,2), T - bars (3,3), Steel cylinders (3,4) and commercially available screwdrivers (3,5).

The rig is shown in figure 3,2. A central pivot was mounted in low-friction, P.T.F.E. bush bearings. A cross beam was set on this pivot which was coupled to a load cell (Pye Ether, type U.F. 2.) which was sensitive in both tension and compression. The load cell output was amplified, recorded and displayed as described elsewhere (Appendix I). The opposite end of the cross beam had facilities for the attachment of calibration weights. A flanged plate mounted on the end of the spindle allowed a handle to be firmly locked to the measuring rig in positions arranged around a circle in 45 degree intervals. Fig. 3,3 is a multiple exposed photograph of a T-bar mounted in these positions.

The rig was mounted on a solid oak base, which in turn was mounted on a scaffolding support such that the position of the rig could be readily adjusted to accommodate subjects of varying height.

Figure 3,2



(A) Sketch of experimental apparatus for measurement of pronation and supination torques.

(B) Anterior view of subject grasping T-bar handle; to show flange for coupling handle to apparatus.

Photograph of experimental apparatus; multiple exposure to show
'-bar in all orientations tested.

A pilot study using this apparatus had shown that the load cell was not sensitive to forces of translation applied to the apparatus and recorded torque alone (O'Neill, 1974).

b) <u>Sign Convention</u>.

When the arm is vertical, the pronated forearm is horizontal and the hand grips a horizontal rod perpendicular to the long axis of the forearm, the posture of the forearm is said to be 0 degrees. Supination then leads to an increase of angle so that when the rod is vertical the angle is 90 degrees and when it is again horizontal the angle is 180 degrees.

When the subject attempts to supinate against resistance, he exerts a positive torque. When he attempts to pronate against resistance he exerts a negative torque.

c) <u>Subjects</u>.

Eight fit young adults took part in the experiment. Five men and three women participated, but sex differences are again considered irrelevant to the experiment and the data has been pooled.

d) <u>Procedure</u>.

All eight subjects were in the laboratory at the same time. They performed in one posture at a time with a bias-free order of presentation. The rig was adjusted before each effort so that the subject could stand facing and grasp the handle with his forearm horizontal and his arm vertical. Standard protocol was observed (Appendix I). Each subject was able to rest for several minutes while the other seven subjects performed.

3,2,2 RESULTS.

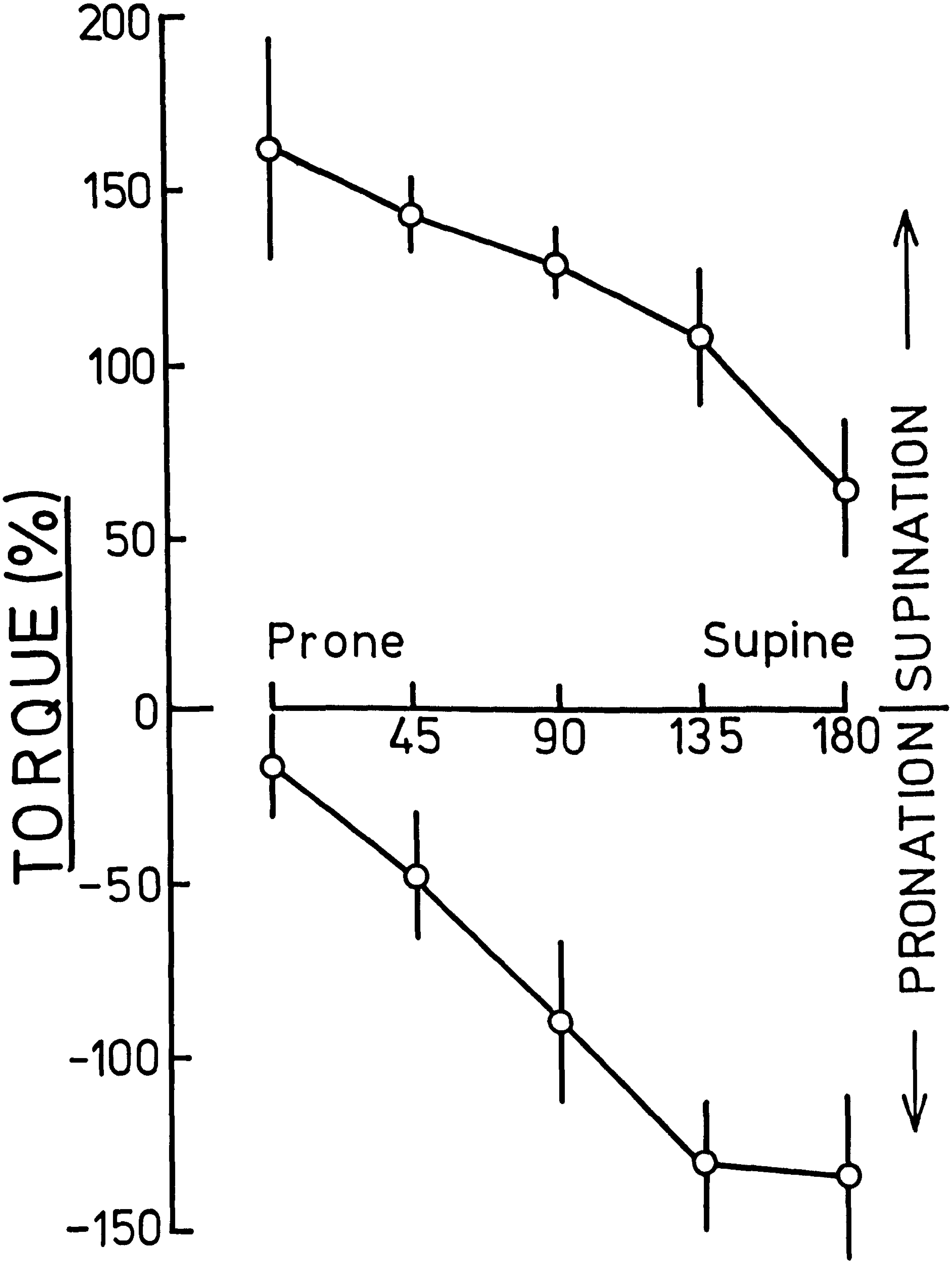
Results of experiment 3,2 are tabulated in table 3,1. The data was normalised with respect to the mean of all performances (pronation and supination efforts being taken together regardless of sign). Normalised data is plotted in figure 3,4 (mean $\pm$ 1 s.d.) and absolute data is plotted in figure 3,5 (a). (mean $\pm$ 1 s.e.m.)

TABLE 3,1.

Position	Supination Efforts				Pronation Efforts			
	Abs.(Nm)		%		Abs.(Nm)		%	
	mean	s.d.	mean	s.d.	mean	s.d.	mean	s.d.
Supine(O)	3.0	0.8	63	20	-7.0	3.0	-132	23
45^{o}	5.2	1.4	106	19	-6.5	2.5	-129	18
90^{o}	6.5	2.2	127	10	-4.5	2.0	- 88	23
135^{o}	7.3	2.6	141	10	-2.6	1.4	- 47	18
Prone (180^{o})	8.3	3.4	161	32	-0.9	0.9	- 16	15

The data was tested using the Student's t test as described in Appendix (2) Levels of significance of differences between all possible pairs of conditions are shown in Fig. 3,6.

Figure 3,4



Results of experiment 3,2. Normalised torques of supination and pronation (mean ± 1 s.d.) plotted against the posture of the forearm.

Figure 3,5

Results of Experiments 3,2 and 3,3.

A. Pronation and supination torque (mean $\pm$ 1 s.e.m.) plotted against forearm posture.

3,2,2

B. Pronation and supination torque (mean $\pm$ 1 s.e.m.) plotted against forearm posture and handle size.

3,3,2

Figure 3,5

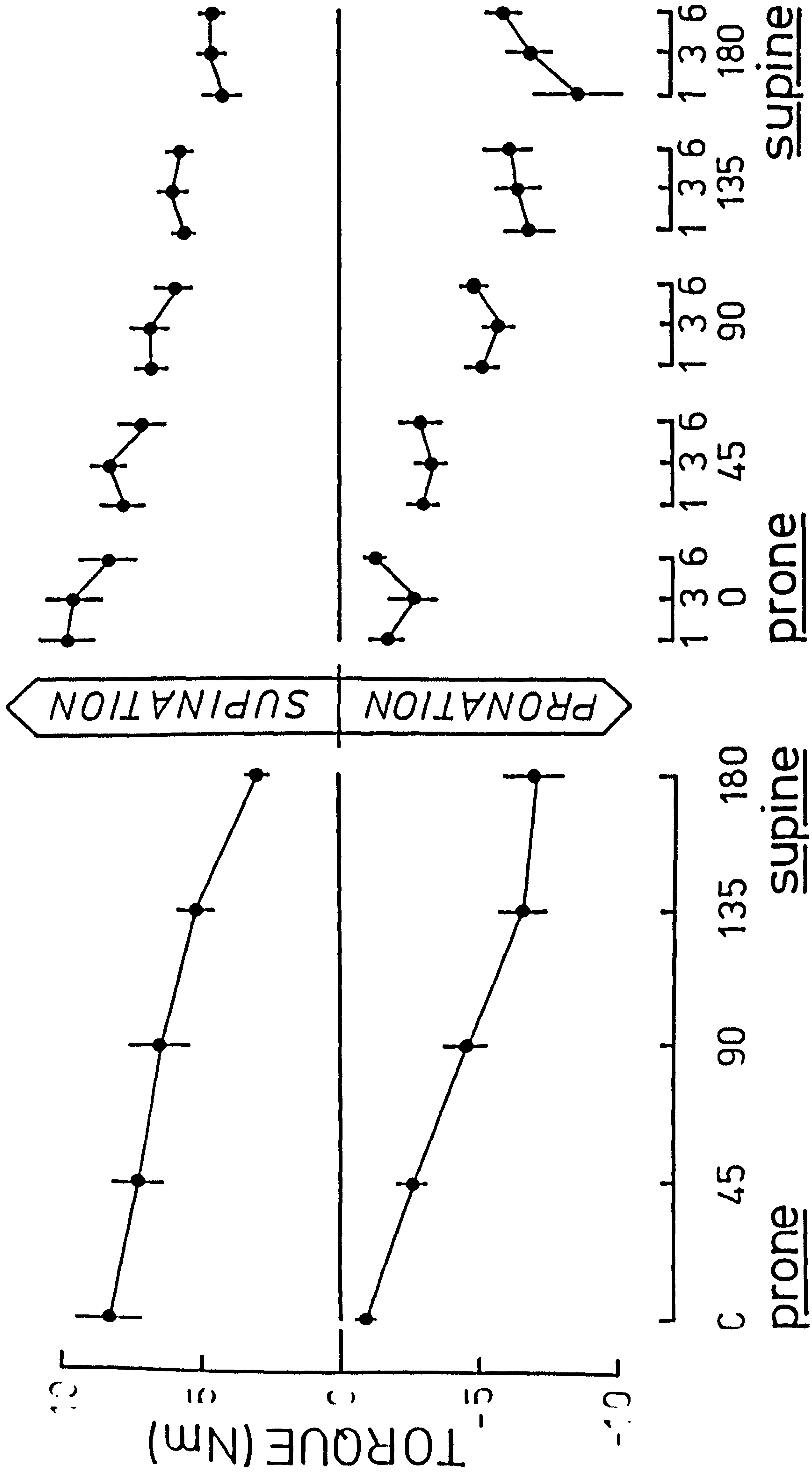
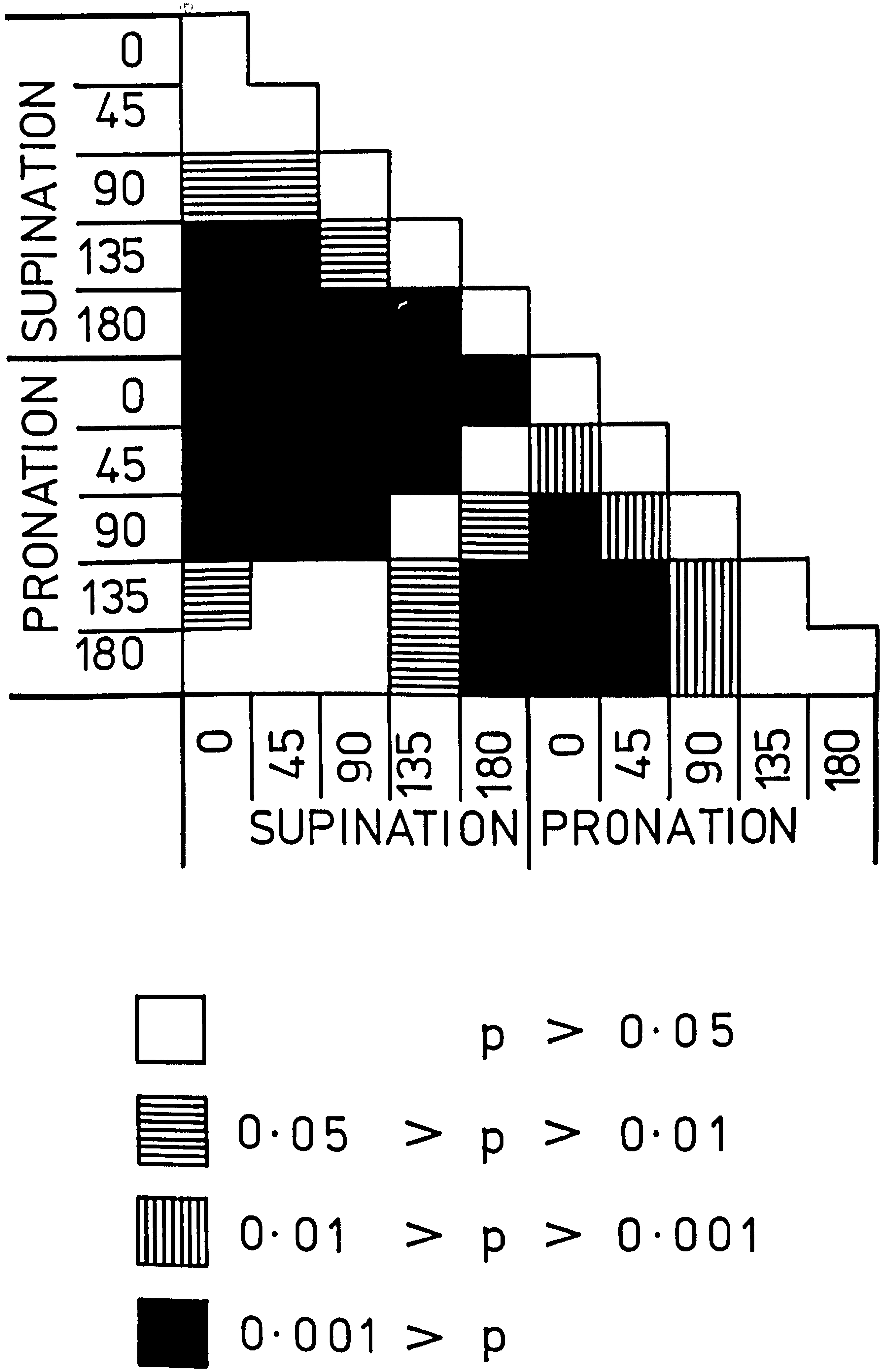

TORQUE (Nm)
10
0
-5
SUPINATION
PRONATION
prone
supine
0
45
90
135
180
1 3 6

Figure 3,6

Levels of significance for t - tests of differences between strength (normalised torque) of pronation and supination in all combinations of forearm posture. The sign convention for torque has been ignored in the analysis of this data.

3,2,2


SUPINATION
0
45
90
135
180
PRONATION
0
45
90
135
180
0
45
90
135
180
0
45
90
135
180
SUPINATION
PRONATION
p > 0·05
0·05 > p > 0·01
0·01 > p > 0·001
0·001 > p

3,3 EXPERIMENT TO INVESTIGATE THE EFFECTS OF FOREARM POSTURE AND THE SIZE OF T-BAR HANDLES ON THE STRENGTH OF PRONATION AND SUPINATION.

Darcus (1951) did not consider the possibility that, under some conditions at least, the capacity of the hand to grasp the handle might limit the forearm torque which could be transmitted to the measuring device. One method of investigating this question is to change the nature of the object gripped. In order to preserve the nature of the experiments of Darcus (1951), the hand must grip a rod whose axis can be presented at any orientation in a vertical plane perpendicular to the axis of the forearm. As the published literature suggests that the strength of grip is greatly influenced by the size of the object gripped, it is reasonable to vary the diameter of the part of the handle which is gripped by the subject.

3,3,1 METHODS.

Three T-bar handles were constructed. The cross bars were fabricated out of steel rod (or tube) of circular cross-section 10, 30, and 60 mm. in diameter respectively. The apparatus and procedure were identical to those described in experiment 3,2. The T-bar was gripped so that its shaft emerged between the middle and ring fingers. The axis of its action was therefore as far as possible consistent with the axis of pronation-supination. In each trial (i.e. for each orientation of the handle) the subject performed with each of the three T-bars, the order of T-bar use being itself randomised. Eight fit young subjects took part in the experiment (5 male, 3 female).

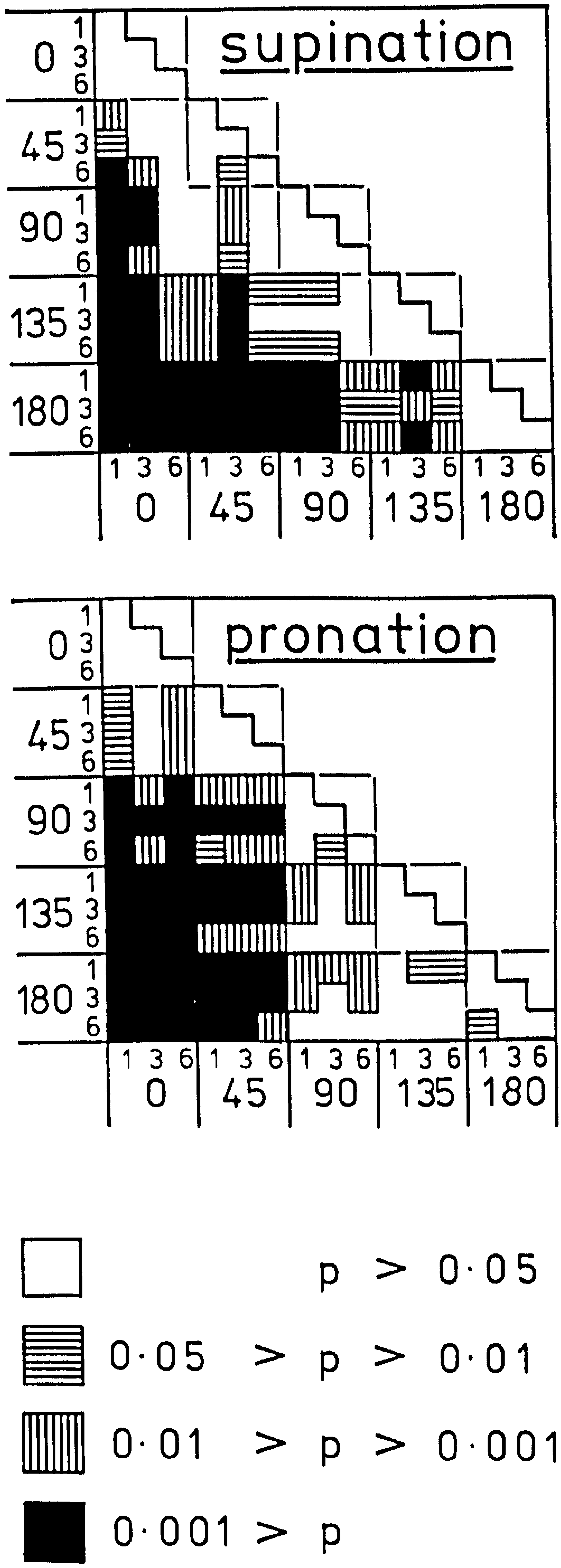
3,3,2. RESULTS.

The results of the experiment are shown in table 3,2. As the objective on this occasion is principally to test for differences between handle sizes, the data for pronation and supination efforts have been normalised separately about their own subject means. The results of t-tests between the data, normalised in this way, for all possible pairs of conditions are shown in fig. 3,7. The absolute results (Nm, mean ± 1 s.e.m.) are plotted in fig. 3,5 (b).

Figure 3,7

Levels of significance for t-tests of differences between strength (normalised torque) of pronation and supination using T-bar handles: all combinations of forearm posture and handle size.

3,3,2


supination
0
45
90
135
180
1 3 6
pronation
p > 0·05
0·05 > p > 0·01
0·01 > p > 0·001
0·001 > p

-104-

TABLE 3,2 RESULTS OF EXPERIMENT No. 3,3

Supination Efforts

ANGLE	HANDLE	ABSOLUTE TORQUE (Nm) (MEAN)	(S.D.)	% TORQUE (MEAN)	(S.D.)
0	S	9.9	2.5	143.4	17.9
	M	9.7	2.7	140.3	22.8
	L	8.4	2.8	119.9	26.8
45	S	7.9	2.3	112.6	16.7
	M	8.4	1.8	122.4	14.8
	L	7.2	2.3	102.8	17.8
90	S	6.9	1.7	98.8	12.5
	M	6.9	1.9	98.1	10.5
	L	6.0	2.1	96.4	22.6
135	S	5.7	1.0	83.3	11.8
	M	6.1	1.5	87.9	10.1
	L	5.8	1.3	85.3	11.8
180	S	4.2	0.9	62.9	14.2
	M	4.6	1.2	67.1	16.8
	L	4.6	1.2	66.0	9.4

Pronation Efforts

ANGLE	HANDLE	ABSOLUTE TORQUE (Nm) (MEAN)	(S.D.)	% TORQUE (MEAN)	(S.D.)
0	S	- 1.7	1.6	34.8	25.2
	M	- 2.7	2.5	50.6	39.2
	L	- 1.3	0.7	29.9	15.0
45	S	- 3.0	1.4	65.3	28.8
	M	- 3.3	1.8	65.5	23.9
	L	- 3.0	1.9	64.0	28.2
90	S	- 5.1	1.8	104.9	13.8
	M	- 5.7	1.7	120.5	21.6
	L	- 4.8	1.5	98.8	17.6
135	S	- 6.9	2.5	142.3	26.1
	M	- 6.4	2.2	130.4	19.1
	L	- 6.1	2.3	124.3	35.3
180	S	- 8.6	4.5	172.9	50.2
	M	- 6.8	2.2	140.0	23.3
	L	- 5.9	2.0	123.5	30.1

3.4 EXPERIMENT TO DETERMINE THE SUPINATOR TORQUES WHICH MAY BE EXERTED USING CYLINDRICAL HANDLES OF VARIOUS DIAMETERS.

A thorough search of the literature in ergonomics revealed no studies in which factors contributing to the effectiveness of handles such as screwdrivers, had been systematically investigated. In such an investigation the first factor to study must be that of handle size, and in order to accomplish this the differences between handles must be eliminated. Cylinders present an obvious starting point, as they combine geometric simplicity, ease of uniform construction and similarity to a range of practical devices.

3.4.1 METHODS.

a) Apparatus.

The measuring rig used was that described in 3,2,1. The handles tested were of 10, 20, 30, 40, 50, 60 and 70 mm. in diameter, and each could be rapidly locked to the central shaft of the measuring rig. The handles were constructed from mild steel tube. The steel tube was initially turned down to size on a lathe and the end portions were brazed in place. The handle was then carefully polished, again using the lathe until the surface finish of the handles was as uniformly smooth as available workshop technology allowed.

b) Subjects.

Twenty four right-handed medical students (10 male, 14 female) took part in this experiment.

c) Procedure.

Procedure in this experiment differed from the previous experiments in certain respects. Since it was believed that the posture of the shoulder and elbow would be less critical for these measurements, the apparatus was set up so that the spindle was always 1 M. from the floor. The subjects took part in the experiment one at a time. Each subject was shown the handles and apparatus and was instructed to grip the handle firmly and make a maximal forceful, steady effort as if tightening a screw. The subjects washed their hands before the experiment and degreased them with an alcohol-soaked rag between each trial. Each subject performed with each handle in turn in a bias-free order of presentation.

After the completion of the experiment, two simple anthropometric measurements were made on the right hands of the subjects.

(i) Hand length - the distance from the proximal skin crease of the wrist to the tip of the middle finger.

(ii) Hand spread - the overall maximum distance between the extreme margins of the thumb and little finger when the hand was placed palm downwards, fully spread on a flat surface.

d) <u>Measurements</u>.

A pilot study of this activity (O'Neill, 1974) showed that the subject's hand inevitably begins to slip around the smooth cylindrical handle and that the point of initiation of slip may be accurately identified on a paper record by the following criteria.

(i) An abrupt decrease in the rate of change of the measured torque,

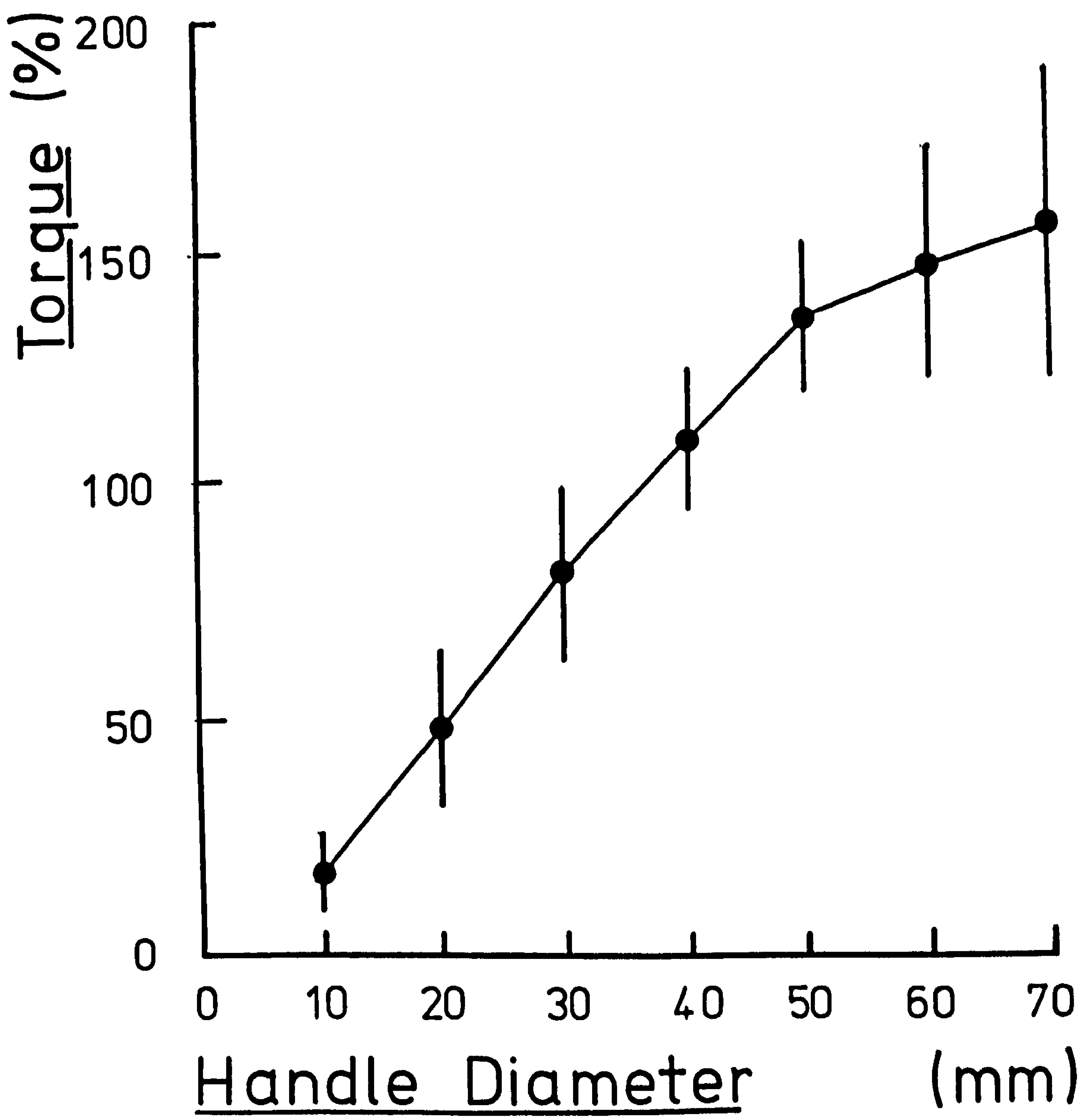
(ii) the onset of oscillations in the torque record as the hand commences to "judder" round the handle. This event was clearly identifiable on all of the force records, and the torque which existed at the onset of slip was recorded. The mechanical conditions may therefore be considered consistent and reproducible.

3.4.2 RESULTS.

An elementary consideration of the mechanics of this test shows that one factor limiting the torque which can be applied is the coefficient of friction between the hand and the handle, (see section 3,6,2). This is an arbitrary factor determined by the precise smoothness of the handle and by the condition of the subject's palm at the time when the measurement was made. Absolute values therefore have no permanent relevance in these conditions. Each result for each subject was normalised against the mean result for that subject using all seven handles. These results are plotted in fig 3,8 (mean $\pm$ 1 s.d.) and tabulated in table 3,3. The grand mean (i.e. 100%) corresponds to a torque of 4.4 Nm.

TABLE 3,3 RESULTS OF EXPERIMENT 3,4.

Handle Diameter (mm)	Torque (% mean)	
	mean	s.d.
10	18	8
20	48	16
30	81	19
40	110	15
50	137	16
60	147	24
70	157	33



Results of Experiment 3,4 Normalised supinator torque (mean ± 1 s.d.) exerted on smooth steel cylinders of different diameters.

3.5 EXPERIMENT TO COMPARE THE SUPINATOR TORQUES WHICH MAY BE APPLIED USING A VARIETY OF HANDLES.

Having shown that supinator torque varies in a systematic way with handle size, it was thought appropriate to consider the influence of other properties of the handle upon performance. A series of handles were therefore tested which fell into families within each of which a range of sizes was represented.

3.5.1 METHODS.

a) Handles tested.

The following handles were tested in the present experiment:

(i) The seven smooth steel cylinders used for experiment 3,4.

(ii) Four similar cylinders which had been knurled to a rough surface. The dimensions of these were 10, 30, 50 and 70 mm.

(iii) The T-bar of 30 mm. handle diameter used in experiment 3,3, this handle was set in the 0 degrees (i.e. optimal) position.

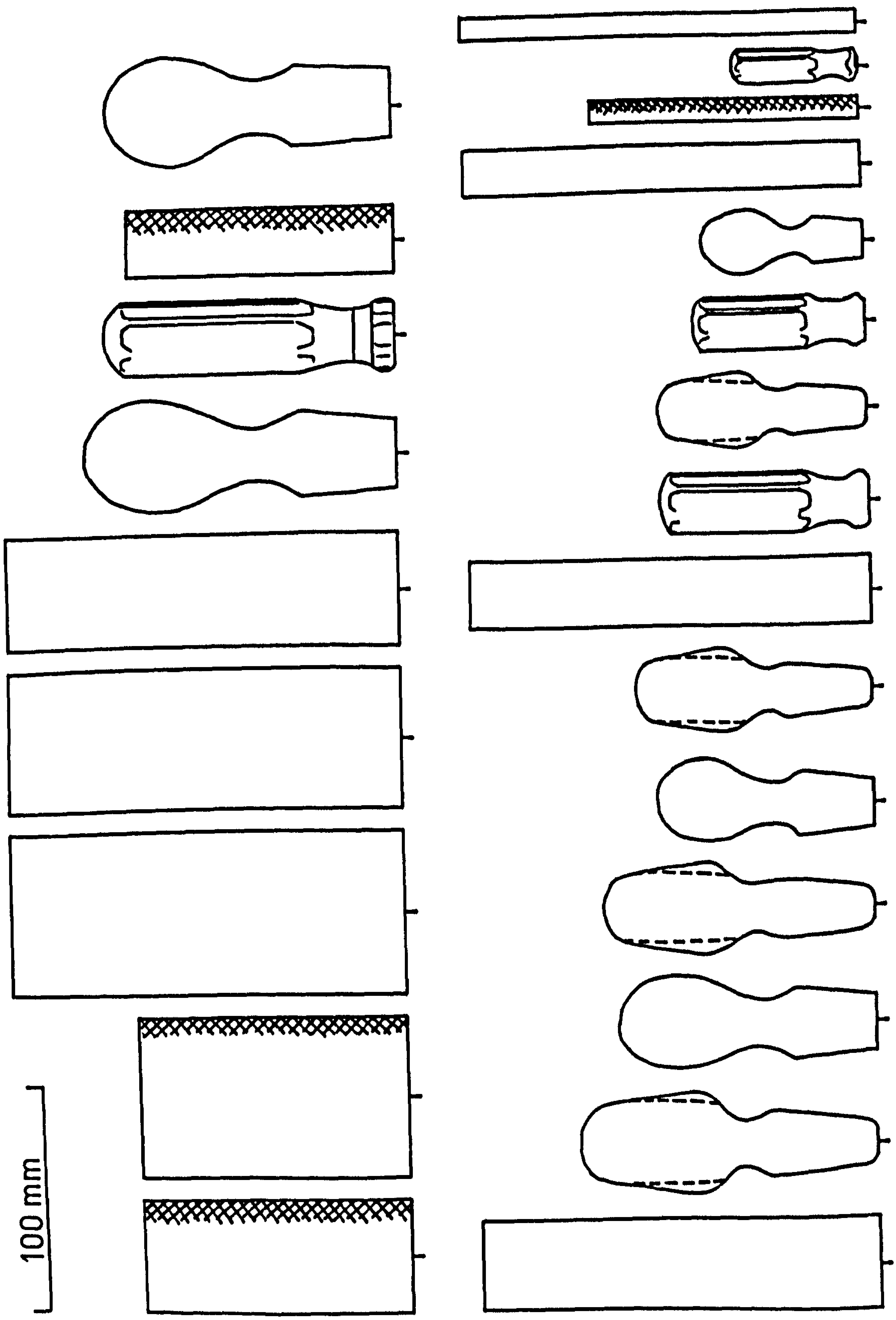
(iv) A range of commercially available screwdrivers of the three following types: Traditional "cabinet makers' turnscrews" with ovoid handles of polished beechwood; plastic handled "engineers" screwdrivers which had six flutes out onto a basically cylindrical handle; a design of screwdriver handle which had the cross-section of a truncated triangle in both axes of symmetry and came to be known as the "lozenge" handle.

The handles are illustrated in fig. 3,9.

In order to compare handles from different families, a size factor must be attributed to the non-cylindrical handles. It is possible to obtain a factor in several ways, e.g. based upon maximum or minimum diameter. The method chosen was to establish a mean diameter which

Figure 3,9

Figure 3,9



Scale drawing of handles used in Experiment 3,5. Handles are arranged in descending order of torque which was exerted in the experiment.

was defined as the diameter of a cylinder having equal volume and length to the handle in question. The handle was immersed in a measuring cylinder of water as far as the commencement of its metal shafts, and its volume was established by displacement. The mean diameter was calculated from the equation:

$$D = 2\sqrt{\frac{V}{\pi L}}$$

Where D is the mean diameter,

V is the volume of the handle

and L is the length of the handle.

The widened and flattened blades of the screwdrivers were cut off. This allowed each shaft in turn to be gripped in a three jaw chuck. The chuck was mounted on a flanged plate and locked onto the shaft of the measuring rig as described above.

b) <u>Subjects</u>.

Fifteen fit young adults (9 male, 6 female) acted as subjects in the experiment.

c) <u>Procedure</u>.

The subjects' hands slipped on some handles but not on others. Therefore, the performance criteria defined in 3,4,1 (d) were not appropriate and a standard maximal steady measurement was taken. Subjects performed first on the seven smooth cylinders then on the remainder of the handles in a randomised order.

3,5,2 RESULTS.

The results of this experiment were normalised with respect to the mean value for each subject's performances on the seven smooth cylinders. The grand mean for all subjects was 4.6 Nm. - a figure which is close to the comparable measure in experiment 3,4 - in spite of the fact that the subjects taking part in the experiment were different and that a slightly different performance criterion was used. The normalised results of the experiment are shown in Table 3,4 and in figure 3,10.

Student's t-test was performed on all possible pairs of handles and the results are shown in figure 3,11.

Figure 3,10

Results of experiment 3,5. Normalised supinator torque (mean $\pm$ 1 s.e.m.) using a variety of handles plotted against the mean handle diameter. Broken lines across top of chart indicate mean $\pm$ 1 s.e.m. of performance using T-bars in their optimum orientation.

