THE FORCES OPERATING AT THE HUMAN ANKLE JOINT DURING STANDING

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INTRODUCTION

In recent years it has become generally appreciated (Clemmesen, 1951 and Fulton, 1949) that the ankle joint is stabilized during standing by two forces operating in harmony. One is that exerted by the postural activity in the posterior crural muscles, and the other is the passive resistance to dorsi-flexion which is exerted by the tissues of the same region. Passive resistance to joint movement by elastic tension in such tissues as joint ligaments, deep fascia, skin and the fibrous framework of muscles itself occurs over a certain terminal range of movement at every joint; its properties have been considered in some detail in a previous publication (Smith, 1956).

The postural activity in the posterior crural muscles during standing has been studied frequently by electromyography, and an analysis of the results of these investigations will be considered in this paper, but it is believed that this method cannot give an absolute quantitative value to the tension exerted by the muscles. On the other hand, the passive resistance of the tissues to dorsi-flexion at the ankle joint has not been previously studied. In the present investigation the contributions made by these two factors to the stability of the ankle joint during standing have been determined.

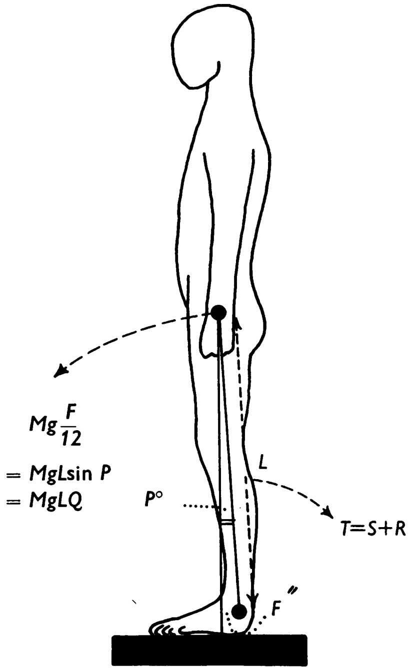
It is well known that during standing the centre of gravity oscillates in the sagittal plane. It moves in an arc around the axes of rotation of the ankle joints with an angular velocity which varies continually both in magnitude and direction. It is evident that such movement must be both motivated and controlled by the 'turning moments' or 'torques' which act on the body about the ankle axes and these torques are indicated in Text-fig. 1. In this diagram two lines have been constructed. One extends downwards and slightly backwards between the centre of gravity of the body and the ankle axis, and is L ft. in length. The other extends vertically downwards from the centre of gravity; it makes an angle of P° with the first line and passes F in. in front of the ankle axis. The force of gravity tends to cause downward displacement of the centre of gravity of the body at an acceleration which is customarily indicated by g and has a value of 32·2 ft. per sec. per sec. A force, by definition, is equal to the product of the mass of the object on which it acts and the linear acceleration it produces, and therefore if the mass of the human body—which is visualized as being concentrated at the centre of gravity—is M lb., the force exerted on the body by gravity is Mg poundals.

The torque or turning moment which a force exerts about a given axis is equal to the product of the force in poundals and the distance of the force from the axis in feet. Thus in Text-fig. 1 the dorsi-flexing torque exerted at the ankle axis by the force of gravity is $Mg\,(F/12)$ poundals ft. From the same figure it will be evident that because $F/12 = L \sin P$, the torque can be alternatively expressed as $MgL \sin P$. Furthermore, in the consideration of angular motion it is often necessary to evaluate the angle P in radians rather than degrees. Transference between the two units is facilitated because during normal standing the angle P which represents body sway never exceeds about 6° (Hellebrandt & Braun, 1939); and consultation of standard mathematical tables shows that for angles of this size the radian measure is numerically equal to the sine of the angle. There are therefore three ways of expressing the dorsi-flexing torque exerted by gravity about the ankle axis, namely

$$Mg(F/12) = MgL \sin P = MgLQ \text{ poundals ft.}$$

when Q is the angle P in radians.

