OBSERVATIONS ON THE POSTURAL MECHANISM OF THE HUMAN KNEE JOINT

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INTRODUCTION

An essential factor in the maintenance of the normal postures of standing (Smith, 1953) is the temporary stabilization of the joints transmitting body weight. It is widely held that the knee joint, in common with other weight-bearing joints, is orientated and stabilized by that low grade and diffuse firing of the antigravitational muscles which has been called postural or tonic contraction. This view is based on the classical work of Sherrington, Magnus and their co-workers, and involves a transference of the results of their animal experiments to the postural mechanism of man.

In the experimental animal the position of the knee joint is controlled during standing by the postural contraction of the quadriceps muscle, but this fact, of itself, gives no clue to the identity of the antigravitational muscles at the human knee joint: this can only be determined by a consideration of the influence exerted by body weight on that joint during standing. In the child who is learning to stand, the angle between the leg and thigh is considerably less than 180° (Pl. 1, fig. 3), and it is apparent that body weight tends to flex the knee joint. Thus in the infant, as in the experimental animal, the muscles which are potentially capable of effecting stability at the knee joint during standing are the extensor group. On the other hand, in the stance of the adolescent and adult, as is now generally accepted (Johnston & Whillis, 1954; Brash, 1951), the centre of gravity of the body falls in front of the axes of the knee joints, and the angle between the leg and thigh exceeds 180° (Pl. 1, figs. 4, 5). Consequently, in these age groups, the quadriceps muscle cannot control the position of the knee joint as has been suggested by Denny-Brown (1949): on the contrary, the ability to perform this function is necessarily transferred to the flexor muscle group. The conditions existing at the adult human knee joint appear to be peculiar to man, indeed, there is considerable evidence to suggest that they are peculiar to modern man (Morton, 1935; Hooton, 1947), and it might therefore be postulated that the ontogenetic change in the posture of the knee joint during standing, and consequently in the identity of the antigravitational muscles, is a recapitulation of a phylogenic change.

The electromyographic studies of the adult lower limb muscles in standing which have been described by Seyffarth (1940, 1941), Weddell, Feinstein & Pattle (1944), Joseph & Nightingale (1952), Smith (1954a), indicate that there is in fact a low-grade activity in the hamstring and posterior crural muscle groups. Attempts to compute the total muscle force represented by such electromyograms are notoriously hazardous, though it has been suggested recently (Lippold, 1952) that it can be accomplished with some accuracy in the case of voluntary contraction in the living subject. In

any event, it seems probable that the electromyographic record of the flexor muscles during standing represents an activity insufficient of itself to stabilize the knee joint. As a result there is to-day increasing support (Wright, 1952; Ralston & Libet, 1953) for the view that the adult knee joint is stabilized during standing by two complementary and harmonious factors. One of these factors is the postural contraction of the flexor muscles, and the other is the passive resistance of the tissues of the part. This view has been clearly expressed by Clemmesen (1951) thus: 'The relaxed standing position...is determined and equilibrium obtained by postural reflex tone in association with the elastic forces and in opposition to gravity.'

The investigation described in this paper is an inquiry into the nature of the *passive* mechanism which resists extension at the knee joint, and the extent of the contribution which it makes to the stabilization of the knee joint in various postural circumstances.

It is postulated at the outset that the passive resistance to extension at the knee joint is divisible into two components which normally operate in harmony with each other. The one component is the resistance which is exerted by the recognized articular mechanism described by Goodsir (1868), and this will be referred to as the articular resistance; the other component is that exerted by all those extra-articular structures which are stretched by the movement of extension, and this will be referred to as the extra-articular resistance.

Before proceeding to an experimental analysis of these two factors, a description is given of the elastic properties of the articular and extra-articular tissues, and of the influence of these properties on the nature of the resistance.

The nature of the articular resistance

The articular mechanism which limits extension at the knee joint was described by Goodsir (1868), and this description has been elaborated and clarified by the recent work of Barnett (1953). Goodsir showed that in osteo-ligamentous preparations of the knee joint, extension is limited by a mechanism which involves the tightening of joint ligaments, and the simultaneous locking of paired articular surfaces of logarithmic spiral curvature; that is, by tensile stresses in joint ligaments and compression stresses in the tissues deep to the articular surfaces. It is widely held that the tissues involved in this mechanism, namely joint ligaments, articular cartilage, and intra-articular fibrocartilages, are practically unchanged in their dimensions by any external force to which they might normally be subjected. Moreover, this conception has led to the assumption that Goodsir's mechanism becomes operative at one particular position of the knee joint, and at that position arrests extension immediately and completely, an assumption which is in conformity with the view of Walmsley (1928).

If this were the true nature of the locking mechanism at the knee joint, then it would have to be accepted that only in exceptional circumstances would the mechanism be a significant factor in the stabilization of the knee joint, for in all common activities, such as standing, walking and running, the maximum extension of that joint is always appreciably short of the full voluntary movement (see below).

However, there is now abundant evidence that the tissues involved in the articular mechanism, far from being rigid, are each appreciably distorted by stress

of normal dimensions. Furthermore, within certain limits, each of these tissues reacts elastically when relieved of stress.

The ready deformability and elasticity of articular cartilage has long been recognized (Braune & Fischer, 1891). Bär (1926) and, more recently, Hirsch (1944) recognized that the elasticity of this tissue has peculiar features: when it is subjected, after death, to a uniform compression stress, instantaneous deformation is followed by a finite rate of deformation which becomes less as the duration of the load is prolonged. When the stress ceases, recoil follows a similar pattern (Text-fig.1), but the extent of recoil, i.e. the elasticity, varies with the duration of the preceding stress. After stress of short duration (less than 1 min.) elasticity is perfect, but with stresses of longer duration the elasticity becomes increasingly imperfect. Ingelmark & Ekholm (1948) confirmed these reactions and proposed on experimental evidence that they were due to a dual mechanism of compression and recoil. They suggested that both processes were due in part to true elasticity (instantaneous deformation and recoil) and in part to a migration of fluid between the cartilage on the one hand
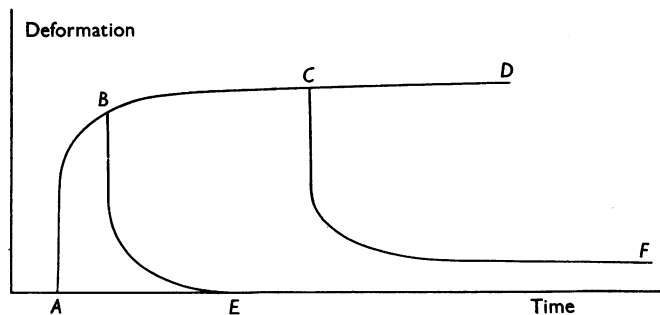
Text-fig. 1. The deformation of articular cartilage caused by a constant compression stress is shown by a line *ABCD*. If the stress ceases after a short time the recoil *BE* is complete, but if the stress is maintained until *C*, the recoil *CF* is incomplete.

and the synovial fluid and the vessels of the marrow cavity on the other. It seems probable, therefore, that the imperfect elasticity noted by Bär and Hirsch is in some measure at least due to the lack, under experimental conditions, of the fluid pool on which cartilage must draw during recoil, and that during life a relatively prolonged compression stress would be followed by a recoil which, although gradual, was eventually complete. At all events it is obvious that in life articular cartilage must normally recoil completely after stress of physiological dimensions, and Fairbank (1948) has demonstrated that the recoil is in fact delayed, so that the articular cartilage of the knee joint is some 2 mm. thinner in the evening than it is in the morning, and recovers its original thickness only after a period of rest. Thus in relation to physiological stress, living articular cartilage is readily deformable and perfectly elastic, but both the extent of the deformation and the duration of the process of recoil are dependent on the magnitude, the duration and the proximity in time of that stress.

It is also worthy of note that as a result of these observations Ingelmark (1950) has put forward the hypothesis that it is by the migration of fluid, brought about by repeated discrete compression stresses, that the nutrition of articular cartilage

is maintained. A corollary to this is, that there must be an optimum duration of compression stress, and in view of Hirsch's observation on elasticity this seems likely to be of the order of one minute.

Intra-articular fibrocartilages may be deformed in a variety of ways by different types of stress, but in relation to the present problem only two of these stresses are of importance. Thus fibrocartilages of annular form may experience a circumferential tensile stress. Fairbank (1948) and Mathur, McDonald & Ghormley (1949) have demonstrated that stress of this kind causes an appreciable deformation, and in my own experience this deformation is elastic in nature. On the other hand, all intra-articular fibro-cartilages may be subjected to compression stress by the neighbouring bone ends and my own observations have shown that they react to such stress in a manner similar to articular cartilage.

The elastic deformation caused in joint ligaments by tensile stress was recognized by Annovazzi (1928), and some features of this property have recently been described (Smith, 1954b). It has to be recognized in the present context that the temporary elongation produced in a joint ligament by a large load of short duration is no infinitesimal change requiring special methods for its demonstration. On the contrary, it is a gross elongation of up to 20 % of the resting length, and it can be readily appreciated with the unaided eye.