3,5,2

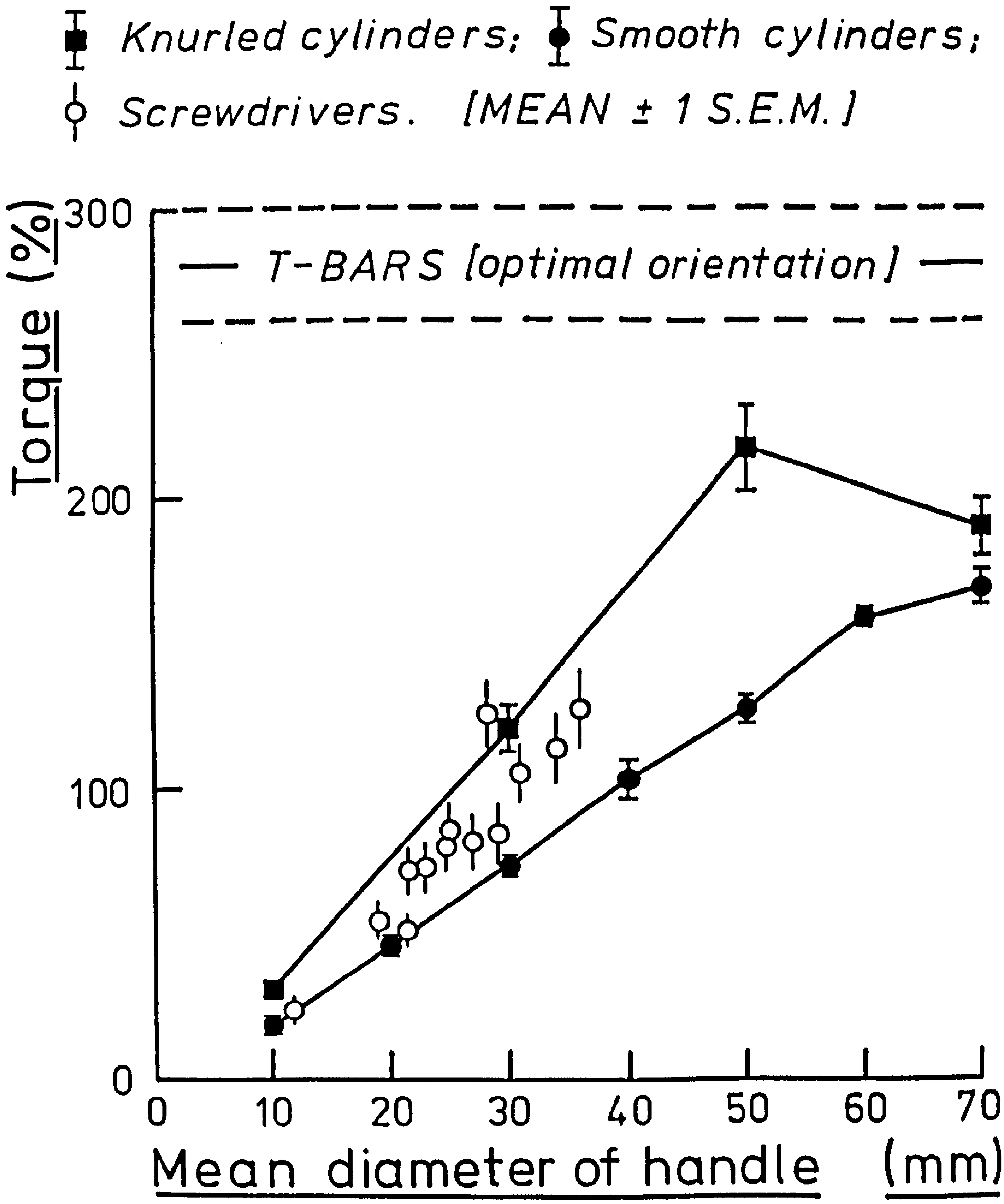

Knurled cylinders; Smooth cylinders;
Screwdrivers. [MEAN ± 1 S.E.M.]
T-BARS [optimal orientation]
Torque (%)
300
200
100
0
0
10
20
30
40
50
60
70
Mean diameter of handle (mm)

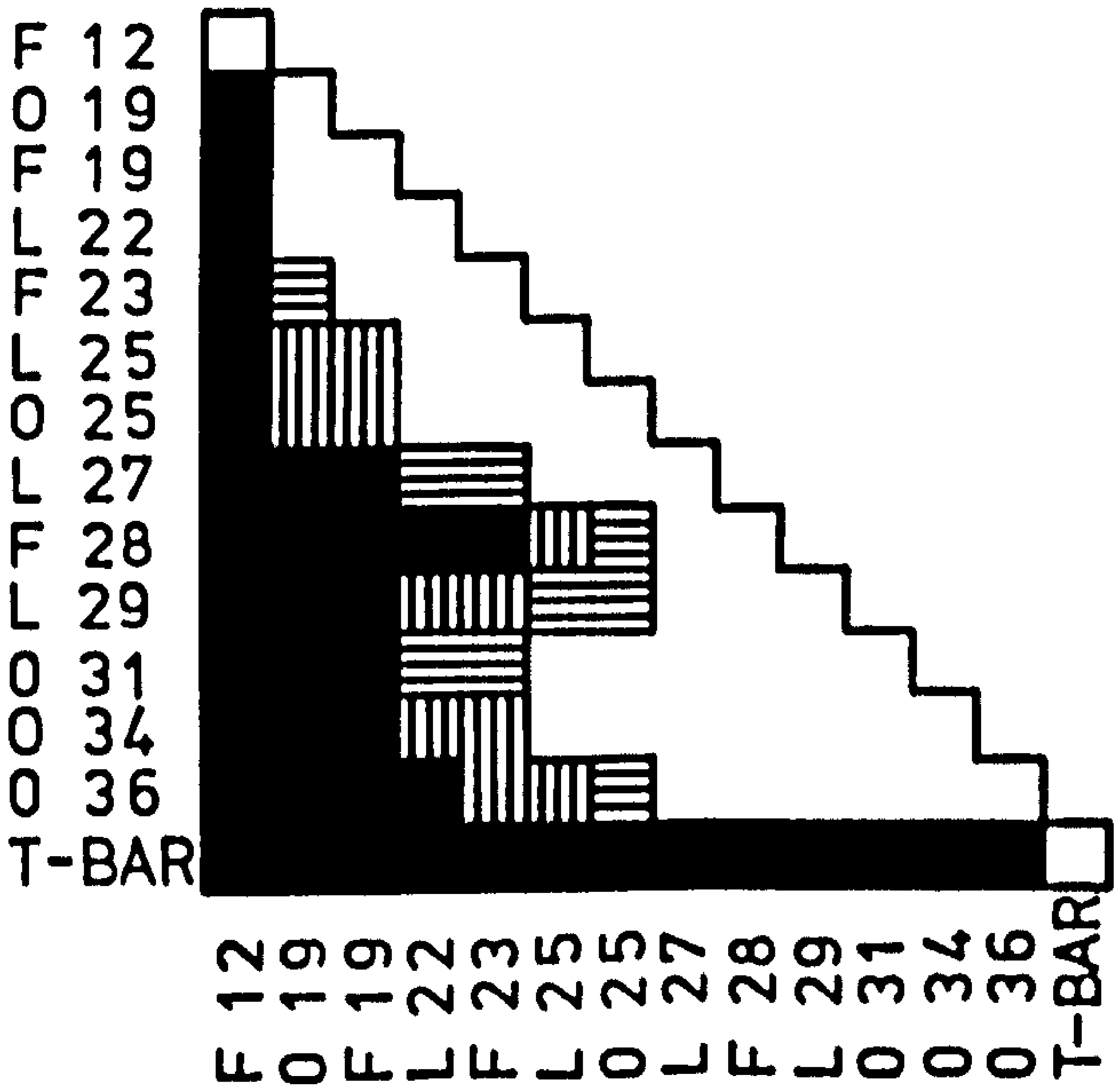
Figure 3,11

Levels of significance in t-tests for differences in strength of supination using a variety of handles.

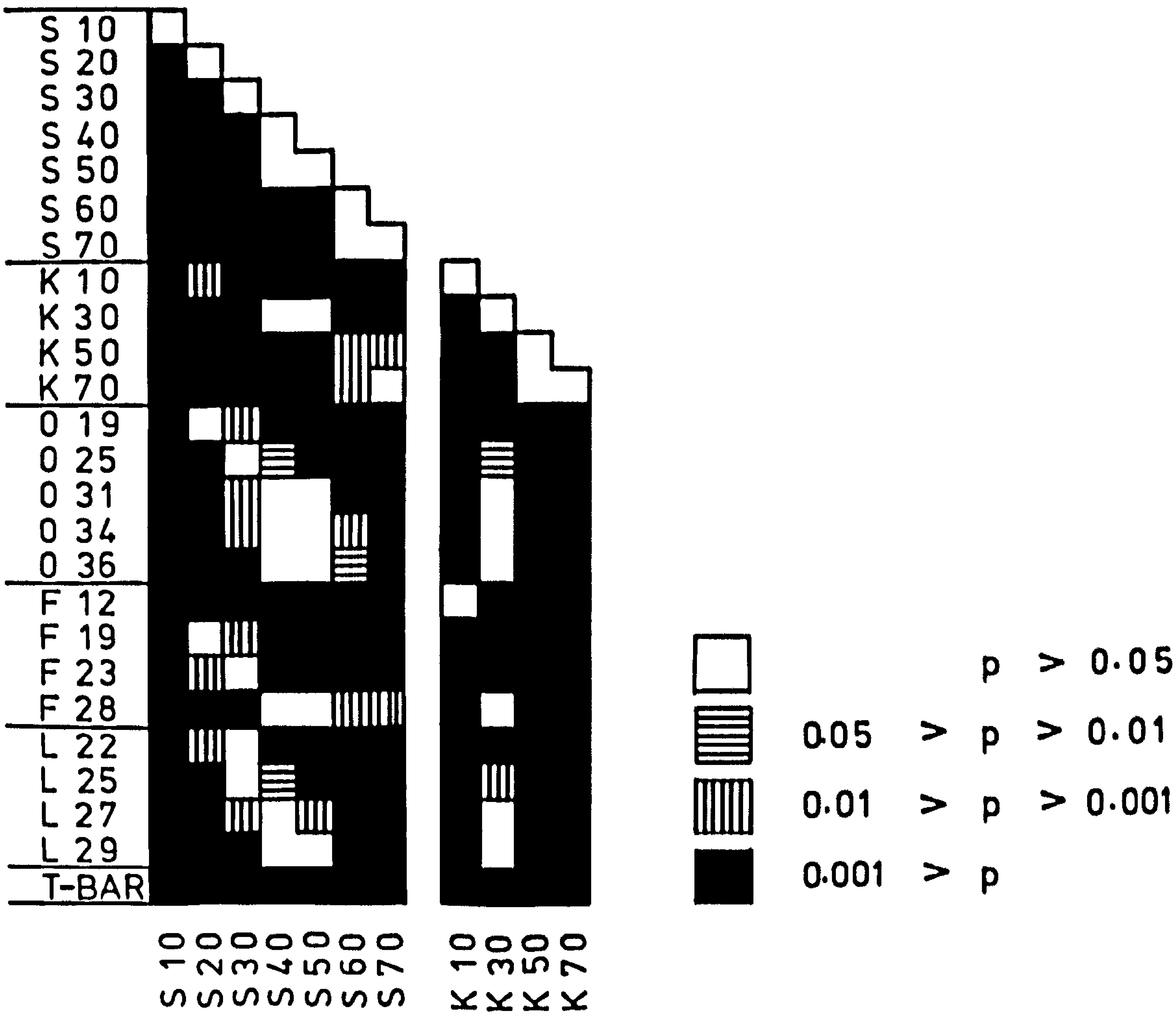
S	-	smooth cylinders
K	-	knurled cylinders
O	-	ovoid screwdrivers
F	-	fluted screwdrivers
L	-	lozenge screwdrivers

Mean diameters of the handles are indicated.

Figure 3,11



F 12
O 19
F 19
L 22
F 23
L 25
O 25
L 27
F 28
L 29
O 31
O 34
O 36
T-BAR



S 10
S 20
S 30
S 40
S 50
S 60
S 70
K 10
K 30
K 50
K 70
O 19
O 25
O 31
O 34
O 36
F 12
F 19
F 23
F 28
L 22
L 25
L 27
L 29
T-BAR
p > 0.05
0.05 > p > 0.01
0.01 > p > 0.001
0.001 > p

TABLE 3,4 SUPINATOR TORQUES WHICH MAY BE APPLIED USING VARIOUS HANDLES.

TYPE	MEAN DIAMETER (mm.)	TORQUE (%) Mean	S. D.
Smooth Cylinders	10	18.2	4.9
	20	45.8	11.5
	30	74.0	11.2
	40	113.6	22.6
	50	128.2	15.7
	60	158.9	13.3
	70	169.5	22.5
Knurled Cylinders	10	31.2	8.9
	30	121.2	32.5
	50	218.2	57.6
	70	191.2	39.0
Ovoid Screwdrivers	19	52.3	19.7
	25	86.8	31.0
	31	106.3	35.8
	34	114.2	42.4
	36	127.6	45.1
Fluted Screwdrivers	12	24.3	13.7
	19	54.9	20.1
	23	73.8	29.1
	28	125.5	40.8
Lozenge Screwdrivers	22	72.4	27.6
	25	84.8	30.0
	27	100.5	32.2
	29	113.4	34.6
T-BAR	-	281.0	71.6

3,6 A STUDY OF HAND-HANDLE CONTACT PATTERNS

In order to further investigate the underlying bases of the above results, a study was made of the way in which the hand engaged cylinders of different diameters.

3,6,1 METHODS.

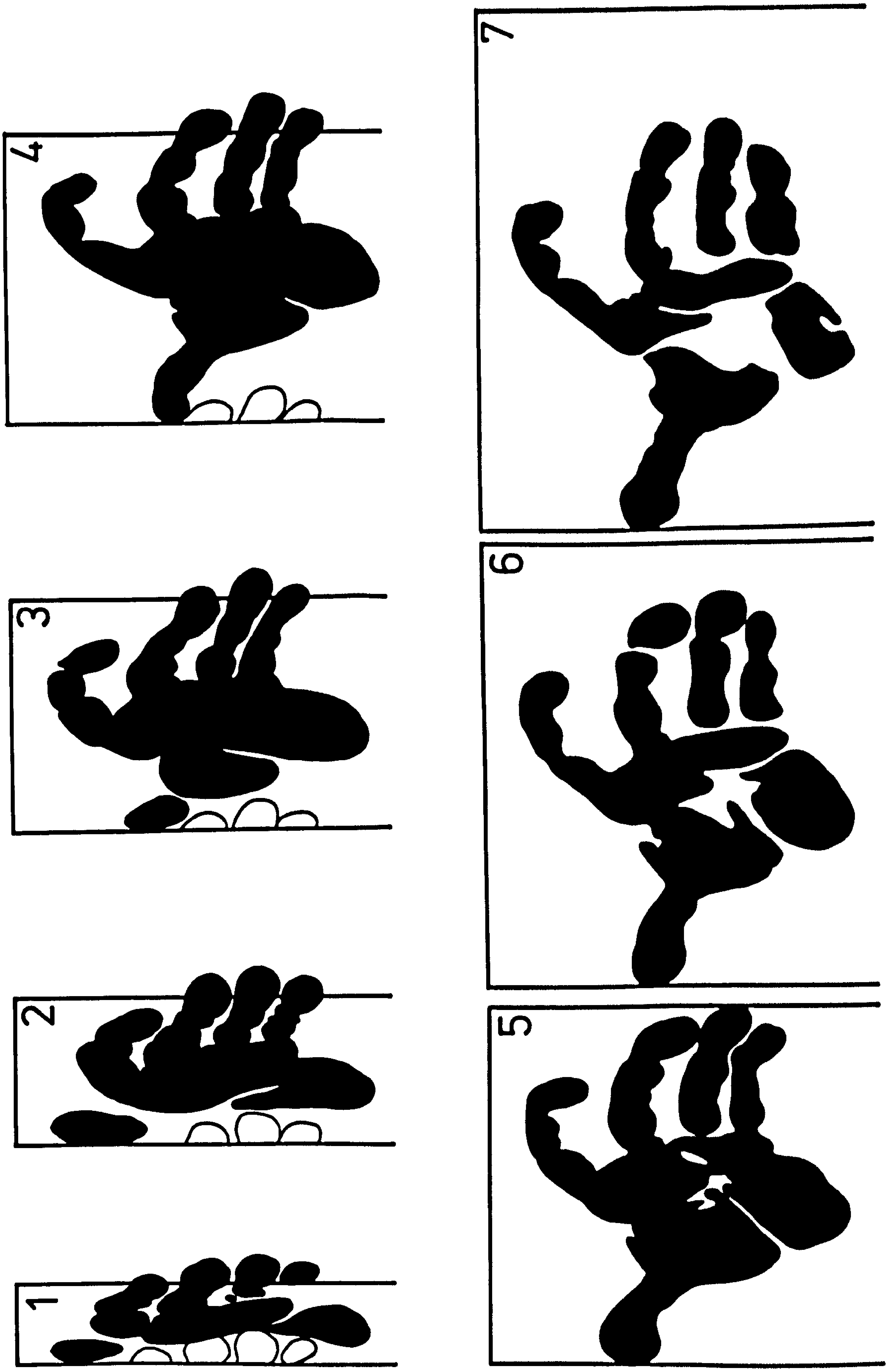
Six subjects for whom torque measurements on the seven smooth cylinders were already available, took part in the experiment. Prints were made of the right hand of the subject gripping each of the cylinders in turn. Photographic paper (Kodagraph P84) was wrapped around the handle. The subject's hand was smeared with vaseline and he was asked to grip the handle firmly as if about to use it in the way he had done during the strength measurement. The subject then carefully disengaged his hand from the cylinder and the photographic paper was stripped from the handle and placed in a bath of developer (May and Baker, Suprol Developer). This gave a high contrast hand print in which the contact areas were quite white and the remaining areas were quite black. The print was then fixed in hypo which at the same time removed the vaseline. It was then washed and dried. The total area of contact of each print was measured using a planimeter.

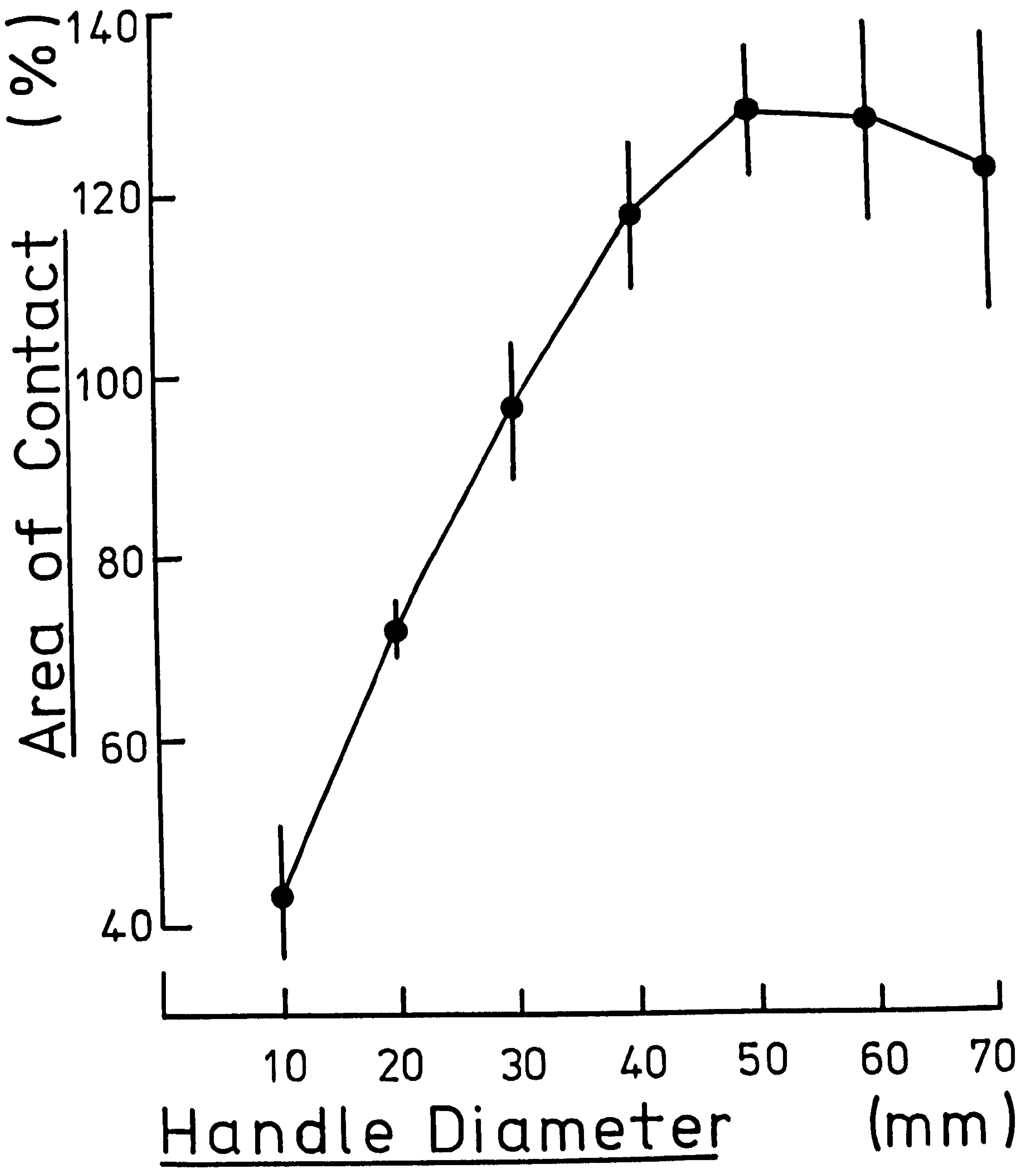
Figure 3,12

Sample set of hand-prints from smooth cylinders of different sizes. Cylinders are numbered by their diameter in centimetres.

3,6,2

Figure 3,12





Total normalised areas of contact between hands and handles (mean $\pm$ 1 s.d.) plotted against handle diameter.

3,6,2

3,6,2 RESULTS.

A specimen set of hand prints is shown in fig. 3,12. (These have been traced and blacked in, but faithfully reproduce the original). The remaining sets of prints were qualitatively similar. The contact areas for each subject were normalised against the mean area for all handles. The results of these measurements are shown in table 3,5, and fig. 3,13. (mean ± 1 s.d.)

TABLE 3,5 TOTAL AREAS OF CONTACT BETWEEN HANDS AND HANDLES.

Handle Diameter (mm)	n	Contact Area (Sq. mm.) mean	s.d.	Contact Area (%) mean	s.d.
10	6	2503	525	43.2	7.4
20	5	4428	700	72.1	3.6
30	5	5562	1078	96.1	7.5
40	6	6796	1149	117.6	8.3
50	6	7490	1347	127.9	6.8
60	5	7248	1525	126.8	11.1
70	5	7108	1890	120.8	14.4

(Out of the total number of forty two prints which were taken, four were deemed unsuitable for quantitative treatment due to poor quality. Some results therefore are based on n = 5)

3,7 DISCUSSION

3,7,1 THE ANGLE-TORQUE RELATIONSHIP FOR PRONATION AND SUPINATION.

The results shown in fig. 3,4 agree with those of Darcus (1951). The action of supination is strongest when the forearm is prone, and the action of pronation is strongest when the forearm is supine. The statistical analysis of these results (Fig. 3,6) confirms that levels of statistical significance increase as the difference of forearm angles increases.

Darcus (1951) measured torques when the elbow and shoulder were held in six different positions. One of his positions; i.e. with the shoulder adducted and the elbow held at a right angle, was repeated in the present experiment. In this position torques about the long axis of the humerus are not required. In contrast, a task demanding rotational torques when the forearm and arm are co-linear (i.e. with the elbow extended) is as much a test of the strength of shoulder rotation as it is of pronation or supination. Furthermore, in the straight limb condition it becomes difficult to define the operating posture as the subject has the opportunity to trade-off rotations at the gleno-humeral joint against rotations about the radio-ulnar axis. The equilibrium conditions then become exceedingly complex. It is theoretically possible that in the present conditions the exerted torque might be limited not by the radio-ulnar mechanism but by the capacity of the shoulder, shoulder-girdle and trunk to transmit the torques. The bulk of the thoraco-humeral, scapulo-humeral and para-vertebral musculature suggests that these possibilities are negligible.

A much more substantial possibility is that the hand, under some circumstances, may fail to transmit the maximum torque of the muscles of pronation and supination, and it is this contingency which is investigated in experiment 3,3. The statistical results of this experiment tabulated in fig. 3.6, require some discussion. The general trends for stirrup handles are confirmed in this case, but the matter of interest is the possible presence of statistically significant differences between <u>handles of different sizes in the same orientations</u>. There are sixty possible comparisons of this type. Of these, two are significant at the level $0.05 > p > 0.01$, and none are significant at higher levels. We must remember however that out of sixty t-tests performed on pairs of samples drawn from the same populations, three of them might be expected to reach this level by chance alone. It is concluded therefore that the present experiment finds no evidence for difference between handle sizes in the same orientation. This is not conclusive proof that the hand is not the weak link in this action, but in the absence of evidence to the contrary, it must be considered a strong indication. The matter could only be finally settled were it possible to rigidly and painlessly couple a transducer system to the forearm itself; and this is clearly not a feasible aim. Fig. 3,5 shows the absolute values of torque measurements made with the stirrup handles and the T-bars, plotted together for comparison. The two sets of results are strikingly similar in spite of the experiments having been conducted on different subject groups. It is therefore safe to neglect the possibility that some special characteristic of the T-bars, such as discomfort caused by sharp edges, is affecting the results.

The conclusion to be drawn from these findings is that we are truly investigating the angle-torque curve of the muscles of pronation and supination and it is worthwhile considering this relationship in greater detail. Darcus (1951) makes the assertion that "by the total range of hand positions investigated, there is a linear relationship between the position of the hand and the isometric torque developed. The correlation coefficients were all significant to below the 0.01 level of confidence ......" It is one thing however to find a significant product moment correlation coefficient but quite another to prove that the relationship between the variables tested is indeed linear (Appendix 2, 4). In the present case the correlation between pronator and supinator torques and forearm posture were 0.90 and 0.85 respectively. Both of these are significantly different from zero at the level of $p < 0.001$.

The more relevant statistical question is whether a straight line gives as good a fit to the experimentally determined points as any other possible curve. In order to test this question, a least squares procedure was used to fit the normalised data with an orthogonal polynomial (Appendix 2,4). In the case of supination data a quadratic term significantly improved the fit; in the case of the pronation data a cubic could also be added. The curves which were derived are plotted in fig. 3,14. The regression equations, quoted with their r.m.s. error were as follows:

$$\tau s = 157.98643 - 0.163450 - 0.00193\phi^2 \pm 19.88$$

$$\tau p = - 16.06607 - 0.31359\phi - 0.00955\phi^2 + 0.00004\phi^3 \pm 19.56.$$

Where τs is the normalised torque of supination,

τp is the normalised torque of pronation

and ϕ is the angular orientation of the forearm.

The anatomico-mechanical basis of the angle-torque curves is obscure. The arguments which may be put forward concerning the various possible interactions have been cited previously in respect of the muscles of plantar-flexion. A formal treatment must await the acquisition of detailed quantitative morphological data. Fick (1911) quotes cross-sectional area and total muscle excusion data for an unknown number of cadavers. These excursions were measured over an arc of movement of 120 degrees and assuming a linear angle-excursion relationship, it is possible to make a best estimate of the leverage and hence of the

Figure 3,14

Regression lines ($\pm$ 1 R.M.S. residual) of normalised supinator and pronator torque on forearm angle.

3,7,1


200 %
supination
100
TORQUE
0
0
45
90
135
180
-100
% -200
pronation

relative importances of the various muscles, as was done in section 2,3,4 for the plantarflexors. The results must be considered as only a first order approximation. It is of interest to note that Fick's data suggests that the strengths of supination and pronation, in their respective optimum positions should stand in the ratio of 1.16 : 1. Experimentally determined values for this ratio were 1.19 in experiment 3,2 and 1.15 in experiment 3,3.

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Table 3,6 ESTIMATED WEIGHTINGS OF MUSCLES OF PRONATION AND SUPINATION BASED ON THE DATA OF FICK (1911)

a) Supinators

Muscle	Excursion (m)	p.c.s.a. (cm^2)	Weighting (%)
Biceps brachii (short head)	0.019	3.22	32.7
Biceps brachii (long head)	0.016	3.33	28.5
Supinator	0.015	2.20	17.6
Abductor pollicis longus	0.004	1.84	3.9
Extensor pollicis brevis	0.004	1.84	3.9
Brachioradialis	0.002	1.86	11.9
Extensor pollicis longus	0.003	0.56	0.9
Extensor indicis	0.003	0.37	0.6

b) Pronators

Muscle	Excursion (m)	p.c.s.a. (cm^2)	Weighting (%)
Pronator teres	0.019	3.6	42.5
Flexor carpi radialis	0.011	2.16	14.4
Brachioradialis *	0.012	1.86	15.9
Pronator quadratus	0.008	2.22	11.0
Extensor carpi radialis longus	0.005	3.14	9.8
Palmaris longus	0.011	0.93	6.4

* Brachioradialis is, according to Fick (1911) a supinator between angles 0 and 20 degrees according to our notation and a pronator between angles 75 and 180 degrees. The excursion measurements have been selected accordingly in the preparation of the final weightings.

No mention has been made in this discussion of the distinction drawn in the case of plantarflexors between active and passive torques. This has been ignored because in the present case the passive torque cannot be measured due to the fact that the coupling between the forearm and handle is not sufficiently rigid unless muscles are activated to make the grip. It is probable that passive elements are under strain in parts of the postural range tested, but the contribution of these elements cannot be determined and we must be content with a torque measurement which is the sum of active and passive components.

3,7,2 THE STATICS OF THE CYLINDRICAL GRIP.

No accepted definition exists either of a gripping action or of the quantitative strength of grip. It is considered here that an act of gripping is performed when the hand forms a closed system of forces in which portions of the digits and/or palm are used, in opposition to each other to exert compressive forces on the object gripped.

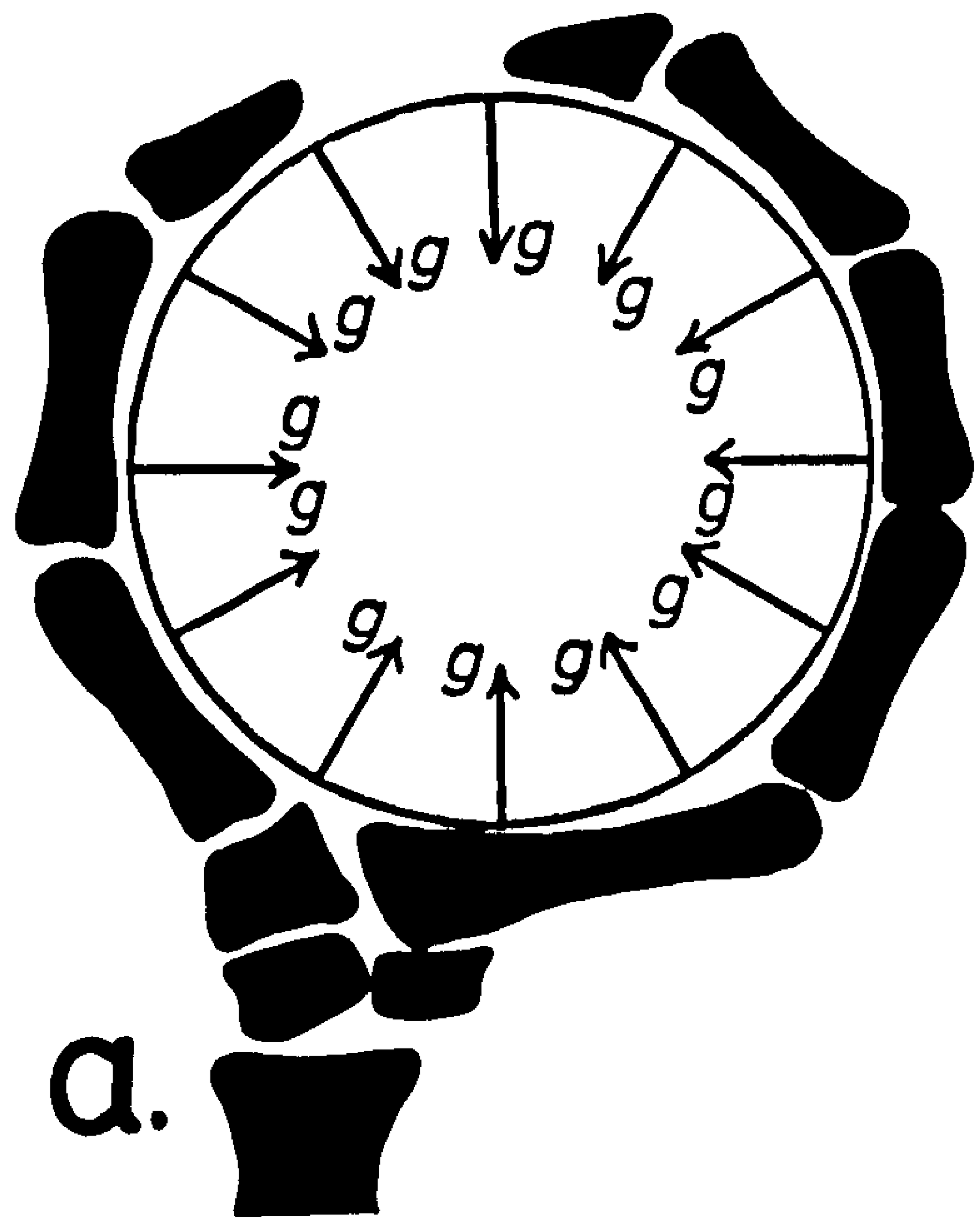
The strength of grip may be defined as the sum of the components of these forces normal to the surface of the handle (Fig. 3,15a). Such a quantity is of limited relevance. It only achieves practical significance with respect to an activity in which the hand-handle complex is used to apply forces or torques to the outside world.

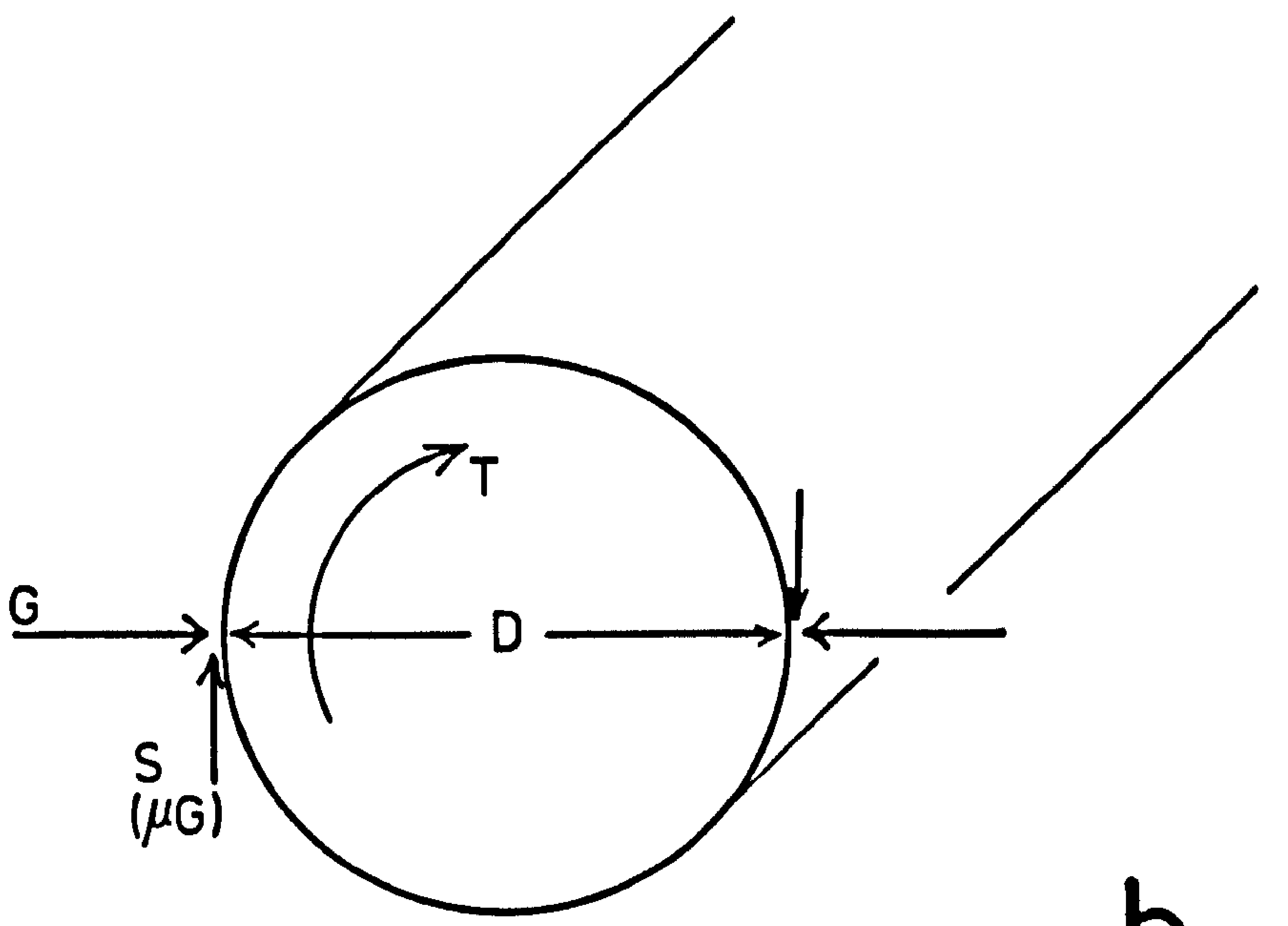
Consider the present case in which the hand grasps a smooth cylindrical handle and twists it against infinite resistance. Forces arising from the musculo-skeletal system of the operator are transmitted as frictional (shear) forces at the hand/handle interface. Fig. 4,15b is a simple resolution of these forces. In the case measured in experiment 3,4, where readings of torque were taken on the point of slip, then the following equations apply:

$$\mathcal{T} = S.\ D.$$

Where $\mathcal{T}$ is the applied torque

S is the total frictional force at the handle-surface, and also





The statics of the cylindrical grip. For explanation see text.

3,7,2

$$S = \mu . G$$

where G is the strength of grip as defined above and μ is the coefficient of limiting friction at the hand-handle interface.