Opposing this dorsi-flexing torque is the plantar-flexing torque exerted by the tissues of the posterior crural regions of both lower limbs. This may be designated T poundals ft., and, as has been stated already, it is the sum of the plantar-flexing torques exerted by postural contraction in the posterior crural muscles (S poundals ft.) and by the passive tension in the tissues of the same region (R poundals ft.) i.e. $T = S + R$. The plantar-flexing and dorsi-flexing torques about the ankle joint operate in different directions and this fact must be indicated by a difference in sign. It is immaterial what convention is employed, but in this paper a dorsi-flexing torque is regarded as negative and a plantar-flexing torque as positive. Thus the resultant torque acting about the ankle joint is $(T - MgLQ)$ poundals ft.



Text-fig. 1. The torques acting about the ankle joint in standing. The upper marker represents the centre of gravity of the body and the lower marker the axis of rotation of the ankle joint.

This resultant torque tends to cause an angular acceleration of the centre of gravity about the ankle axis which may be designated (B) radians per sec. per sec. The actual value of (B) depends on the value of the factor known as the moment of inertia of the body about the ankle axis (I_a). The name of this factor has frequently given rise to misunderstandings: in the present context the moment of inertia is best regarded as being the equivalent, in relation to angular motion, of the mass in relation to linear motion. Thus it has already been noted that in linear motion, force = mass × linear acceleration: in the same way in angular motion, torque = moment of inertia × angular acceleration. That is

$$T - MgLQ = I_a B$$
$$S + R = MgLQ + I_a B \text{ poundals ft.}$$

In this equation the factors (S) and (R) are expressed in poundals ft. By dividing the terms on the right hand side of this equation by g, (S) and (R) can be expressed in the more familiar non-gravitational units of pounds ft., 1 lb.ft. being the torque produced by a mass of 1 lb. acting at a distance of 1 ft. from an axis of rotation,

i.e.
$$S+R=MLQ+\frac{I_a B}{g}\text{ lb.ft.},$$

or
$$S=MLQ+\frac{I_a B}{g}-R\text{ lb.ft.} \tag{1}$$

In this equation, M, L, I_a and g are constants for any one subject whereas S, Q, R and B vary with time. The solving of the equation necessitates the measurement of M and L and the experimental determination of I_a, R, Q, B and S.

METHODS

(1) *Determination of the position of the centre of gravity of the body*

In the present investigation it was not necessary to know the exact position of the centre of gravity in three dimensions; what was required was the distance between the centre of gravity and the ankle axis (L ft.).

This distance was determined by placing each subject supine on a board tilting freely about a transverse axis, so that the feet were at right angles to the legs and the arms were by the side. The position of the body in relation to the tilting axis was then varied until the position of balance was obtained.

(2) *Experimental determination of the moment of inertia of the body about the axis of the ankle joint (I_a)*

One of the standard methods for determining the moments of inertia of objects of irregular outline and composition is to regard the object as a compound or irregular pendulum. The duration in seconds of one complete oscillation of such a pendulum is

$$t=2\pi\sqrt{\frac{I}{MgD}} \tag{2}$$

I being the moment of inertia about the point of suspension and D the distance in feet between the point of suspension and the centre of gravity of the body. Thus if an object is suspended and allowed to swing through a small angle, the value of t can be determined and the moment of inertia about the axis of suspension calculated.

Direct determination of the moment of inertia of the living body about the ankle axis by this method would require suspension of the subject upside down about this axis. Use of the theorem of parallel axes makes this procedure unnecessary. This theorem states that if the moment of inertia about a given axis is I and the distance of that axis from the centre of gravity is D ft., then the moment of inertia about the centre of gravity itself is $I_c=I-MD^2$.