Furthermore, the stress/strain relationship gives no support to the view that elongation of a ligament is dependent on stretching of so-called elastic fibres and straightening of sinuous collagen bundles: rather it corroborates that it is dependent on elongation of the collagen fibres which are themselves deformable and elastic.

On the basis of the tissue properties described above it is postulated that the articular mechanism at the knee joint operates in the manner illustrated by the three diagrams in Text-fig. 2. These illustrations represent the knee joint in three different positions: they are essentially schematic and some features have been intentionally exaggerated. In each diagram the femur above articulates with the tibia and a meniscus below. The opposed femoral and tibio-meniscal surfaces are spirally curved, the curvature becoming progressively more acute from before backwards. The deformable tissues deep to these surfaces are indicated as stippled areas. The single joint ligament is so orientated as to tighten during extension. It is generally accepted that when the joint is extended from a position such as that in Text-fig. 2a, neither the articular surfaces nor the joint ligaments offer any resistance to the movement. Thus Haines (1941) has pointed out that, although both cruciate ligaments are then taut, the tension in one is in equilibrium with that in the other, and Barnett (1953) has demonstrated that although the lateral tibio-meniscal surface may be congruent with its femoral counterpart well before full extension is reached, there is no locking, because its curvature in the sagittal plane is subsequently rendered less acute by forward movement of the meniscus on the tibia (Text-fig. 2b). At or about the position indicated in Text-fig. 2b the curvature of the lateral tibio-meniscal surface becomes maximal, the paired femoral and tibio-meniscal surfaces become congruent, and the ligamentous apparatus of the joint becomes taut, but the advent of these factors does not necessarily limit extension. On the contrary, in proportion to the magnitude of the extending force, the joint ligaments will be stretched and the articular surfaces will be moulded to a new form, so that the joint will move, say,

to the position indicated in Text-fig. 2*c* before the stresses in the deformed tissues
are sufficient to arrest movement.

Thus it is postulated that the exact limit of extension at the knee joint varies with
the magnitude of the extending force. Furthermore, it is suggested that in relation
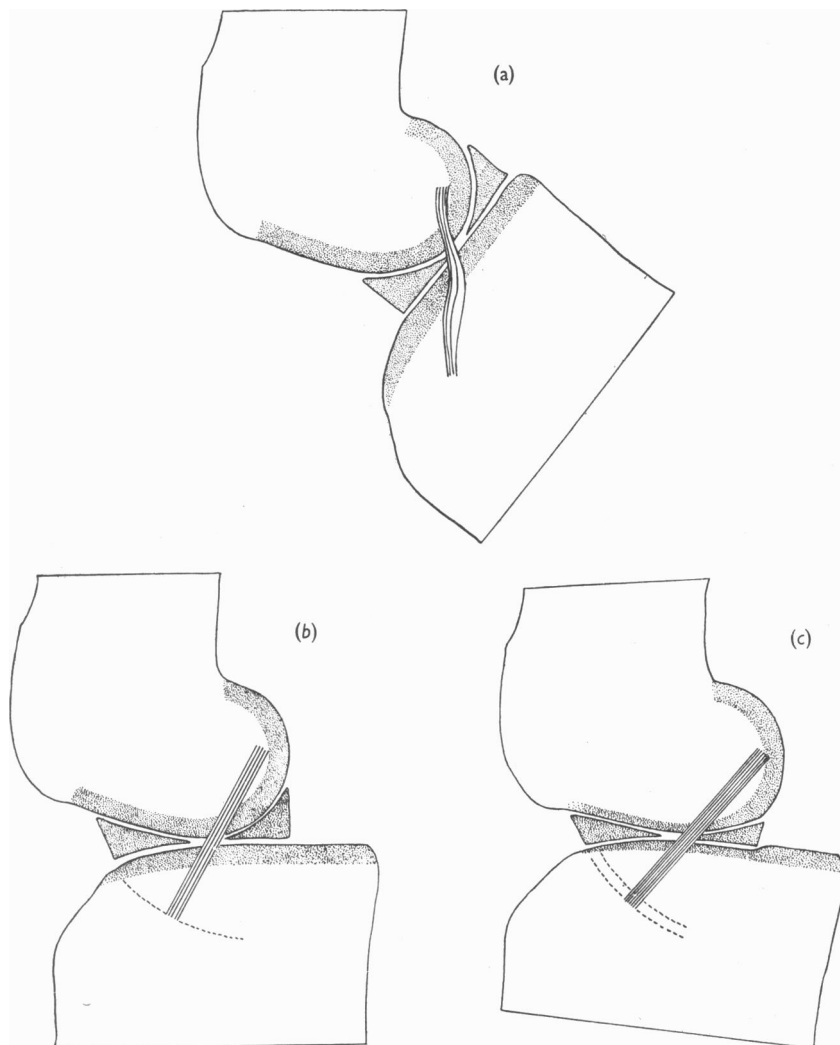
Text-fig. 2. Diagrammatic representation of movement at the knee joint. The interrupted lines in
(*b*) and (*c*) indicate the elongation of the ligament which occurs during movement between
these two positions.

to all but the smallest extending forces, the terminal degrees of joint movement are
only permitted by tissue distortion, and are therefore resisted in ever increasing
measure by the distortion stresses in those tissues. This hypothesis is in effect a
compromise between the view of Strasser (1914) and Fick (1911) who believed that
the deformability of articular cartilage was such as to prevent any effective locking

of joint surfaces, and the more recent concept of Walmsley (1928) who held that any flexibility in the locking mechanism was negligible. It receives some support from the observations which MacConaill (1945) has made on the function of the talo-calcaneo-navicular joint, and it is in keeping with certain observations of Brantigan & Voshell (1941), who noted that the position of full extension at the knee joint varied with the extending force, and that the menisci were compressed—that is the tibio-meniscal surfaces were remoulded—in the 'hyperextended' position.

The nature of the extra-articular resistance

It is considered that the extra-articular tissues which, by their resistance to stretch, exert an appreciable passive resistance to extension at the knee joint, are the flexor muscles, and the deep fascia and skin over the popliteal fossa.

It has long been recognized that an inactive or denervated muscle is elongated by load; that the extent of the elongation is related to the duration of the load; and that, within certain limits, the muscle recovers its original length when the load is removed (Langelaan, 1915). It is now considered (Fulton, 1949) that this elasticity is referable, not to the contractile actomyosin, but to the sarcolemma and fibrous framework of the muscle. As joint movement separates the attachments of an inactive muscle, the muscle becomes taut before its resting length is reached, and thereafter the tension—and consequently the resistance to the movement—increases at an increasing rate until the joint movement is complete (Fulton, 1949). The joint movement may of course stimulate contraction of the muscle through the myotatic reflex, but even if this occurs the passive tension is added to, and may be greater than, the active tension (Creed, Denny-Brown, Liddell & Sherrington, 1932). Thus in extension at the knee joint a terminal range of the movement is presumably resisted by passive tension in the flexor group of muscles.

The functions which are generally attributed to the deep fascia of the limbs are those of protection, the maintenance of shape, the promotion of venous return and the provision of extra-osseous muscle origins, but the additional function of acting as a remote ligament to the joints which it crosses is usually recognized only in the plantar aponeurosis. It is considered, however, that this ligamentous function is attributable to many other areas of fascia, and in particular to that covering the popliteal fossa. This fascia is attached proximally to the iliac crest and distally to the medial tibial condyle, the head of the fibula, the malleoli and the calcaneum. When the knee joint is fully extended a tension can be palpated in the tissues covering the central part of the popliteal fossa—a tension which can be accentuated by dorsiflexing the ankle or flexing the hip. It is considered that this tension is referable in large measure to the deep fascia, but the skin and the popliteal neuro-vacular bundle must also be regarded as minor contributory factors. Gratz (1931) showed that deep fascia was readily and proportionately elongated by stress and that it was, moreover, highly elastic, returning to its original dimensions when relieved of stress. It is probable therefore that the deep fascia becomes taut before full extension of the knee joint is reached, and resists movement in steadily increasing measure over the subsequent terminal range.

It is a matter of common experience that skin is stretched by tensile stress and returns elastically to its original dimensions when relieved. The simple manœuvre

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of pinching up the skin over the popliteal fossa shows that the skin tension increases progressively over a wide terminal range of extension, and it follows that this tension exerts some restraint on the movement.

It is considered therefore that a terminal range of extension at the knee joint is resisted collectively by passive tension in the flexor muscles and in the relevant areas of skin and fascia.