It is tempting to suggest that we might attempt to measure grip strength from the relationship

$$G = \frac{\tau}{D . \mu}$$

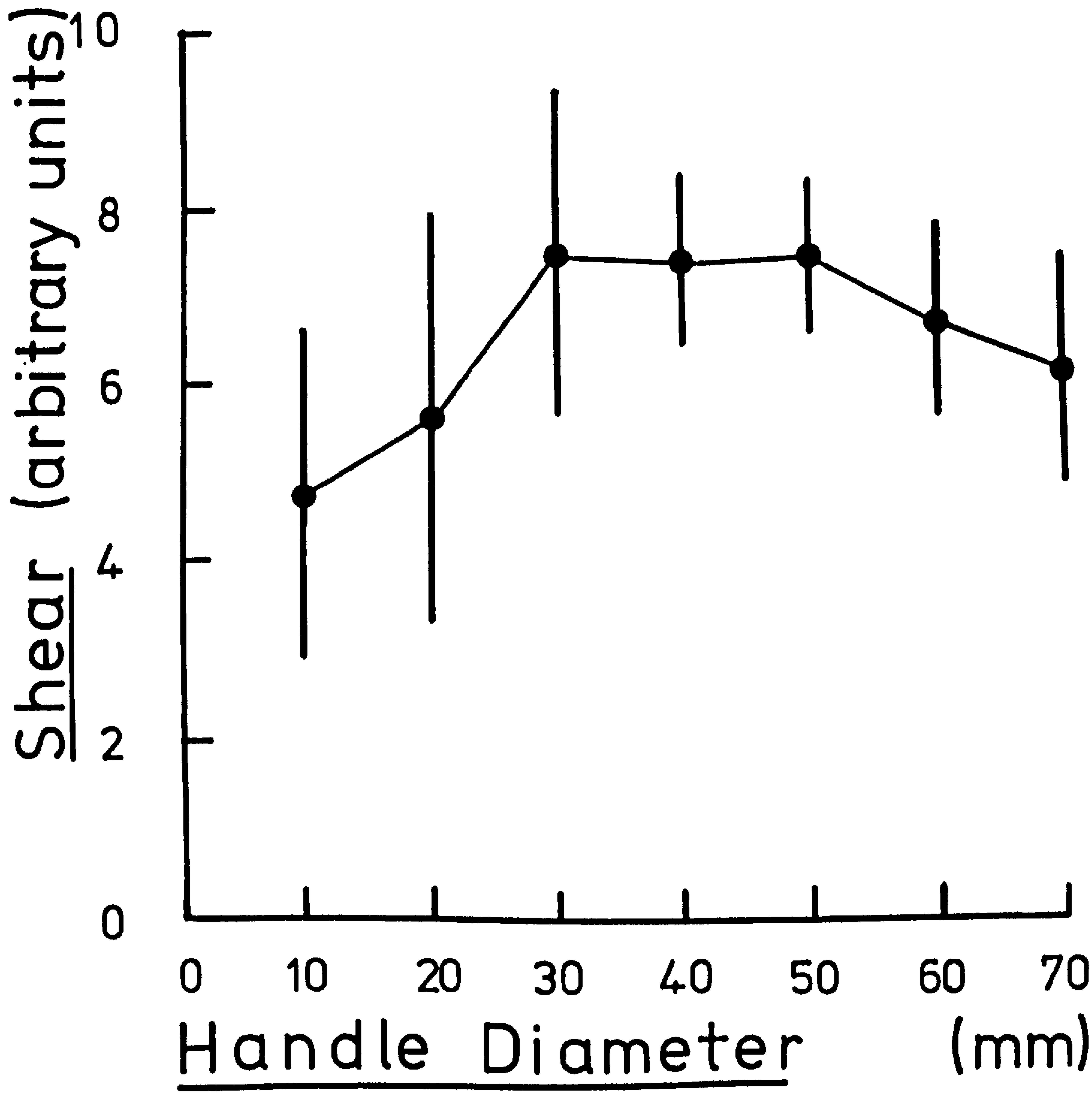
but further investigation of this possibility suggests that a numerical value for μ is exceedingly elusive. An exact value for μ depends upon the precise condition of the handle and cutaneous surfaces. Furthermore, were it determined by such a direct method such as pressing and pulling on a flat plate, we could not be certain of its validity - the complex ways in which the skin of the palm might deform are in no way predictable.

The analysis is continued on the assumption that, given the precautions described in 3,4,1, for a single subject on the occasion of one experiment, the value of μ will be constant throughout measurements on all seven cylindrical handles. Calculation of the shear force at the handle surface may be considered to yield comparative values which are directly proportional to the strength of grip as defined. Table 3,7 shows the absolute and normalised shear forces exerted on handles of different sizes as calculated from the data of experiment 3,4.

Table 3,7 SHEAR FORCES EXERTED ON SMOOTH CYLINDRICAL HANDLES.

Handle Diameter mm.	Absolute Shear (N)		Normalised Shear (Arbitrary Units)	
	mean	s.d.	mean	s.d.
10	75	36	17.5	7.7
20	103	39	23.0	7.9
30	122	52	27.1	6.3
40	123	50	27.4	3.8
50	124	57	27.4	3.1
60	110	48	24.4	4.0
70	101	48	22.0	4.7

These results are plotted (mean $\pm$ 1 s.d.) in figure 3,16. Students t-test was performed on all possible pairs of the normalised data and the matrix of significance levels is shown in fig. 3,17. The shear force may be seen to go through a clear and statistically significant optimum in the region of 20 - 50 mm handle diameter. There is a marked reduction in strength at diameters above and below this range. Referring to the values from the literature shown in fig. 3,1 it may be seen that these findings are compatible with those of Ayoub and Lo Presti (1971) and O'Neill (1974) but substantially dissimilar to Hertzberg (1955). No firm conclusions may be drawn from these comparisons as the different tasks involved in the measurement methods demand markedly different muscular loadings.



Frictional Force (Shear) exerted on smooth cylindrical handles, plotted in arbitrary units (mean ± 1 s.d.) against handle diameter.

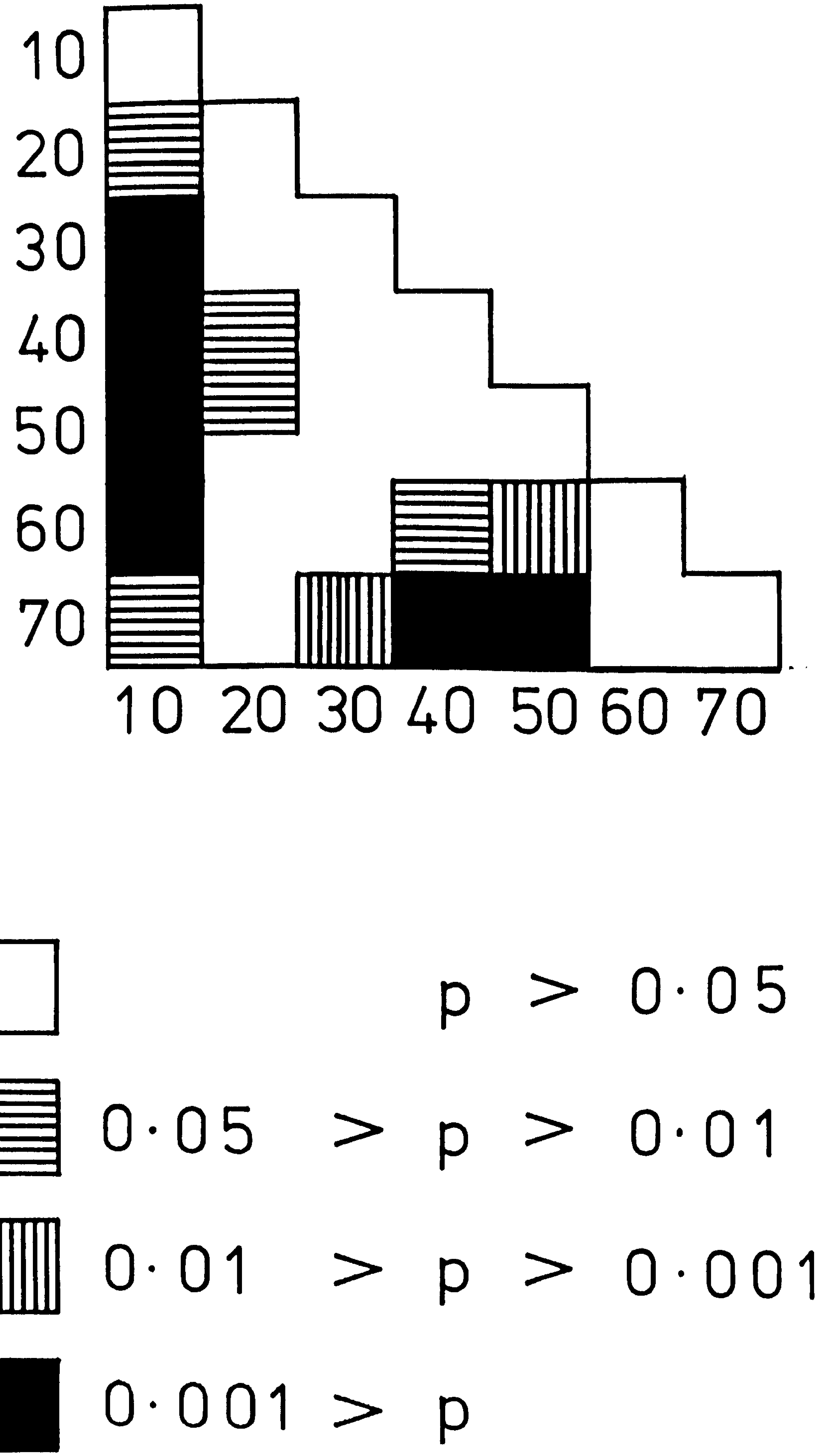
3,7,2

Figure 3,17

Levels of significance for differences in normalised shear force exerted on smooth cylindrical handles of different diameters.

3,7,2

Figure 3,17



10
20
30
40
50
60
70
10 20 30 40 50 60 70
p > 0·05
0·05 > p > 0·01
0·01 > p > 0·001
0·001 > p

The relationship between the shear force and the handle diameter is therefore quite different from the relationship between measured torque and diameter. In this latter case, the increased leverage about the centre axis of the handle due to handle size, far outweighs the decline in gripping strength and the output torque continues to increase throughout the range studied. It is appropriate at this point to consider the factors which determine the shape of the relationship between shear force and handle diameter.

3,7,3 A DISCUSSION OF FACTORS UNDERLYING THE FORM OF THE RELATIONSHIP BETWEEN SURFACE SHEAR FORCE AND DIAMETER IN CYLINDRICAL HANDLES.

a) Dimensions of the hand.

Anthropometric data was available for a total of 25 subjects (14 male and 11 female) amongst those who took part in experiments 3,4 and 3,5. For these subjects, correlation coefficients were calculated between the two hand dimensions defined in section 3,4,1 (b), and the diameter of handle on which the greatest shear force was exerted. The results of the statistical analysis were as follows:

Variable	mean (mm)	s.d. (mm)	Correlation with optimal handle size
Hand length	188	13	0.05
Hand spread	209	21	0.20

Neither of these correlation coefficients is significantly greater than zero ($p > 0.1$) and it must therefore be concluded that in the present group of subjects there is no detectable relationship between the size of the hand and the handle size which was found to be optimal for the exertion of shearing forces, and hence by inference for strength of grip. This result is not what would intuitively be expected. One would reasonably anticipate that subjects with large hands would find large handles satisfactory whereas the statistics suggest that the optimum is random in its relationship, at least to simple linear measures of the hand. It is possible that a relationship would be discerned if the range of hand sizes were increased to include for example the hands of juvenile subjects.

b) <u>Hand-handle contact area</u>.

The results plotted in figure 4,12 show that the contact area between hand and handle has a relationship with handle size which is rather different from that of shear force. Contact area continues to increase throughout the 10 to 50 mm. range and thereafter shows an apparent modest decrease between 50 and 70 mm. which is statistically non-significant ($t = 0.93$, $p > 0.1$). Although it is possible to hypothesise that hand-handle contact is responsible for the rising portion of the shear force/handle diameter relationship, it cannot be held to affect the descending portion.

c) It is necessary to search therefore for a further unmeasured factor to account for the experimental findings. The most likely explanation is that at handle sizes of greater than 50 mm. one or more of the several groups of muscles involved in the task enter a less advantageous portion of their range, either in terms of length-tension curve or of leverage. The complexity of hand mechanics is such, that at the present time it is impossible to identify this phenomenon with any greater accuracy.

3,7,4 A SPECULATION CONCERNING POWER AND PRECISION GRIPS.

The terms power and precision grip were introduced in section 3,1. To date these have not been studied in a systematic quantitative fashion. Sharpe (1962) studied the supinator torques which could be exerted on knobs of different diameters and surface qualities. Sharpe's subjects were instructed to grasp the knobs with their fingertips, so it seems likely that a precision grip within the limits of the usual definition was used. Sharpe's data for smooth knobs was subjected to the same analysis as the data of the present study to derive values for mean surface shear in Newtons. The results of both studies are plotted for comparison in figure 3,18. It is probable that the subject population in Sharpe's studies (American males) was intrinsically stronger than that of the present study (which included a proportion of female subjects); furthermore, there is no basis for comparison in the surface finishes of the handles. The locations of the curves plotted in fig. 3,18 are therefore questionable, but it seems likely that the forms of the curves are correct. It is of considerable interest that the data for knobs continues to climb throughout the range studied, showing no signs of reaching a plateau or tailing off. The anatomico-mechanical basis of these observations is again obscure.

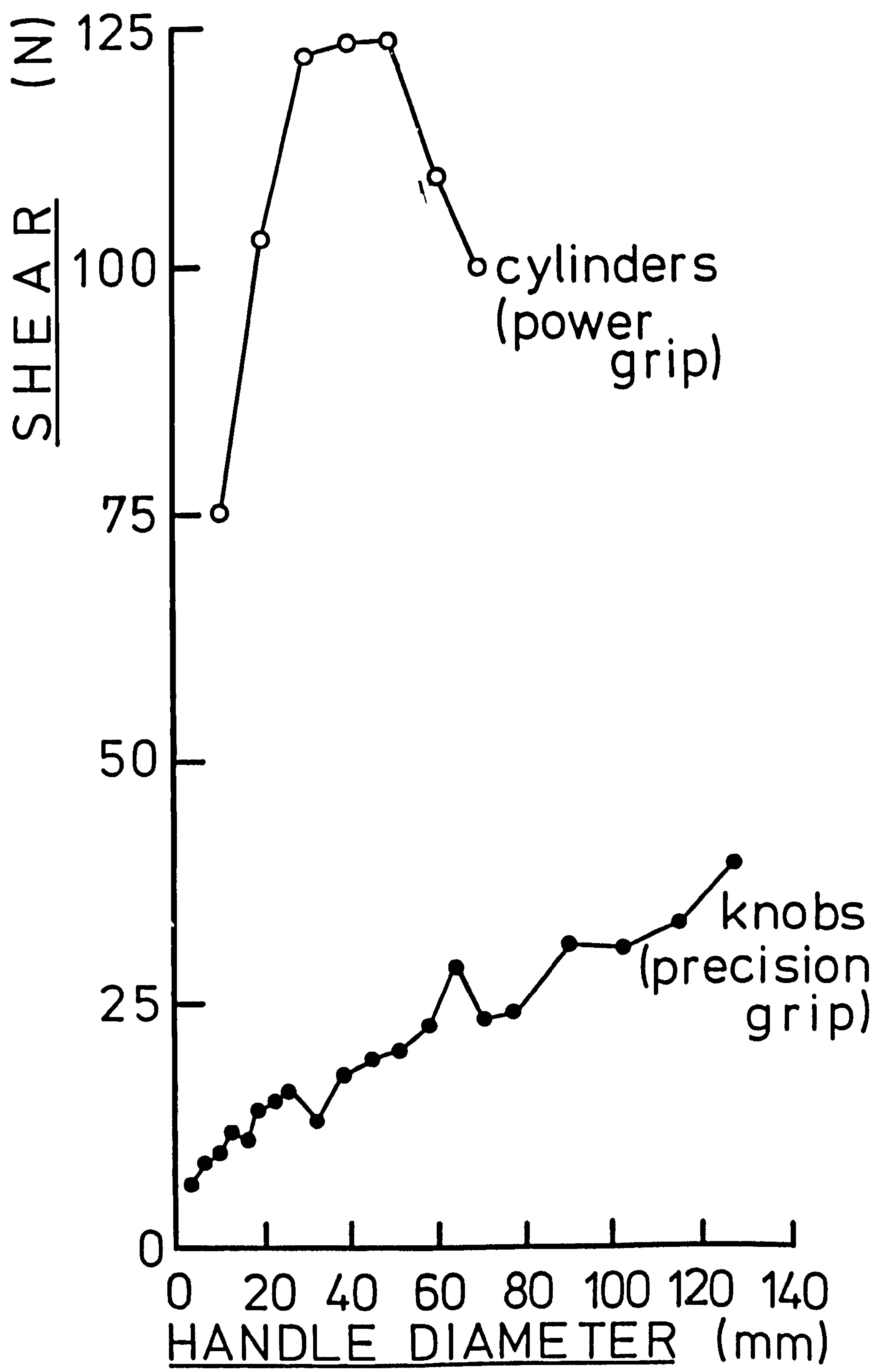
<u>FOOTNOTE</u>:

Subsequent to the completion of the present thesis experiments have been conducted on a population of sixteen subjects which have confirmed both the forms and relative locations of the curves shown in Fig. 3,18.

Figure 3,18

Comparison of shear forces exerted on smooth cylinders with forces exerted on smooth knobs. Latter set derived from the published data of Sharpe (1962)

3,6,4


SHEAR (N)
125
100
75
50
25
0
0 20 40 60 80 100 120 140
HANDLE DIAMETER (mm)
cylinders
(power
grip)
knobs
(precision
grip)

3,7,5 THE COMPARISON OF HANDLES AND THE DESIGN OF SCREWDRIVERS.

The ability to exert torques using screw-drivers of various designs has been compared in two published studies (Murphy, 1968; Which, 1974). In both cases an arbitrary selection of screwdrivers was compared and no attempt was made to account for the differences. The statistical analysis of the present data (fig. 3,10 and 3,11) shows some findings of potential practical importance.

For handle diameters of less than 60 mm. the knurled handles show a significantly higher torque than the smooth handles of comparable diameter - this finding is to be anticipated from the equations of section 3,7,2. At the 70 mm. handle size there is no significant difference between knurled and smooth cylinders. Although some screw-drivers are better for the exertion of torque than smooth cylinders of similar diameter, none are better than knurled cylinders. It must be concluded that shape has no relevance to the exertion of torque and that size is of overwhelming importance provided that the hand is in some way given "purchase" to prevent it from slipping round the handle. The largest commercially available screwdriver tested was a cabinet maker's turnscrew of 36 mm. mean diameter. The author has not been able to locate a larger one. The author has heard anecdotes of craftsmen who construct their own monstrous handles which are believed to be very effective. The data suggests that for the exertion of maximal torque a knurled cylinder of 50 mm. diameter is as good as any other possible design. Hand-handle contact area is also maximal in this range and this is desirable in a hand tool as large contact areas minimise shear stress on the skin for a given output. The 50 mm. knurled cylinder is only marginally ($0.05 > p > 0.01$) weaker than a T-bar placed in its optimal orientation. The forearm posture was not controlled for torque measurements using cylindrical handles and it is not likely that all of the subjects adopted the extreme (optimal) position. It is concluded therefore that the 50 mm. knurled handle allows the transmission of close to, if not all, the torque-exerting capacity of the muscles of supination. The reduction in torque between knurled handles of 50 and 70 mm. probably reflects the same process as is observed in the reduction of shear forces in this range (3,6,3).

In the light of the above findings a simple knurled cylinder of 50 mm. diameter seems an attractive proposition for the design of a screwdriver. It must be borne in mind that the exertion of torque is not the only relevant criterion in the ergonomic design of a screwdriver - speed and precision of use and long-term comfort to the user are of at least equal importance.

CHAPTER IV - Frontispiece

from Borelli (1680)

CHAPTER 4

SYMMETRICAL PULLING ACTIONS IN THE SAGITTAL PLANE

"When you wish to represent a man in the act of moving some weight ------- one should remember that a man's weight drags in proportion as the centre of his gravity is distant from that of his support, and to this must be added the force exerted by his legs and bent spine as he straightens himself".

Leonardo da Vinci

Mss. E. Institut de France

after MacCurdy (1939).

The above quotation demonstrates that the role played by body weight and its leverage in certain activities was perceived by Leonardo in the sixteenth century. The frontispiece to this Chapter is taken from Borelli's "De Motu Animalium" of 1680. Borelli was aware that when a man statically supports a load, many muscle groups are called into play to maintain equilibrium. The first detailed studies of these questions were those of the late W.T. Dempster (Dempster, 1955; Gaughran and Dempster, 1956; Dempster, 1958) who studied the forces which could be exerted in symmetrical pulling tasks in the sagittal plane. Dempster analysed his results graphically by means of free-body diagrams and stated categorically that in both the standing and sitting positions it was the weight and mass distribution of the subject's body which limited the force applied rather than the strength of any specific combination of muscles.

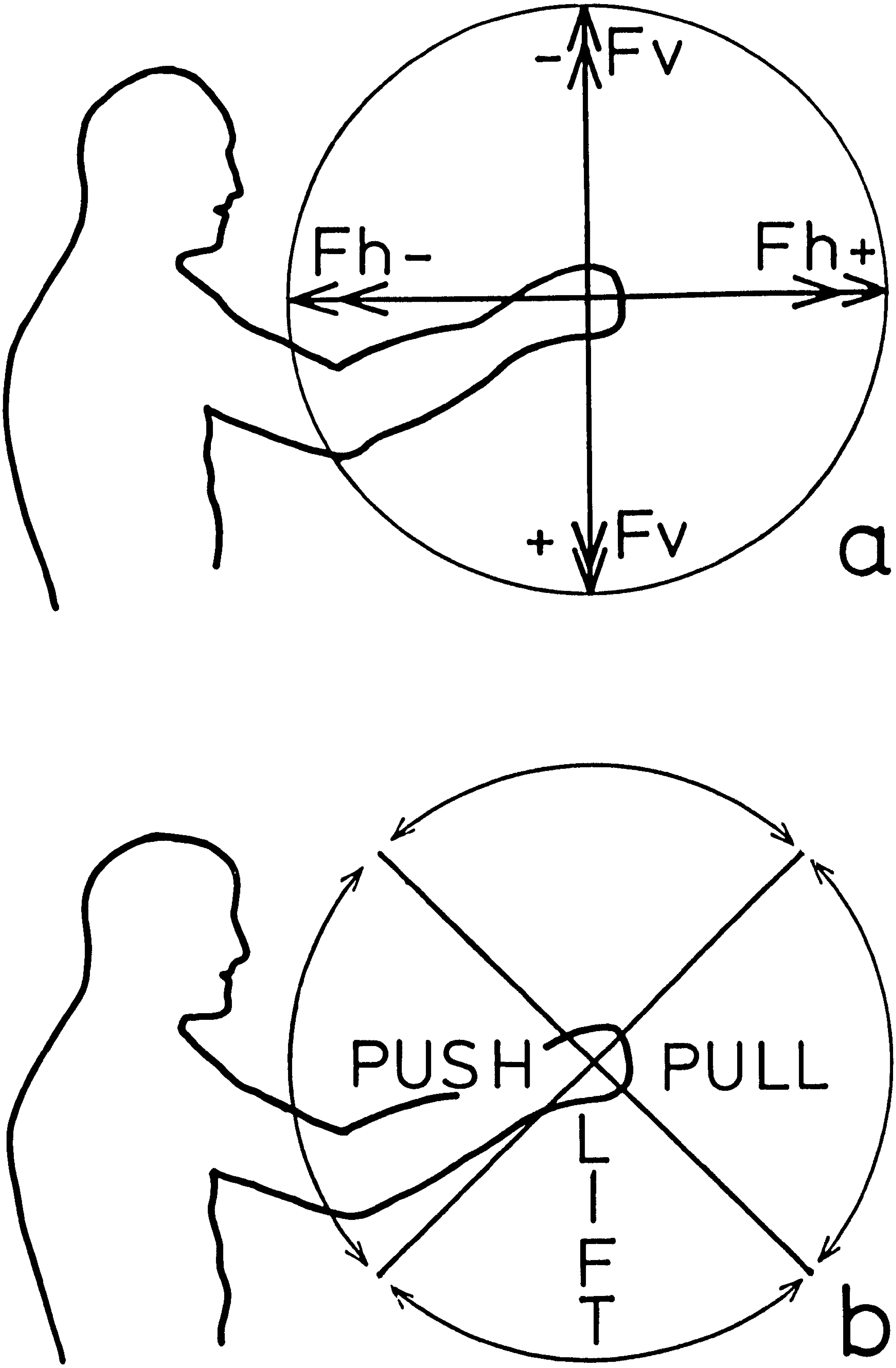
Whitney (1958) conducted a study of the "lifting" action and came to conclusions that were similar to those of Dempster. Whitney presented an algebraic formulation of the statics of the lifting task and derived a predictive equation for the vertical force with two unknown constants which could be determined by a linear regression analysis of the experimental data. He concluded that in cases where the point of foot support was close to the lifting axis, "the maximum force of the extending musculature will become the limiting factor".

Troup and Chapman (1969) measured the maximum pulling and pushing forces exerted by 230 subjects in standing and sitting postures in which the trunk was erect, the arms were horizontal and the pelvis was restrained both before and behind. The subjects used both hands to grip a dynanometer which recorded forces in the horizontal direction only. The authors made the reasonable assumption that in such a posture the vertical component would be negligible. The vertical distance between the hands and the pelvic restraint was multiplied by the measured force to give a value of the maximum torque which could be exerted by the flexor and extensor muscles of the trunk at the level of the pelvic restraint. It was acknowledged that intra-abdominal pressure could act as an additional extensor mechanism in such an activity (Davis, 1956; Bartelink, 1957; Morris _et al_, 1961; Eie and Wehn, 1967; Davis and Troup, 1964a, 1964b; Eie, 1966; Stubbs, 1975; Grieve, 1977).

Other authors (Asmussen and Heebøll-Nielsen, 1961; Poulsen, 1971) have measured the extensor thrust which a subject can exert on a harness placed around the shoulders, the pelvis being restrained. The latter authors showed that this measure of "back strength" was correlated with the maximum weight which could be lifted from floor to table height (male subjects r = 0.72, female subjects r = 0.78) whereas the correlations between lifting capacity and body weight were non-significant.

Opinions expressed in the literature suggest that a two-handed pulling force is limited by body weight and mass distribution if the hips are unrestrained, whereas if the hips are restrained, it is the truncal extensor mechanism (i.e. the paravertebral muscles assisted by intra-truncal pressure changes) which is tested. The present Chapter will further examine these propositions.

Figure 4,1



Sign convention and definition of terms. For explanation, see text.

4,1 THE STATICS OF THE SYMMETRICAL EXERTION OF FORCES IN THE SAGITTAL PLANE.

The first necessity in the analysis of the tasks under consideration is to derive equations in the general case for the force which a man may exert on an external object by two-handed symmetrical efforts in the sagittal plane. This derivation is achieved here by the method of free-body diagrams and their algebraic solution. (Dempster, 1958)

The following sign convention will be used. (Fig. 4,1(a)

a) All forces or their components acting on the man in the same direction as gravity are positive, and forces in the opposite direction are negative.

b) All horizontal forces or horizontal components of forces acting on the man away from his ventral surface are positive, and forces in the opposite direction are negative.

c) Force vectors of pushing, pulling and lifting act from the hands into the quadrants indicated in Fig.4,1(b)

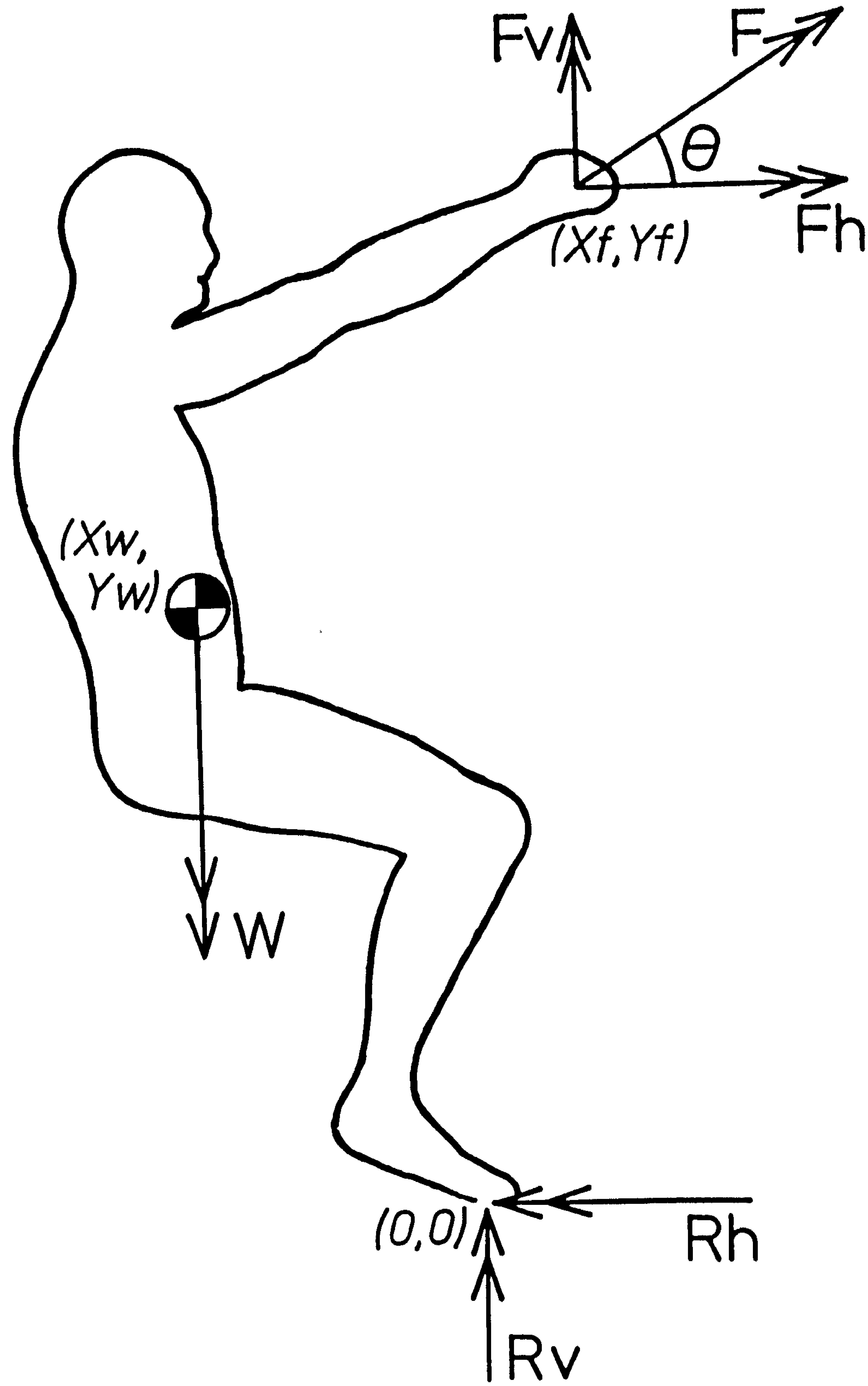
Consider a man of body weight W exerting a force F (Fig. 4,2) the location of the force and the man's centre of gravity may be expressed as x and y co-ordinates with respect to an origin located at the centre of pressure (pivot point) of the feet. The force F may be resolved into vertical and horizontal components, Fv and Fh respectively, and reactive forces Rv and Rh act at the foot pivot

Figure 4,2

Free-body diagram of a man exerting
a symmetrical pulling force in the
sagittal plane.
See text for explantion.

4,1

Figure 4,2



point 0,0. By Newton's second law, the man being in a state of equilibrium (i.e. no accelerations are occurring), the following equations may be written.

In the vertical direction

$$Fv + Rv + W = 0 \qquad (1)$$

In the horizontal direction

$$Fh + Rh = 0 \qquad (2)$$

(It must be noted that in the special case drawn in fig. 4,1, the quantities Fv, Rv, Rh and Xw all have negative values). Taking moments about the origin, where Υ is any pure torque exerted on the body by its environment,

$$F_V . x_F + F_H . y_F + W . x_W + \Upsilon = 0 \qquad (3)$$

It will be assumed hereafter that $\Upsilon = 0$, and hence

Rearranging equation (3),

$$Fh = - \frac{1}{Yf} (Fv. Xf + W. Xw) \qquad (4)$$

and

$$Fv = - \frac{1}{Xf} (Fh. Yf + W. Xw) \qquad (5)$$

The latter of these equations is similar to the one derived, and partially verified in experiment by Whitney (1958).

If the force exerted by the man is truly horizontal, i.e. Fv is xero; then from equation (4)

$$Fh = - \frac{W. Xw}{Yf} \qquad (6)$$

and it may be seen that if Xw is negative (i.e. the man's centre of gravity is behind his foot pivot), Fh is positive (i.e. the man is"pulling"). Whereas if Xw is positive (i.e. the man's centre of gravity is in front of the foot pivot), then Fh is negative (i.e. the man is pushing). These conditions would of course be reversed in the unusual condition that Yf was negative - i.e. that the man was pulling or pushing an object located beneath the floor. Similarly, if the force exerted by the man is truly vertical, i.e. Fh is zero; then from equation(5)

$$Fv = - \frac{W.\ Xw}{Xf} \qquad (7)$$

and hence if $\frac{Xw}{Xf}$ is positive, the man is "lifting", whereas if $\frac{Xw}{Xf}$ is negative, the man is "pulling downwards". Equations (6) and (7) taken alone suggest that the horizontal and vertical forces become infinitely great when Yf and Xf are zero, but this possibility must be modified to satisfy equations (1) and (2).

From equation (2)

$$Fh = - Rh$$

but RH is the frictional resistance to the movement of the man's foot on the floor, hence:

the limiting magnitude of Fh is given by

$$|Fh'| = |\mu .\ Rv| \qquad (8)$$

where μ is the coefficient of limiting friction between the floor and the feet.

and hence from equation (1)

$$|Fh'| = |\mu \cdot (W + Fv)| \qquad (9)$$

Similarly, from equation (1)

$$Fv = - (W + Rv) \qquad (10)$$

but since Rv can never be positive, then the negative limit of Fv is - W; i.e. when the man is freely suspended by his hands. There are, however, no positive limits to Fv since in theory Rv can reach $- \infty$

Returning now to equations (4) and (5) it may be seen that Fv may only be specified in terms of Fh and vice versa (except for the special conditions described in equations 6 - 10) and it is this relationship which must now be examined.

From (4) and (5)

$$\frac{Fv}{Fh} = \frac{Yf}{Xf} \left(\frac{Fh.\ Yf + W.\ Xw}{Fv.\ Xf + W.\ Xw} \right) \qquad (11)$$

and $\theta = \tan^{-1} \left(\frac{Fv}{Fh} \right)$

where θ is the angle that the total Force F makes with the horizontal

Equations (4) and (5) may be simplified to eliminate either Fv or Fh but not both, hence for example substituting (5) into (10).

$$\frac{Fv}{Fh} = \frac{Yf}{Xf} \left(\frac{Fh.\ Yf + W.Xw}{-\left[Fh + W.Xw\right] + W.Xw} \right)$$

$$= \frac{Yf}{Xf} \left(\frac{Fh.\ Yf. + W.Xw}{Fh} \right)$$

and

$$\theta = \tan^{-1} \left\{ \frac{Yf}{Xf} \left(\frac{Fh.\ Yf + W.Xw}{Fh} \right) \right\}$$

The above considerations show that neither the force F, nor its vertical or horizontal components Fv and Fh can be totally specified simply from knowledge of the weight of the man and the points of action of forces at the hands and feet. (If, however, Fv, Fh or θ are known in addition, then the remainder can of course be calculated.)

Let us now analyse the same situation in a slightly different way by resolving the force F into two mutually perpendicular components Fa and Fb, such that component Fa is in the direction of a line drawn from the point of action of the hands to the foot pivot, and let this line have a length AF and make an angle of ϕ with the vertical. (Fig. 4,3)

The foot is supported by reaction forces Ra and Rb and body weight may be resolved into components Wb and Wa, such that:

$$Wa = W \cdot \cos\phi \qquad (14)$$

and $$Wb = W \cdot \sin\phi \qquad (15)$$

Forces in acting in the same direction as gravity are positive, and distances are marked on the diagram. For simplicity of analysis let the surface on which the man stands be a rough one so that slipping cannot occur.

Writing equations for equilibrium we have:

$$Fa + Ra + W. \cos\phi = 0 \qquad (16)$$

$$Fb + Rb + W. \sin\phi = 0 \qquad (17)$$

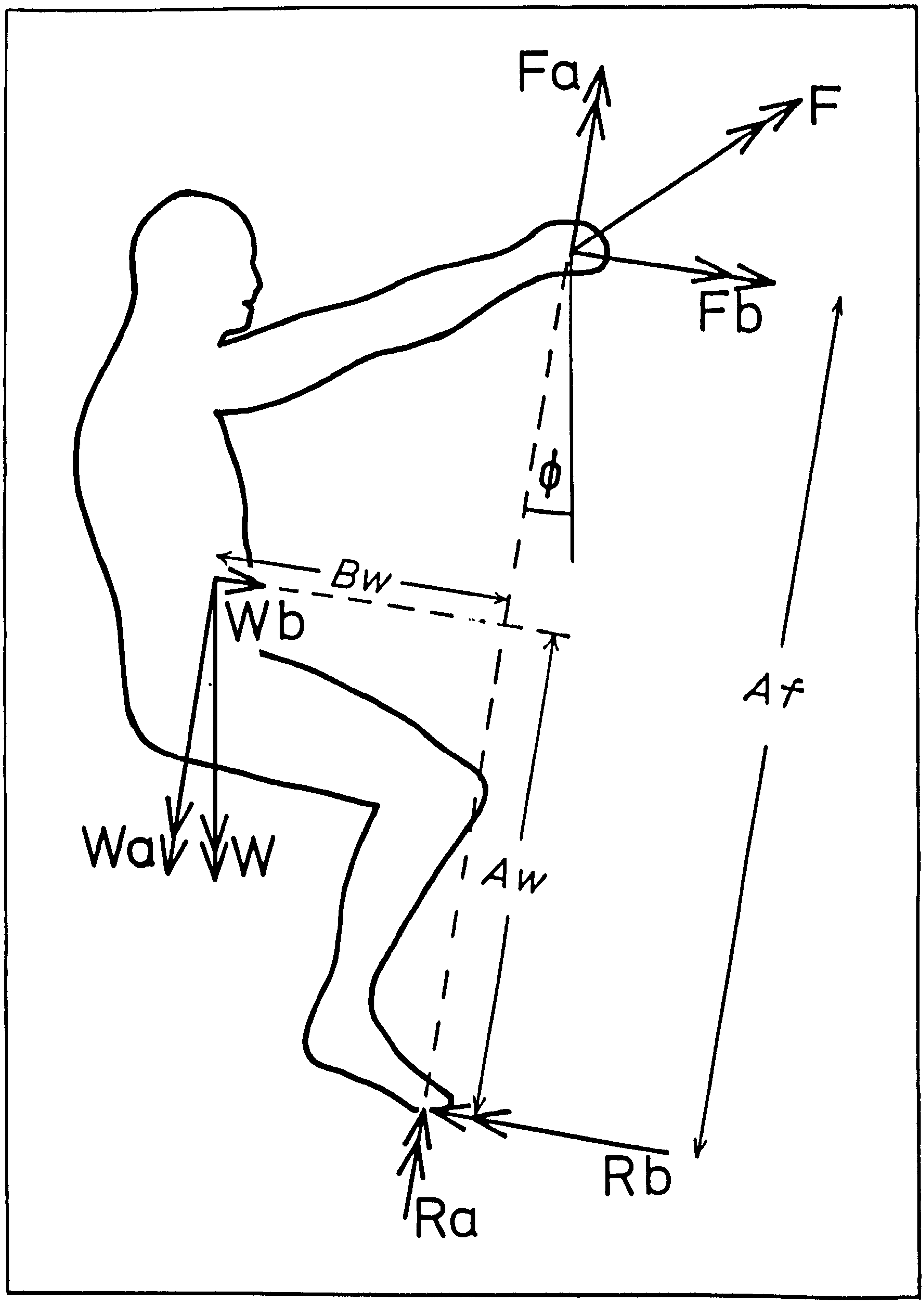
and taking moments about the foot pivot point

$$Fb. \ Af + W. \ \sin\phi.Aw - Wa. \cos\phi.Bw. = 0 \qquad (18)$$

then

$$Fb = \frac{1}{Af} (W. \ \cos\phi \ . \ Bw - W. \ \sin\phi. \ Aw) \qquad (19)$$

Figure 4,3



Free-body diagram of a man exerting a symmetrical pulling force in the sagittal plane. The force has been resolved along the live and dead axes. See text for explanation.