Furthermore, by the same theorem the moment of inertia about the ankle axis is

$$I_a=I_c+ML^2$$

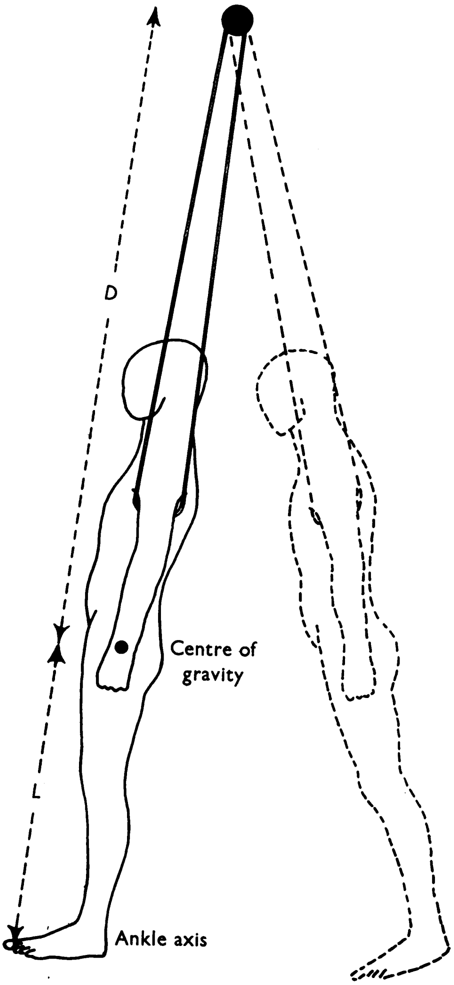
and therefore
$$I_a=I+M(L^2-D^2). \tag{3}$$

Thus each subject was suspended as in Text-fig. 2 from a beam by ropes passing beneath the axillae, and allowed to swing through a small arc. The average value of 't' for fifty oscillations was observed. The values of I about the axis of the beam was then determined from equation (2) and subsequently the value of I_a was calculated from equation (3).

(3) *The determination of the plantar-flexing torque which is exerted at the ankle joints during standing by the passive tension in the tissues in the posterior crural regions of both lower limbs* (*R*).

It is apparent that the value of this torque varies with the position of the ankle joint, increasing in dorsi-flexion and decreasing in plantar-flexion. Its values throughout the greater part of the range of movement at the ankle have been determined by the following method.

Each subject lay supine with the leg to be examined suspended clear of the table in a broad sling passing beneath the calf muscles (Text-fig. 3). The position of full voluntary dorsi-flexion at the ankle joint was then determined. Subsequently, a large sphygmomanometer cuff was applied to the mid-thigh and a pressure of 200 mm. Hg maintained in it until sensory and motor paralysis below the calf were complete. It was assumed that in these circumstances the posterior crural muscles reacted to stretch in the same way as denervated structures, an assumption which is in conformity with observations of Magun (1940) and Magladery, McDougal & Stoll (1950).



Text-fig. 2. Determination of the moment of inertia of the body about the ankle axis (I_a)

The foot was now moved slowly from plantar-flexion into dorsi-flexion by pulling on a calibrated tensile spring (*C*) attached to the foot in the region of the metatarsal heads (*D*). The movement was stopped some distance short of full voluntary dorsi-flexion to avoid injury to the tissues which had been rendered insensitive. During the movement a cine film was taken from the lateral side, the whole foot and leg and the calibrated spring being included in the field, and subsequently enlarged tracings were made of every fourth frame. On each tracing the lines *AM* and *DM* (Text-fig. 3) were constructed so that they passed from the anterior aspect of the patella and from the attachment of the spring to the foot respectively, to the tip of the lateral malleolus which was regarded as lying on the axis of rotation of the ankle joint.

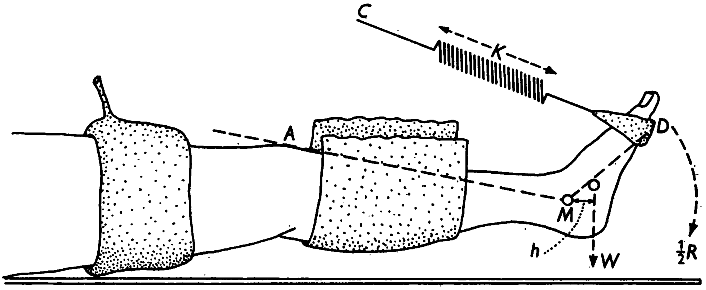
In these circumstances three torques acted about the axis of the ankle joint. If the tension in the spring is denoted by K, the dorsi-flexing torque exerted by the spring was the product of that tension and the minimum distance of the spring from the ankle axis, that is $K.DM.\sin \angle CDM$. The plantar-flexing torque due to the passive tension in the posterior crural tissues of the one limb was $\frac{1}{2}R$. The similar torque due to the weight of the foot was (Wh), W being the weight of the foot and h the horizontal distance between the centre of gravity of the foot and the ankle axis. At any moment during the experiment these torques were in equilibrium and therefore