FINDINGS

The nature of the articular and extra-articular resistance to extension at the knee joint and of their sum—the total passive resistance—to extension can be substantiated and placed on a quantitative basis by the following experiments.
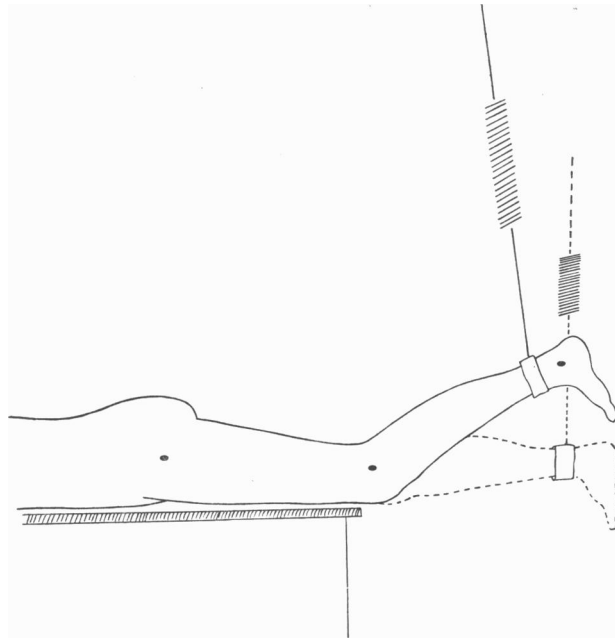
Text-fig. 3. Determination of the total passive resistance to extension at the knee joint. The leg is shown in two of the positions which it occupies during the experiment.

Experimental assessment of the total passive resistance to extension

The experiment illustrated in Text-fig. 3 was performed on the knee joints of three young adult subjects. In each case the experiment was carried out while the subject was anaesthetized and under the influence of a relaxant before an operation for a mild chronic condition. Dr Shearer, to whom acknowledgements are made later in this paper, informs me that anaesthesia was induced by 3·75 g. thiopentone. An intravenous injection of 5·0 mg. decamethonium iodide (Eulissin) was then given and anaesthesia was continued by rhythmic inflation with gas and oxygen (2:1) with carbon dioxide absorption. The patient was completely apnoeic and all muscles were completely flaccid on manipulation. On this evidence, together with the dose

of relaxant administered, it was assumed that all limb muscles were completely relaxed.

In each experiment black markers were placed over the tip of the lateral malleolus, the lateral femoral epicondyle and the tip of the greater tronchanter. The first two were regarded as marking the axes of the knee and ankle joints. The relationship of the third marker to the centre of the head of the femur is variable: in forty-two femora examined the distance of the centre of the femoral head in front of the greater tronchanter varied from zero to 1 in. with an average of 0·6 in. This average figure has been used in the three subjects examined. It is appreciated that its use involves an approximation but the maximum resulting error in absolute measurements of the angulation of the knee joint is only about 2°.

In each experiment the subject was placed in the prone position so that the knee joints lay just beyond the edge of the trolley. The leg to be examined lay passively in a sling which was connected above to a calibrated tensile spring. A loop was attached to the upper end of the spring and by this means an assistant could raise or lower the leg at will. Commencing with an angle of about 140° between the leg and thigh, the leg was gradually lowered until extension was naturally limited. While the movement was in progress the relative positions of the leg and thigh and the length of the spring were recorded on cine film at sixteen frames per second. Subsequently an enlarged tracing was made of every fourth frame, and from these tracings the several quantities detailed below were measured.

The experiment was then repeated with the subject in the supine instead of the prone position, the knee joint being raised in stages in this instance between about 140 and 175°.

The forces operating at the knee joint during these experiments are shown in Text-fig. 4. B is the angle between the leg and thigh and A the angle between the leg and the horizontal. S is the tension of the tensile spring in pounds and L the minimal distance in feet from the line of the spring to the knee axis. W is the weight of the leg and foot in pounds and D the distance in feet of the centre of gravity of these parts from the knee axis. R is the total passive resistance to extension and is regarded arbitrarily as a torque acting at a distance of one foot from the knee.

In the first experiment (Text-fig. 4a) the torque resisting extension when the angulation of the knee joint is $B°$ is

$$R_B = WD \cos A_1 - S_1 L_1. \tag{1}$$

The resistance at the same angulation in the second experiment (Text-fig. 4b) is

$$R_B = S_2 L_2 - WD \cos A_2. \tag{2}$$

In both equations the quantities W and D are unknown, and neither can be directly measured. However, the value of the factor (WD) can be calculated because the passive resistance to extension must be the same in the two experiments for equal angulations of the knee joint. In other words, for any value of B

$$WD \cos A_1 - S_1 L_1 = S_2 L_2 - WD \cos A_2,$$

or
$$WD = \frac{S_2 L_2 + S_1 L_1}{\cos A_1 + \cos A_2}. \tag{3}$$

Once the value of (WD) is known R can be calculated for all values of B from equation (1).

Thus in the first subject the values of A_1, A_2, (S_1L_1) and (S_2L_2) for different values of B were as shown in Text-fig. 5.

Substituting these values in equation (3) the values of WD were calculated for values of B between 145 and 175°, and it was found to vary between 6·9 and 7·1 with an average of 7·0 lb. ft. It is considered that this method of assessing the value
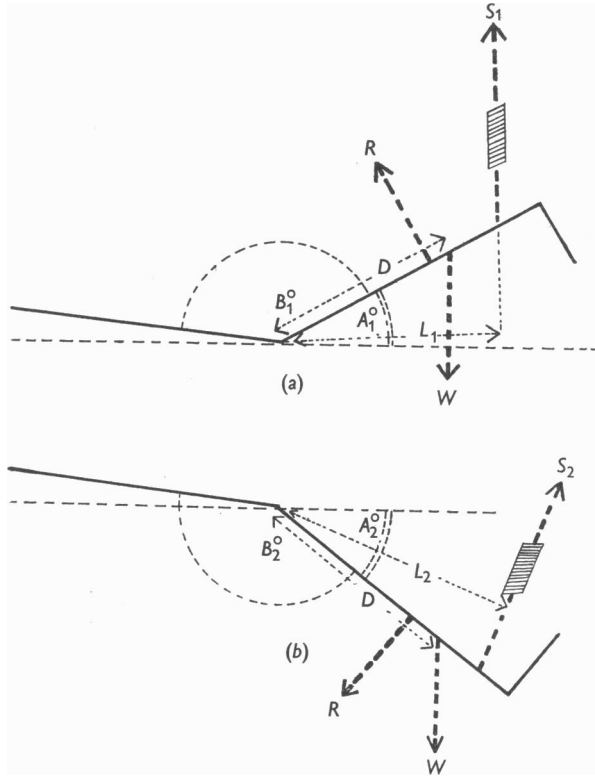
Text-fig. 4. The forces operating during the experiments described in the text, (a) with the subject prone, and (b) with the subject supine. In each case the solid lines indicate the positions of the thigh, leg and foot.

of WD is more accurate than the use of the formulae of Braune & Fischer, quoted by Fenn (1930a), which are dependent on observations on the cadaver and necessarily take no account of individual variations.

Using this value of WD and the values of A_1 and S_1L_1 in Text-fig 5, the total passive resistance to extension (R) at different angulations of the knee joint (B) can be calculated from equation (1).

The values of the total passive resistance to extension between 140° and full extension in the three subjects examined are shown in Text-fig. 6a. The results in the three individuals are closely similar, and their average is shown in Text-fig. 6b. It is apparent that at 150° the resistance is a negative quantity, that is, the stresses in the tissues tend to *promote* extension and resist flexion. At about 160° the tissue

stresses are in equilibrium, promoting neither flexion nor extension. Thereafter, the passive resistance to extension increases until, when the leg and thigh are in alignment, it has a value of 1·4 lb. ft. Beyond this angulation the resistance increases at a progressively increasing rate and at 192°, the position at which extension was naturally limited, it amounts to 7·0 lb. ft.

Experimental assessment of the articular resistance to extension

The experiment was carried out on four lower limbs which had been amputated through the mid-thigh not more than 1½ hr. earlier. The skin and fascia around the knee joint, and all the muscles directly associated with the joint, were divided at convenient levels but not removed. The leg was then divided at approximately the junction of the upper and middle thirds. Thereafter steel pins were driven into
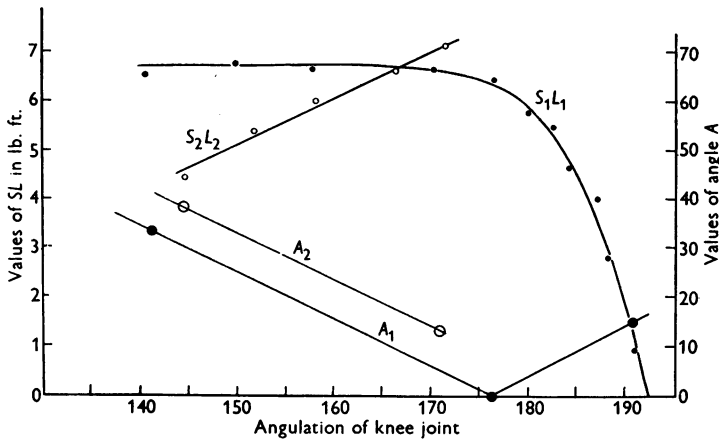

Text-fig. 5. Graphs showing the values of A_1, A_2, (S_1L_1) and (S_2L_2) for different values of B in the first subject.

the posterior aspects of the femur and the tibia close to the joint and in a sagittal plane, so that any change from the initial angulation of the knee joint could be recorded by photographing the limb from the side. The femur was fixed with its anterior aspect facing downwards, and after the knee joint had settled into extension the limb was photographed (Pl. 1, fig. 7). Subsequently weights were applied, in increments of 2 lb. to a maximum of 20 lb., to the stump of the leg at a measured distance from the axis of the knee joint (0·48 ft.) and after each addition the limb was re-photographed and the relative angulation of the knee joint measured. The upper third of the leg was then detached from the femur and its weight (2·6 lb.) and the distance of its centre of gravity from the knee axis (0·35 ft.) determined: the same factors were then determined for the segment consisting of the lower two-thirds of the leg and the foot (3·8 lb. and 1·24 ft.). The figures given in brackets apply to one of the four specimens examined.