We may therefore conclude that the component of the force F which we have called Fb is completely specified by the weight and distribution of mass of the body. On the other hand the component Fa appears only in equation (16) and is limited only by the fact that Ra cannot become negative. Hence Fa has a negative limiting value of $- W.\cos\phi$ but no positive limiting value may be determined on theoretical grounds.

The components into which the force F has been resolved seem to differ in a very fundamental way, Fb being totally determinate and Fa being indeterminate by the application of statics. The magnitude and sign of the force Fa must in fact be determined by the muscular exertions of the man and its positive limit must be set by the maximum loading which can be sustained by the musculo-skeletal system of the man. No convenient terms exist to describe the directions in which F has been resolved. It is therefore proposed to name the direction of Fa the <u>live</u> axis and the direction of Fb the <u>dead</u> axis.

In a symmetrical exertion in the sagittal plane the live axis is a line joining the point of application of the force to the centre of pressure of the floor support. The dead axis is a line drawn perpendicular to this through the chosen origin.

The categorical statement (Dempster, 1958) that "Whenever the body exerts forces on its environment, forming a closed chain system of forces, limb and trunk muscles do not directly affect pull forces; instead, they maintain joint postures which permit body weight to exert an effective moment" must therefore be considered suspect. Dempster's subject was allowed to choose his own direction of pulling and it may be suggested that in cases where the line of pull is close to the dead axis, Dempster's statement is true, whereas if the line of pull is close to the live axis, then Dempster's statement is likely to be false. Whitney's findings concerning lifting actions in situations where the position of the foot pivot is close to the axis of the lift are an example of the axis of measurement approaching the live axis and his suggestion that a muscular limitation operates is to be supported.

Let us now consider a situation in which a man exerts a pulling force when a restraint is placed in front of his pelvis. (Fig. 4,4).

In this case additional reaction forces Pv and Ph are present located at a centre of pressure Xp, Yp.

Equations (1), (2) and (3) must now be modified thus:

$$Fv + Rv + Pv + W = 0 \qquad (20)$$

$$Fh + Rh + Ph = 0 \qquad (21)$$

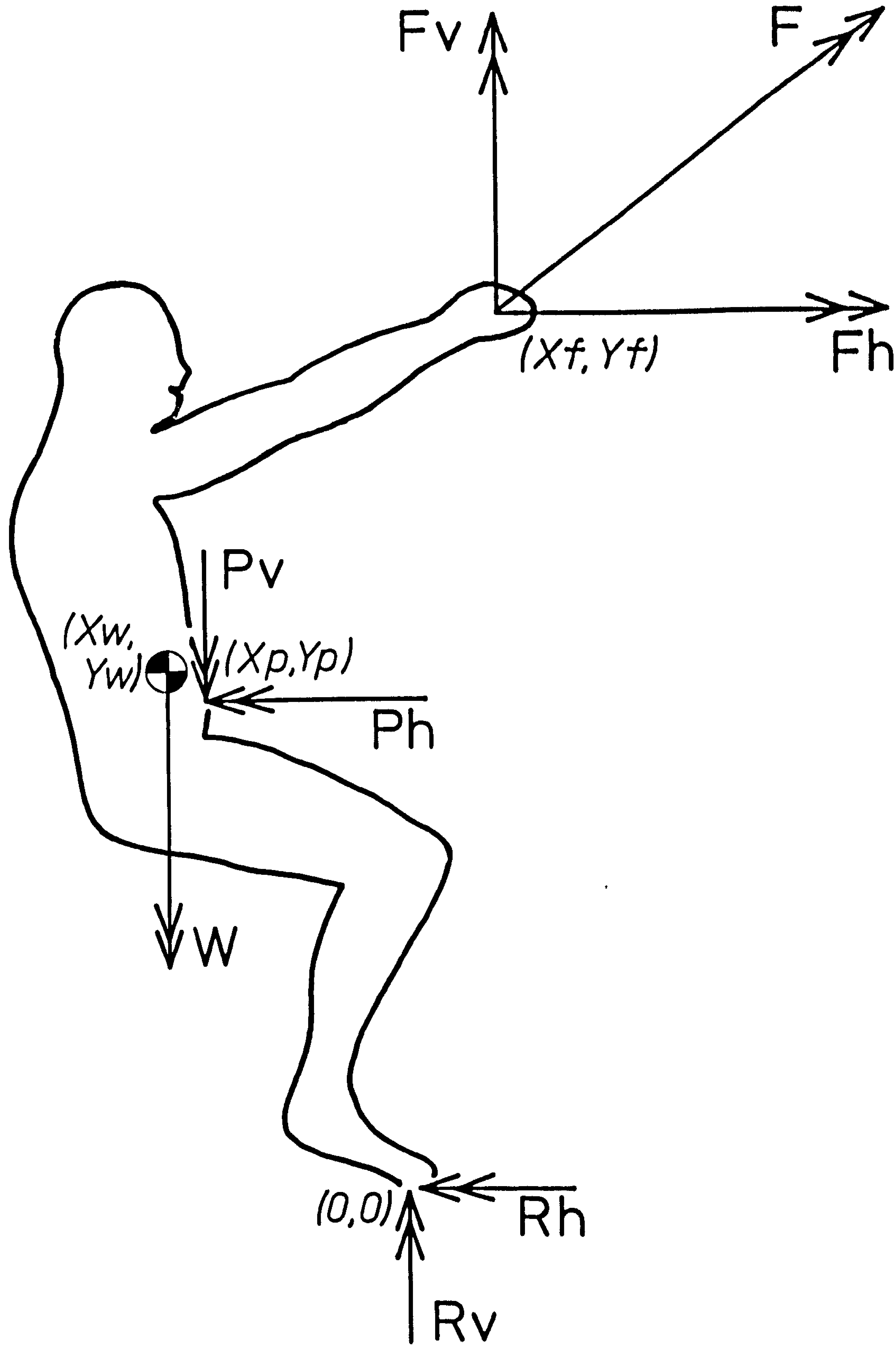
$$Fv.Xf + Fh.\,Yf + W.Xw + Pv.Xp + Ph.\,Yp = 0 \qquad (22)$$

Figure 4,4

Free-body diagram of man exerting
a symmetrical pulling force in the
sagittal plane using a pelvic support.
See text for explanation.

4,1

Figure 4,4



In Troup and Chapman's measurements Ph could be either positive or negative (the pelvis was restrained both before and behind) and with a suitable choice of restraint Pv could be either positive or negative. In theory both Ph and Pv can be infinite compressive forces; in practice they are possibly limited by pain thresholds. Therefore, in principle, any value of Fv or Fh can have an equal and opposite value of Pv or Ph, and the force which may be exerted ceases to be related to the weight of the man's body, to his mass distribution or to conditions at the foot floor interface. In this situation the effective limits of strength will be determined by a critical maximum loading somewhere within the musculo-skeletal system, and the lumbar spine is the most likely choice for a weak link.

In the light of these theoretical propositions, it seemed appropriate to conduct certain experimental studies, in order to enlarge upon the characteristics of the gravitational or muscular limitations of strength which may apply in different varieties of two-handed lifting or pulling actions. The experiments to be described were as follows:-

i) A pilot study (4,2) in which the strength of ninety-two medical students was measured. The subjects performed two-handed horizontal pulls on a bar set at 0.5, 1.0 and 1.5 metres from the ground. They also took part in tests of pronation and supination strengths and their statures and weights were measured. A statistical analysis of these results lead to the formulation of an hypothesis concerning the relative importance of gravitational and muscular limitations.

ii) A second more detailed study (4,3) in which seventy-three medical students acted as subjects. The tasks performed by these latter subjects were chosen so as to test the hypothesis derived from experiment 4,2.

iii) An experiment (4,4) in which eight male subjects exerted maximum pulling forces in twelve different combinations of body weight and body support. These results lead to the conclusion that the limiting factor in supported pulling was a muscular one, but a subsidiary experiment (4,5) was required to eliminate the possibility that the weak link in such an action was the subjects' strength of grip.

iv) In a final experiment (4,6) an attempt was made to demonstrate an angle-torque relationship for the trunk extensor mechanism which is believed to be the weak link in supported pulling tasks.

4,2 THE STRENGTHS, WEIGHTS AND STATURES OF NINETY-TWO MEDICAL STUDENTS.

Studies of relationships between "muscular" strength and many anthropometric and demographic variables have been reviewed above (1,2,4). It may be hypothesised that such correlations as exist between strength and either weight or stature would be more pronounced for "gravitationally" limited strength tests, than for tests limited by muscular capacity (1,3). A correlational study of stature, body weight, and five measures of strength was performed using as subjects ninety-two medical students (49 male, 43 female). These students ranged in age from 18 to 27 years and were of diverse racial and cultural backgrounds. The group measured represented virtually the whole of one year's intake of students and were not selected in any way. The age distribution of the students showed a very pronounced positive skew.

4,2,1 METHODS.

The variables measured were stature, body weight, maximum voluntary pronator torque, maximum voluntary supinator torque, and maximal horizontal forces which could be exerted on bars set 0.5, 1.0 and 1.5 metres above the ground.

a) Apparatus.

Stature was measured using a standard stadiometer. Measurements were rounded to nearest half centimetre.

Body weight was measured using standard weighing machines, the calibration of which had been recently checked. Measurements were rounded to the nearest half kilogram force.

Pronation and Supination torques were measured using the apparatus described in section 3,2,1(a). A stirrup handle was set in vertical orientation (i.e. the subject's hand was mid-way between prone and supine). The load cell was powered from a variable D.C. supply (of local construction) and the bridge output voltage was measured using a moving coil galvanometer (Tinsley SS6.45) calibrated to give a direct reading in kilogram-force centimetres. The measurement was recorded to the nearest kilogram-force centimetre, the maximum steady reading being estimated by eye.

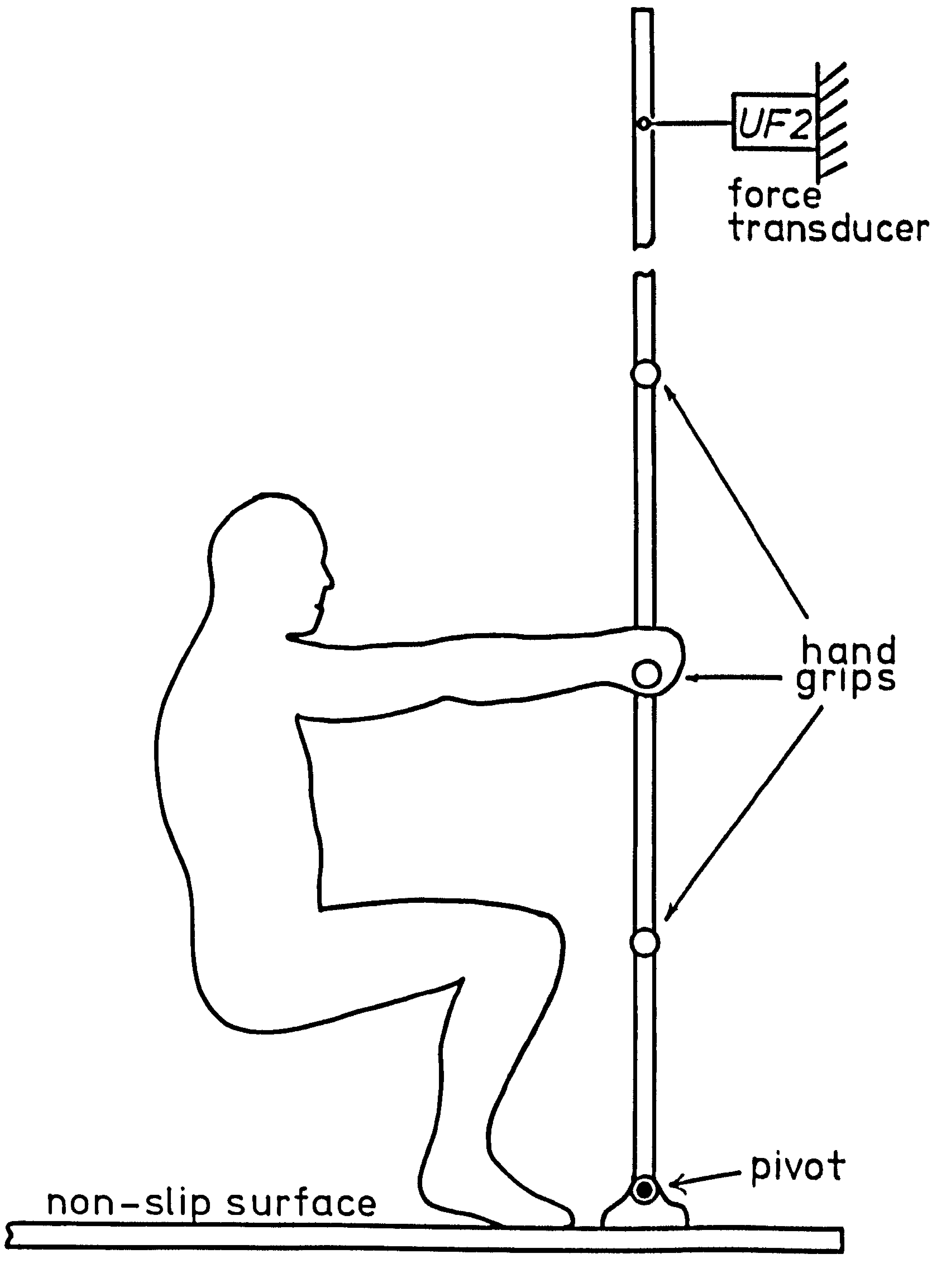
Horizontal pulling forces were measured using the rig shown in Fig. 4,5. The framework constructed of "Dexion" pivotted on a Picador speed shaft, the twin bearings of which were mounted on a base board. An area of the base board 0.25 M X 0.8 M was covered in coarse emery paper to provide a non-slip floor surface. Three hand grips of iron pipe covered with rubber were set on the framework at heights of 0.5, 1.0 and 1.5 metres from the ground and the top of the structure engaged the load cell (Pye ether type U.F.2) via a toggle coupling. Load cell output was amplified using a simple solid state D.C. amplifier of local design and recorded using a U.V. galvanometer (S.E. labs type 3006). Each hand grip bar was calibrated directly for pulling forces by means of weights acting over a pulley.

b) Procedure.

The author and two assistants, recorded data from all 92 subjects during a single working day. The subjects entered the laboratory, and initially took part in either the pronation/supination, or the pulling experiment on a random (self-selecting) basis.

The pronation/supination test was performed as follows. The subject approached the rig, and it was adjusted on its scaffolding plinth so that the subject's arm was vertical and his forearm horizontal. The subject

Figure 4,5



Sketch of apparatus for measurement of horizontal pulling forces.

4,2,1 (a)

then made four maximal voluntary efforts, each against a count of five. The efforts were pronation with the right hand, pronation with the left hand, supination with the right hand and supination with the left hand. The four efforts were made in a bias-free order.

The subject taking part in the horizontal pulling test was asked to pull "as hard as possible, in a horizontal direction, steadily without jerking". The subject was given a free choice as regards to posture and placement of feet. The subject performed on each handgrip against a count of five. The order of presentation of hand grips was bias-free.

Rest pauses were necessarily of a short duration, but as the tests were not arduous it is unlikely that fatigue had a significant effect on the results. Sources of variance in the results will be discussed again at a later stage.

4,2,2 RESULTS.

In the case of both the pronation and the supination torques, the mean performance for the right and left hands was calculated. In the present analysis, both pronation and supination torques are considered positive. The seven variables for each subject were punched onto Hollerith cards together with the sex of the subject. The entire data set could then be interrogated by computer. Many questions could be asked of such a data set. However, the present investigation was limited to factors closely related to the central line on enquiry of this thesis. A FORTRAN programme was therefore written by the author which performed the following statistical investigations:

i) Means and standard deviations of each variable.

ii) Product-moment correlation coefficients between all possible pairs of variables.

iii) Least squares linear regression between all possible pairs of variables.

iv) First order partial correlation coefficients with first stature and then weight held constant.

v) Second order partial correlation coefficients with both stature and weight held constant.

Details of these statistical methods are to be found in Appendix 2.

The programme was run three times, so as to analyse the data for male and female subjects first separately and then together. The following abbreviations will be used in the tables below:

St	=	Stature in metres
W	=	Weight in Newtons
F1.5	=	Horizontal force in Newtons exerted at 1.5 metres from the ground.
F1.0	=	Horizontal force in Newtons exerted at 1.0 metres from the ground.
F0.5	=	Horizontal force in Newtons exerted at 0.5 metres from the ground.
Pr	=	Torque of pronation (mean of left and right hands) in Newton metres.
Su	=	Torque of supination (mean of left and right hands) in Newton metres.

Tables 4,1 to 4,3 give the results of the simple statistical breakdown of the data.

Table 4,1 Means and standard deviations of Variables

	Males		Females		Males + Females	
VARIABLE	Mean	S.D.	Mean	S.D.	Mean	S.D.
St	1.77	.05	1.65	.07	1.71	.09
W	663	79	562	68	616	89
F1.5	251	35	176	38	216	52
F1.0	391	59	268	57	333	84
F0.5	564	103	350	84	463	143
Pr	7.18	1.95	4.20	0.99	5.78	2.17
Su	8.65	1.62	4.91	1.05	6.90	2.32
Age	20.3	1.9	19.7	1.7	20.1 ~~12.0~~	1.8

Table 4,2 Zero Order Product-Moment Correlation Coefficients.

A. Male Subjects

	St.	W.	F1.5	F1.0	F0.5	Pr	Su
St.							
W.	.48						
F1.5	.48	.73					
F1.0	.52	.74	.83				
F0.5	.36	.57	.68	.76			
Pr	.29	.34	.28	.31	.11		
Su	.17	.41	.37	.30	.18	.53	

* n = 49 ∴ r = 0.29, p = 0.05
r = 0.37, p = 0.01

B. Female Subjects

	St	W.	F1.5	F1.0	F0.5	Pr	Su
St.							
W.	.58						
F1.5	.61	.73					
F1.0	.46	.60	.85				
F0.5	.24	.32	.58	.76			
Pr	.16	.31	.41	.46	.56		
Su	.24	.43	.46	.36	.37	.61	

* n = 43 ∴ r = 0.30, p = 0.05
r = 0.38, p = 0.01

C. Male and Female Subjects

	St.	W.	F1.5	F1.0	F0.5	Pr	Su
St.							
W.	.71						
F1.5	.79	.82					
F1.0	.76	.80	.92				
F0.5	.68	.68	.83	.89			
Pr	.61	.58	.65	.67	.63		
Su	.67	.66	.74	.72	.70	.79	

* n = 92 ∴ r = 0.21, p = 0.05

r = 0.27, p = 0.01

* Significance levels quoted refer to the hypothesis that r = o.

4,2,3 DISCUSSION OF STATISTICAL ANALYSIS.

The sex differences shown in table 4,1 are of the order which would be anticipated from published data (1,2,4,2). Considering the correlation matrices of table 4,2, we see substantially higher correlation coefficients for the grouped male and female data than we do for either sex taken alone. This is to be expected if we consider the proposal that the relationship between these variables is independent of sex. Correlation coefficients are always reduced in a sample in which one or other of the variables is limited to part of its total range. Before progressing with more sophisticated analysis of the data, it is necessary therefore to establish whether the relationship between relevant pairs of variables follows a common trend for both sexes. An appropriate test in this case is to find whether significant differences exist between the slopes of the regression lines for the male and female data of a given pair of variables. (As the means are substantially different and r is less than 1, the regression lines cannot be coincident, but at best parallel if a constant relationship between the variables holds regardless of sex.) The t-test used for such a purpose is described in Appendix 2,3,1. Results of these tests are given in Table 4,3.

Table 4,3

Tests for Difference in Slope Between Regression Lines for Male and Female Subjects.

Variables: Dependent	Independent	t			
F1.5	Wt	1.06			
F1.0	Wt	0.37			
F0.5	Wt	1.50			
Pr	Wt	0.86			
Su	Wt	0.44			
F1.5	St	0.07			
F1.0	St	1.17			
F0.5	St	1.33	♀	*	
Pr	St	1.59	♀	*	
Su	St	0.37	♀	*	♂*
St	Wt	0.74			

In all cases $p > 0.05$

* Variable pairs marked thus have regression coefficients which are not significantly different from zero ($p > 0.05$)

In no case does the difference in slopes of regression lines reach the 5% level of significance and we may say with certainty that the regression lines for the male and female subjects are parallel. However, the above statistical test does not investigate the possibility that the two regression lines may be parallel, but separated to an extent which may not be explained by the differences in means of the independent variable. The appropriate test of this possibility is the analysis of covariance (Appendix 2,3,1). This procedure was performed and the results are shown in table 4,4.

Table 4,4 Analysis of Covariance in Regression Lines for Male and Female Data.

Variables: Dependent	Variables: Independent	t		
F1.5	Wt.	5.98		
F1.0	Wt	6.35		
F0.5	Wt	10.99		
Pr	Wt	5.71		
Su	Wt	9.19		
F1.5	St	3.86		
F1.0	St	4.34		
F0.5	St.	8.97	* ♀	
Pr	St	4.81	* ♀	
Su	St	8.00	* ♀	* ♂
Wt	St	0.12		

t = 1.99; P = 0.05

t = 3.42; P = 0.001

* Variable pairs marked thus have regression coefficients which are not significantly different from zero. (P > 0.05)

We may deduce therefore that for a given stature or weight the male subjects are stronger than the female subjects on all five tests of performance. There is no difference in weight for a given stature. Armitage (1971) discusses the pitfalls of drawing separate regression lines through sub-samples when a common trend exists. Visual inspection of scatterplots of this data (Figs. 4,6 - 4,10) gives the strong impression that such trends do exist.

It has been claimed (Asmussen and Heebøll-Nielsen 1961) that strength varies with the square or cube of stature. This contention was argued on theoretical grounds although evidence based on the very different case of adolescent growth curves was cited. It was therefore deemed appropriate to perform a power curve regression on the data by means of the logarithmic transformation of both variables. A best fit equation of the form $y = ax^b$ was thus generated for each pair of variables. In no case did this procedure produce a better fit to the data than did a simple linear regression and in some cases the curvilinear fit was slightly worse. The possibility of a curvilinear relationship between strength and either stature or weight may be eliminated from the present data set. In a population which included juveniles and adolescents the result of course might have been different, a wider range of the variables having been considered.

Figures 4,6 - 4.10

Results of Experiment 4,2 - Measurements made on 92 medical students.

Figures 4,6 and 4,7

Scatter plots of the strengths horizontal pulling against body weight and stature respectively.

A - at 0.5 metres from ground

B - at 1.0 metres from ground

C - at 1.5 metres from ground.

Figures 4,8 and 4,9

Scatter plots of the strengths of pronation and supination against body weight and stature respectively.

Figure 4,10

Scatter plot of body weight and stature.

In each case the male subjects are denoted by filled circles and the females by open circles. The regression lines for the pooled male and female data are shown.

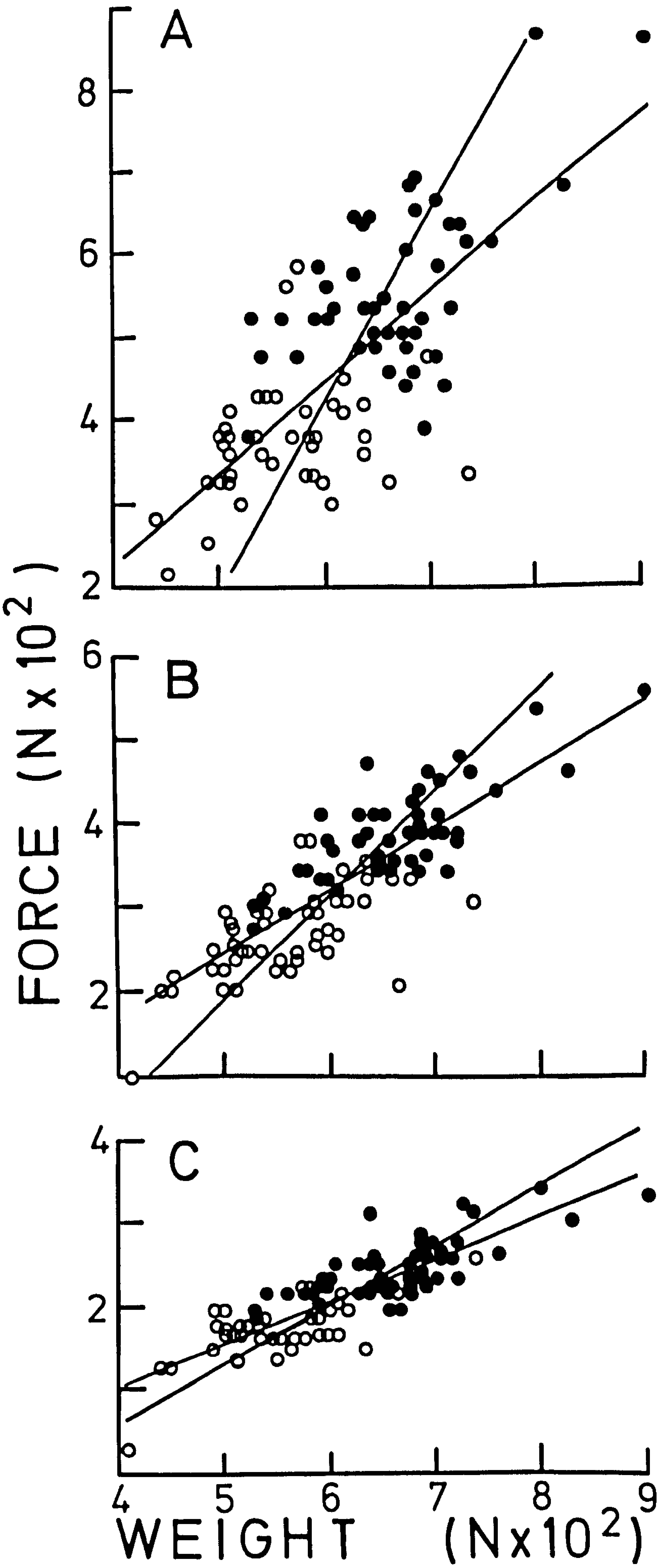
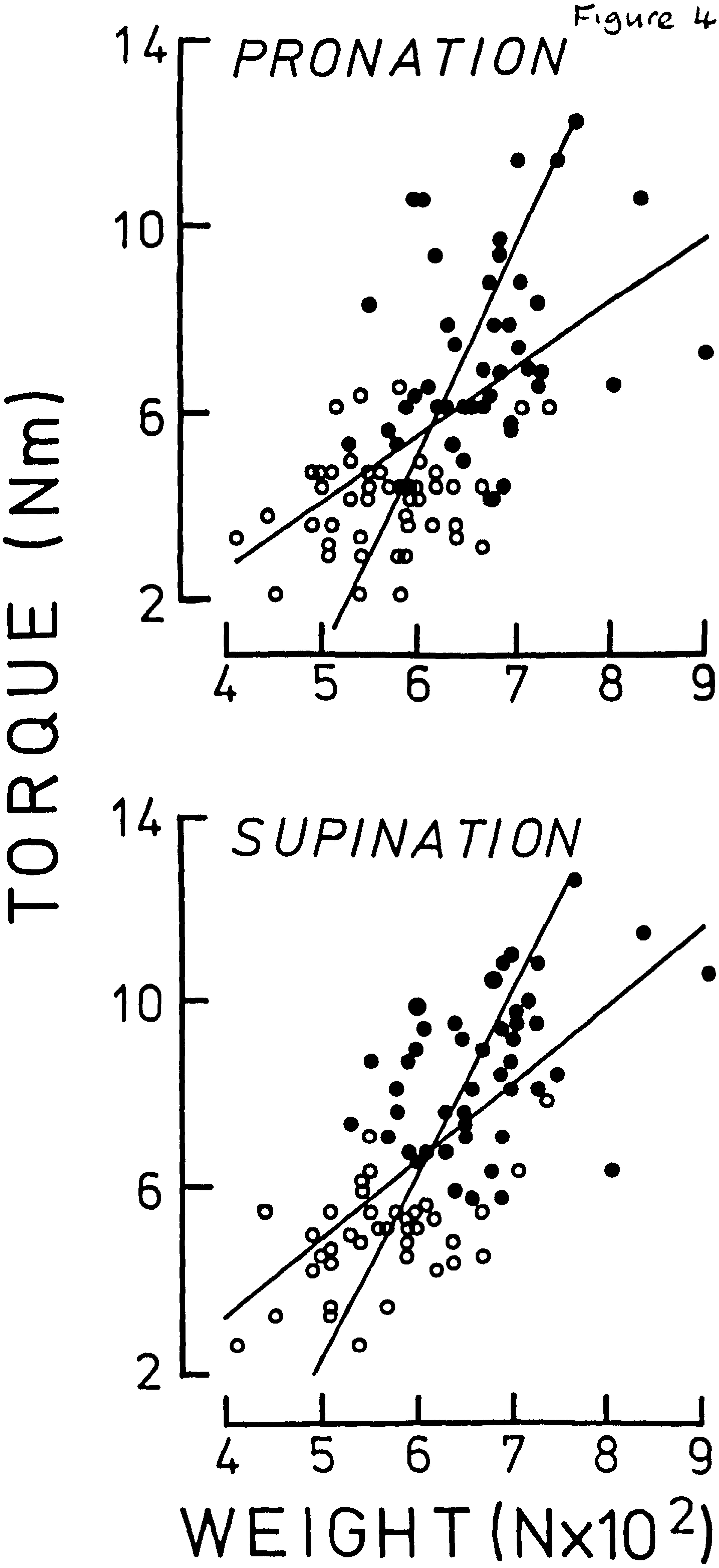

A
B
C
FORCE (N x 10²)
WEIGHT (N x 10²)
8
6
4
2
6
4
2
4
2
4
5
6
7
8
9

Figure 4,7

Figure 4.8



PRONATION
SUPINATION
TORQUE (Nm)
14
10
6
2
4
5
6
7
8
9
WEIGHT (Nx10 2)

Figure 4,9

Figure 4.9

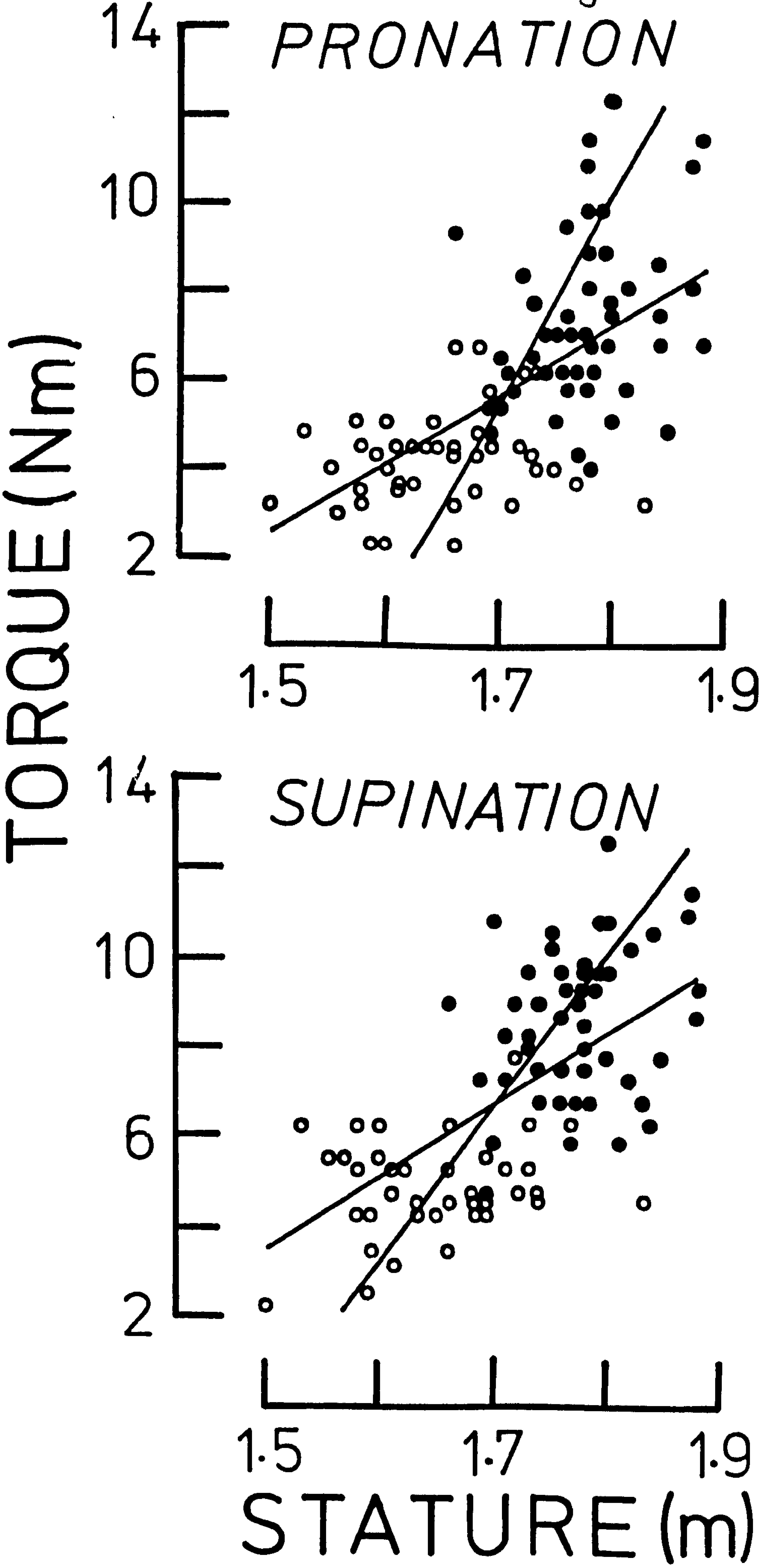
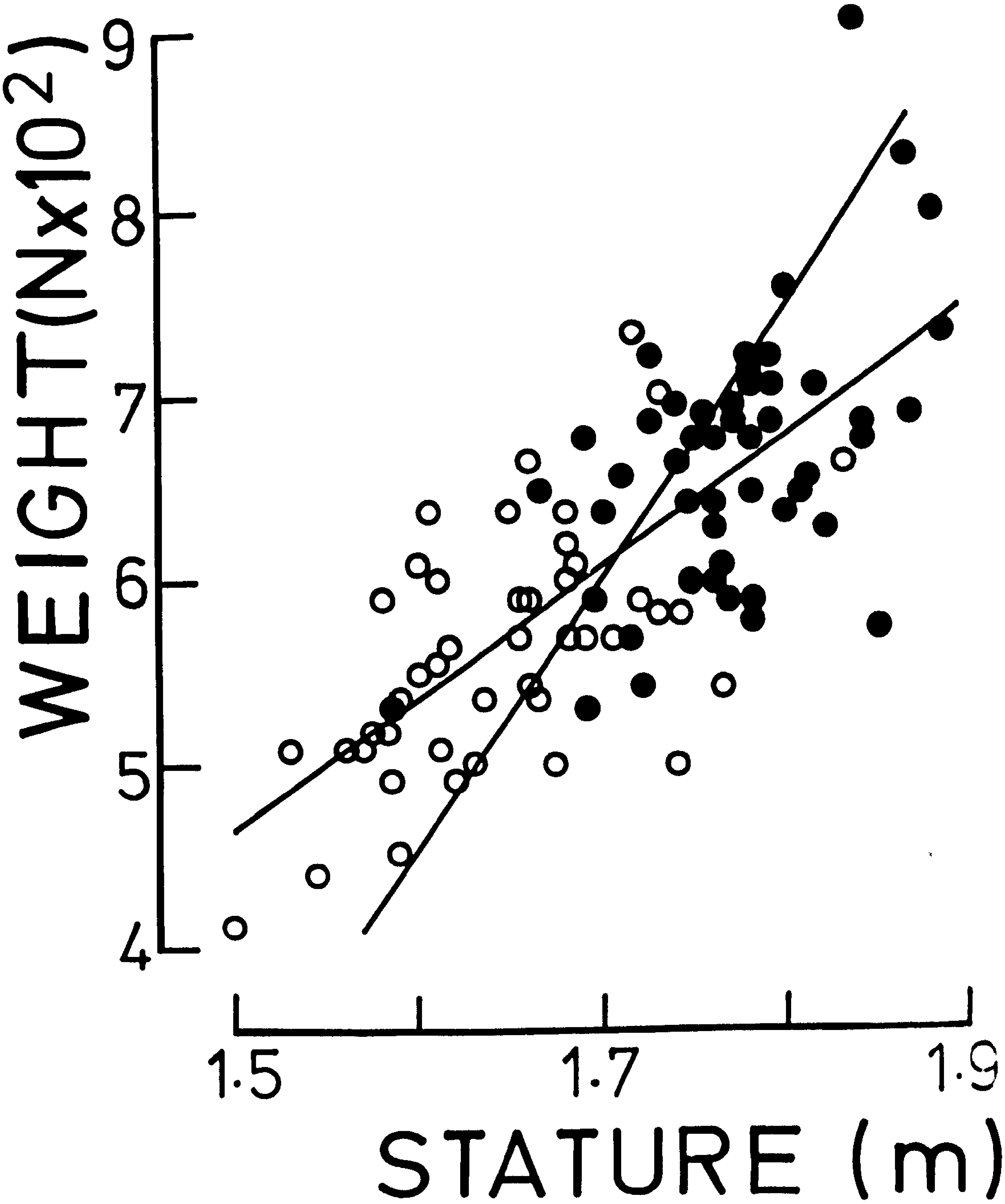


Figure 4,10



In the situation described by Dempster in which the force a man may exert is determined by body weight and leverage, it is reasonable to anticipate a high correlation between body weight and applied force. Furthermore, stature confers an advantage by giving the subject greater possibilities for the effective disposition of his centre of gravity. Several authors have described modest but significant correlations between strength and weight in situations where the strength limitations were certainly muscular ones (1,2,4). It has been postulated therefore that strength, weight and stature are all reflections of a non-specific factor of bodily development. (Roberts, Provins and Morton,(1959) refer to this as the "common size factor"). Initial inspection of the zero order correlation matrices confirms that there are higher correlations between the pulling forces and stature and weight than there are between the torques and stature and weight. However, these relationships are confounded by the fact that stature and weight are themselves significantly correlated. It is necessary therefore to extract partial correlation coefficients to clarify this relationship (Appendix 2,3,2). Tables 4,5 - 4,8 show the results of these calculations.