$$K.DM \sin \angle CDM = \tfrac{1}{2}R + Wh,$$

or

$$\tfrac{1}{2}R = K.DM \sin \angle CDM - Wh. \tag{4}$$

Now the value of the factor Wh in this equation varies with the angulation of the ankle joint. It is evident from Text-fig. 3 that in plantar-flexion the factor Wh is comparatively large because the distance h is large, but that, as the foot is dorsi-flexed the factor becomes progressively smaller until, when the foot is at right angles to the leg it approaches zero. In the present study the significant values of R are those operative during standing, when the angle between the leg and foot is rather less than 90°, and when the value of the factor Wh is consequently approximately zero. In this investigation, therefore, the factor Wh can be ignored and equation (4) can be approximated to



Text-fig. 3. Method for determination of values of *R*.

$$R = 2K.DM.\sin \angle CDM. \tag{5}$$

The values of K, DM, $\angle CDM$ and $\angle AMD$ were therefore measured on each of the tracings described above and a graph was then constructed showing the values of R for successive values of $\angle AMD$.

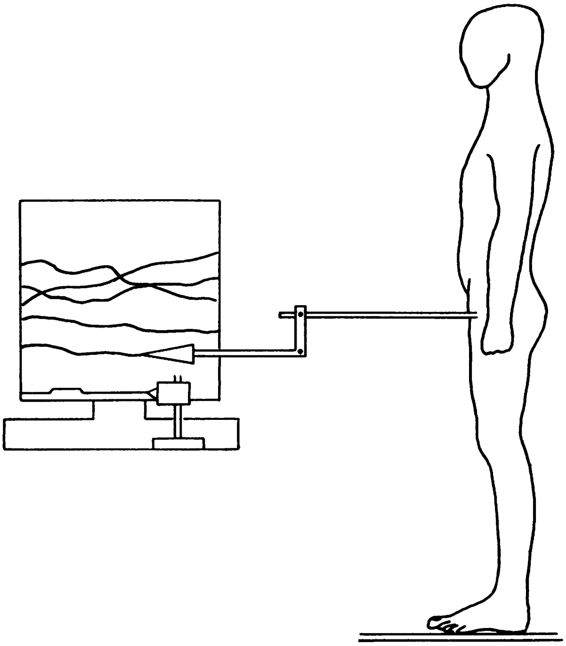
In standing the angle AMD is variable—it is in fact an inverse function of the angle P in Text-fig. 1. However, the variation is always small, being of the order of about 3°, and it has been found in practice that the associated variation in the value of R is insignificant. It is, therefore, justifiable to determine a single value of $\angle AMD$ from a lateral photograph of the standing subject, and to read off the

$R/\angle AMD$ graph the corresponding single value of R. This value of R, with only a slight degree of error, is applicable to all phases of standing in that subject.

(4) *Determination of variations in the angle P and in the angular acceleration B during stance*

The manner in which the terms MLQ and I_aB/g in equation (2) vary with time can be determined by the one set of experiments.

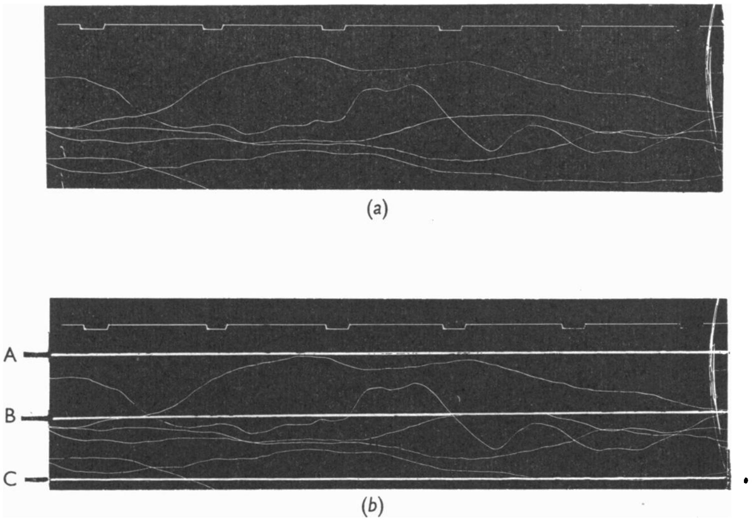
In each subject a magnified kymographic record of the antero-posterior movements of the centre of gravity during standing was prepared with an accompanying appropriate time tracing. The arrangement is shown diagrammatically in Text-fig. 4.