As indicated in Text-fig. 7, the total extending torque operating on the knee joint at any stage of the experiment is

$$\text{torque} = \cos A(W_1D_1 + W_2D_2) \text{ lb. ft.} \tag{4}$$
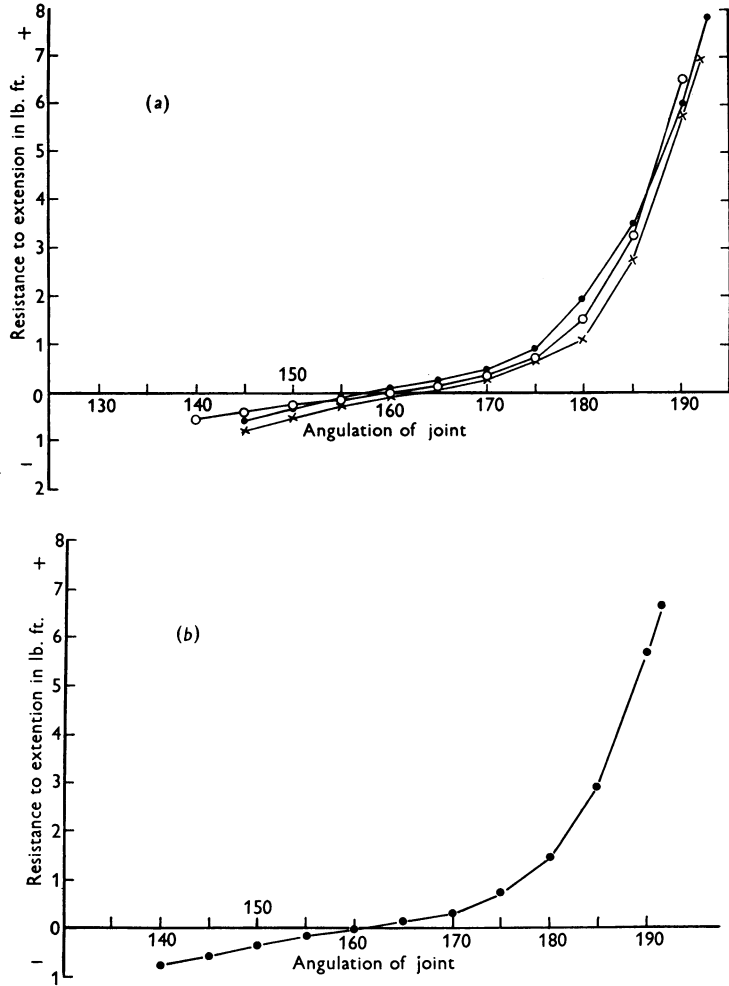
Text-fig. 6. The total passive resistance to extension at the knee joint (a) in the three subjects examined, (b) the average value.
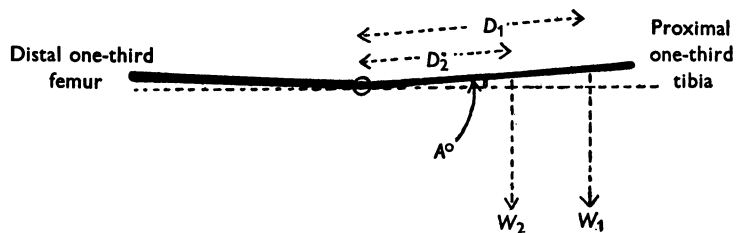
Text-fig. 7. The lower part of the femur is indicated by the thick line on the left of the diagram and the upper third of the leg by the thick line on the right. The circle shows the position of the knee axis.

W_2 is the weight of the upper third of the leg in lb. acting downwards from the centre of gravity of that part, which is situated D_2 ft. from the knee axis. W_1 is the applied load acting at a distance of D_1 ft. from the same axis. A is the angle between the leg and the horizontal. Throughout these experiments A was small (less than 8°), and therefore cos A was almost unity; furthermore, in the specific experiment being described $W_2 D_2 = 0.91$ and $D_1 = 0.48$, and therefore equation (4) becomes

$$\text{torque} = 0.91 + 0.48 \ W_1 \text{ lb. ft.}$$

In Text-fig. 8 the total extending torque acting at the knee joint is plotted against the relative angulation of the joint. The point X on this graph represents the position in which the limb would have settled if the lower two-thirds of the leg and the foot had not been detached and no loads had been applied, for in those circumstances the total extending torque would have been

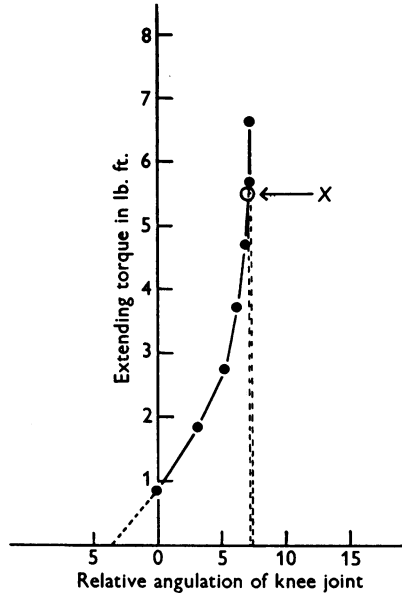$$\text{torque} = 0.91 + (3.8) \ (1.24)$$
$$= 5.62 \text{ lb. ft.}$$

In each stage of the experiment described above the extending torque acting at the knee joint is equalized by the passive resistance of the articular mechanism, so that the graph in Text-fig. 8 indicates the magnitude of this passive resistance over the terminal range of extension.

Thus it is apparent that the function of the articular mechanism at the knee joint is not confined to one position of the joint. On the contrary, the movement of extension is resisted by the mechanism over a range of about 10°, and may be limited by it anywhere in that range depending on the magnitude of the extending torque. When the knee joint is extended by the weight of the leg and foot it

Text-fig. 8. The relationship between the extending torque and the angulation, in a stripped preparation of the knee joint.

is practically at the limit of its range: increase in the extending torque does produce some additional extension, but this is very slight. The specimens in Pl. 1, figs. 6 and 7, illustrate this property of the articular mechanism.

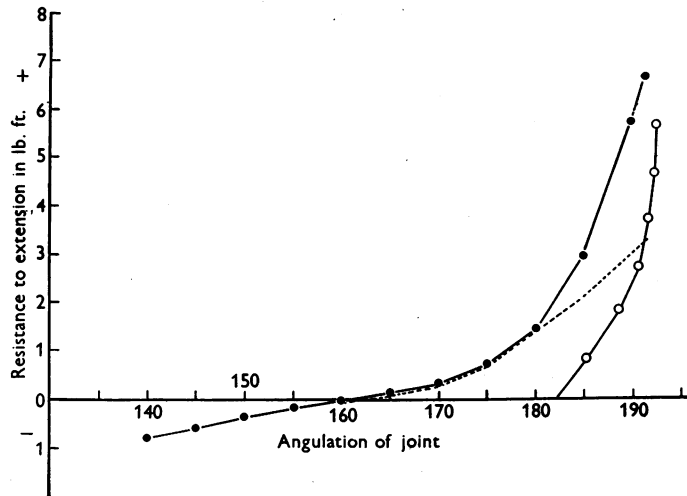### Assessment of the extra-articular resistance to extension

I have been unable to find any method of measuring the extra-articular resistance directly. In the living subject it is impossible to eliminate the articular mechanism, and in amputated lower limbs—apart from the rare hindquarter amputation—many of the extra-articular tissues which would normally resist extension have no proximal attachment. However, despite this difficulty, an indirect assessment can be reached owing to the fact that the resistance exerted by the extra-articular tissues alone, is equal to the difference between the total passive resistance (Text-fig. 6b) and the resistance of the articular mechanism (Text-fig. 8). Unfortunately, the graphs in

these two figures cannot be exactly correlated; they refer to the knee joints of different individuals of different ages, and in the experiment expressed in Text-fig. 8, the exact angulations of the knee joint were unknown because the specimen necessarily lacked the upper half of the femur. Nevertheless, it is considered that the tentative correlation made in Text-fig. 9 is sufficiently exact for it to be said that from about 160° to about 180° passive resistance to extension is due entirely to the extra-articular tissues, and that thereafter the movement is resisted by the articular and extra-articular tissues together.