Table 4,5 <u>First Order Partial Correlation Coefficients with Stature Held Constant</u>

A. Male Subjects.

	W	F1.5	F1.0	F0.5	Pr	Su
W						
F1.5	.65					
F1.0	.65	.78				
F0.5	.49	.62	.71			
Pr	.24	.17	.19	0.0		
Su	.38	.34	.25	.13	.51	

$r = 0.29$; $p = 0.05$
$r = 0.37$; $p = 0.01$

B. Female Subjects.

	W	F1.5	F1.0	F0.5	Pr	Su
W						
F1.5	.58					
F1.0	.46	.80				
F0.5	.22	.57	.75			
Pr	.27	.40	.44	.55		
Su	.38	.41	.29	.33	.60	

$r = 0.30$; $p = 0.05$
$r = 0.39$; $p = 0.01$

C. Male + Female Subjects.

	W	F1.5	F1.0	F0.5	Pr	Su
W						
F1.5	.61					
F1.0	.56	.81				
F0.5	.39	.65	.78			
Pr	.27	.34	.40	.36		
Su	.36	.47	.43	.45	.65	

Table 4,6 First Order Partial Correlation Coefficients with Weight Held Constant

A. Male Subjects

	St.	F1.5	F1.0	F0.5	Pr	Su
St.						
F1.5	.21					
F1.0	.27	.63				
F0.5	.12	.47	.61			
Pr	.15	.06	.09	- .11		
Su	-.03	.12	.00	- .06	.46	

B. Female Subjects

	St.	F1.5	F1.0	F0.5	Pr	Su
St.						
F1.5	.34					
F1.0	.18	.74				
F0.5	.07	.54	.75			
Pr	- .03	.28	.36	.52		
Su	- .02	.23	.44	.27	.55	

C. Male + Female Subjects.

	St.	Fl.5	Fl.0	F0.5	Pr	Su
St.						
Fl.5	.51					
Fl.0	.46	.78				
F0.5	.38	.65	.78			
Pr	.35	.37	.42	.38		
Su	.38	.47	.42	.46	.66	

r' = .21 ; p = .05

r' = .27 ; p = 0.01

Table 4,7 Second Order Partial Correlation Coefficients with Both Height and Weight Held Constant.

A. Male Subjects.

	F1.5	F1.0	F0.5	Pr	Su
F1.5					
F1.0	.61				
F0.5	.46	.60			
Pr	.03	.05	- .13		
Su	.13	.01	- .06	.47	

B. Female Subjects

	F1.5	F1.0	F0.5	Pr	Su
F1.5					
F1.0	.74				
F0.5	.55	.75			
Pr	.31	.37	.52		
Su	.25	.14	.27	.55	

C. Male + Female Subjects.

	F1.5	F1.0	F0.5	Pr	Su
F1.5					
F1.0	.71				
F0.5	.57	.74			
Pr	.23	.31	.29		
Su	.34	.30	.36	.61	

$r = .21$; $p = 0.05$

$r = .27$; $p = 0.01$

In the case of the mixed sex data, all first and second order partials are significant, whereas in the single sex matrices, areas of insignificance appear. The overall pattern present in the zero order matrices is in all cases confirmed, and as the anthropometric variables are eliminated, the separation of the two hypothetical classes of strength test progresses.

Rather than attempting a detailed analysis of these matrices as they stand, it is of greater relevance to the present study to estimate the importance of stature and weight in determining the results of the five different tests of strength. The square of the correlation coefficient is equal to the proportion of the total variance of the independent variable which can be explained by variance in the dependent variable. If we calculate the sum of the squares of the zero order correlation of strength against weight and the first order partial correlation of strength against stature with weight held constant (or vice-versa) a quantity results which is an unbiased and unconfounded estimate of the proportion of the variance in strength which may be accounted for by weight and stature together. This quantity has been provisionally named the dead-weight factor (D-W.F.).* Its values for the present data set are shown in table 4,8.

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* The Shorter Oxford English Dictionary (3rd Edit.) defines the word "dead-weight" thus: "The heavy unrelieved weight of an inert body."

In the case of the mixed sex data, stature and weight account for substantial proportions of the variance in F1.5 and F1.0, but much less of the variance in other tests. The same overall tendency is displayed in the single sex data but the figures are smaller for the reasons discussed above.

Table 4,8 Dead-Weight Factors in Measures of Strength

	A	B	C
F1.5	.57	.65	.93
F1.0	.62	.39	.85
F0.5	.33	.10	.60
Pr	.14	.10	.46
Su	.17	.18	.58

A - Males alone

B - Females alone

C - Males + Females

These results therefore strongly suggest that there is a fundamental difference between the pulling forces and the forearm torques. As an horizontal pulling force is applied nearer to the floor the extent to which it is determined by weight and stature is reduced. This reduction is systematic for the female and mixed data, but less systematic for the males. If the force is applied 0.5 m from the ground, weight and stature play a role in the determination of performance which is only marginally greater than they do for pronation and supination by virtue of the "common size factor". The dead-weight factor is a potentially important parameter describing tests of strength. Although calculations of this precise nature have not, to the author's knowledge, appeared in the literature, there are many published studies in which tables of zero or first order correlations may be used to calculate dead weight factors for different tests. Table 4,9 shows a selection of such sources from the literature together with the present data. All studies located by the author have been performed on male subjects, with the exception of Troup and Chapman (1969) who quoted data for male and female subjects separately. (The data of Troup and Chapman which has been used is that of forces as these are more obviously comparable to the present tests). F1.5 and F1.0 stand out clearly from all other investigated strength tests, but F0.5 falls within the range of the other tests.

Table 4,9 Dead-weight Factors in Published Tests of Strength.

Source	Test Performed	Subject Group	D-W. F
Present Study	F1.5	49 male students	.57
	F1.0		.62
	F0.5		.33
	Pr		.14
	Su		.17
Troup & Chapman (1969)	Trunk Flexion (Standg.)	98 male students & student teachers	.26
	Trunk Extension "		.24
	Trunk Flexion (sitting)		.29
	Trunk Extension "		.39
Lanbach (1969)	Trunk Flexion	48 male students & Air Force Personnel	.28
	Trunk Extension		.39
	Hip Flexion		.45
	Hip Extension		.22
Roberts, Provins & Morton (1959)	Grip	62 Male Naval Ratings	.30
	Elbow flexion		.30
	Elbow Extension		.49
Present Study	F1.5	48 Female students	.65
	F1.0		.39
	F0.5		.10
	Pr		.10
	Su		.18

(contd.)

Table 4,9 (contd.)

Source	Test Performed	Subject Group	D-W. F
Troup & Chapman (1969)	Trunk Flexion (standg.)	132 Female students	.07
	Trunk Extension "		.15
	Trunk Flexion (sitting)		.08
	Trunk Extension "		.18

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Consider the hypothesis of section 4,1 described in equations 11 to 14. Although the location of the foot pivot point is not known in these measurements, the subjects having a free choice in this respect, it is reasonable to assume that as the hand grip is lowered, the live axis will become more nearly horizontal. Force measured in the horizontal direction therefore has a smaller stature-weight contribution. The experimental findings therefore support the hypothesis of live and dead axes.

It is interesting to speculate as to the sources of variance in performance other than the stature-weight contribution. In the cases of the pulling tasks these must include foot-placement high on the list. Ayoub and McDaniel (1974) have shown foot placement to be a major determinant in measures of this nature, confirming the predictions of the theoretical analysis. Although the subjects were requested to pull horizontally it is likely that an unmeasured vertical component existed, and as the analysis of 4,1 shows, the interaction between vertical and horizontal components is indeterminate. A further source of variance is the extent to which each individual approaches the optimal deployment of his or her body mass - a factor which represents skill and experience in the performance of the task.

In the case of tasks in which the contribution of stature and weight is low, a major portion of the remaining variance must be such elements of muscular development as are not associated with the "common size factor", e.g. the development of specific muscle groups through training.

In all tests it is likely that some variance is attributable to motivation and other such Type IV or random noisy factors. It is of interest that in most tests the dead-weight factor is smaller for the females than it is for the males. This is possibly due to a greater presence of noise of this kind in the female data. A generally higher motivational level in the males may also account for the parallel shifting of the regression lines.

Some attention must be given to the ages of the subjects. The range is such that the bulk of the subjects have reached a height very close to their maximum (Tanner, 1962; Stoudt et al, 1960) whereas weight is still increasing (ibid) as is the strength of many muscle groups. (Asmussen and Heebøll-Nielsen, 1961). Age differences must therefore have some effect on the distribution of weight and strength and of any interactions in which they take part. As the distribution of age is so highly skewed in the subject sample, it does not seem justifiable to introduce it into the analysis as a separate variable.

4,3 THE STRENGTHS, WEIGHTS AND STATURES OF SEVENTY THREE MEDICAL STUDENTS.

The results of the experiments described in section 4,2 are in general agreement with the theoretical analysis of section 4,1. It seemed desirable to make more detailed measurements to check this correspondence. The availability of a new intake of medical students made such a study feasible.

4,3,1 METHODS.

The most conspicuous source of uncontrolled variance in the previous data set was the uncontrolled placement of the feet in the pulling tasks. In the present study, the subjects were required to line up the anterior margins of their great toes with lines marked on the floor. (Although this procedure standardises experimental conditions, it does not fully control the effective centroid of foot contact. The location of this centroid may be measured using a force platform of the type described by Whitney (1958) or by a variety of simpler devices generally referred to as balance boards or stabilometers. A platform for use in the present experiments would have to be sufficiently robust to tolerate considerable forces exerted tangential to its surface. It is regretted that such a device was not available. An alternate procedure is to control this centroid by locating the subjects' feet upon a pivot. It was considered that such a procedure would be unacceptable to naive subjects. It was therefore concluded that the approximate control adopted was the most desirable procedure under the present circumstances.)

The following quantitites were measured on each of the subjects. Thirty subjects were male and 43 were female. The male and female groups of subjects both ranged in age from 18 to 27 years.

Stature and weight were measured as described in section 4,2,1.

Trochanteric height was measured as the distances from the ground to the most prominent palpable points of the greater trochanters of the right and left femurs when subject was standing fully upright. The two measurements were made using the stadiometer, the mean was taken and corrected to the nearest half centimetre.

Strengths of Horizontal Pulling were measured using the apparatus described in section 4,2,1 and shown in figure 4,4. The subjects each exerted a total of six maximal steady pulling efforts. The hand grips were placed as before at 1.5, 1.0 and 0.5 metres above the ground and the subjects' toes were located 0.25 and 0.5 metres behind the pivot of the apparatus.

Strengths of vertical lifting were measured on the force bar apparatus used by Whitney (1958). This apparatus will be described in greater detail in a subsequent section (4,4,1). In the present study the subjects were instructed to exert a maximal steady force which was as nearly vertical as they were able. Both vertical and horizontal components of the applied force were measured. Subsequent discussion refers to the vertical components only, unless otherwise specified. The force bar was set 0.5 metres above the ground and the feet were located directly beneath it and at distances of 0.25 and 0.5 metres behind it. The transducer outputs were amplified using the Medelec AD6 amplified and recorded on light sensitive paper using the Medelec system. These force traces were measured using standard procedure (Appendix 1).

Experimental procedure was as described in section 4,2,1.

4,3,2 RESULTS.

A FORTRAN programme was written which calculated the mean and standard deviation of each variable; the zero order product moment correlation coefficients of each variable with stature, weight and trochanteric height; the first order partial correlation coefficient of each strength variable with weight and trochanteric height when stature was held constant; and the dead-weight factor for each strength variable. These analyses were deemed most appropriate in the light of the findings of section 4,2. The programme was run for the 43 females, for the 30 males and for the total group of 73 subjects. These results are shown in Table 4,10.

The means and standard deviations of the horizontal components of the attempted vertical lifting actions were also calculated, together with the angles between the calculated resultant direction of force exertions and the vertical. The results of these calculations are shown in Table 4,11.

All forces in tables 4,10 and 4.11 are expressed according to the sign convention of section 4,1. Angles or deviations of the measured force from the vertical towards the subject are positive.

Table 4,10 A. Male Subjects

Variable	Grasp Height	Foot Placement	Mean	S.D.	Correl. With Stature	Correl. With Weight	Correl. With Troch. Ht.	D.W.F.
Horizontal	1.5	0.25	452	71	.50	.85	.47	.88
Pulling	1.0	0.25	634	111	.27	.73	.30	.59
(N)	0.5	0.25	885	199	.09	.60	.15	.42
	1.5	0.5	404	66	.58	.68	.56	.63
	1.0	0.5	559	85	.47	.76	.53	.69
	0.5	0.5	683	126	.28	.60	.41	.39
Vertical	0.5	0.	2123	433	.10	.44	.19	.21
Lifting	0.5	0.25	751	109	.25	.57	.22	.35
(N)	0.5	0.5	315	79	.17	.37	.15	.14
Stature (N)			1.77	.06		.52	.83	
Weight (N)			651	62	.52		.55	
Troch. Ht. (M)			.93	.04	.83	.55		

n = 30: p = .05, r = .36,
p = .01, r = .46

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Table 4,10 B. Female Subjects

Variable	Grasp Height	Foot Placement	Mean	S.D.	Correl. With Stature	Correl. With Height	Correl. With Troch. Ht.	D.W.F.
Horizontal	1.5	0.25	325	79	.70	.63	.45	.74
Pulling	1.0	0.25	464	98	.65	.66	.46	.72
(N)	0.5	0.25	621	139	.49	.52	.31	.39
	1.5	0.5	278	63	.71	.56	.50	.65
	1.0	0.5	394	95	.58	.56	.39	.51
	0.5	0.5	484	108	.52	.58	.27	.48
Vertical	0.5	0.	1290	273	.25	.30	.09	.11
Lifting	0.5	0.25	530	146	.32	.31	.32	.14
(N)	0.5	0.5	220	78	.42	.51	.14	.34
Stature (M)			1.64	.06		.44	.75	
Weight (N)			569	81	.44		.28	
Troch. Ht. (M)			.86	.04	.75	.28		

n = 43

$p = 0.05$, $r = .30$
$p = 0.01$, $r = .39$

Table 4,10 C. Male and Female Subjects.

Variable	Grasp Height	Foot Placement	Mean	S.D.	Correl. With Stature	Correl. With Weight	Correl. With Troch.Ht.	D.W.F.
Horizontal	1.5	0.25	377	98	.80	.78	.71	.98
Pulling	1.0	0.25	534	133	.73	.76	.67	.85
(N)	0.5	0.25	730	211	.62	.66	.58	.58
	1.5	0.5	330	90	.84	.71	.77	.88
	1.0	0.5	462	122	.77	.73	.72	.83
	0.5	0.5	566	151	.70	.70	.65	.70
Vertical	0.5	0.	1632	538	.65	.56	.62	.49
Lifting	0.5	0.25	621	171	.63	.57	.62	.48
(N)	0.5	0.5	259	91	.57	.59	.46	.46
Stature (m)			1.69	.09		.63	.90	
Weight (N)			603	84	.63		.58	
Troch. Ht. (m)			.89	.05	.70	.58		

n = 73

p = 0.05, r = .23

p = 0.01, r = .30

Table 4,11 HORIZONTAL COMPONENT FORCES (N) OF ATTEMPTED VERTICAL LIFTING ACTIONS.

Foot Position	Males Mean	Males S.D.	Females Mean	Females S.D.	Males + Females Mean	Males + Females S.D.
0	271	223	196	113	227	175
0.25	135	172	122	113	127	139
0.5	76	208	113	173	98	188

ANGLE (deg.) BETWEEN EXERTED FORCE AND VERTICAL

Foot Position	Males Mean	Males S.D.	Females Mean	Females S.D.	Males + Females Mean	Males + Females S.D.
0	8	7	9	5	9	6
0.25	10	12	13	12	12	12
0.5	12	29	20	32	17	31

4,3,3 DISCUSSION.

In earlier sections, an hypothesis has been put forward suggesting that the performance of a group of subjects in a test of strength involving symmetrical exertions in the sagittal plane, will be most highly correlated with body weight and stature in situations where the axis of force measurement is close to the "dead axis" which has been previously defined. The current set of measurements was designed to test this proposition as precisely as was possible within the limitation of available apparatus and facilities. It may be argued that the length of the lower limbs makes a greater contribution to the leverage of the body's centre of gravity than stature *per se*. A subsidiary aim of the experiment therefore was to test whether trochanteric height had a higher correlation with performance than did stature.

For each of the nine conditions tested, it is possible to calculate an angle ϵ which is defined as that angle subtended by the axis of measurement (i.e. vertical or horizontal) and a line in the mid-sagittal plane of the subject joining the handgrip to the marker line for foot placement. This angle is approximately equal to the angle ϵ' between the axis of measurement and the live axis. (As the centre of pressure of the feet is always some small distance behind the line of foot placement, then for pulling tasks ϵ slightly over estimates ϵ' and for lifting tasks ϵ slightly under estimates ϵ'. However, since in both of these activities the centre of foot pressure is commonly near to the toes, then these discrepancies must be modest).

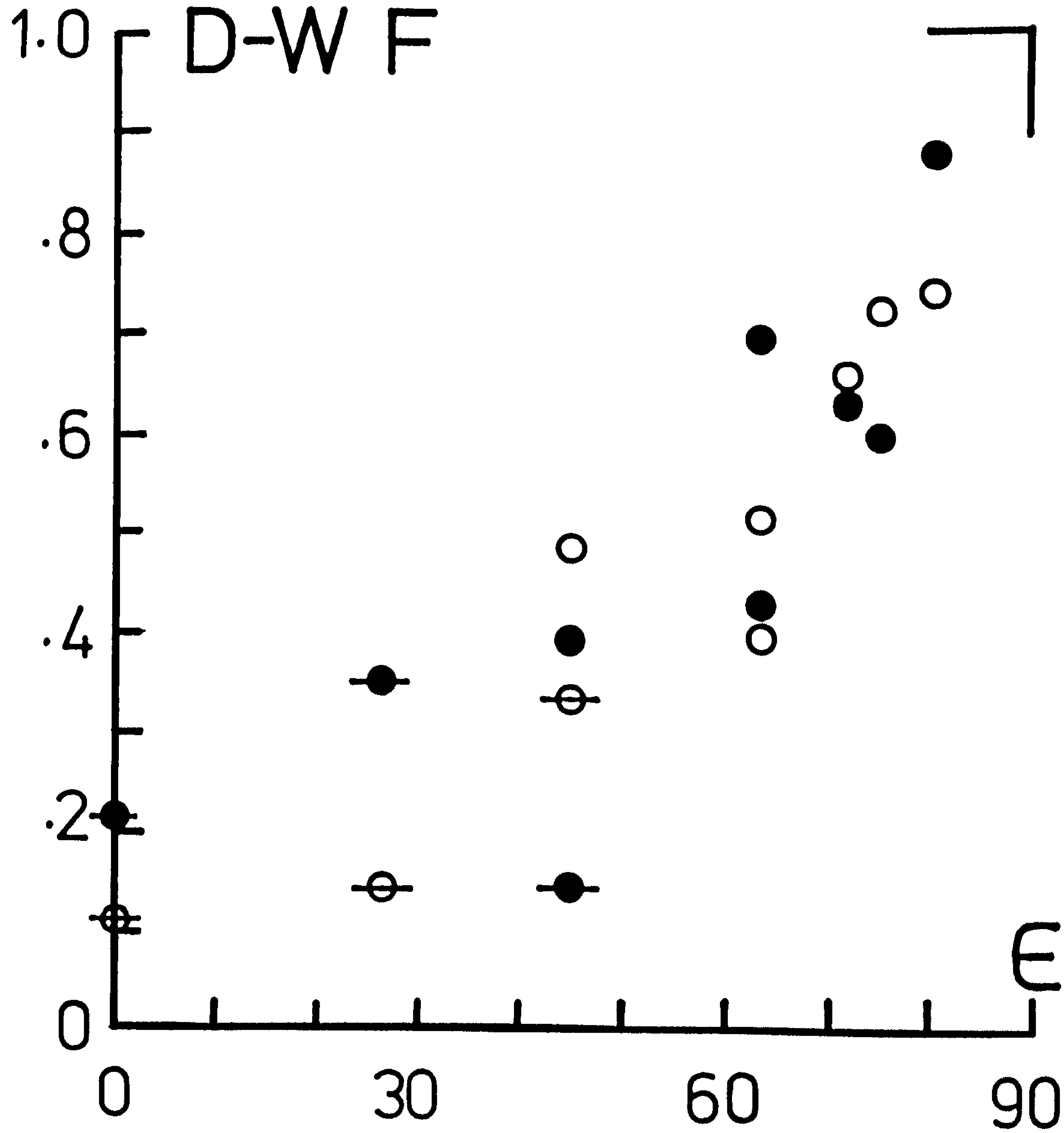
The results of the present experiment enables us to investigate the way in which the dead-weight factor varies with the angle ϵ. Fig. 4,11 is a scatter-plot of these two quantities against each other, and the data on which this figure is based is repeated for convenience in table 4,12. The relationship expected from the theoretical analysis and from the previous study emerges. It is concluded therefore that when the angle ϵ is large (i.e. the axis of measurement is close to the dead axis), then the dead-weight factor is large also. There are, however, certain cases where two task conditions have the same angle ϵ (i.e at 45° and 63°). In these conditions considerable disparities in dead-weight factor exist. It is likely that these disparities may be best explained in terms of constraints placed upon the effective disposition of body weight by the location of the hands and feet. Hence there is a tendency for the pulling tasks with backward foot placement to show a higher D.W.F. than the pulling tasks with forward foot placements and the same angle ϵ.

Figure 4,11

Calculated values of dead-weight factor (D-W.F) plotted against the angle ($\in$) as defined in the text (4,3,3).

●	horizontal pulling male subjects.
○	horizontal pulling female subjects.
—●—	vertical lifting male subjects.
—o—	vertical lifting female subjects.

Figure 4,11



The lifting task in the 0.5 m foot placement shows a very low D-W.F. This may be due to the fact that such an action is perceived by the subjects as being "awkward and unnatural" and that the effective disposition of body weight is inhibited for this reason. An alternative explanation is that in such a posture the externally applied force has a considerable leverage about the articulations of the trunk and shoulder girdle, and that a genuine muscular limitation may supervene at one of these sites. Another pronounced source of variance in performance of the lifting tasks is differences in the true direction of force application, the means and standard deviations of which are given in table 4,11. No explanation of these results will be attempted at the present time, but their existence must be noted along with their high variability.

In no case did the partial correlation of strength against trochanteric height,with stature held constant,reach the 0.05 level of significance. The results have therefore been omitted from Table 4,10. It is concluded that length of lower limb confers no measurable advantage in these tasks over and above the advantage of stature.

Table 4,12

Action	Grip Height	Foot Placement	Angle $\in$ (deg)	D-W.F. Males	D-W.F. Females
Horizontal	1.5	0.25	81	.88	.74
Pulling	1.0	0.25	76	.59	.72
	0.5	0.25	63	.42	.39
	1.5	0.5	72	.63	.65
	1.0	0.5	63	.69	.51
	0.5	0.5	45	.39	.48
Vertical	0.5	0.0	0	.21	.11
Lifting	0.5	0.25	27	.35	.14
	0.5	0.5	45	.14	.34

4,3,4

In summary of this section, it may be stated that the extent to which a man's maximal pulling or lifting force is determined by body-weight and leverage varies with the precise location of the point of force application with respect to the position of his feet. This variation may be partially explained by the hypothesis of live and dead axes, but it must be conceded that other factors are also involved.

4,4 MAXIMUM PULLING FORCES WITH DIFFERENT MODES OF BODY RESTRAINT

A second way in which the limitations of pulling force may be investigated is by varying the manner in which the body of the subject is restrained with respect to the outside world. An experiment was therefore conducted in which the interfaces between the man and his environment were varied in a systematic way.

4,41 METHODS.

a) Apparatus

The measuring device used was the horizontal bar dynamometer which was built in Oxford in the fifties and described by Whitney (1958). This essentially consists of a bar, on which the subject pulls, mounted at either end on blocks carrying steel blades on which strain gauges are mounted. The whole apparatus runs on vertical rails so that the bar may be fixed at a chosen height above the ground. The strain gauges are mounted in such a way as to resolve the applied force into vertical and horizontal components. The signals from the strain gauges were amplified and recorded by two AD6 channels of the Medelec system. The measurement device was calibrated by means of weights in a scale pan which was suspended on a long cord which ran over a pulley attached by a short cord to the transducer bar. Equal vertical and horizontal loads could thus be applied to the bar. The system was found to be linear and free of cross-talk between channels. Furthermore, it did not respond to applied torques, and the output was independent of the point of loading along the axis of the bar. A full description of the transducer system may be found in Whitney (1958).

The entire transducer system was mounted on a cage-like framework of tubular steel scaffolding which was securely bolted to the walls and floor of the laboratory and had a rigid timber base. Two different surface boards, each 1.82 m. x 0.61 m. in area, could be clamped to the timber base of the frame. These were designated the "rough floor" which was covered in course emery paper and the "smooth floor" which was faced with "Formica" plastic veneer. Provision was also made to mount a horizontal bar, parallel to the transducer bar. This bar could be adjusted to any vertical and horizontal location and acted as a pelvic restraint for the subject. It was well padded so as not to cause discomfort to the subjects.

b) Subjects

Eight fit young adult male subjects acted as subjects. The subjects had a mean weight of 670 N with a standard deviation of 113 N and had a mean stature of 1.75 m. with a standard deviation of .08 m.

c) Procedure

Each subject was required to make a total of twelve efforts. The transducer bar was set at 1.5 m., 1.0 m. and 0.5 m. from the floor and the subject pulled unrestrained with the rough floor, unrestrained with the smooth floor, restrained at the pelvis with the rough floor and restrained at the pelvis with the smooth floor. The order of performance was bias free, and the efforts were made according to standard protocol (Appendix 1). The subjects came to the laboratory in two groups of four, and each subject performed either his six restrained tests or six unrestrained tests before resting while the other subjects exerted themselves. The pelvic restraint was set so that the padded portion engaged the anterior superior iliac spines of the subjects and great care was taken to adjust it until the subject pronounced it comfortable.

In all cases the subjects were given free choice with respect to posture, foot placement and direction of pull. The subjects were instructed to pull in whatever direction they considered to be most natural in the given conditions, and it was clearly explained that they should pull as hard as they could without concern for direction.

In order to standardise frictional conditions between feet and floor, all subjects wore their outdoor shoes, but covered them with the same pair of thick woollen socks.

The subjects were all instructed to pull with an overhand grip.

4,4,2 RESULTS.

The sign convention used throughout this section is that described in 4,1. The vertical force (Fv) and the horizontal force (Fh) were measured directly from the experimental paper records and for each individual effort. The direction (θ) and magnitude (F) of the resultant force were calculated by the following equations:-

$$\theta = \tan^{-1}. \frac{Fv}{Fh} \quad ; \qquad (1)$$

$$F = Fh \;.\; \cos\theta \quad . \qquad (2)$$

In addition, for the unrestrained efforts, the apparent coefficient of friction (μ) of the floor and feet was calculated from the equation:

$$\mu = \frac{Fh}{Fv + W} \quad , \qquad (3)$$

where W is the subject's body weight. The relevance of this quantity will be discussed at a later stage.

The mean and standard deviation of the above quantities were calculated for all subjects in each condition. In addition, each subject's values of total force were normalised as a percentage of the mean of his six performances on the rough floor and the group means and standard deviations of the normalised data were calculated. The results of these calculations are shown in tables 4,13 and 4,14 and Figures 4,12 and 4.13.

Figure 4,12

Results of experiment 4,4. Mean values of magnitude (F) and direction (θ) of the resultant force exerted on the transducer bar, plotted against the height of the bar.

——●——	rough floor; pelvic restraint.
——X——	smooth floor; pelvic restraint.
——O——	rough floor; no restraint.
——+——	smooth floor; no restraint.

4,4,2

Figure 4,12

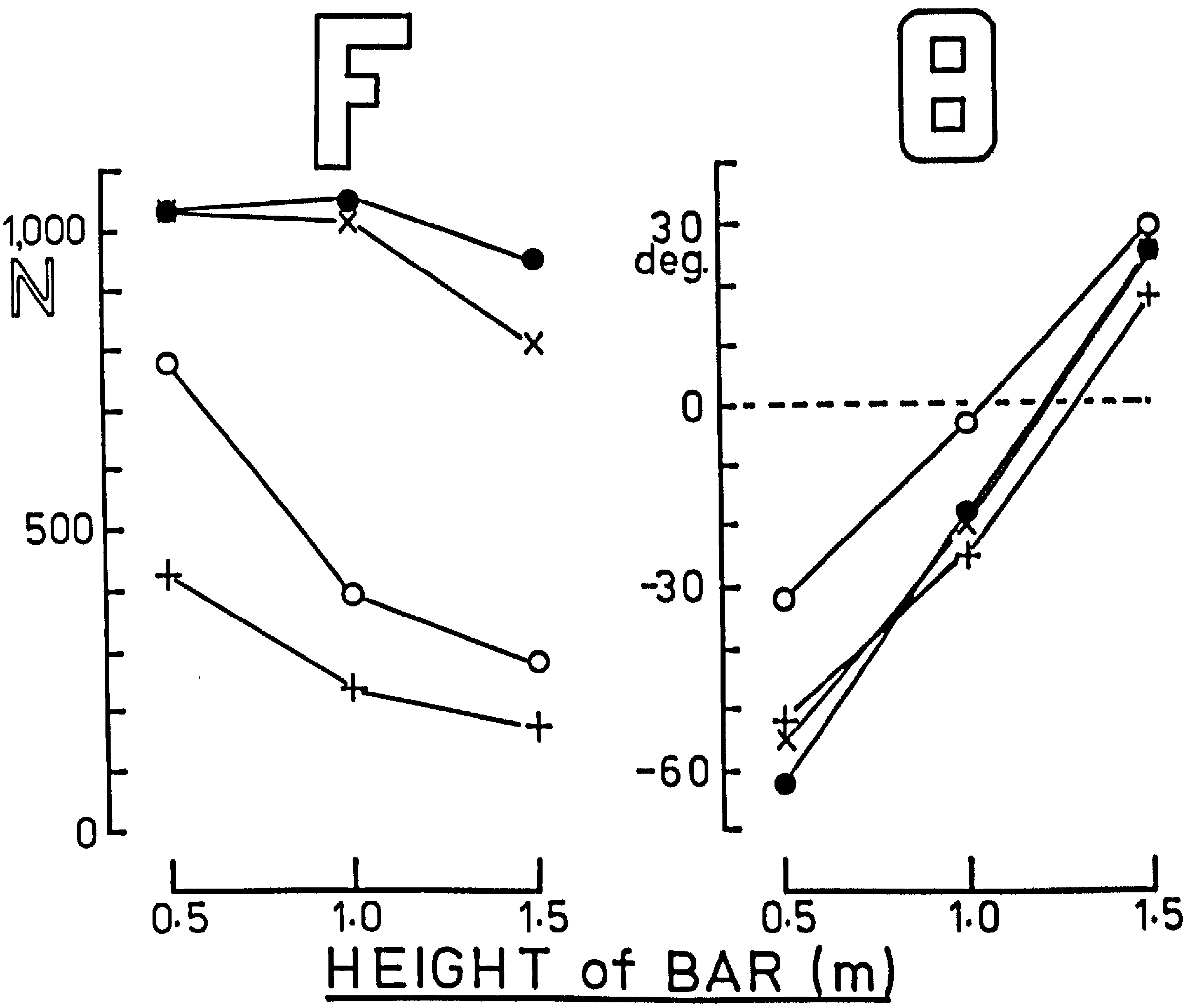
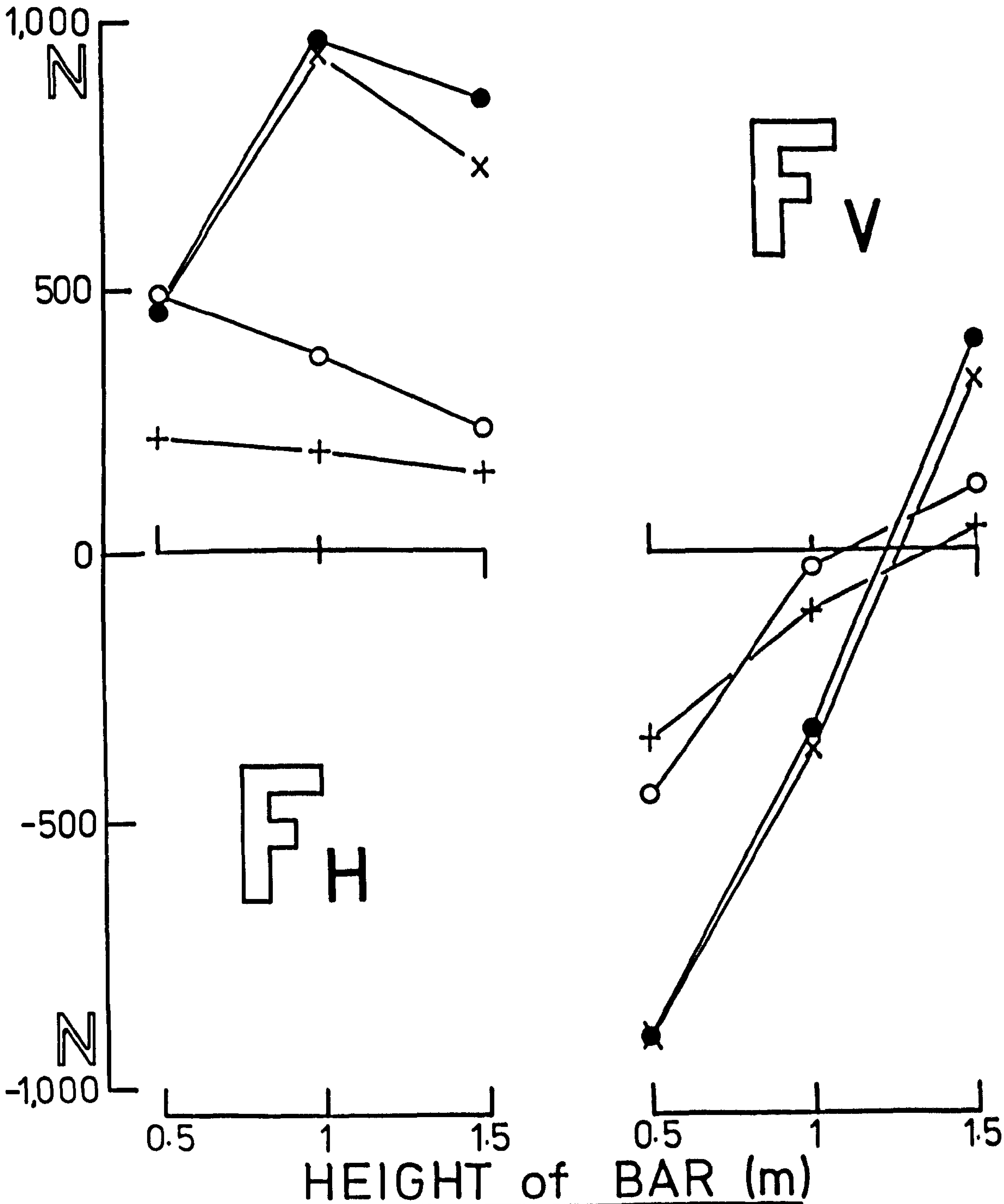


Figure 4,13



Horizontal (Fh) and Vertical (Fv) components of pulling forces exerted in experiment 4,4. Mean forces plotted against height of transducer bar. Plotting symbols as for figure 4,12.

Student's t - test was performed on all possible pairs of values of the resultant force F. The matrix of these results is shown in Fig. 4,14. The following salient points emerge from this analysis.

(a) In the unsupported condition there is always a significant difference between the rough and smooth floors at a given bar height.

(b) In the supported condition there is only a significant difference between the smooth and rough floors when the bar is set at 1.5 m.

(c) In the rough floor supported conditions, differences between bar heights are insignificant except in the case of the 1.5 m., 1.0 m. combination where the difference is marginal.

Student's t- test was performed on the apparent coefficients of friction between feet and floor for all possible pairs of conditions. For a given surface, in no case did the difference between bar heights reach the 0.1 percent level of significance. The data for the three bar heights was therefore pooled for each surface condition and the apparent coefficient recalculated.

Figure 4,14

Levels of significance for t-tests of differences in total force (F) in all combinations of bar height, floor surface and restraint.

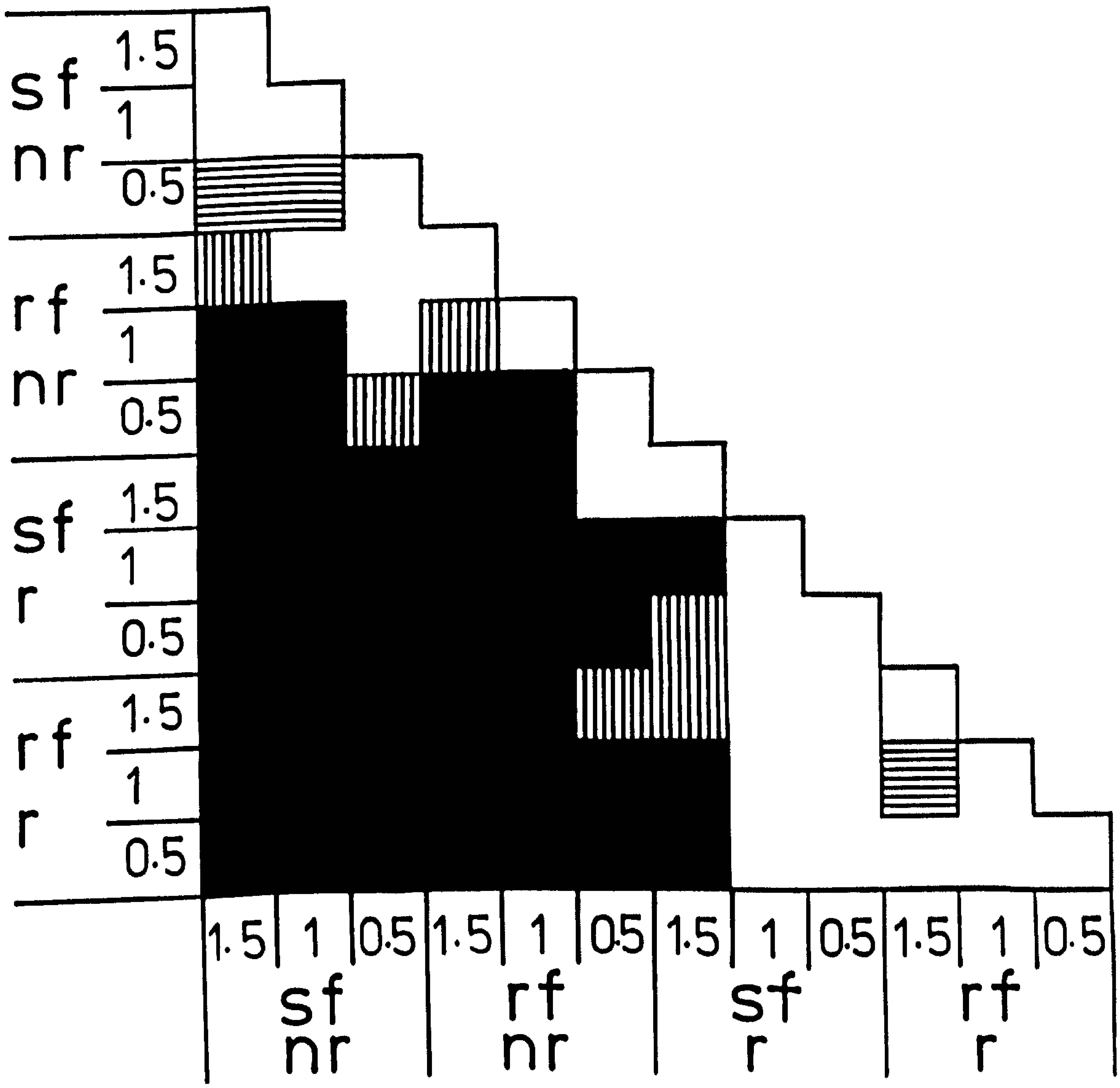
sf - smooth floor

rf - rough floor

r - pelvic restraint

nr - no pelvic restraint.