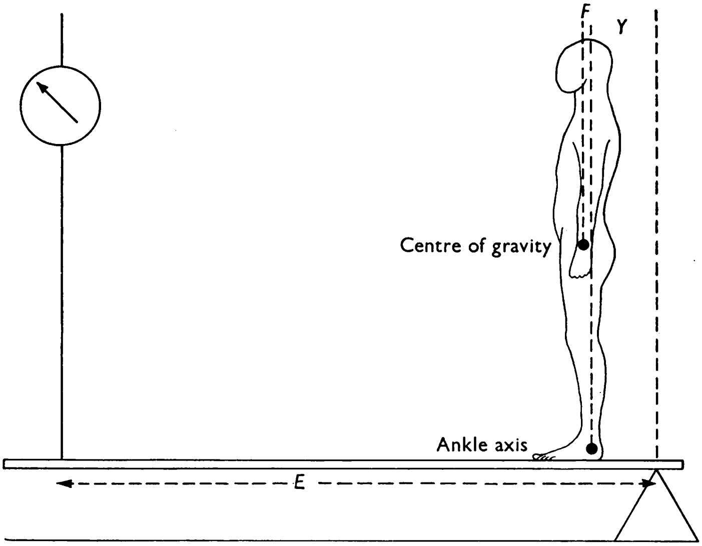
Text-fig. 4. Method for constructing kymogram of body sway.

The recording was continued for about 2 min., and as this involved several revolutions of the drum, the tracing became repeatedly superimposed on itself. Thus the finished kymogram had the appearance shown in Text-fig. 5*a*. On this kymogram the horizontal lines *A*, *B* and *C* were constructed (Text-fig. 5*b*) so that *A* passed through the highest point on the tracing, *C* through the lowest and *B* lay mid-way between *A* and *C*. The line *B* thus indicates the mid-position of the centre of gravity during the period of the recording: displacement of the tracing above the line *B* indicates a backward displacement of the centre from its mid-position, whereas displacement below the line indicates a forward displacement of the centre from the mid-position. Subsequently, the kymogram was photographed and the negative was projected at a high magnification on to 1 by $\frac{1}{10}$ in. graph paper on which random segments of the tracing plus the line *B* were drawn. In a typical example the final magnification of the displacements of the centre of gravity was × 92·8 and the time tracing was 1 sec. to 25·4 in.

In this form the tracing shows accurately the antero-posterior displacements of the centre of gravity in relation to the mid-position (line *B*), but it gives no absolute information because the relationship between this mid-position and the ankle axis is unknown. This relationship was established by use of the apparatus shown diagrammatically in Text-fig. 6.



Text-fig. 5. (*a*) Kymogram of body sway. 1 sec. time tracing. (*b*) The same kymogram with the lines *A*, *B* and *C* constructed.



Text-fig. 6. Method of determining the relationship of the line *B* in Text-fig. 5*b* to the axis of the ankle joint.

A broad wooden plank about 6 ft. long was supported close to one end by a fulcrum and suspended at the other from a spring balance. The distance from fulcrum to suspension may be designated E inches and the reading on the spring balance H_1 lb. The subject of body weight W lb. stood on the plank close to the fulcrum and facing the suspension so that the tip of the lateral malleolus (representing the ankle axis) was Y in. from the fulcrum.