Furthermore, it indicates that over the greater part of the range in which the extra-articular and articular mechanisms operate together, the former is the greater of the two factors.

Text-fig. 9. The passive resistance to extension at the knee joint. The total passive resistance is represented by the upper solid line, the extra-articular resistance by the interrupted line, and the articular resistance by the lower solid line.

It is apparent from these observations that whenever extension at the knee joint is arrested, or prevented, at an angulation of more than about 160°, this is achieved, in part at least, by the passive resistance which is automatically exerted by the tissues of the part. The movement may be limited by the passive resistance alone, or by the passive resistance operating in conjunction with active contraction of the flexor muscles, and the only factors which influence the choice of mechanism are the magnitude of the extending torque which has to be resisted, and the angulation at which the knee joint must be stabilized.

The total passive resistance to extension, and the articular and extra-articular resistances which are illustrated in Text-fig. 9, are those operating in one particular set of circumstances, namely when the hip joint is practically in full extension and the foot is at right angles to the leg. It will be apparent that flexion at the hip or dorsiflexion at the ankle, although having no effect on the articular mechanism at the knee joint, must tighten the tissues involved in the extra-articular mechanism

and thus increase the range and magnitude of extra-articular resistance. But despite this variation in the resistance to extension at the knee joint with variations in the positions of the hip and ankle joints, it is considered that the positions of these joints in standing make the resistance illustrated in Text-fig. 9 approximately applicable to the study of the stability of the knee joint during that act.

It is also to be noted that the resistance of the articular limiting mechanism illustrated in Text-fig. 8 is that exerted in the non-weight bearing limb. It is apparent that weight bearing will *per se* cause compression of the tissues deep to the articular surface, and that if it were possible to measure the resistance of the articular mechanism alone in the living subject after prolonged weight bearing, it would be rather less and would operate over a slightly smaller range.

There is another aspect of the passive limiting mechanism which is of more academic interest. It has already been observed that in relation to two of the tissues involved in the mechanism, namely cartilage and muscle, the relationship between a given stress and the resulting deformation is not constant but that, within certain limits, it varies with the duration of the stress. For this reason it is to be presumed that when a prolonged extending torque of constant magnitude is passively limited, the limitation will not be absolute. On the contrary, it is to be expected that, as long as the extending torque continues to act, extension at the knee joint will proceed at a slow and gradually decreasing rate until equilibrium is finally established. This phenomenon can be observed in the living subject. When a subject lies passively in the supine position with no support between the sacrum and the lower third of the leg (Text-fig. 10), it is found that after 10 min. the anterior surface of the patella has descended half a centimetre, a displacement which represents an increase in extension of about 1·7°. In other words, prolonged stress acting on the knee joint creates a new limit of extension, and the normal limit, if such a term may be applied to it, is re-established only after a period of rest.

The influence exerted by the passive limiting mechanism on the knee joint in a number of common circumstances is discussed below.

THE STABILITY OF THE KNEE JOINT IN SYMMETRICAL STANDING

In the symmetrical form of standing (Smith, 1953), illustrated in Pl. 1, fig. 1, each lower limb transmits half the weight of that part of the body above the knee joints (K). The extending torque operating at each knee joint in this posture is the product of K and F, when F is the horizontal distance which separates the line of the centre of gravity of that part of the body above the knee joints from the transverse axis of the knee joint (Text-fig. 11).

Certain approximations have to be made before this extending torque can be calculated. Thus the weight of the leg and foot cannot be measured in the living subject, and in the subsequent calculations it has been regarded as being equal to the measurable quantity WD (see p. 243), because D is approximately equal to 1 ft. in most subjects. Secondly, the position of the centre of gravity of that part of the body above the knee joints cannot be determined in the standing subject, and it has consequently been assumed to lie in the same coronal plane as the centre of gravity of the whole body.

Thus far in this paper the transverse axis of the knee joint has been consistently identified with the lateral femoral epicondyle because a slight error in its position has been of little significance. In the present problem, however, the exact position of the axis must be known. Its position at various angulations of the knee joint has therefore been determined from a series of lateral photographs of the lower
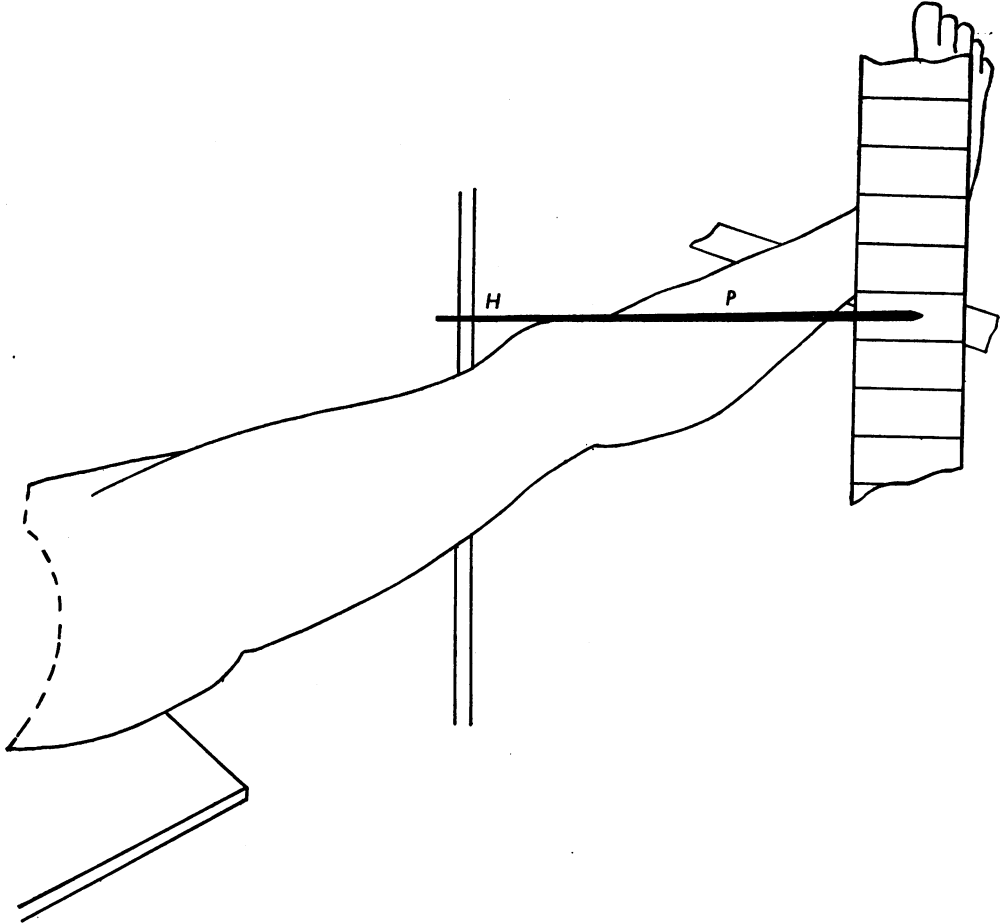
Text-fig. 10. The lower limb, supported only at the sacrum and the lower third of the leg. A pointer *P*, lying at right angles to the long axis of the limb is hinged at *H*: it rests on the anterior surface of the patella and its free end moves over a vertical scale.

limb taken at different angulations. These photographs were superimposed and the position of the knee axis constructed from the arc traversed by a marker on the lateral malleolus.