4,4,2



p > 0·05

0·05 > p > 0·01

0·01 > p > 0·001

0·001 > p

Table 4,13 RESULTS OF MAXIMUM PULLING EXPERIMENTS

Restraint Condition	Bar Height (m)	Vertical Force (N)		Horizontal Force (N)		Angle of Resultant (deg)		Resultant Force (N)		Resultant Force (%)	
		Mean	S.D.	Mean	S.D.	Mean	S.D.	Mean	S.D.	Mean	S.D.
Rough Floor;	1.5	125	83	230	78	28	20	275	76	37	6
No Pelvic	1.0	-36	125	373	119	-3	22	394	117	54	11
Restraint	0.5	-457	412	494	223	-32	30	775	284	93	23
Smooth Floor;	1.5	47	78	150	52	18	24	173	61	24	8
No Pelvic	1.0	-112	113	189	81	-25	22	232	113	31	11
Restraint	0.5	-349	289	219	118	-52	14	421	299	54	28
Rough Floor;	1.5	391	110	852	197	25	8	944	188	130	14
Pelvic	1.0	-333	224	977	224	-19	11	1049	248	143	8
Restraint	0.5	-916	244	457	82	-63	5	1029	225	142	16
Smooth Floor;	1.5	338	97	721	234	26	7	806	220	108	14
Pelvic	1.0	-354	179	934	251	20	7	1011	286	137	12
Restraint	0.5	-916	243	469	59	55	21	1034	227	143	24

Table 4,14 APPARENT COEFFICIENTS OF FRICTION BETWEEN FEET AND FLOOR

Floor Surface	Bar Height	Coefficient for Single Bar Heights.		Coefficient for All Bar Heights.	
		Mean	S.D.	Mean	S.D.
	1.5	0.24	0.06		
Smooth	1.0	0.23	0.06	0.23	0.06
	0.5	0.21	0.05		
	1.5	0.42	0.14		
Rough	1.0	0.48	0.17	0.48	0.20
	0.5	0.54	0.27		

4,5 AN ADDITIONAL INVESTIGATION OF MAXIMUM PULLING FORCES WITH DIFFERENT MODES OF BODY RESTRAINT.

The results described in section 4,4,2 raise several points concerning the limitations of strength in these types of task. These matters will be discussed in a later section (4,7). In the light of the finding that maximum pulling force was largely independent of bar height, it was deemed necessary to perform a second investigation to clarify this single particular point. The above fact strongly suggests that the weak link in a supported pull might be an articulation along the line of action of the pull. The most likely candidate for such a linkage would be the hand grip, with the shoulder girdle as a second possibility. If this were the case, such a method would be quite inappropriate to the study of the intrinsic strength of the trunk. If it could be demonstrated that the upper limb can transmit greater forces than those involved in a supported pulling action it would then be possible to investigate the trunk in this way rather than, for example, by the much less convenient technique of slings around the upper thorax which interfere with respiratory mechanics and thus introduce new possible sources of error. The action of lifting vertically in a posture such that the load is at knee height and the feet are close to the load is known to be a strong one (Table 4,10). This action was therefore directly compared with the former supported pulling action.

4,5,1 METHODS.

The apparatus used in this experiment was identical to that described in 4,4,1. Eight fit young male subjects made two maximal exertions each. The exertions made were a pulling action with the transducer bar set at 1 metre from the ground and a lifting action with the transducer bar set at knee height. The order of presentation of these two tasks was alternated and hence bias free. The subjects were given the same instructions as before for the supported pull and were recommended to keep their feet close to the bar and their trunks as near vertical as possible for the lift. Vertical and horizontal components of the exerted forces were recorded. Exertion of efforts followed standard protocol.

4,5,2 RESULTS.

Vertical and Horizontal components were measured, and the resultant force was calculated. The values of the resultant force were normalised by taking the mean of the two conditions for each subject and expressing each result as a percentage of that mean. The means and standard deviations of the normalised results were as follows:

	Mean (%)	S.D. (%)
Restrained Pull	87.4	11.5
Lift	112.5	11.2

Student's t - test was performed on these results and they were found to be different with a significance level of $p = 0.001$. ($t = 4,14$, d.f. $= 14$). It is concluded therefore that the upper limb linkage is capable of sustaining greater loads than those it transmits in restrained pulling tasks.

4,6 THE RELATION BETWEEN THE EXTENSOR STRENGTH OF THE TRUNK AND ITS POSTURE.

The results of experiments 4,4 and 4,5 indicate that in a pulling task in which the pelvis is restrained, the limitation of strength is a muscular one. It is implicit in the experimental techniques of Troup and Chapman (1969) that the weak link in such an activity is the extensor mechanism of the trunk and the results described above produce no evidence to the contrary. It is therefore of some interest to describe the properties of this weak link which is believed to be located in the lumbar spine.

Extension of the spine is carried out by the erector spinae muscles, assisted by an extensor torque due to increases in intra-abdominal pressure. The erector spinae muscles must exhibit both passive and active length tension characteristics and may change in mechanical advantage with changes in lumbar posture. In addition, ligamentous and other passive structures may exert an extensor counter-torque and support the weight of the head, arms and trunk in fully flexed postures (Floyd and Silver, 1955). Davis and Troup (1964b) have suggested that with an increase in flexion of the spine the extensor muscles have a reduced leverage whereas the centroid of the intra-abdominal pressure has an increased leverage. These two factors therefore in some measure offset each other. Mayer and Greenberg (1942) observed that the actions of extension, flexion and lateral flexion of the trunk were all stronger when the muscles involved were in a "stretched" position than when they were in a "neutral" position, but these terms were not defined in any quantitative way.

The object of the current experiment was to attempt to establish an angle torque relationship for trunk extension analogous to those previously described for plantarflexion, pronation and supination. Because of the much greater anatomical and mechanical complexity of the spine both the conduction and interpretation of this experiment present special problems.

When a man exerts a steady effort, equilibrium conditions must be satisfied at each articulation in the body. The counter-torque demanded from the trunk extensor mechanism has a postural component, and therefore in order to calculate equilibria about any given vertebral level, information is required covering both externally applied forces and the disposition of body segments.

4,6,1 METHODS.

a) Apparatus.

External forces were measured using the apparatus exactly as described in section 4,3 with the padded pelvic restraint in position. The postures of the subjects were analysed from a photographic record.

The present experiment was simplified by the fact that a previous study using videotape (Grieve and Pheasant, 1976) had shown that during such activities the subject's posture is constant within the limits of experimental accuracy. It was therefore decided that a single frame of 35 mm. film was a satisfactory record of the subject's posture during the effort.

A 35 mm single lens reflex camera was set up on a very rigid stand. The camera was an Edixa Reflex B with a 50 mm lens using fine grain film (Panatonic X). The wall behind the apparatus had been painted black and so gave good contrast in the photographs. Previous experience had shown that a major problem causing waste of time and resources and potential major sources of error are inaccuracies in the alignment of the camera. Ideally the optical axis of the camera must be truly normal to the desired plane of analysis, i.e. the sagittal plane of the subject. In the present case this means that the optical axis of the camera must be parallel to the horizontal bar. The following procedure was adopted and found to be exceedingly effective.

Two pre-stretched strings with small weights on the end were suspended precisely 1.5 metres apart in a plane behind that of the subject and perpendicular to the horizontal bar. Plastic markers were attached to these strings so that they formed an exact 1.5 metre square at a known height above the floor. These markers were arranged in such a way as to "frame" the estimated position of the subject. The positioning of the markers was regularly checked for accuracy and constituted a final proof of camera alignment - if they appeared square on the film the camera must have been aligned. A helium-neon gas laser was set up in the estimated centre of the subject and accurately pointed (by means of triangulation) in a direction perpendicular to the plane of analysis as defined by the marker strings. The camera was then placed in the laser beam, its aperture was closed to its smallest stop, its back was opened and a small front silvered mirror was placed in the place of the film. The camera was then adjusted so that the reflected laser beam had the same path as the incident laser beam.

A second well known problem of photogrammetry is that of parallax error. The film was calibrated by suspending a graduated marker rod from the point of the horizontal bar, i.e. in the estimated mid-sagittal plane of the subject. A test film was then taken. On analysis (see below) it was found that the "frame" of four markers showed no detectable deviation from a true square. It was also found that after a simple similar triangles correction for parallax the measured distances between the four markers were equivalent to the measured graduations on the suspended marker rods within the limits of measuring accuracy.

b) Subjects and Procedure.

Eight healthy male volunteers between the ages of 21 and 30 years acted as subjects for the experiment. None of these had any history of injury to or disorder of the vertebral column. The mean height of the subjects was 1.77 m. with a standard deviation of 0.07 m. The mean weight of the subjects was 698 N. with a standard deviation of 107 N.

Each subject took part in the experiment on a separate occasion. After arrival in the laboratory he changed into briefs and was marked up for the experiment as described below. The height of the pelvic support was adjusted until the subject considered it to be comfortable and it was explained to him that his task was to pull steadily on the bar as hard as he could with no concern for the direction of the pull. Each subject performed in the following five conditions:-

i) with the transducer bar set at 75% of the subject's stature and the pelvic restraint 0.45 m. behind the bar:

ii) as (i) but with the restraint 0.3 m. behind the bar:

iii) with the transducer bar at 50% of the subject's stature and the pelvic restraint 0.3 m. behind the bar.

iv) with the transducer bar at 25% of the subject's stature and the pelvic restraint 0.3 m. behind the bar.

v) as (iv) but with the pelvic restraint 0.45 m. behind the bar.

The vertical location of the pelvic restraint was not altered except at the request of the subject.

The subject's posture was not controlled except in as much as he was requested to place his feet equidistant on either side of a white line marked down the middle of the emery floor and his hands equidistant on either side of a mark placed upon the centre of the transducer bar above the line on the floor. This line therefore was held to represent the mid-sagittal plane of the subject for purposes of parallax correction. A point directly below the centre of the transducer bar was clearly marked on the floor to aid analysis of the photographic record.

The subjects performed in each posture in a bias-free order following standard procedure. On the count of three during the five second effort a photograph of the subject was taken.

c) Description of posture.

In the previously described cases of the ankle and forearm the relevant axes of rotation about which to calculate both torque and posture are relatively well defined. In the case of the trunk articulating about the vertebral column no such immediately apparent reference points exist. Davis (1957) and Troup (1968) have discussed the range of movement in the sagittal plane at different intervertebral articulations. A full description of the posture of the trunk would need to take all of these into account. Clearly such a procedure is exceedingly complex and not to be contemplated at the present time. Several authors (Davis and Troup, 1964; Davis *et al*, 1965; Troup *et al*, 1968; Kumar, 1974; Grieve, 1974) have described the posture of the vertebral column by means of protruding marker rods placed on the skin above spinous processes of specified vertebrae. Troup (1968) performed a detailed evaluation of this procedure. For the present study three marker rods were constructed as follows. A base 5 c.m. x 3 c.m. was cut out of thin copper sheet and trimmed and smoothed for comfort. The points were removed from small wood screws and these were mounted in the centre of the base. A 15 c.m. length of white paper drinking straw could then be mounted providing a marker rod which was long enough to accurately define the angle in the photograph, but light enough to remain in position when the base was taped to the subject's skin.

Principal interest in the present study centred on the posture of the lumbar spine. In order to define this the marker rods were placed normal to the skin, overlying the vertebral spines of S1 and T12. A third rod was placed overlying the vertebral spine of C7, but this was not used in the final analysis. The term, lumbar angle, as used in the present discussion is defined as the angle, measured from the photographic record, subtended by the S1 and T12 markers. This angle was positive when the rods were divergent (in flexion) and negative when they were convergent (in extension). It is noted that this sign convention is not the same as that used by Davis *et al* (1965) and by Troup *et al* (1969).

d) *Calculation of Torque*.

Torque was calculated by the method of segmental analysis due to Dempster (1955). This requires a knowledge of the masses and centres of gravity of the body segments above the level of analysis. The data used is taken from Dempster's (1955) published findings for the mean of eight male cadavers, with certain modifications due to the unpublished calculations of R.J. Whitney and D.W. Grieve. Table 4,12 shows the data used in the present calculations. Torques were calculated about a point in the mid-lumbar region as defined in a previous study (Grieve and Pheasant, 1976). This point was marked on the subject with a small piece of black tape 6 c.m. anterior to the tip of the 3rd lumbar spinous process and was held to represent the centre of the L3 - 4 intervertebral disc. The estimated centres of rotation of the gleno-humeral, elbow and wrist joints were marked on the subjects together with the estimated centres of gravity of the head and trunk segments (the subject wore a white bathing cap).

After development of the film, the individual frames were carefully traced from the screen of a P.C.D. Ltd. digital trace reader fitted with an Aldis Tutor projector and film carrier. The tracings were measured using a simple but effective drawing board mounted draughting instrument. Small parallax errors could have resulted from the fact that the skin markers were not precisely in the mid sagittal plane of the subject. A correction was therefore made corresponding to 1/3 of the subject's biacromial width. This figure was considered to represent the mean plane of the markers to within 1 c.m. and it was deemed that further parallax errors were trivial.

Figure 4,15 is a sample tracing from the experiment and the co-ordinates of the measured points are marked taking the mid lumbar marker as the origin. The lumbar torque is given by the equation:

$$\begin{aligned} \mathcal{T}l = {} & 0.081\ W.\ X_H + 0.248\ W.\ X_T + \\ & 0.056\ W\left[0.436(X_E - X_S) + X_S\right] \\ & + 0.032\ W\left[0.43\ (X_W - X_E) + X_E\right] \\ & + (0.012\ W + Fv)\ X_F + F_H\ .\ Y_F \end{aligned}$$

Figure 4,15

Sample tracing of photographic record of subject in experiment 4,6.

L - Mid-lumbar reference point. (approx. L3/4 disc).

T - Centre of Gravity of trunk above L3/4 level.

S - Centre of rotation of shoulder joint.

E - Centre of rotation of wrist joint.

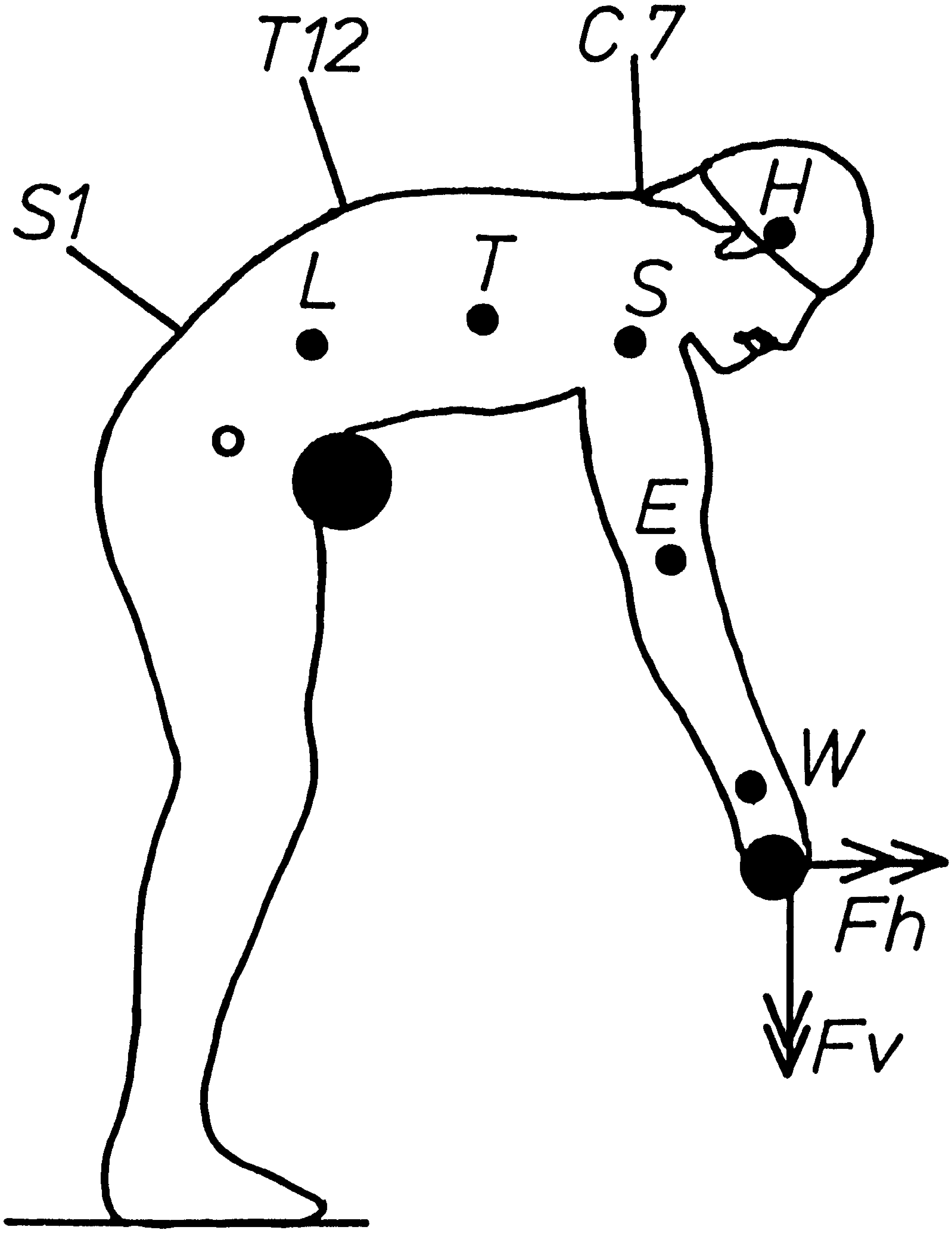
H - Centre of gravity of Head.

C7, T12, S1 - marker rods placed above spinous processes of named vertebrae.

4,6,1 (c)

(d)

Figure 4,15



Where X_H, X_T, X_S, X_E, X_W are the horizontal distances in metres (after parallax correction) between the lumbar marker and the head, trunk, shoulder, elbow and wrist markers respectively.

X_F, Y_F are the co-ordinates of the known centre of the force bar expressed in metres distance from the lumbar marker. Y_F is positive if it is above the lumbar marker and negative if it is below.

W is the body weight in Newtons;

and, $\mathcal{T}\ell$ is the lumbar torque (in Nm) which is positive if it tends to extend the trunk.

Table 4,15

Weights and Centres of Gravity of Body Segments.

Segment	Weight (% Body Weight)	Location of Centre of Gravity
* Head	8.1	6.3% of stature down from Vertex, midway between the glabella and occiput.
* Trunk (above L3-4 disc)	24.8	Mid-way between the L3/4 marker and shoulder joint height in the standing posture. Mid-way between sternum and the tips of the spinous processes at this level.
Arms	5.6	43.6% of arm length from the glenohumeral joint on the line of the link.
Forearms	3.2	43.0% of forearm lenth from the elbow joint on the line of the link.
Hands	1.2	In the position of the horizontal bar.

* These values are modifications of Dempster (1955) based on the unpublished calculations of Grieve and Whitney.

4,6,2 RESULTS.

The data tabulated from the above experiment consisted of five readings of lumbar torque and lumbar angle taken on each of eight subjects. Considerable variation existed in the lumbar angle for any given set of external conditions, and for that reason the methods of data processing adopted in other parts of this study were inappropriate. Three different types of normalisation were attempted. The individual performances of a subject were expressed as a percentage of the mean of all that subject's performances; as a percentage of an interpolated value of torque for zero degrees lumbar angle and as a percentage of the product of the subject's stature and weight. A polynomial least squares fit procedure was then attempted with normalised lumbar torque ($\mathcal{T}\ell$) and the dependent variable and lumbar angle (λ) as the independent variable (Appendix 2,4). In all cases there was no significant departure from linearity in the polynomial equations. A simple linear regression was therefore accepted and the regression equations $\pm$ 1 r.m.s. residual are tabulated below (table 4,16). Figures 4,16 and 4,17 are scatter plots of the data with the regression line $\pm$ 1 r.m.s. residual shown.

Table 4,16 Regressions of Lumbar Torque (τ_l) and Lumbar Angle (λ).

$$\tau_l \text{ (N.m)} = 198.9 + 1.97\,\lambda \pm 54.3$$
(r = 0.73)

$$\tau_l \text{ (\% zero)} = 97.0 + 1.08\,\lambda \pm 19.7$$
(r = 0.85)

$$\tau_l \text{ (\% mean)} = 107.4 + 1.11\,\lambda \pm 18.6$$
(r = 0.86)

$$\tau_l \text{ (\%St. W)} = 16.57 + 0.174\,\lambda \pm 4.03$$
(r = 0.78)

Figure 4,16

Scatter plot of absolute lumbar torque (Nm) against lumbar angle. The regression line of torque on angle ($\pm$ 1 R.M.S. residual) is indicated.

4,6,2

Figure 4,16

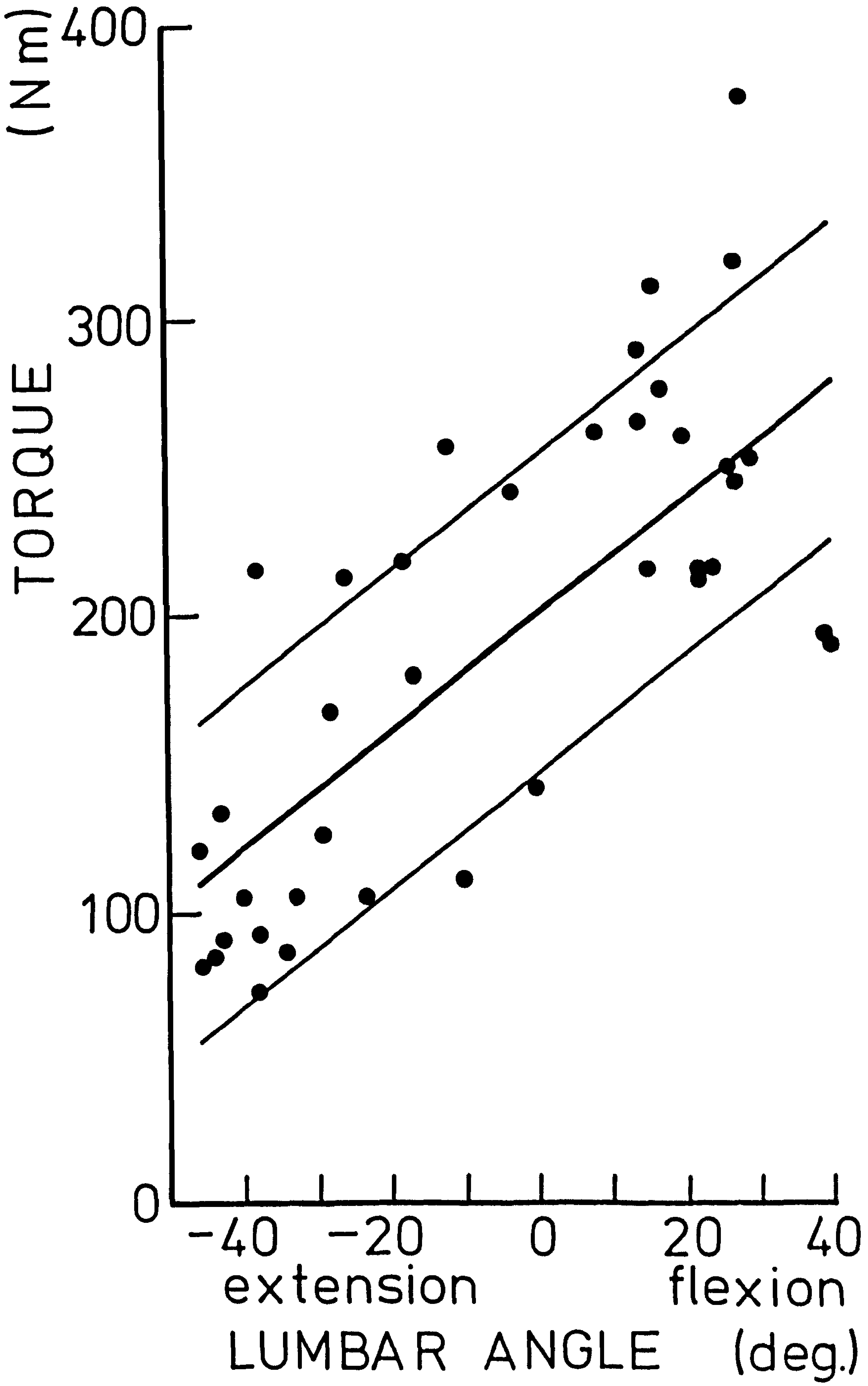
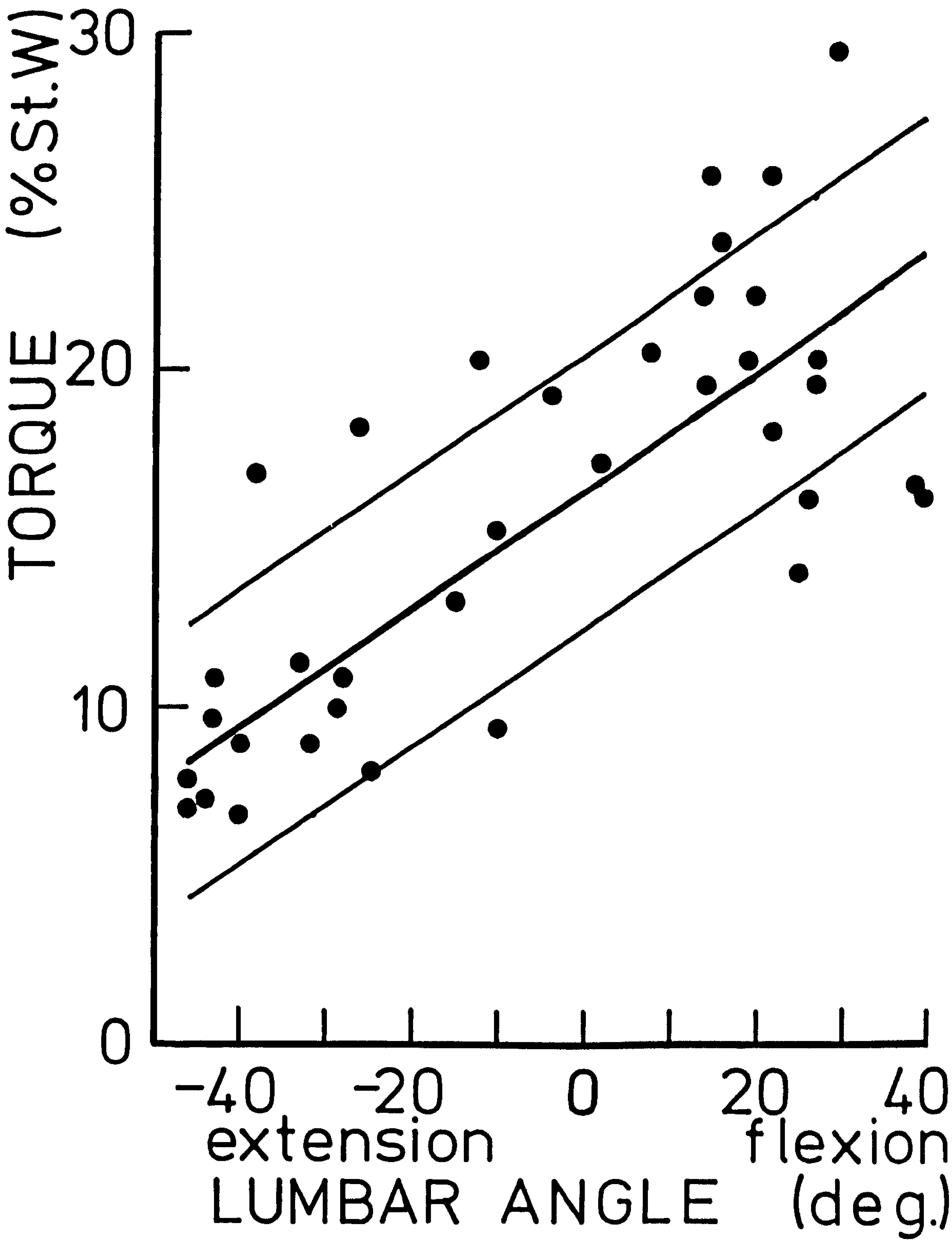

(Nm)
TORQUE
400
300
200
100
0
-40
-20
0
20
40
extension
flexion
LUMBAR ANGLE (deg.)

Figure 4,17

Scatter plot of normalised lumbar torque (expressed as a percentage of the product of stature and weight) against lumbar angle. The regression line of torque on angle ($\pm$ 1 R.M.S. residual) is indicated.

4,6,2


TORQUE (%St.W)
30
20
10
0
-40
-20
0
20
40
extension
flexion
LUMBAR ANGLE (deg.)

4.7 DISCUSSION

Experiment 4,4 illustrates the several ways in which the results of a test of strength might be determined. In the case of smooth-floor pulls with no pelvic support it seems indisputable that the horizontal force component is limited by the frictional resistance at the foot-floor surface (4,1 eq.8). In these conditions most of the measurements were made when the subject was sliding, and the calculated coefficients of friction must closely reflect the true coefficient of sliding friction between the feet and floor ($\mu = 0.23$). Even in this limiting condition, the subjects do have a choice in the way they exert vertical forces, and the resultant angle of pull shows substantial variation. The low-bar positions show substantial negative values of Fv, indicating the tendency of the subjects to compress their feet against the ground and thus gain greater purchase on the slippery floor. In the lowest position these forces are, on the average, 52% of body weight.

In one case of the unsupported pulls with a rough floor a different situation appertains. Table 4,11 might suggest that the limit was again frictional, but as the subjects showed no signs of slipping on this floor the "apparent coefficient of friction" as calculated was less than the true coefficient of friction. The only statement that can be made in this respect is that the coefficient of friction between feet and floor is greater than 0.48 and it is necessary to look elsewhere for the factor which determines the force exerted. The horizontal component Fh changes systematically with the bar height, and the change is in the direction which would be anticipated from theoretical considerations (4,1 eq.4). The vertical component also shows a systematic variation. The resultant cannot be predicted from theoretical considerations alone, and indeed it may be seen that the subjects show very considerable variation in this respect, confirming the findings of section 4,3. Dempster (1958) considered this question without coming to any clear conclusions, recording only that in standing pulls "the pull vector was directed to various levels of the lower limb". The factors underlying the individual subject's choice of direction of pull remain unknown.

Provision of a pelvic restraint allows the exertion of forces which in the case of the 1.5 m. and 1.0 m. bar heights are substantially in excess of those exerted in the unsupported condition. In the case of the 0.5 m. height, the horizontal forces are not significantly different ($p > 0.05$) although the vertical forces differ at the level $0.05 > p > 0.01$. This result supports the proposal of 4,2 and and 4,3 that for unrestrained pulling in a low position, the horizontal force is not limited entirely by body stability, but partially at least by muscular strength. It is therefore not improved by the provision of a pelvic restraint. (The restrained and unrestrained conditions are not entirely comparable, and the provision of the pelvic restraint modifies the posture of the subject, and as this was unrecorded the above conclusion is tentative). When the pelvis is restrained, the nature of the floor makes very little difference to either the vertical or the horizontal forces. As suggested in the theoretical analysis (4,1 eq. 15-17) the bar may bear any reaction forces demanded and the feet are effectively uncoupled from the action. It is of interest to note that in restrained conditions the directions of resultant force show much less variation than they do in unrestrained conditions.

It seems certain that forces in the restrained condition are limited by the strength of an articulation above the level of the pelvis. The finding that the total resultant forces are very similar for all three bar heights suggests that this weak link is located along the line of action of these forces (i.e. within the upper limb or shoulder girdle), because the posture of the trunk, and by inference its strength, changes with the experimental conditions whereas that of the upper limb does not. The results of the additional experiment of section 4,5 enable us to discount the possibility that the capacity of the subject to grip the bar is the limiting factor. This does not conclusively prove that the weak link is in the lumbar region. The pulling and lifting tasks of experiment 4,5 impose quite different types of loading on the shoulder girdle, and it could for example be the muscles which retract the scapula which determine the limit; the bulk of the available musculature of the shoulder region makes this seem unlikely.

The results of experiment 4,6 are superficially simple in that a strong relationship has been shown to exist between the total extensor torque calculated about the mid lumbar region and the posture of the lumbar spine and hence, by inference, the length of the post vertebral muscles. Further consideration shows that these results are not at all simple in their interpretation.

There appears to be no generally accepted convention for the description of lumbar posture. The system used here (T12 and S1 markers) is similar to that of Davis and Troup (1964b) although it has an opposite sign convention. The choice of the reference point for the calculation of torque in the L3-4 region is also arbitrary. The results of the experiment describe an angle-torque relationship for this level alone. It may be that the weak link is lower or higher than this, but such a question would be exceedingly difficult to investigate and answer with certainty.

The calculation of postural torques must be considered as no more than a formalised procedure for the approximate solution of a complex problem. The location of a trunk centre of gravity is little better than guess-work. Dempster (1955) provides data for the combined trunk and head segment and separately for the head, shoulder girdles, thorax and abdomen plus pelvis. Liu and Wickstrom (1974) have supplied more detailed data for serial sections of the trunk.

Neither of these sets of data is adequate for an accurate calculation of gravitational torques about a given vertebral level, as the trunk in truth consists of a flexible framework from which deformable fluid-filled bags are suspended in an unknown manner. Less uncertainty is associated with the limb segments although even in this case the variability of human proportions introduces a source of error of unknown magnitude. However, in the presence of maximal external forces, relatively large errors in the calculations of the gravitational torques will still only result in relatively small errors in the final result.

The final measured torque is dependent upon the interaction of a large number of factors all of which may vary with lumbar posture. The following factors may be listed:

i) Active tensions in the muscles which are dependent upon muscle lengths,

ii) The leverages of the muscles,

(iii) Passive tensions in various soft tissues,

(iv) The effective leverages of these elastic structures,

(v) Intra-abdominal pressure,

(vi) The centroid of the action of the intra-abdominal pressure.

In the presence of so many complex variables, further speculation seems futile. The measured angle torque relationship must remain an empirical finding only. None-theless, it is of some importance to record that the strength of the trunk is posturally dependent in a manner which is, in general terms at least, similar to the way in which the strengths of the limb articulations which have been described are posturally dependent.

CHAPTER 5 - FINAL DISCUSSION

In an early part of this thesis (1,3), it was suggested that tests of strength and their results, could be divided into three main categories according to the biomechanical nature of the supposed limiting factor operating in each case. These categories were named as follows:

Type I	- Musculo-skeletal
Type II	- Gravitational
Type III	- Interface

In addition, it was deemed necessary to acknowledge the existence of a large number of non-biomechanical factors, which, if allowed to can give rise to a fourth group of limitations (Type IV) which must not be allowed to bias the test situation.

These three biomechanical categories are exemplified in the experimental section of this thesis. The ways in which performance in tests of strength is dependent upon posture has received particular attention. The characteristics of, and interactions between, musculo-skeletal, gravitational and interface limitations of human strength will now be discussed briefly in the light of the experimental findings.

5,1 MUSCULO-SKELETAL LIMITATIONS

Examples of musculo-skeletal limitations of strength have been demonstrated in tests of plantarflexion, in pronation and supination using stirrup handles and in extension of the lumbar spine. In each case the dependence of the strength of these actions on posture was described in terms of an angle-torque curve. In the case of plantarflexion it was possible to measure separately the active and passive components of the torque and an attempt was made to attribute portions of the active torque to specific muscles (2,3,4). In the case of pronation and supination, the separation of active and passive elements was not possible since synergistic muscle activity is necessary for the transmission of the measured torque. In the case of lumbar extension it was recognised that the anatomico-physiological basis of the experimental findings was sufficiently complex and obscure to make any detailed analysis impossible at the present time (4,7).

Each of these angle-torque curves has the same general form. Maximum voluntary torques increase with the lengths of the muscles responsible for the action.

In the cases of plantarflexion, pronation and supination, torques reached a plateau in the postures of greatest muscle lengths, but in none of these cases was a statistically significant downward trend observed beyond the plateau. In the case of lumbar extension, the plateau was absent, but the postures tested in this experiment did not represent as great a fraction of the total range as did those tested in the previous experiments.

Section 2,3,3 showed that, for the action of plantarflexion, changes in leverage have a relatively small effect upon changes of torque, and the length-tension relationships of the muscles must be the dominant factor underlying the form of the angle-torque curve. It is probable that the physiological range of lengths for most muscle groups extends from the mid-part of the rising portion of the length-tension curve up to and including the optimal condition. (Fig. 1,1(b)). The evidence suggests that the physiological range intrudes very little into the descending portion of the curve that can be demonstrated in isolated muscle preparations.

Similar conclusions may be drawn from the published data concerning abduction and adduction of the hip (Clarke et al, 1950; Murray and Sepic, 1968; Olsen et al 1972) and flexion of the knee (Clarke et al, 1950; Smidt, 1973). Lindahl et al (1969) and Smidt (1973) have described a reduction in the maximum voluntary torque of extension of the knee when the knee is fully flexed. This appears to be contrary to the rules suggested above. However, the analysis of Smidt (1973) shows that this decrement of strength is mainly due to a marked reduction in muscle leverage.