Thereafter, if at any given moment the reading on the spring balance was H_2 lb., then the horizontal distance of the centre of gravity in front of the ankle at that moment was

$$F=\frac{E\,(H_2-H_1)}{W}-Y.$$

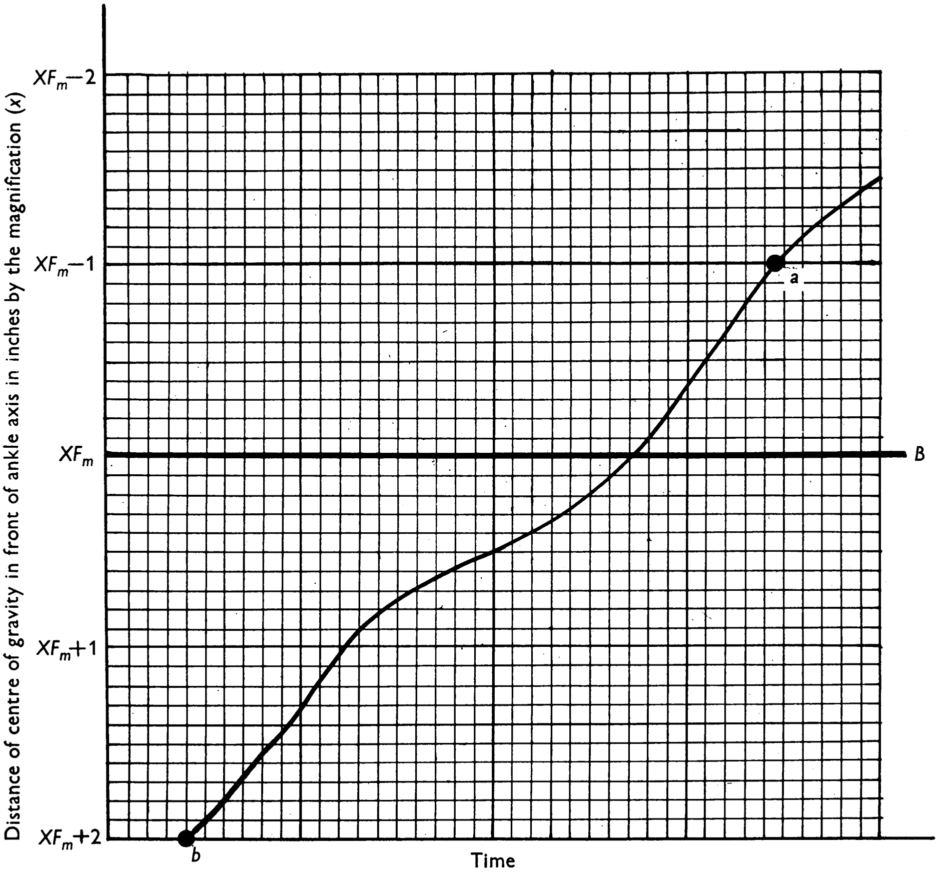
Each subject stood comfortably on this apparatus for about 2 min. and during this period the reading on the spring balance was continually recorded on cine film. From this film the maximum and minimum readings were noted (i.e. the maximum and minimum values of H_2) and from these the mid-reading was calculated and the mid-position of the centre of gravity in front of the ankle axis (F_m in.) was determined.

It has been noted by Hellebrandt & Braun (1939) that the mid-position of the centre of gravity remains practically constant during a stance of any appreciable duration, and it can therefore be said that the line B on the enlarged tracing of the kymogram represents a position of the centre of gravity F_m in. in front of the ankle axis. However, the magnification of this tracing must be borne in mind and it will be evident that if this magnification is denoted by X, then the position of the line B would, in this tracing, be XF_m in. from the ankle axis (Text-fig. 7). Furthermore, it has been noted that a displacement of the tracing above the line B represents a proportionate backward movement of the centre of gravity and vice versa. Thus, a point on the tracing such as (*a*) in Text-fig. 7 is 1 in. above the line B; consequently, at the magnification of the tracing it denotes a position of the centre of gravity (XF_m-1) in. in front of the ankle axis. Similarly, a point such as (*b*) in Text-fig. 7 is 2 in. below the line B; it therefore represents, at the magnification of the tracing, a position of the centre of gravity which is (XF_m+2) in. in front of the ankle axis. In this way the whole tracing was calibrated.

Thereafter the vertical positions of successive points on the tracing—which represent the distances of successive positions of the centre of gravity in front of the ankle axis (F) at a magnification X—were read off at intervals of 2·5 small squares, that is at intervals of the order of $\frac{1}{100}$ sec. The distances were tabulated as in column 2, Table 1.