To determine the extending torque on the knee joint, black marks were placed over the greater trochanter, the lateral femoral epicondyle and the lateral malleolus, and thus prepared each subject stood symmetrically on a broad board which rested at one end on a fulcrum and was suspended at the other through a spring balance (Text-fig. 12).
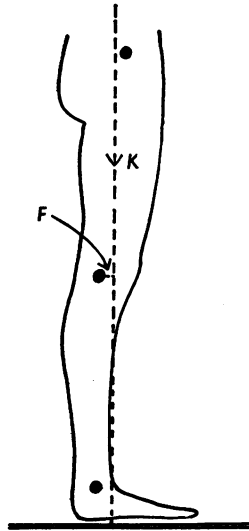
Text-fig. 11. The lower limb in symmetrical standing. The black markers indicate
the positions of the axes of the hip, knee and ankle joints.
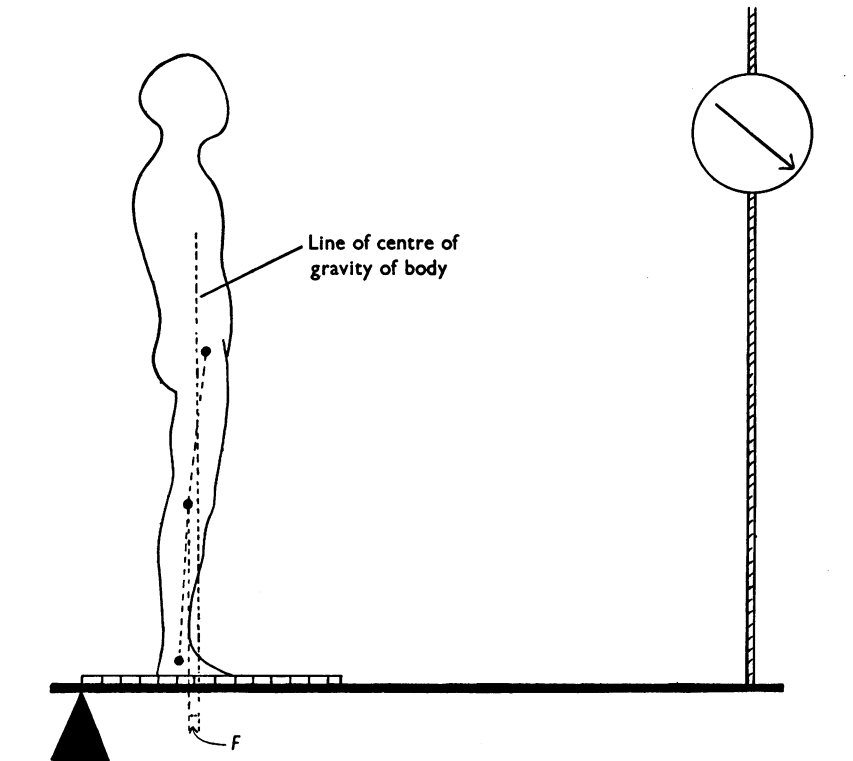
Text-fig. 12. Determination of the extending torque acting at
the knee joint during symmetrical standing.

An inch scale was applied to the side of the board extending from the fulcrum towards the suspension. While the subject was standing a photograph was taken from one side to include the subject, the fulcrum, the inch scale and the spring-balance dial. On this photograph, the previously determined position of the knee axis was marked so that the distance of this axis in front of the fulcrum and the angulation of the knee joint could be measured. Thereafter the line of the centre of gravity of the body was constructed from the fact that its distance in front of the fulcrum is equal to the product of the balance reading and the distance between fulcrum and suspension, divided by the body weight.

In one subject (J. S.) (Pl. 1, fig. 4) the following values were noted:

Body weight $= 161$ lb.

Half the weight of that part of the body above the knee joints $\left(K = \dfrac{\text{body weight}}{2} - WD\right)$ $= 72\cdot3$ lb.

Distance from fulcrum to suspension $= 5$ ft.

Reading on spring balance $= 12\cdot5$ lb.

Distance of centre of gravity of body in front of fulcrum $= \dfrac{5 \times 12\cdot5}{161}$
$= 0\cdot39$ ft.

Distance of knee axis in front of fulcrum $= 0\cdot32$ ft.

Distance of centre of gravity of body in front of knee axis (F) $= 0\cdot07$ ft.

Extending torque on knee joint $(K \times F)$ $= 5\cdot1$ lb. ft.

Angulation of joint $= 186°$

Thus in this subject during symmetrical standing, the knee joint is 6° short of full extension and the extending torque which must be resisted for the knee joint to be stable amounts to 5·1 lb. ft. It is apparent that the data expressed in Text-fig. 9 cannot be exactly applied to the present subject, but nevertheless there is sufficient uniformity between different subjects for the correlation to be regarded as approximately correct. It is considered, therefore, that in the present subject during symmetrical standing the knee joint is within the sphere of influence of both the articular and the extra-articular passive limiting mechanisms. In these circumstances the articular mechanism exerts a resistance to extension of 0·9 lb. ft., the extra-articular tissues exert a passive resistance of 2·7 lb. ft., making a total passive resistance to extension of 3·6 lb. ft. Thus, what is required from the postural activity of the flexor muscles to stabilize the knee joint is a flexing torque of 1·5 lb. ft. In other words, of the total flexing torque which is necessary to counteract the force of gravity and stabilize the knee joint during symmetrical standing, about 50% is derived from the passive resistance of the extra-articular tissues, about 30% from the postural activity of the flexor muscles and about 20% from the passive resistance of the articular mechanism.

THE STABILITY OF THE KNEE JOINT IN ASYMMETRICAL STANDING

In the asymmetrical form of standing (Smith, 1953), illustrated in Pl. 1, fig. 2, one lower limb (the supporting limb) transmits the greater part of body weight, while the other (the balancing limb) acts as a prop controlling the forward sway of the

body. The weight transmitted through each limb can be measured if the subject stands asymmetrically with the feet on separate weighing machines.

In this posture the tendency of body weight is not to carry the body and thigh forwards over the knee joint as in symmetrical standing, but to force the knee joint of the supporting limb backwards in relation to both the hip and ankle joints. Thus assuming that the hip joint is stabilized and that the resistance to plantarflexion at the ankle joint is negligible, the extending torque operating at the supporting knee joint is the product of W_s and C (Text-fig. 13) when W_s is the weight transmitted through the supporting limb and C the vertical distance from the knee axis of a line joining the axes of the hip and ankle joints.

In the subject J. S. (Pl. 1, fig. 5) the weight transmitted by the supporting limb was $W_s = 141$ lb. and that transmitted by the balancing limb was $W_B = 20$ lb. The distance C, as measured from a photograph, was 0·11 ft. and therefore the extending torque at the knee joint was $W_s C = 15·5$ lb. ft.

The angulation of the supporting knee joint in this posture was 190°, which as would be expected from the greater extending torque, is appreciably greater than the angulation of either knee joint in symmetrical standing. The data in Text-fig. 9 which can be approximately applied to the present subject shows that at this angulation the total passive resistance to extension is 5·7 lb. ft., of which 3.4. lb. ft. is due to the extra-articular mechanism and 2·3 lb. ft. to the articular mechanism. It is considered therefore that of the total flexing torque necessary to counteract the force of gravity and stabilize the knee joint of the supporting limb during asymmetrical standing, about 65% is derived from the postural activity of the flexor muscles, 20% from the passive resistance of the extra-articular tissues and about 15% from the resistance of the articular mechanism. Moreover, it is to be noted that although in asymmetrical standing the knee joint is only 2° short of full extension, and is thus well within the range of movement affected by the articular mechanism, that mechanism still contributes comparatively little to the stability of the knee joint.

THE ANGULATION OF THE KNEE JOINT
IN WALKING AND RUNNING

The angulation of the knee joint during walking and running has been studied on cine films by Elftman (1939) and Fenn (1930 b), and the graphs in Text-fig. 14 have been compiled from the data in their papers. The pattern of movement is similar in the two activities: peaks of extension at the beginning and end of the stance phase are separated by the maximal weight-bearing period during which the joint is rather less extended; during the swing phase the joint is flexed and the limb shortened.

Two features are relevant to the present paper. In both walking and running the knee joint moves into the passively resisted range of extension during the stance phase, but in neither activity does it enter the zone of influence of the articular mechanism. Thus the joint cannot be regarded as being locked at any phase of either walking or running (Saunders, Inman & Eberhart, 1953). Secondly, it will be noted that in running the extension at foot contact is less than at 'toe off'. It is considered that this difference is due, in part at least, to the different position of the

hip joint on those occasions. At 'toe off' the hip joint is extended and the ankle joint plantarflexed so that, as already noted on p. 248, the passive extra-articular resistance to extension at the knee joint is minimal. At foot contact, on the other hand, the hip joint is flexed so that the passive extra-articular resistance to extension at the knee joint is augmented in its effective range and magnitude.
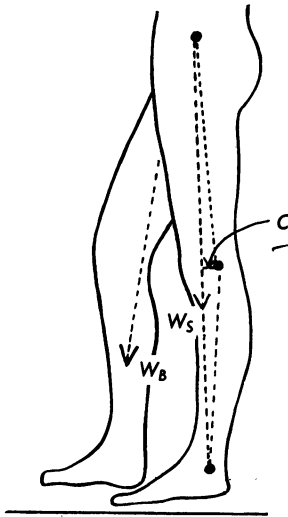
Text-fig. 13

Text-fig. 14

Text-fig. 13. The lower limbs in asymmetrical standing. The black markers indicate the positions of the axes of the hip, knee and ankle joints. W_S represents the weight transmitted by the supporting limb and W_B that by the balancing limb.
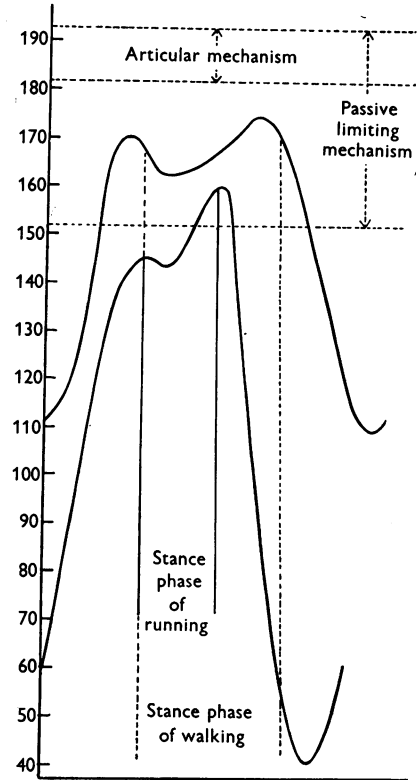
Text-fig. 14. Variations in the angulation of the knee joint during walking and running.