The strength of elbow flexion is greatest in the middle part of the range of the joint and is markedly reduced in either extreme position (Clarke *et al*, 1950; Doss and Karpovitch, 1965). This may again be seen as the interaction of the length-tension relations of the muscles with changes in their leverage. Fick (1911) showed that all of the elbow flexor muscles have a very much greater leverage in positions of flexion than in positions of extension where their intrinsic strengths are presumably greatest.

In summary, we may make the following statement. In voluntary isometric tests, muscles exhibit their greatest strength in those postures in which they are at their greatest length. In a minority of cases this relationship may be masked by posture-dependent differences in leverage.

5,2 GRAVITATIONAL LIMITATIONS

The early sections of Chapter 4 were devoted to an analysis of the pulling and lifting actions described by Dempster (1958) and by Whitney (1958), who concluded that in these tasks the weight and leverage of the body imposed the effective limits on measurable strength. It was demonstrated that the extent to which this was true was a function of both the posture of the subject and the precise nature of the pulling or lifting task. In conditions in which the direction of pulling was close to the "live axis" (as defined in section 4,1) the contribution of weight and stature (and hence by inference leverage) was no more than would be expected in strength tests in which musculo-skeletal limitations operated. It must of course be acknowledged that although it is possible in the statistical analysis to calculate the proportion of variance in strength which can be attributed to weight and stature, it is not possible to attribute a specific source to the remaining variance. Part of this remaining variance must be associated with the miscellaneous "type IV" factors which have received little discussion or analysis in this thesis.

Whitney (1958) noted that his predictive equation for strength became unreliable in lifting tasks where the foot placement was close to the axis of the lift. The present research has analysed the mechanical basis for that finding and is, to the author's knowledge, the first attempt which has been made to separate in a formal way the situations in which the dead-weight of the body and the action of muscles respectively dominate the strength of an action.

5,3 INTERFACE LIMITATIONS.

The preceding chapters contain two examples of situations in which the strength of an action is limited by the characteristics of an interface between the subject and the outside world.

In Chapter 4 it was shown both theoretically and experimentally that the strength of an unsupported pulling action may be limited by the low frictional resistance offered to the feet by a smooth floor. In a similar way, as shown in section 3,6,2, the torque which may be exerted upon a smooth cylindrical handle is limited by the frictional properties of the hand-handle interface. The experimental study of commercially available screwdrivers showed that, at least within the relatively small range of handle forms tested, the shaping of the handle is equivalent to the roughening of its surface in that the torques transmitted by rough cylindrical handles of the same mean diameter are never exceeded. The size, shape and texture of the handle determine the proportion of the total muscular capacity for supination which can be transmitted to the dynanometer and it may be supposed that an interface limitation arises.

5,4 INTERACTIONS BETWEEN THE LIMITING FACTORS IN TESTS OF STRENGTH.

Any discussion of these three major categories of limitation in strength test must take into account the interactions between them. The limiting frictional force at an interface is given by the product of the coefficient of limiting friction and the compressive force which may be brought to bear between the surfaces. In the case of the smooth handles discussed in Section 3,6,2 the limiting value of this compressive force (which may be thought of as the strength of grip) is a function of the complex musculo-skeletal mechanics of the hand. In the case of the pulling actions described in Chapter 4 the compressive force is determined by the weight of the body and by the vertical component of the exerted force. The factors which influence or determine the precise direction in which the subject "chooses" to pull remain obscure. It is possible that a subject may increase his purchase on a slippery floor by the skilful application of vertical forces.

As Dempster (1958) pointed out, in order for a man to exert a force by the exploitation of his dead-weight, he must use his muscles to lock his articulations into a suitable posture. Situations may arise therefore in which the musculo-skeletal limitation of this locking action may prevent the optimal deployment of body weight.

It must again be noted that a variety of miscellaneous influences may intervene. A sharp edged handle will not be strongly gripped. The optimal deployment of body weight is to some extent a matter of skill. The effects of fatigue and motivation may be present in any type of strength test.

The principal interactions between the limiting factors in tests of strength are summarised in Fig. 5,1.

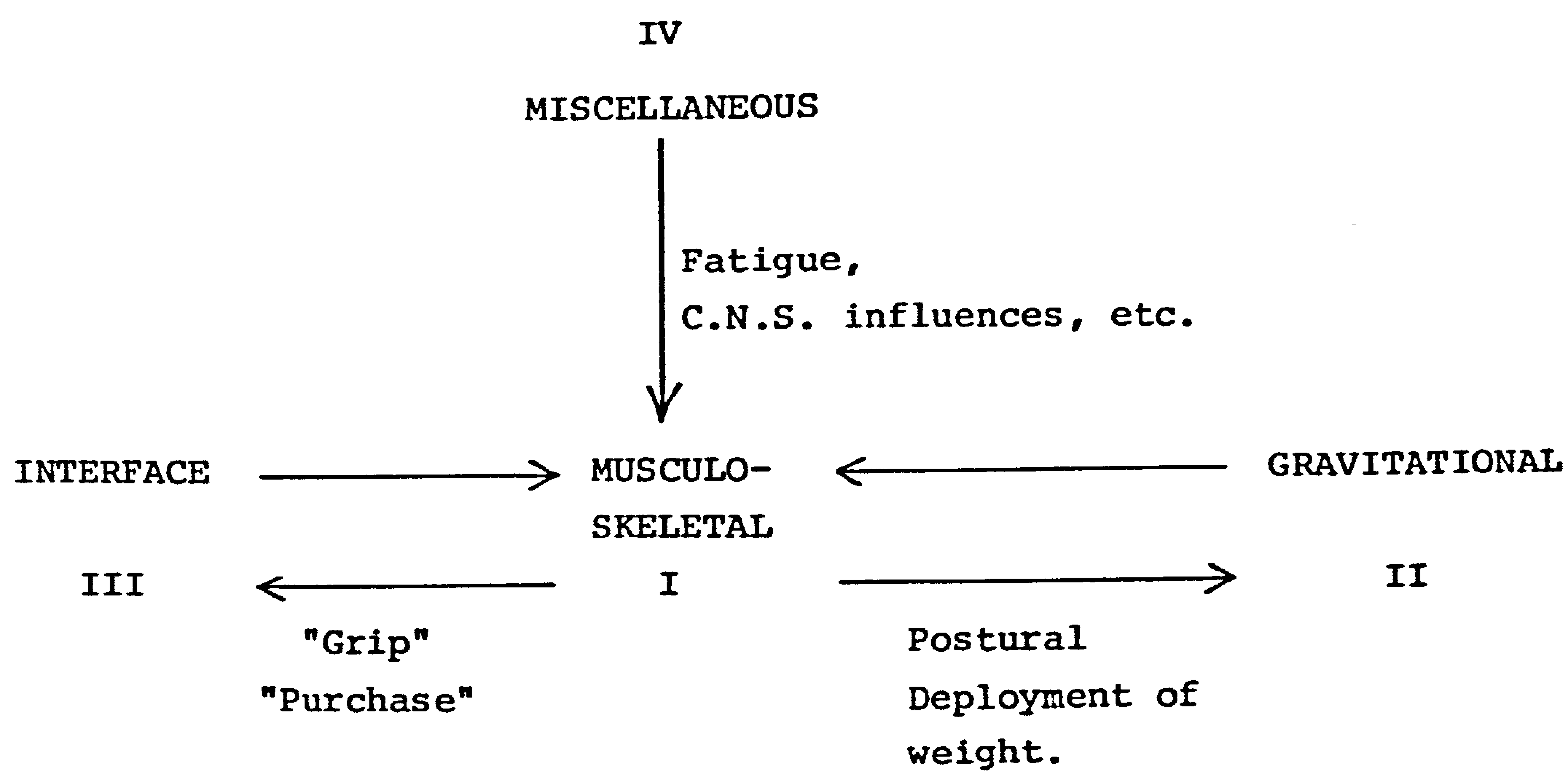

IV
MISCELLANEOUS
Fatigue,
C.N.S. influences, etc.
INTERFACE
MUSCULO-
SKELETAL
GRAVITATIONAL
III
I
II
"Grip"
"Purchase"
Postural
Deployment of
weight.

Fig. 5,1

5,5 PROPOSALS FOR FURTHER RESEARCH

It is appropriate at the conclusion of a work of this nature to consider the areas in which the present findings may lead to further research in the immediate future. Two such projects are already in hand.

The cadaveric data concerning muscle lengths (2,2,2) is being employed in an analysis of the function of the gastrocnemius muscle in walking at different speeds. This is the subject of a paper in preparation at the present time (Cavanagh, Grieve and Pheasant, 1977).

An experimental investigation is in progress which is based on, and has initially confirmed, the speculations of Section 3,6,4 concerning power and precision grips. Subjects are being tested using disc shaped handles with knurled edges and a range of diameters. These handles may be gripped using the distal phalanges only (precision grip) or the distal phalanges and portions of the palm, thenar and hypothenar eminences (power grip). It is hoped that quantitative information concerning the ways in which the hand engages objects with simple geometrical shapes may ultimately lead to a better understanding of the complexities of hand function.

The concepts of the live and dead axes and the dead-weight factor clearly need further investigation. The next stage in this line of enquiry should be a study similar to those described in Sections 4,2 and 4,3, but with the additional information which could be provided by a photogrammetic analysis of the subjects' postures, and the use of an instrumented force platform. By these means a greater understanding may be achieved of the ways in which a subject optimises the deployment of his body bulk and actively exerts his muscles in the task. A complementary approach to these questions is by means of computer simulation, pilot studies of which have been undertaken.

Recent research in basic muscle physiology (Edman, et al, 1976) suggests that the mechanical behaviour of active muscle under dynamic conditions is more complex than was hitherto fully appreciated. The extension of the present work into the area of dynamic strength testing presents analytical and experimental problems of a new order of complexity. The results of such future experiments will however bring us nearer to an understanding of human performance under conditions found outside the laboratory.

CHAPTER 6

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APPENDIX 1

GENERAL METHODOLOGY OF TESTS OF STRENGTH

A standard protocol was observed for the vast majority of experiments reported in this study. This will be described at this stage in order to avoid repetition in the text. The few instances where non-standard techniques were necessary have been separately described.

The force transducer used was a commercially available load cell (Pye Ether type U.F.2) which contains unbonded strain gauges mounted in a full Whetstone bridge configuration. The majority of recordings were made using the Medelec M.S.6 modular electrophysiological recording system, included in which was an AD6 strain gauge amplifier unit which supplied power to the bridge, enabled zero offsets to be corrected and amplified deflection voltages. The system was linear within the range employed. Force traces were made by the fibre- optic ultraviolet recorder of the Medelec system which wrote directly onto self developing light sensitive paper. Permanent records of the experiments were therefore available for analysis at leisure.

A certain controversy has revolved around the "correct" length of time over which an isometric effort should be held in a test. Mougdil and Karpovitch (1969) maintained that a peak maximum can only be held for less than a second, and apparently consider that this transient phenomenon is biologically relevant. As Kroemer and Howard (1970) have pointed out, such jerky peaks in a force record reflect reactions to the accelerations of body parts. It may be further claimed that the size and shape of such transients are a function also of the elastic, frictional and inertial properties of the recording system which may interact with the man in a complex and unpredictable fashion. Kroemer (1970) recommended that measures of impulse (integral of force with respect to time) be used for the testing of strength. This seems to be a slightly extreme reaction to the recording of transients and a compromise was adopted.

In each experiment careful explanations as to the nature of the desired force or torque were given to the subject and he was allowed several practice trials. In the majority of cases where the independent variable was a postural one, the practice runs were conducted in a posture close to the middle of the range. The subject was instructed to build up his force to a maximum <u>rapidly but without jerking</u> and to hold this maximum force until told to relax. The following verbal command was given for each recorded effort:- "Are you ready ...(pause), Go! one, two, three, four, five, relax." As the numbers were counted at one second intervals, the entire force record was between six and seven seconds in duration and had a central portion during which, in most cases, the force oscillated about a plateau. In order to get a numerical reading from the force record, the experimenter established a mean value of this central plateau by visual inspection such that this mean value could be held to represent the maximal steady force which could be sustained for at least three seconds. The errors inherent in such an assessment are certainly small in comparison with the intrinsic variability of strength measurements.

The majority of the experiments reported are designed to compare trial conditions, rather than to acquire absolute data for the performance of a given task. Such experiments require relatively small numbers of subjects, but it is necessary to take some care in "designing out" the effects of such confounding factors or sources of error as fatigue, learning or changes in level of motivation. This may be achieved by the construction of a bias-free order of performance. It is important to distinguish between random and systematic influences on experimental results. The latter are much less tolerable than the former and they may be removed by a design based on the reasonable assumption that systematic changes due to fatigue, learning or motivation are occurring in a progressive fashion throughout a subject's performance of a sequence of trial conditions. By giving a different temporal sequence of trial conditions to each subject, it is possible to arrange for subjects to compensate for each other so as to tend to cancel out any systematic errors.

A bias-free design is one in which, taking a group of subjects, any given trial condition has the same *mean overall position* in temporal sequence as any other trial condition. In the final statistical analysis of such a set of results the residual effects of learning, fatigue and motivation become sources of random rather than systematic error and have the effect of widening confidence limits rather than introducing bias. Bias-free designs were used in all experiments except where otherwise stated. In these latter cases the complexity of the number of conditions required the slightly less advantageous experimental design in which the sequence of trial conditions is chosen at random.

APPENDIX 2

STATISTICAL METHODS

The statistical methods used for the analysis of data in this thesis are listed below. They were derived from the following sources.

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In the present study the initial descriptive breakdown of the data was executed on a hand calculator. The more detailed statistical analyses were performed using FORTRAN programmes written by the present author especially for the purpose. Polynominal least squares routines were performed using a library programme (see below).

A2,1 DESCRIPTIVE STATISTICS

The statistical methods used in this thesis are all parametric, i.e. they are all based on the assumption that measurements which have been made are distributed within the subject population according to a Gaussian ("Normal") distribution. In all cases therefore the first step in analysis is to use the sample data to estimate the descriptive parameters (mean and standard deviation) of the hypothetical normal distribution of the parent population. These sample estimate parameters are given by the following equations:

$$\bar{x} = \frac{\sum_{i=1}^{n} (x_i)}{n} \qquad (1)$$

$$S = \sqrt{\frac{\sum_{i=1}^{n} (x_i - \bar{x})^2}{n - 1}} \qquad (2)$$

Where $\bar{x}$ is the mean of each individual value x_i of the variable x,

S is the standard deviation of x,

and n is the number of subjects.

In some cases, for convenience of graphical presentation, the standard error of $\bar{x}$ has been presented. This quantity is given by the equation:

$$SE\ (\bar{x}) = \frac{S}{\sqrt{n}} \qquad (3)$$

The assumption of normality in the present data is suspect. It is commonly accepted,however, that the inferential tests which have been used (see below) are "robust" in the presence of modest deviations from normality. Formal tests for skewness and kurtosis exist. Such tests have not been performed as they require larger data sets than have been acquired in the present study.

In most data sets of the present study, a process of "normalisation" has been carried out on the strength measurements. In this process, each performance score of a single subject is expressed as a percentage of the mean performance score of that subject. This procedure eliminates that portion of the variance in the original data set which is due to differences in absolute strength between individual subjects. This massive source of variance tends to obscure the differences between experimental conditions which are the important aspect of the study. An example may be found in Section 2,1,2 of a data set which has been processed both with and without normalisation. Fig. 2,6 shows the levels of statistical significance which were found in each case. (It should be noted that in most of the experiments described in the present thesis the number of experimental conditions is too great for a formal analysis of variance to be feasible.)

A2,2 STUDENT'S t - TEST

The test used for establishing levels of significance in differences of strength measured in different experimental conditions (e.g. postures) has been the two-sample Student's t - test. (Although the Student's t - test for paired comparisons was at first sight a more obvious choice, the un-paired test was selected as between subject differences in absolute strength had already been extracted from the data set by the process of normalisation). The test assumes that the variances of the two samples are equal and establishes the probability that they are drawn from the same parent population. (The assumption of equal variances is reasonable for all the present data sets. In the small minority of comparisons where the variances are unequal the means are so widely different that the outcome of the test is not subject to doubt. The alternative test which does not assume equality of variance is based on the so-called d-distribution which is described by Armitage (1971) as "both more complex than the t - distribution and more contentious".)

If $\bar{x}_1$ and $\bar{x}_2$

S_1 and S_2

and n_1 and n_2

are the means, standard deviations and sizes of the two study samples, then the pooled estimate of variance (S^2) is given by the equation:

$$s^2 = \frac{(n_1 - 1)s_1^2 + (n_2 - 1)s_2^2}{(n_1 - 1) + (n_2 - 1)} \tag{4}$$

The statistic t is given by the equation:

$$t = \frac{\bar{x}_1 - \bar{x}_2}{\sqrt{\left\{s^2\left(\frac{1}{n_1} + \frac{1}{n_2}\right)\right\}}} \tag{5}$$

To test the null hypothesis that the two samples are drawn from the same population, the table of t is consulted for $n_1 + n_2 - 2$ degrees of freedom.

A2,3 LINEAR REGRESSION AND CORRELATION

If two variables, x and y are observed or measured on a sample of n individuals, then an empirical equation may be derived which predicts the most probable mean value of y, (Y) for any given value of x. This is commonly achieved by a least squares analysis, i.e. one which minimises the residual sum of squares, $(y_i - Y_i)^2$ where Y_i is given by estimated regression equation

$$Y_i = a + b\ x_i \tag{6}$$

The Y intercept (a) and slope (b) of the regression line are given by the formulae

$$a = \bar{y} - b\ \bar{x} \tag{7}$$

$$b = \frac{\sum (x_i - \bar{x}) . (y_i - \bar{y})}{\sum (x_i - \bar{x})^2} \tag{8}$$

The situations in which both x and y are random variables (i.e. we are investigating a bivariate normal distribution), then it is as valid to establish a regression line of x or y as it is to establish a regression of y or x as performed above. These regression lines will coincide if the two variables are perfectly dependent on each other, and will be perpendicular if the variables are totally independent. The degree of association between the variables is given by the product moment correlation coefficient of Pearson which is defined by the fomula:

$$r = \frac{\sum(x_i - \bar{x}) \, . \, (y_i - \bar{y})}{\sqrt{\left\{\sum(x_i - \bar{x})^2 . \sum(y_i - \bar{y})^2\right\}}} \qquad (9)$$

when x and y are totally independent r = o; when x and y are totally dependent r = $\pm$ 1. Significance tables exist for the null hypothesis that r = o; the tables are entered at n - 2 degrees of freedom. The proportion of the total variance in one variable which may be attributed to variance in the other is given by r^2. It is to be noted that if the sample on which the study is made is selected in a restrictive way by reference to one variable then the calculated correlation coefficient will be low, (Armitage, 1971) and any deductions made from the numerical value of r under such circumstances will be unreliable.

A2,3,1 SIGNIFICANCE TESTS OF LINEAR REGRESSION EQUATIONS.

In the study described in Chapter 4 it was necessary to determine whether the strengths of male and female subjects had common relationships with weight and stature (i.e. were (a) the slopes and (b) the intercepts of the male and female regression lines of strength on stature or weight significantly different?)

(a) Differences in slopes.

If SS_1 and SS_2 are the residual sums of squares of each group taken separately about their own regression lines, then a pooled estimate of the residual variance Sr^2 is given by

$$Sr^2 = \frac{SS_1 + SS_2}{n_1 + n_2 - 4} \qquad (10)$$

If the regression slopes are b_1 and b_2, then the variance of the difference, $V(b_1 - b_2)$, is given by

$$V(b_1 - b_2) = Sr^2 \left[\frac{1}{\sum_1 (x - \bar{x})^2} + \frac{1}{\sum_2 (x - \bar{x}_2)^2} \right] \qquad (11)$$

The suffixes behind the summation signs indicating that summation is being carried out only through the sub-group of the whole sample indicated.

The difference in the slopes is tested by

$$t = \frac{b_1 - b_2}{\sqrt{\left\{V(b_1 - b_2)\right\}}} \qquad (12)$$

entering the table of the student's t distribution at $n_1 + n_2 - 4$ degrees of freedom.

(b) Differences of intercept (The analysis of Co-variance).

If no difference exists between the slopes b_1 and b_2 it becomes pertinent to ask whether the separation of the regression lines (now deemed to be parallel) is greater than would be expected from the difference between $\bar{x}_1$ and $\bar{x}_2$. (i.e. whether the intercepts a_1 and a_2 are significantly different).

A common regression slope b may be estimated thus

$$b' = \frac{\sum_1 (x - \bar{x}_1)\cdot(y - \bar{y}_1) + \sum_2 (x - x_2)\cdot(y - \bar{y}_2)}{\sum_1 (x - \bar{x}_1)^2 + \sum_2 (x - \bar{x}_2)^2} \tag{13}$$

The difference between the two values of Y at a given value of x is given by

$$d = \bar{y}_1 - \bar{y}_2 - b'(\bar{x}_1 - \bar{x}_2) \tag{14}$$

The standard error of this difference, SE (d) is given by

$$SE\ (d) = \sqrt{\left\{ S_z^{\,2} \left[\frac{1}{n_1} + \frac{1}{n_2} + \frac{(\bar{x}_1 - \bar{x}_2)^2}{\sum_1 (x-\bar{x}_1)^2 + \sum_2 (x-\bar{x}_2)^2} \right] \right\}} \tag{15}$$

Where $S_z^{\,2}$ is the pooled residual mean square about the parallel lines, given by

$$s_z^2 = \frac{\sum_1 (y-\bar{y}_1)^2 + \sum_2 (y-\bar{y}_2)^2 - \dfrac{\left\{\sum_1 (x-\bar{x}_1)\cdot(y-\bar{y}_1) + \sum_2 (x-\bar{x}_2)\cdot(y-\bar{y}_2)\right\}^2}{\sum_1 (x-\bar{x}_1)^2 + \sum_2 (x-\bar{x}_2)^2}}{n_1 + n_2 - 3} \qquad (16)$$

(n.b. $s_z^2 \neq s_r^2$)

This data may be tested using

$$t = d \,/ SE\,(d) \qquad (17)$$

and consulting the table of Student's t at $n_1 + n_2 - 3$ D.F.

Further details of the above tests may be found in Chapter nine of Armitage (1971).

A2,3,2 PARTIAL CORRELATION.

If variables A, B, C and D are measured on a population of n subjects, then the correlation coefficients between all possible pairs of variables may be calculated and a matrix may be set out as follows:

	A	B	C	D
A	1.0			
B	$r_{a\,b}$	1.0		
C	$r_{a\,c}$	$r_{b\,c}$	1.0	
D	$r_{a\,d}$	$r_{b\,d}$	$r_{c\,d}$	1.0

In order to investigate the interactions between the variables, first and second order partial correlation coefficients may be calculated from this table of "zero-order" correlation coefficients. For example, the first order partial $r_{ab.\,c}$ is the estimated correlation between variables A and B in a cross-section of individuals all having the same value for variable C. It is a measure of that part of the correlation between A and B which is independent of C. It may be calculated from the formula

$$r_{ab.\,c} = \frac{r_{ab} - r_{ac} \cdot r_{bc}}{\sqrt{(1 - r_{ac}{}^2)\ (1 - r_{bc}{}^2)}} \qquad (18)$$

and its significance may be determined by entering the normal table of r at n - 3 degrees of freedom.

The second order partial $r_{ab.cd}$ is the correlation between A and B with both C and D held constant. It is given by the formula:

$$r_{ab.cd} = \frac{r_{ab.d} - r_{ac.d} \cdot r_{bc.d}}{\sqrt{(1 - r_{ac.d}{}^{2})(1 + r_{bc.d}{}^{2})}} \qquad (19)$$

and its signficance is tested by entering the usual table of r at $n - 4$ degrees of freedom.

A2,4 NON-LINEAR REGRESSION

The above techniques all assume that the relationship between a given pair of variables is a linear one. In situations where it was deemed necessary to test such an assumption and possibly provide an alternative, a polynomial regression was derived of the form

$$y = b_0 + b_1 x + b_2 x^2 + b_3 x^3 \quad - - - - - - \quad b_i x^i \tag{20}$$

This computation was performed using the library programme No. BMD 05R in the BMD Statistics Package (Dixon, 1968).

The addition of each successive order in the polynomial reduces the residual variance about the regression line. The significance of each reduction may be established using the F test of variance by calculating $F = c/d$, such that c is the reduction in the residual sum of squares about the regression line achieved by the addition of the present polynomial order, and d is the mean squared residual deviation at this level.

The table of F is entered at 1 degree of freedom for c and at n - 1 - the polynomial order degrees of freedom for d.

Alternatively, a curve of the form $y = a x^b$ may be fitted by performing a simple linear regression on the logarithms of the values of x and y to give an equation of the form $\log y = a + b \log x$.

APPENDIX 3 - RELATED PUBLICATIONS

MYOELECTRIC ACTIVITY, POSTURE AND ISOMETRIC TORQUE IN MAN

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Introduction

Relationships between voluntary isometric tension or torque and myoelectric activity have been described by many authors, including Lippold (1952), Inman et al. (1952), Zuniga & Simons (1969), Vredenbregt & Rau (1973), Bouisset (1973) and Moeller (1966). The relationships are entirely empirical and depend upon the subject, the quantitative measure of the electromyogram used, the electrode separation and the postures of the joints over which the muscles act. The latter phenomenon, partly reflecting changes with muscle length, deserves wider recognition by many who employ the electromyogram as a means of comparing the economies of working postures. Erector spinae, triceps surae and biceps brachii muscles have all been monitored in published ergonomic assessments of posture or hand tools. All three groups have been used in this paper for the further exploration of relationships between myoelectric activity, torque and posture.

The effects of change of length upon the properties of human skeletal muscle are ill-understood and poorly documented. The best known effect is that upon the maximum exertable tension. Despite its importance, it does not account for all the effects of posture upon the EMG-isometric torque relationships. The quality of the surface electromyogram e.g. its frequency spectrum (Sato, 1966) and other measurable features of the waveform (Troup & Chapman, 1972), is affected by posture. Some changes may be due to a shift of spatial arrangements deep to the electrode sites rather than to intrinsic changes associated with muscle length. Although normal patterns of efferent activity, conduction velocities in muscle fibres and time-courses of individual action potentials may all be affected by muscle length, none appear to have been studied in a postural context in man. Jewell & Wilkie (1960) reported an increase in the duration of the active state with length in isolated amphibian muscle. The time-

Electromyogr. clin. Neurophysiol., 1976, 16, 3-21

courses of active states in man could also be important determinants of the electrical activity necessary to produce a given tension. Evidence for a change of active state with length was sought and found in partially summated twitches of soleus and tibialis anticus muscles.

Methods

Subjects and procedure

Five healthy adults were used in the experiments. All measurements of torque and electromyogram were made under as nearly isometric conditions as could be achieved, and the same procedure was followed for each muscle group. The subjects graded their efforts from zero to maximum in ten seconds (against a spoken count of ten), held at maximum for two seconds and relaxed to zero in a further, counted ten seconds. Postures were randomised and ample rest periods were given between efforts. A pilot study showed that this procedure yielded results equivalent to the more time-consuming process of exerting a series of steady-state efforts, and were free of systematic hysteretic phenomena which would have indicated the presence of fatigue. The forces and electromyograms were both recorded on galvanometer traces. These were measured at points corresponding to each second of real time throughout the duration of the subject's effort.

Electrical recording

The electromyogram was obtained from Ag-AgCl surface electrodes (Medelec Ltd.),amplified via Medelec AM6 type amplifiers and fed to a signal processor. The processor consisted of a distortion-free full-wave rectifier, followed by an R-C smoothing circuit and buffer stage which drove optical galvanometers in an S.E. 3006 U.V recorder. The mean galvanometer deflection was flat within 1 dB in the frequency range 10Hz to 5kHz with 7 dB/octave roll-off. The time constant of the smoothing circuit was 0.15 sec.

Electrode placement

Erector spinae: Electrodes were placed 6.8 cm apart, one on each side of the mid-line at a level of the third lumbar spinous process.

Biceps brachii: Electrodes were placed vertically, 5 cm apart on the anterior surface of the arm, centred mid-way between the gleno-humeral joint and the lateral epicondyle of the humerus.

Triceps surae: Electrodes were placed vertically, 5 cm apart, over the lateral head of gastrocnemius, centred 1/3 of the distance between kne and ankle. A second pair were placed vertically, 5 cm apart, centred mid-way between the ankle and knee joints, over soleus on the posterolateral surface of the leg.

Electrical stimulation

Square pulses (300 V, 0.2 msec) from a Medelec NS6 stimulator were applied to a pair of electrodes (0.5 x 2 cm) situated 2 cm apart. Soleus muscle was stimulated at the same site that was used for recording. Tibialis anticus muscle was stimulated at a site 3 cm lateral to the anterior border of the tibia, 10 cm inferior to the tibial tubercle. The stimuli were submaximal. Care was taken to maintain the electrode positions and stimulus parameters unchanged during an experiment. Maximal stimulation of triceps surae via the tibial nerve in the popliteal fossa was tried and abandoned because it was unacceptably painful and the results were not reproducible.

Determination of torque

a) Torque of supination

The standing subject exerted torque on a T-bar handle at elbow height (Fig. 1b) with the arm vertical and the forearm horizontal. The torque was transmitted via a coupling flange to a spindle mounted in low-friction bearings. A symmetrical lever arm (10 cm) on the spindle was attached on one side to a force transducer (UF2, Ether Ltd) and could be calibrated by suspension of weights to the opposite side (Fig. 1a). The shaft of the handle passed between the 2nd and 3rd digits of the right hand, ensuring that the spindle was approximately coaxial with the axis of supination-pronation movements. Independent experiments with this equipment showed that the measured torques were unaffected by the presence of axial thrust or by the elimination of the possibility of lateral thrust on the handle (O'Neill, 1974). Experiments were performed in five postures of the forearm from the fully prone to the fully supine positions.

b) Torque of trunk extension

The subject pulled on a horizontal strain-gauge bar (Whitney, 1958) and the horizontal and vertical components of the force were courses of active states in man could also be important determinants of the electrical activity necessary to produce a given tension. Evidence for a change of active state with length was sought and found in partially summated twitches of soleus and tibialis anticus muscles.

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The subject pulled on a horizontal strain-gauge bar (Whitney, 1958) and the horizontal and vertical components of the force were

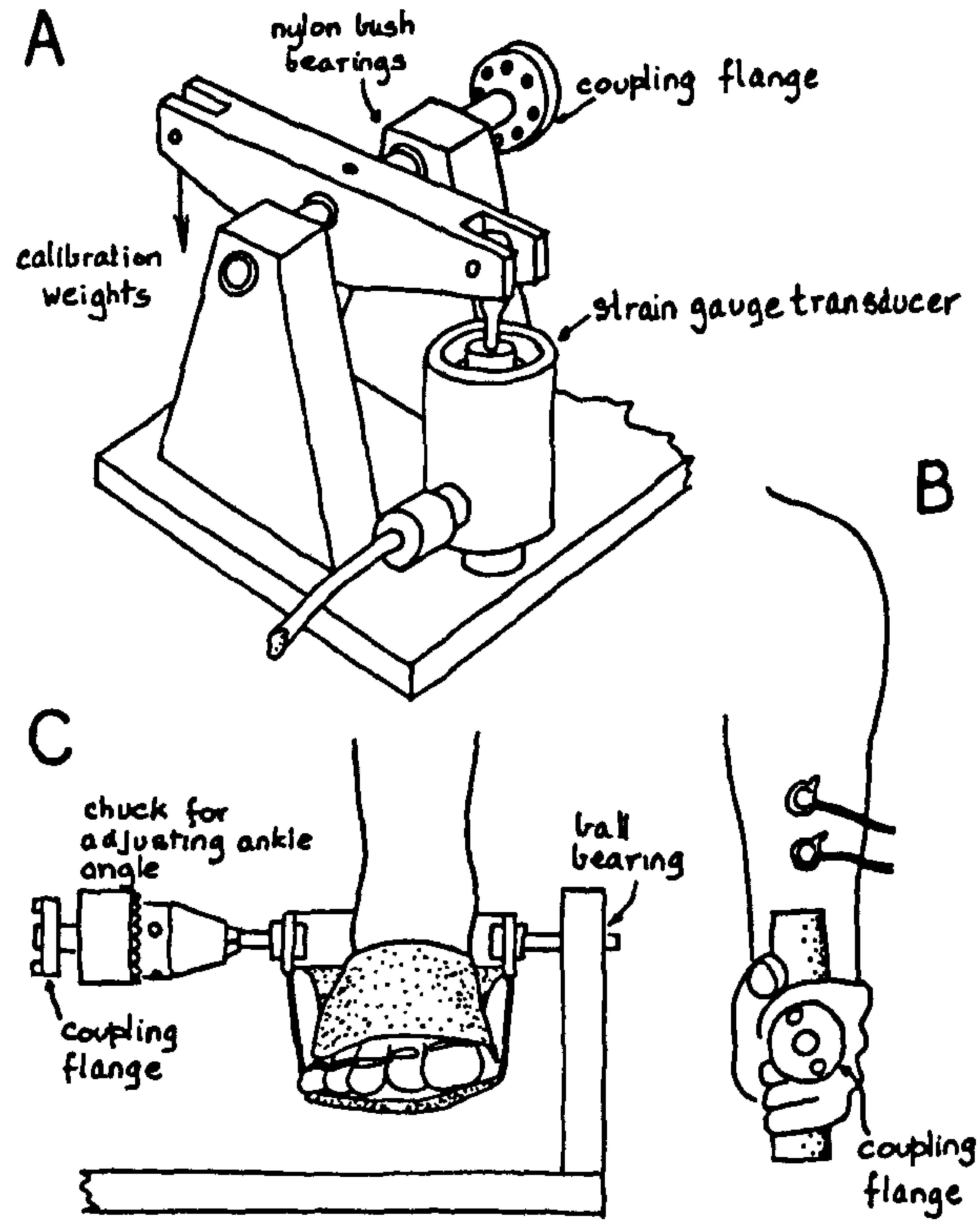

Fig. 1

a) Rig for measurement of torque. b) Use of T-bar handle for measuring isometric torques of supination at same time as the surface E.M.G. from biceps brachii. c) Foot cradle for transmission of twitch-torques of soleus and tibialis anterior muscles to the measuring rig.

determined. The tip of the spinous process of the 3rd lumbar vertebra was identified, together with the vertical height of the estimated centre of rotation of the gleno-humeral joint, when the subject was standing erect. Black adhesive markers were placed on the skin to represent the locations of the centres of gravity of various body segments as follows (Fig. 2a):

i) C.G. of trunk above L3-4 intervertebral disc.
Mid-way between the L3 marker and shoulder height in the erect stance and mid-way between the sternum and the tips of the spinous processes at this level. Assumed weight of segment W_a, 24.8 % of body weight.

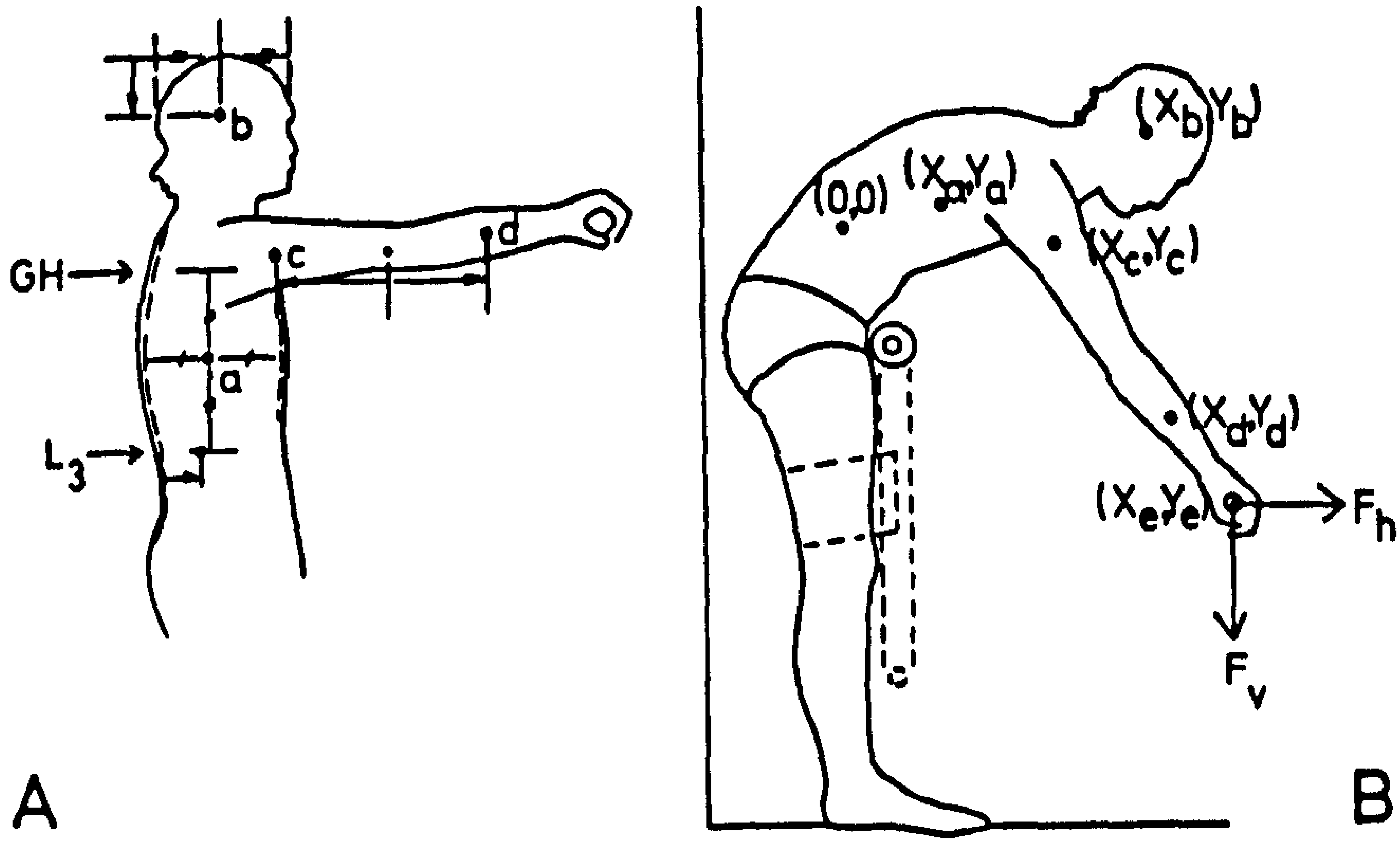

Fig. 2

a) Lateral view of subject during measurements of pull and of associated signals from erector spinae muscles. Body markers and the constructions of their locations are shown (see text). b) Coordinates and forces measured during the pulls. See text for details of calculations.

ii) C.G. of head

7.3 % of stature below the vertex, mid way between the glabella and the occiput (a bathing cap was worn by the subject for the attachment of the marker. Assumed weight of segment W_h, 8.2 % of body weight.

iii) C.G. of arms

The postures were symmetrical. Marker placed along the axis of the arm, 9.5 % stature proximal to the lateral epicondyle of the humerus. Assumed weight of paired segments, W_a, 5.4 % of body weight.

iv) C.G. of forearms and hands.