DISCUSSION

It is believed that the passive limiting mechanism which has been described in connexion with extension at the knee joint is not unique. On the contrary, because it depends on physical properties which are common to tissues in all regions of the body, it seems certain that a terminal part of every joint movement is resisted by a similar mechanism. Preliminary observations have been made on the passive mechanisms resisting dorsiflexion at the ankle joint and flexion of the vertebral column, and these taken in conjunction with the observations of Silver (1954) on the latter part have confirmed this view. However, despite the essential similarity

of these mechanisms, it is to be presumed that they will vary considerably both in the magnitude of the resistance they exert, and the range over which they operate, and certain propositions can be laid down which govern these variations. In this context the tissues involved in a passive limiting mechanism are most conveniently divided into those resisting tensile stresses, and those resisting compression stresses, and these groups will be referred to as 'tensile tissues' and 'compression tissues' respectively.

In relation to a tensile tissue, the magnitude of the resistance which it exerts to a movement will be proportional to Young's modulus and the cross-sectional area of the tissue.

The range of joint movement over which a tensile tissue is capable of exerting resistance depends on three factors. It depends first on the elastic extensibility of the tissue (E). In Text-fig. 15a ABC is a joint which is hinged at B and has its two arms AB and BC joined by a tensile tissue HK. If this tensile tissue is capable of an extension of 10% of its original length before rupturing, such an extension would allow the joint to move to the position shown in (b) of the same figure. On the other
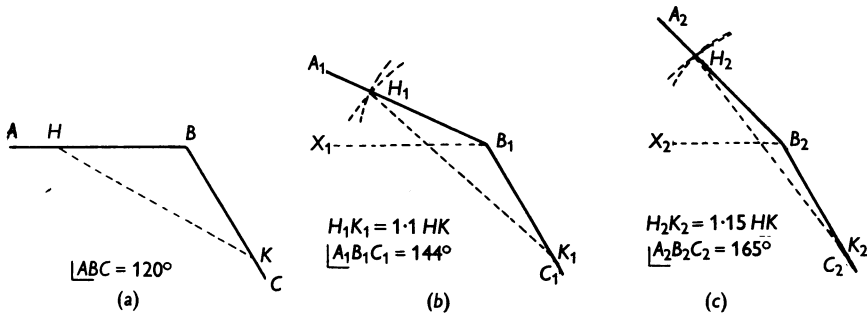

Text-fig. 15. The influence of the extensibility of a tensile tissue on the range of passively resisted joint movement.

hand, an extension of 15% would allow movement to the position shown in (c). It will be noted that $\angle X_2 B_2 A_2$ is greater than $\angle X_1 B_1 A_1$, that is, the range of passively resisted movement at a joint is a function of the extensibility of the related tensile tissues.

In Text-fig. 16a the arms of a joint ABC are joined by a tensile tissue HK, which lies at a distance D from the axis of the joint. A 10% extension of HK allows the joint to move to the position shown in (b), that is, it allows a passively resisted movement of 19°. In (c) of the same figure the arms of the same joint are linked by a tensile tissue $H_2 K_2$, which is longer than HK but lies at the same distance D from the joint axis. A 10% extension of $H_2 K_2$ allows the joint to move to the position shown in (d), that is, it allows a passively resisted movement of 29°. Thus the range of passively resisted movement at a joint is a function of the length of the related tensile tissues.

The third factor is the distance of the tensile tissue from the axis of rotation of the joint (D). In Text-fig. 17a the tensile tissue HK links the arms of the joint ABC and lies at a distance D_1 from the joint axis B. A 10% extension of HK allows the joint to move to the position shown in (b) of the same figure, that is, it allows a

passively resisted movement of 20°. In (c) the arms of an exactly similar joint are linked by a tensile tissue H_2K_2, which is the same length as HK but lies at a shorter distance D_2 from the joint axis. A 10% extension of H_2K_2 allows a passively resisted movement of 42° as shown in (d). Thus the range of passively resisted movement permitted by a tensile tissue is an inverse function of the distance of that tissue from the joint axis.
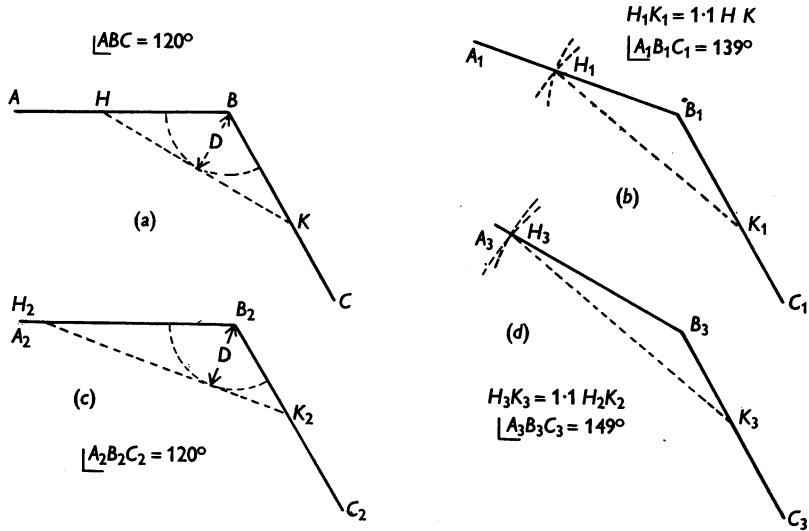
Text-fig. 16. The influence of the length of a tensile tissue on the range of passively resisted joint movement.
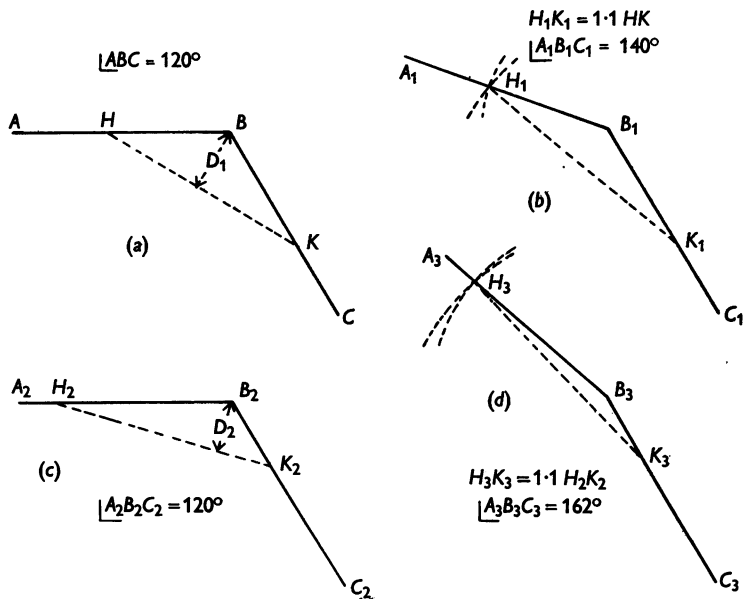
Text-fig. 17. The influence of the distance of a tensile tissue from the axis of rotation, on the range of passively resisted joint movement.

The range of passively resisted movement (M) which can occur at any joint is thus dependent on certain properties of the tensile tissues: it is a function of their extensibility (E), their length (L) and an inverse function of their distance from the joint axis (D). Furthermore, E, L and D are the only properties of the tensile tissue which influence the value of M.

In Text-fig. 18 *a* the joint ABC has an angulation of 73°. The arms of the joint are linked by the tensile tissue HK lying at a distance D from the joint axis. A 10% extension of HK allows a passively resisted movement of 7° (see *b*). In (*c*) of the same figure, the joint LMN has an angulation of 56°. Its arms are linked by the
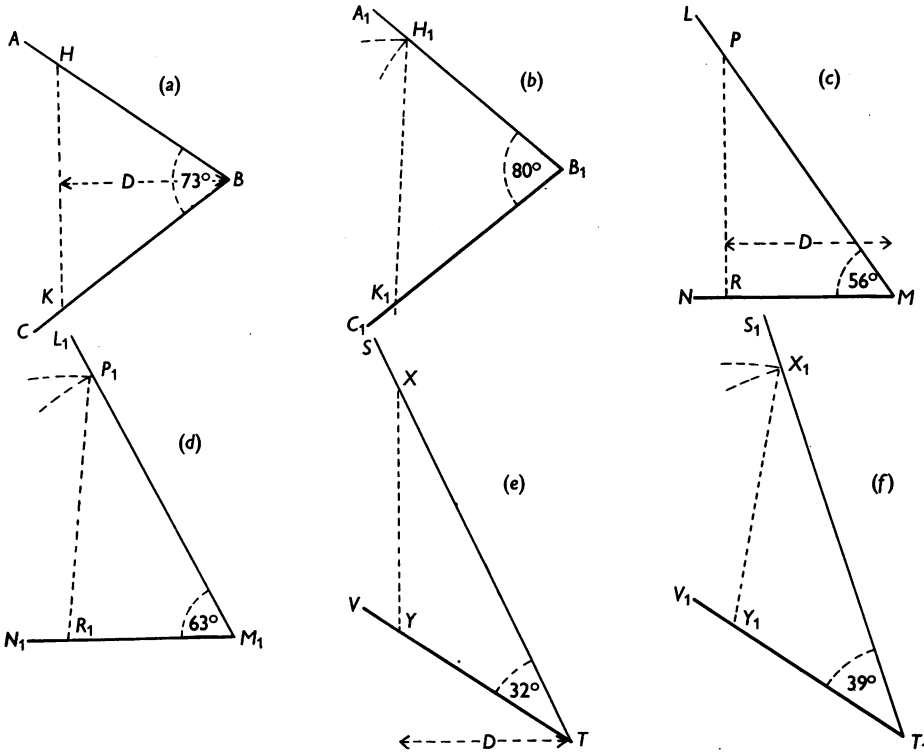
Text-fig. 18. Proof that the range of movement resisted by a tensile tissue depends only on the extensibility and length of the tissue, and its distance from the axis of rotation.

tensile tissue PR, which is equal in length to HK and lies at the same distance D from the joint axis. A 10% extension of PR again permits a passively resisted movement of 7° (see *d*). Similarly, a 10% extension of the tensile tissue XY in (*e*) allows a passively resisted movement of 7° in the joint $STV(f)$.

In relation to compression tissues, the range of resisted movement at any joint is a function of the deformability of the articular surfaces, and this in turn depends on two factors: first, the compressibility of the tissues deep to the articular surfaces, and secondly, the rigidity of the union between different parts of those tissues.

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The compressibility of the tissues deep to a joint surface is a function of their thickness, and is greater in relation to a thick layer of articular than to a thin layer, greater at those joints possessing intra-articular fibro-cartilages than in those devoid of those structures, and greater in relation to an incompletely ossified epiphysis than to a mature bone.

When an articular surface extends over a continuous layer of articular cartilage which is supported by a single bone, that surface can be appreciably deformed only by distortion of the cartilage. But all articular surfaces do not conform to this pattern. In the child, the proximal surfaces of the shoulder and hip joints consist of areas of articular cartilage supported by a number of discrete bony elements, which are themselves united by cartilage. And in the adult, the proximal surfaces at the ankle joint and most of the costo-vertebral joints, and the distal surface at the talo-calcaneo-navicular joint consist of areas of articular cartilage supported by bones which are themselves joined by ligaments. It is apparent that in such instances relative displacement of the separate supporting bones will give rise to changes in the form of the articular surface quite apart from any distortion of the compression tissues. Furthermore, as Barnett (1953) has shown at the knee joint, an intra-articular fibro-cartilage may be capable of a displacement in relation to the under-lying articular cartilage, which profoundly alters the form of the total articular surface.

SUMMARY

1. A terminal range of extension at the knee joint is associated with distortion of certain articular and extra-articular tissues, and the movement is consequently passively resisted by distortion stresses in those tissues. Thus the total passive resistance to extension has been regarded as the sum of the resistances exerted by an articular and a passive extra-articular mechanism.

2. The exact nature of the articular mechanism is discussed.

3. The magnitude and the range of operation of the total passive resistance and of its articular and extra-articular parts are determined.

4. Measurements are made of the contribution of postural muscle activity and the passive resistance to extension to the stabilization of the knee joint in symmetrical and asymmetrical standing.

5. The influence of passive resistance to extension of the knee joint on walking and running is discussed.

6. The factors influencing the magnitude and range of operation of mechanisms passively resisting joint movement are discussed.
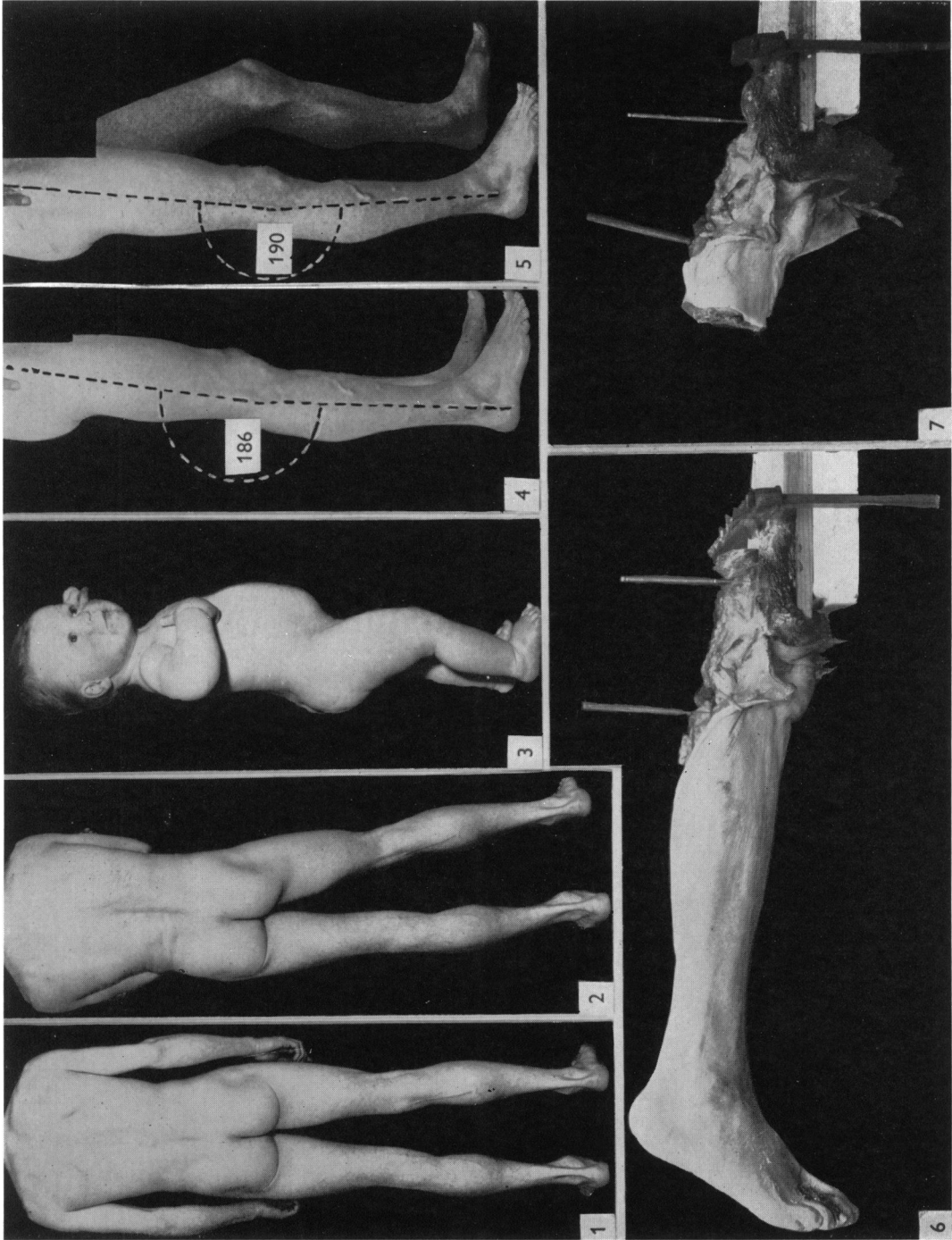
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EXPLANATION OF PLATE

Fig. 1. Symmetrical standing in the adult.

Fig. 2. Asymmetrical standing in the adult.

Fig. 3. Standing in a child of 14 months.

Fig. 4. The adult lower limb in symmetrical standing. The finger marks the tip of the greater trochanter of the femur.

Fig. 5. The adult lower limb in asymmetrical standing. The finger marks the tip of the greater trochanter of the femur.

Fig. 6. Stripped knee joint of amputated lower limb. The joint is here extended by the weight of the whole leg and foot.

Fig. 7. The same knee joint as that illustrated in fig. 6. The joint is here extended by the weight of the proximal third of the leg only. The reduced extension at the joint is indicated by the altered position of the pins.