Along the axis of the forearm, 10 % of stature distal to the lateral epicondyle. Assumed weight of paired segments, W_{fl}, 4.4 % body weight.

The centre of the L3-4 intervertebral disc was marked for analysis in the sagittal plane as a point 5 cm anterior to the tip of the 3rd lumbar spinous process.

The subject stood with his anterior superior iliac spines against a horizontal, padded bar. In some experiments, in order to inhibit sagittal rotation of the pelvis, the thighs were immobilised by straps against a vertical board which extended below the knees, as shown by dashed lines in Fig. 2b. The strain-gauge bar was located so that the subject could comfortably grip it with horizontal outstretched arms when standing erect. Only the vertical height of the bar was altered thereafter in order to record pulls at shoulder, hip and knee heights.

A closed circuit television recording was made of each experimental run (Top Rank TRC5 camera, Top rank Nivico KV820 videorecorder with Top Rank KGM222 profesional monitor). The positions of the projections of the markers in the sagittal plane relative to the L3-4 disc marker were measured with the help of calibration scales in the field of the camera and in the optical plane containing the markers. Parallax corrections were unnecessary.

The total torque about the L3-4 disc axis had a gravitational component and a second component due to the forces at the hands. Referring to Fig. 2b, the total torque was calculated from the expression

$$W_a.X_a + W_b.X_b + W_c.X_c + W_d.X_d + F_h.Y_e + F_v.X_e$$

The postural analysis was checked for each instant of observation of the forces and electromyogram. A Racal counter-timer in the field of view and a galvanometer in the U.V. recorder were both driven by a 2Hz oscillator to ensure synchrony of the observations.

c) Torque of plantar flexion

The subject lay prone on a table with his foot strapped to a plate which was hinged to rotate coaxially with the ankle joint (Fig. 3). The active plantarflexor torques, relative to the torque which existed when the limb was apparently electrically silent, were measured in 5 postures of the ankle between the extremes of plantarflexion and dorsiflexion. The knee was fully extended.

d) Twitch torques

A sock was placed on the subject's greased foot. A mould of plaster was built around and incorporating the sock, leaving a narrow channel in the sagittal plane for cutting and removal in two parts. The reformed mould, after smoothing and greasing. was used to

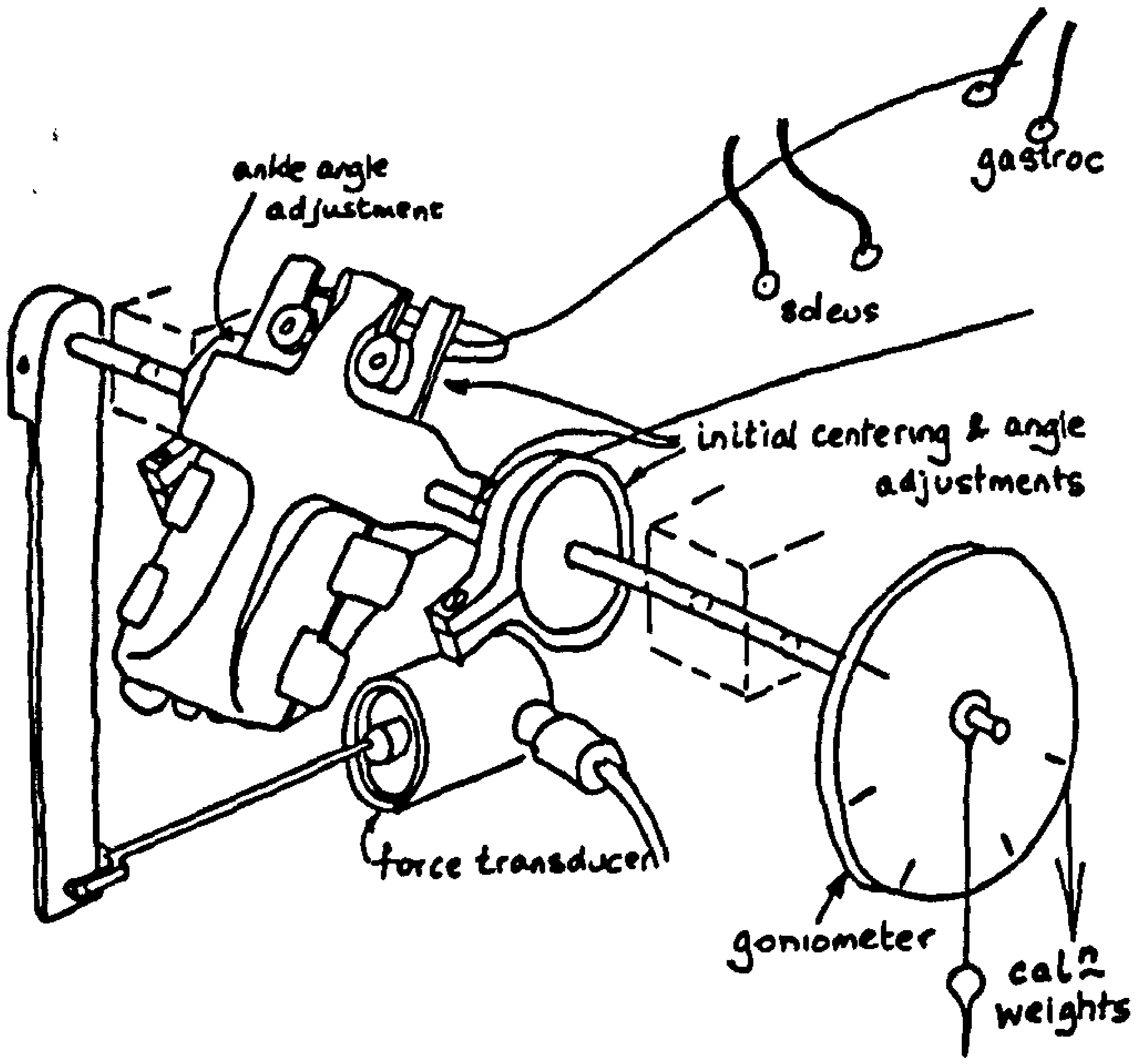

Fig. 3

Rig used to measure isometric torques of plantarflexion and associated signals from triceps surae at various postures of the ankle.

make a plaster cast of the foot. The position of the ankle axis was identified by inspection, guided by the anthropometric data of Isman and Inman (1968). A sole and heel was moulded from Isopon resin, filler and metal mesh, integral with a frame which carried bolts coaxial with the ankle joint. A dorsum was separately moulded which could be strapped to the frame to hold the foot securely. The toes were left clear (Fig. 1c). The torque-measuring rig (Fig. 1a) was coupled to the ankle bolt. Partially summated, double and triple twitches were recorded at various ankle postures between the extremes of plantarflexion and dorsiflexion.

Results

Biceps brachii

The results of experiments on two subjects are shown in Fig. 4. The electromyogram increased most rapidly with torque in the fully supinated posture and progressively less rapidly as the forearm was rotated towards the fully prone posture. The maximum torque increased with the degree of pronation. The maximum electromyograms increased with the degree of supination.

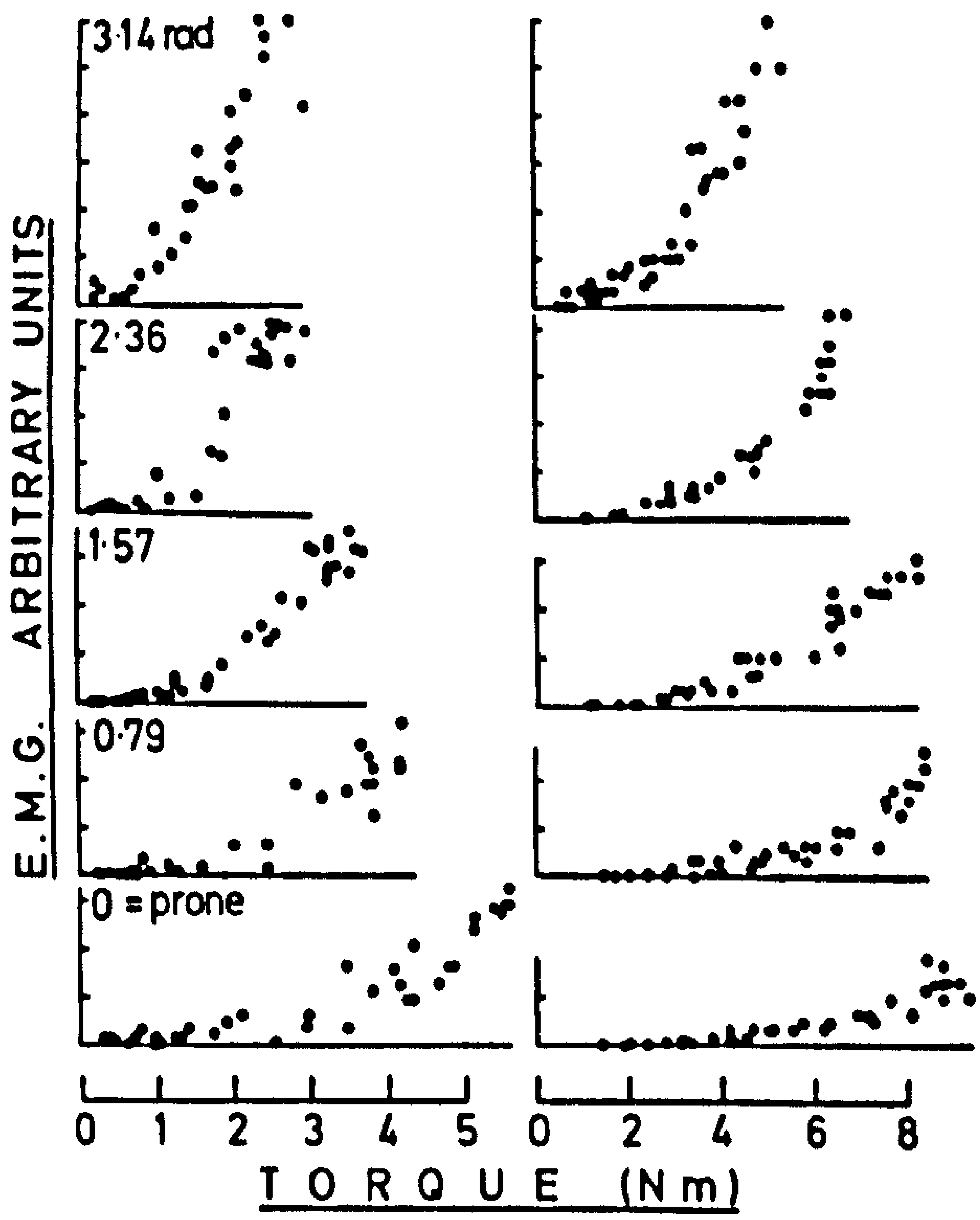

Fig. 4

Relationships between rectified and smoothed surface-electromyograms from biceps brachii and the isometric torque of supination in 5 postures of the forearm. Zero forearm angle represents full pronation. Left and right refer to two subjects.

Erector spinae

The results of experiments with two subjects are shown in Fig. 5. The electromyograms increased most rapidly with torque in the extended posture and least rapidly in the most flexed posture. The maximum torque increased with the degree of flexion. The maximum electromyogram increased with the degree of extension.

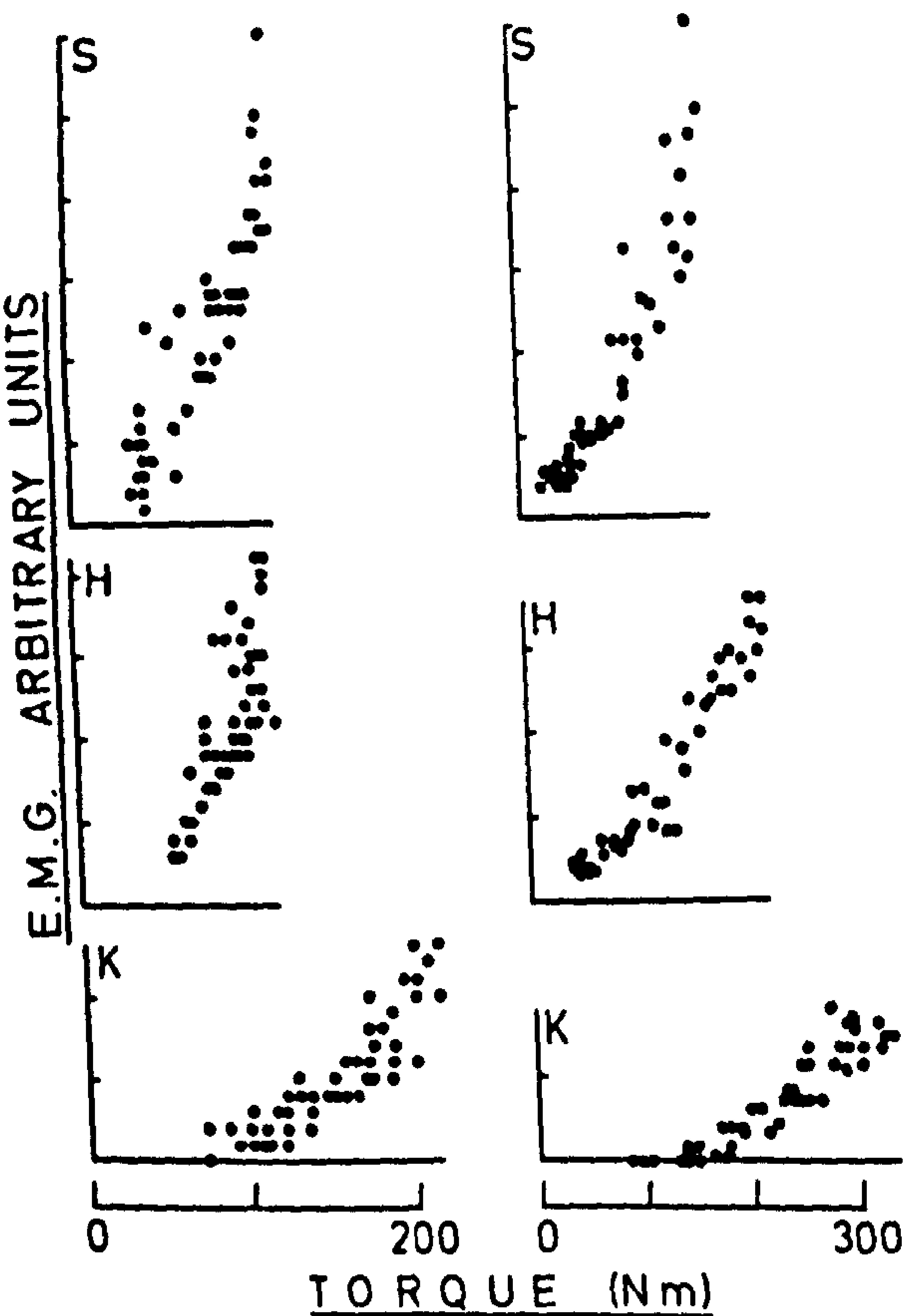

Fig. 5

Relationships between rectified and smoothed surface-electromyograms from erector spinae (L3 level) and the computed torque about the L3 vertebral body in the sagittal plane during pulls at shoulder (S), hip (h) and knee (K) height. Left and right refer to two subjects.

Triceps surae

Log-linear regression lines were usually more satisfactory than log-log regressions of the electromyographic activity against the plantarflexor torques in a given posture of the ankle. The maximum plantarflexor torque was greatest at an ankle angle between 1.57 and 1.92 radians, when the knee was fully extended. The log-linear regression equations were used to estimate the EMG levels that corresponded to 25, 50 and 75 % of the maximum possible torque at each ankle angle. The resulting curves, i.e. for constant torque, for soleus and gastrocnemius signals are shown in Fig. 6.

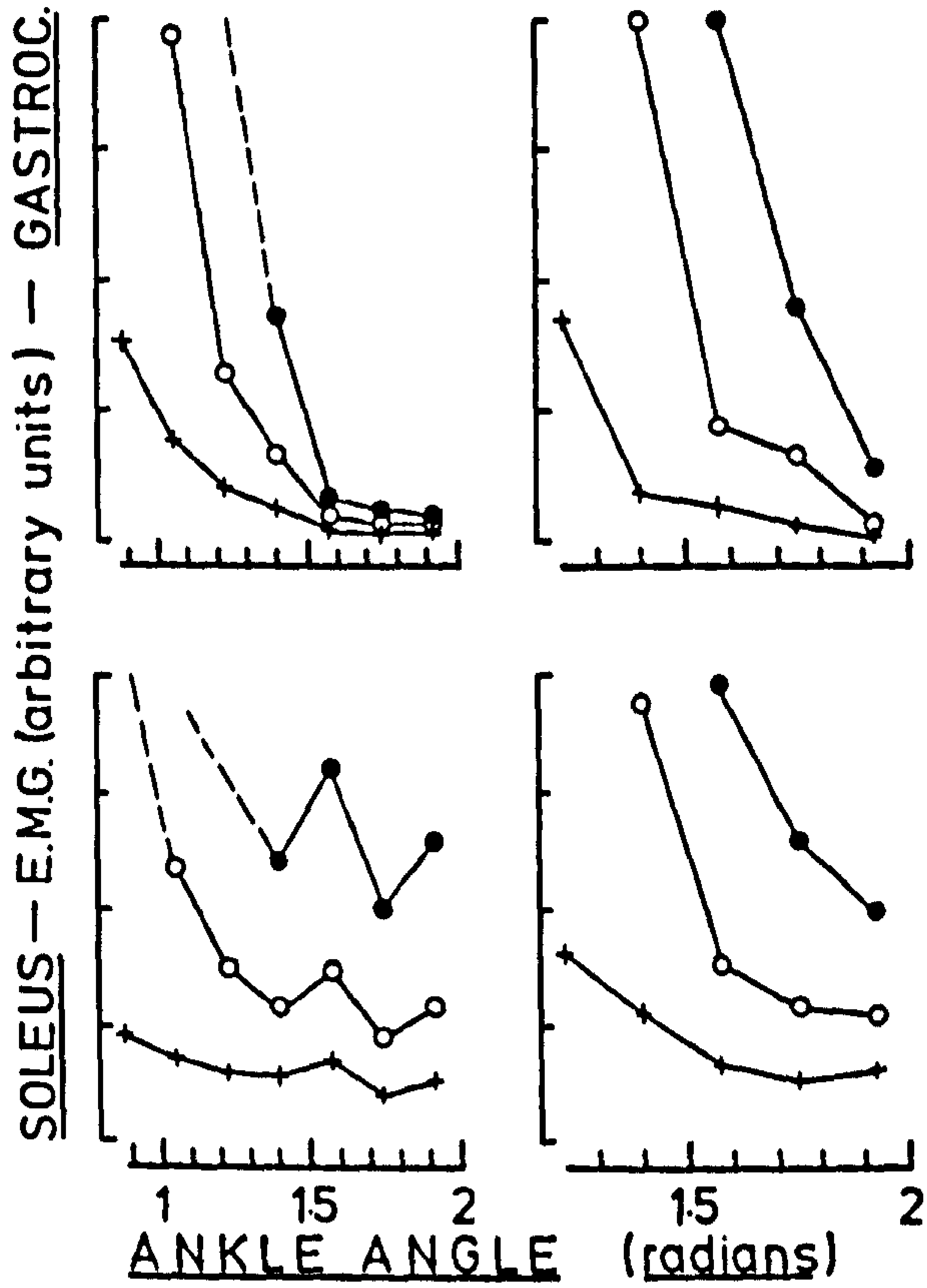

Fig. 6

Relationships between the rectified and smoothed surface electromyograms from gastrocnemius and soleus muscles and the posture of the ankle at 25 (+). 50 (0) and 75 (0) % of maximum possible torque. Left and right hand refers to two subjects.

Both subjects showed a more rapid rise of EMG with torque when the ankle was in a plantarflexed position than when it was in a dorsiflexed position. The experimental data is similar, in relation to muscle length, to the data for the back and biceps brachii. The reason for presenting the data for triceps surae in modified form is to emphasise the very large difference in the electromyograms associated with a given isometric torque in different postures, usually showing an increase of maximum EMG activity as the postures with shortest muscle lengths are approached. While this was consistently true for erector spinae and biceps brachii, one subject showed very little change of maximum EMG in the calf muscles except in extreme dorsiflexion.

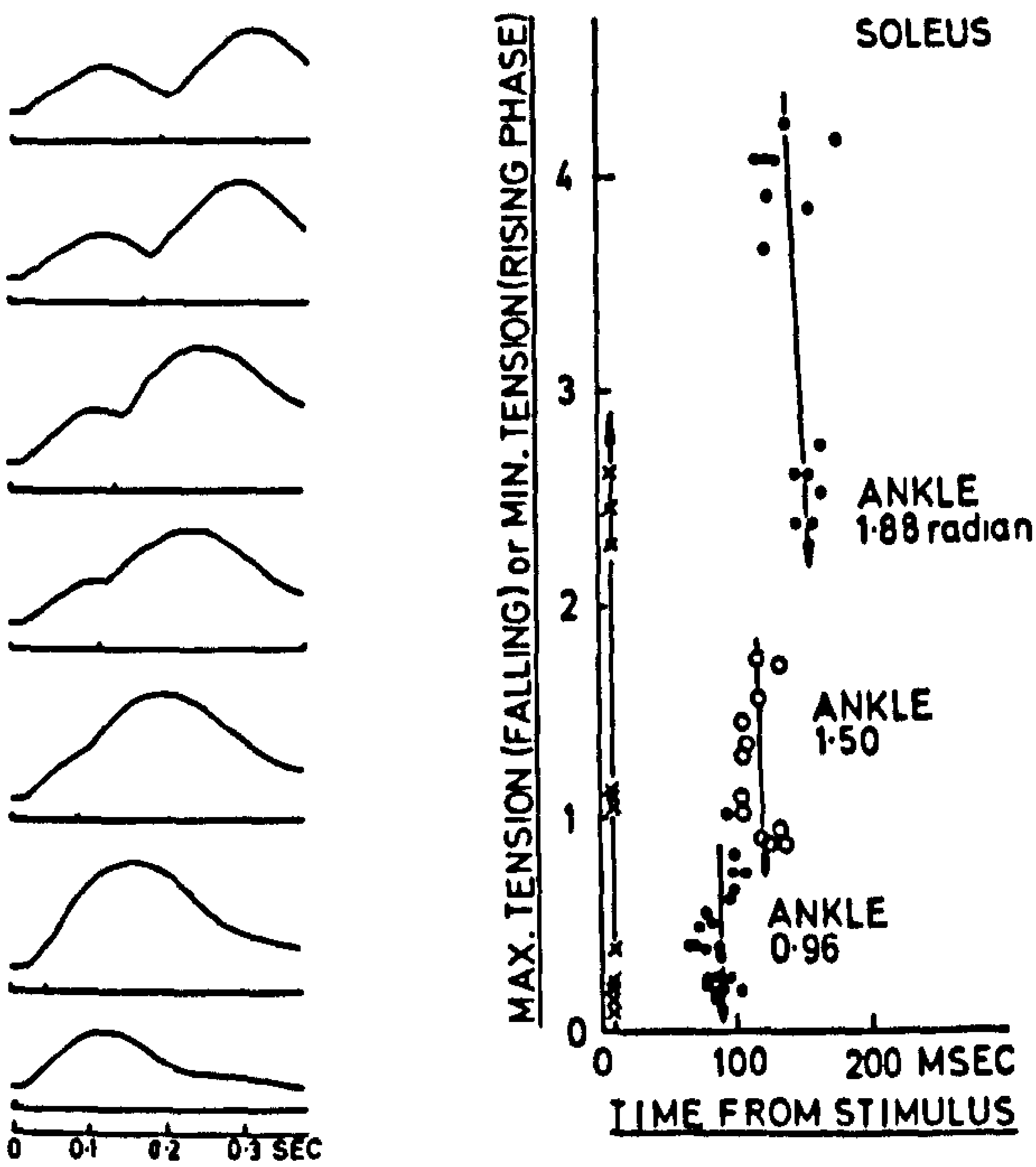

Fig. 7

a) Series of double-twitches from soleus (reducing delay between stimuli from above down). b) Times of occurrence of minimum and maximum tension (torque) relative to the time of the preceding stimulus for three postures of the ankle, obtained from records similar to those shown in (a).

Twitches

A series of partially summated twitches obtained with various delays between the stimuli, are shown in Fig. 7a. The peaks and troughs of the soleus torque record for three postures of the ankle are shown in Fig. 7b, with the time of occurrence of the preceding stimulus and resting torque as common origin. The troughs, regardless of the absolute torque, all occurred at 10 msec after the stimulus in the case of soleus and at 8 msec in the case of tibialis anticus.

The times of the peaks in a given posture fell in a vertical linear manner with time rather than sigmoid fashion. The technique did not permit the upper regions of the «contraction time» curve near tetanic torque to be established.

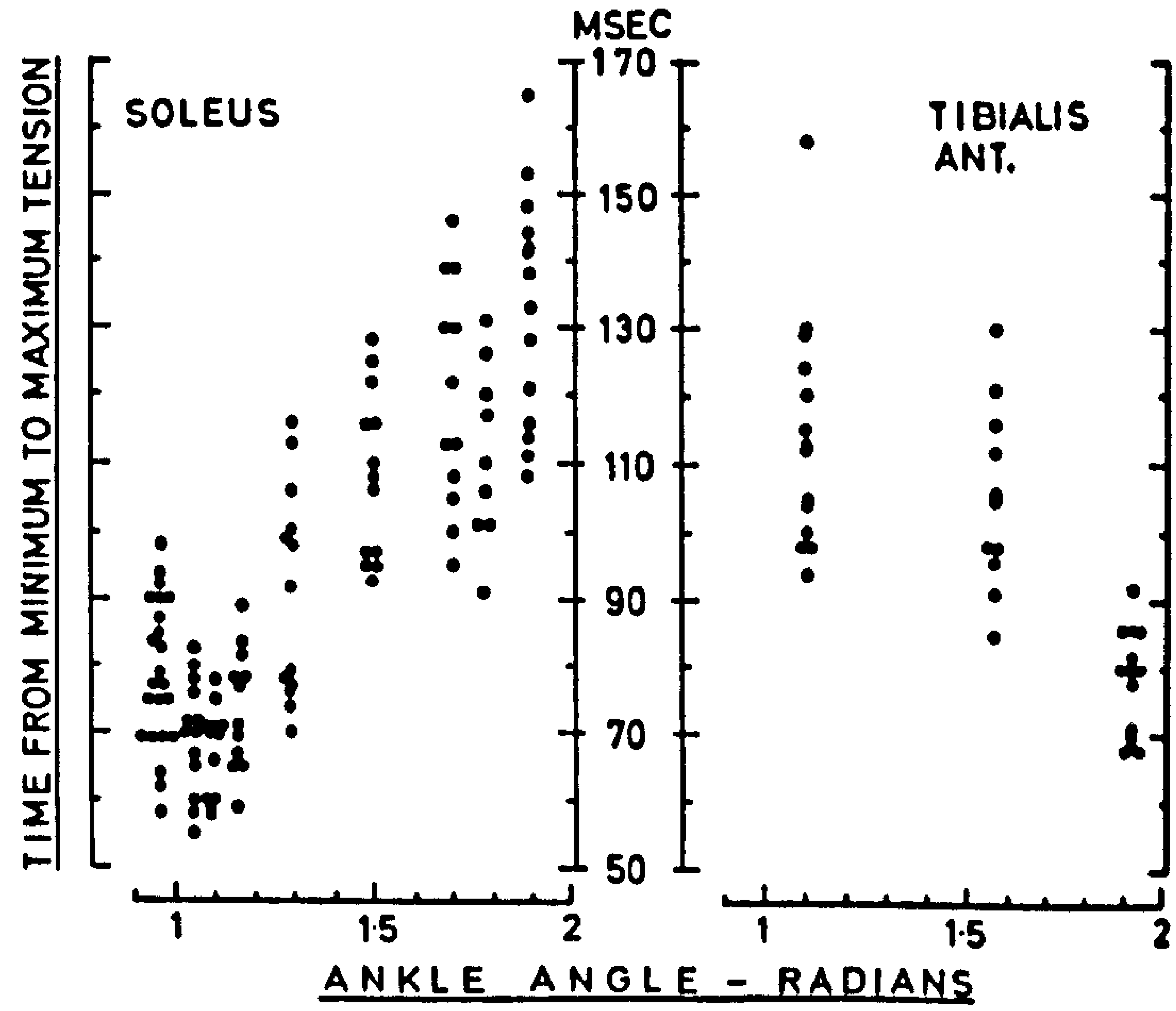

Fig. 8

Relationships between the intervals between minima and succeeding maxima of torque in partially-fused twitches as a function of ankle angle. Results for soleus and tibialis anterior in the same subject are shown on left and right respectively.

The distribution of trough-to-peak time-intervals for soleus and tibialis anticus are shown in Fig. 8 for various postures of the ankle. The range of times, which are strictly neither active state times nor contraction times (see Discussion) varied rapidly with ankle posture. In extreme dorsiflexion, when the largest twitches were obtained from soleus (partly because of the tension-length characteristics of that muscle), the trough-to-peak times were of the order of 130 msec (range 108 - 165 msec). The times were almost halved (70 msec, range 55 - 98 msec) in extreme plantraflexion. A mirror image of the soleus characteristic with ankle posture was found for tibialis anticus which of course changed in length in a reciprocal manner to that of soleus.

Discussion

In all the present measurements of voluntary effort, curvilinear relationships were found between the electrical activity and the torques about associated joints. Evidence of fatigue during the 22 seconds of a single graduated increase and decrease of effort was sought by comparison of EMG-torque relationships in the incremental and decremental phases. Although differences sometimes existed, they appeared to be random and did not show a consistent hysteretic lag of electrical activity in the decremental phase that would have suggested a progressive fatigue. All 22 observations in each effort were therefore considered together.

It is virtually impossible to avoid anatomical complexity in studies of human muscle. In the present experiments, the surface electromyograms sample from an unknown fraction of the fibre populations that are contributing to the mechanical outputs. In every case, some of the population is acting across more than one joint and other muscles may be acting synergistically or antagonistically at the same time without contributing to the recorded electrical activity.

Experiments with triceps surae in which two pairs of electrodes were sampled predominantly from soleus and gastrocnemius respectively suggested an interplay of the contributions of each muscle to the tension in the tendocalcaneum. A closer correlation was found between the torque about the ankle and the sum of the gastrocnemius and soleus activities than with either of the activities alone.

The curvilinear relationship between EMG and torque was to be expected from other reports. Where linear relationships have been found (with rectified and smoothed signals) it is likely that the observations were confined to a restricted part of the total range of torques, as Zuniga and Simons (1969) have discussed.

The present experiments extended from zero to maximal voluntary effort and in spite of the uncertainties (due to motivation for example) a dramatic and consistent feature of all the experiments was that maximum electrical activity decreased when the muscles were used in lengthened postures. This is not a consistent finding of other authors using other muscle groups although present results are compatible with those of Simons and Zuniga (1970) and Jorgensen and Bankov (1971). Vredenbregt & Rau (1973) state unequivocally that the maximum electrical activity was the same at all lengths of the biceps brachii muscle, while Haffaser et al. (1972) report activity which decreased with muscle length in quadriceps femoris. We cannot therefore claim that the explanation of our present results set out below applies to all muscle groups, although it would be surprising if the active state changes with length were not universally present even though masked by other factors.

The maximum voluntary torque in all experiments was a function of joint posture which is to be expected from the length tension relationships of active muscle. If this were the only length-dependent change of relevance, the associated maximum electrical activity might have been expected to be independent of length. Since the maximum electrical activity is diminished by increase of muscle length, a reasonable hypothesis is that more action potentials are required per unit time in order to cause maximal contraction when the muscles are short than when they are long. This would occur if the duration of the active state increased with muscle length. An alternative and not exclusive hypothesis is that the magnitude or width of the individual unit action potential changes with muscle length. Ralston and Libet (1953) found an increase in the amplitude of the action potential of frog muscle when stretched, but found that the phenomenon was absent in guinea pigs and rats. The ways in which unit potentials may summate to form the interference pattern of the surface electromyogram in the 1st dorsal interosseus muscle of the human hand has been studied by Milner-Brown and Stein (1975). It cannot be assumed that these results are directly applicable to the present study. More complex reasons might be sought in which the time for propagation along the muscle fibres altered in relation to the duration of the active state at any particular points along them.

If a muscle is considered as mechanically equivalent to a contractile component, which possesses length-tension and tension-velocity characteristics and a particular time course of active state (ability to develop tension or to shorten) when stimulated, in series with an elastic component which is devoid of viscous elements and does not creep, it is possible to determine the time course of part of the active state by timing the peaks and troughs of tension during a series of partially fused isometric twitches. The maxima and minima of tension occur when the series elastic element is momentarily in static equilibrium with the contractile element. There is little ambiguity in the result if an isolated muscle fibre is studied and the fibre is stimulated simultaneously in all parts of its length (Edman et al., 1966). When a whole excised muscle with a population of mixed fibre types is studied in this way, the peak of a twitch represents an instant when the sum of the tensions in all the elements reaches a peak, although none of the elements are necessarily in static equilibrium. If the time course of some or all of the active states of the individual fibres in the stimulated population are shortened, the time to peak twitch tension will be shortened. The comparison of the twitch measurements at different muscle lengths, in which different twitch tensions are produced, must take into account that the descending phase of an active state curve is sigmoid.

In the present experiments, the difference between the stimulus-to-trough and stimulus-to-peak torque times can only be interpreted as due to a shortening of the duration of at least some of the active states of individual fibres when muscle length is decreased. Nothing can be said about the shape of the trough-to-peak durations near maximal tension by means of this technique. The observed changes near the base of the curve suggest that the active state curves in these muscles are greatly affected by muscle length in a similar manner to that suggested by Rack and Westbury (1969) for cat, soleus muscle.

The twitch experiments were performed on only part of the fibre population of tibialis anterior and soleus muscles. The stimulus spreads indefinitely as the voltage is increased. It is possible that the stimulated population changes with posture although it seems highly unlikely that this would lead to the observed changes in contraction time.

The electrical activity at the shortest muscle lengths often reaches levels where it is increasing with almost no concomitant increase of torque. It is possible that under thee conditions the fibres are receiving nerve impulses at higher frequencies than are necessary for producing full mechanical activation. At greater muscle lengths this does not occur and it is possible that higher rates of discharge would lead to further increase of tension. Possibly, in spite of the subjects subjectively maximal efforts, there are inhibitory mechanisms at the longer lengths and higher tensions, which prevent further activation of the muscles. There is normally reckoned to be a reserve of effort available, if the subject can be sufficiently aroused and motivated. The greatest possible output and electromyographic activity were possibly not recorded, at least at the greater lengths. Due to the bulk of the muscle groups studied in the present experiments it was not feasible to stimulate all the contributing muscles to determine whether a torque could be developed which was greater than maximal voluntary effort, as described by Merton (1954) for the adductor pollicis muscle.

The present results have disturbing consequences for the in vivo classification of motor units on the basis of their contraction times. The changes with length are enough to radically alter the distribution of fibre bundle contraction times reported by Buchthal & Schmalbruch (1970) for the mixed populations of human soleus and tibialis anterior muscles.

In their early studies of back muscles, Floyd and Silver (1955) concluded that in the fully flexed posture, the trunk is supported against gravity by ligamentous tension alone. Morris, Brenner and Lucas (1962) have, however, found iliocostalis dorsi to be active in most of their subjects in the fully flexed posture. The subjects in the present study were by no means fully flexed when pulling on the bar at knee-height: it is therefore reasonable to assume that in more extreme postures very small levels of myoelectric activity might have substantial effects. Caution is hence required when judging stretched muscles to be «silent», especially when using surface electrodes.

The findings of the current study are of practical significance in relation to the use of the electromyogram in the assessment of posture for ergonomic purposes. Studies of this nature are too numerous to list, but share the common assumption that the intensity of myoelectric activity reflects the mechanical loading of the muscle group concerned. (The number of active motor units per se is of little interest in this context as the energy expenditure involved in the maintenance of these postures is in most cases trivial). Since posture itself radically effects the calibration of e.m.g. against mechanical output, to use the former as a measure of the mechanical load induced by a posture would seem to be a singularly misleading pursuit.

Anderson and Ortengren (1974) have recently carried out extensive studies on the effects of various parameters of chair design on the myoelectric activity of the back muscles of seated subjects. Figure 5 of the present study shows that in postures of truncal flexion (kyphosis) very modest levels of electrical activity can coexist with considerable mechanical loads, whereas in extended postures (lordosis) modest loads elicit substantial activity. The reduction in the level of myoelectric activity which may be achieved by encouraging kyphosis does not necessarily indicate a reduction of the mechanical stresses on the soft tissues of the back. It is therefore possible to overestimate the importance of reduction in e.m.g. activity achieved by, for example, the increase in seat-back angle (Andersson and Örtengren, 1974). Furthermore, the slight increase in activity that Andersson found with the addition of a lumbar support can be readily explained by the alteration of the e.m.g./load relationship due to increased lordosis. In the light of this information, the negative correlation between myoelectric activity and intra-discal pressure, which Andersson et al. (1974) found when varying lumbar support at constant chair-back angle, is not surprising.

Ward (1971) includes electromyography amongst a range of techniques as a means used in the determination of an optimal kitchen working-surface height. Current findings for gastrocnemius, soleus and erector spinae, all of which were used in Ward's study suggest that such a procedure is of dubious value unless some allowances are made for posturally dependent changes in e.m.g./load relationship.

The integrated electromyogram of biceps has been used by Tichauer (1968) and subsequently by Saran (1973) in the comparative evaluation of hand tools in a pronation-supination task. A consideration of the angular ranges in which these tools were used would seem essential in the interpretation of these results. The same reservations must be expressed with regard to Tichauer's (1971) investigations of lifting and carrying activities.

In the light of, firstly, the e.m.g./load relationship and, secondly, the fact that substantial forces may be found in association with small electrical activities, a strong case may be made out for the use of e.m.g. amplifiers with a logarithmic voltage gain characteristic. Such devices, now used in some laboratories, have the dual advantage of being sensitive to the low level events associated with postural maintainence, but at the same time not subject to overloading during more intense activity.

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