It was previously stated (p. 546) that the term MLQ is equal to $ML \sin P$ which in turn is equal to $MF/12$ lb.ft. Thus, if each figure in column 2 is multiplied by $M/12X$ the values of MLQ are obtained at the same intervals of $\frac{1}{100}$ sec. and these are tabulated as in column 3.

Angular velocity is the angle traversed by a rotating body in radians divided by the time in seconds and is expressed in radians per second. In the present experiment it will be evident from Text-fig. 8 that at the end of the first $\frac{1}{100}$ sec. of the tracing the sine of the angle P_1 is $F_1/12L$, and that P_1 has therefore a value of $F_1/12L$ radians. Similarly, at the end of the second $\frac{1}{100}$ sec. the angle P_2 has a value of $F_2/12L$ radians and therefore the angle traversed during the second $\frac{1}{100}$ sec. is (P_2-P_1) which equals $(F_2-F_1)/12L$ radians. Thus the average angular velocity of



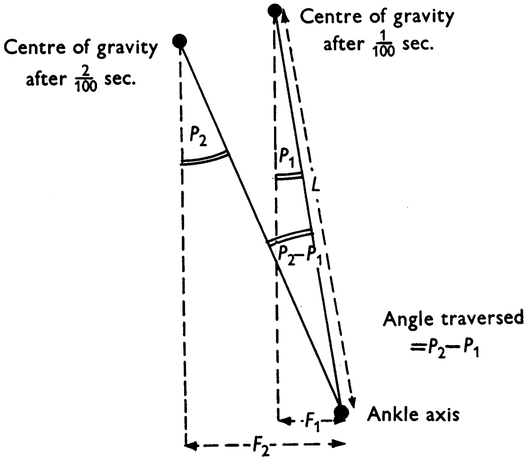
Text-fig. 7. Segment of kymogram of body sway and the line *B* enlarged and traced on to 1 in. by $\frac{1}{10}$ in. graph paper. Method of calibration.

Table 1. *Method of calculation and tabulation of the values of the terms MLQ and* I_aB/g *at intervals of the order of* $\frac{1}{100}$ *sec.*

1 Time in $\frac{1}{100}$ sec.	2 Distance of centre of gravity in front of ankle axis by the magnification (X)	3 $MLQ = \frac{FX.M}{12X}$	4 Angular velocity of centre of gravity about the ankle axis	5 Angular acceleration of centre of gravity about ankle axis (B)	6 $\frac{I_aB}{g}$
1	F_1X	$\frac{MF_1}{12}$			
2	F_2X	$\frac{MF_2}{12}$	$\frac{100(F_2-F_1)}{12L}$	$\frac{100^2(F_3-2F_2+F_1)}{12L}$	$\frac{100^2(F_3-2F_2+F_1)\,I_a}{12Lg}$
3	F_3X	$\frac{MF_3}{12}$	$\frac{100(F_3-F_2)}{12L}$	$\frac{100^2(F_4-2F_3+F_2)}{12L}$	$\frac{100^2(F_4-2F_3+F_2)\,I_a}{12Lg}$
4	F_4X	$\frac{MF_4}{12}$	$\frac{100(F_4-F_3)}{12L}$		

the centre of gravity about the ankle axis during that period is $[100\,(F_2-F_1)]/12L$ radians per sec. The average angular velocity during successive periods of $\frac{1}{100}$ sec. can therefore be calculated and tabulated as in column 4, Table 1.

Angular acceleration is the rate of change of angular velocity and is expressed in radians per sec. per sec. Thus the angular acceleration (B) of the centre of gravity around the ankle axis between the middle of the second $\frac{1}{100}$ sec. and the middle of the third $\frac{1}{100}$ sec. is



Text-fig. 8. The angle traversed by the centre of gravity about the ankle axis during $\frac{1}{100}$ sec.

$$\frac{\dfrac{100\,(F_3-F_2)}{12L}-\dfrac{100\,(F_2-F_1)}{12L}}{\frac{1}{100}}=\frac{100^2\,(F_3-2F_2+F_1)}{12L}\ \text{rad./sec.}^2.$$

In this way the angular acceleration during successive periods of $\frac{1}{100}$ sec. can be calculated and tabulated as in column 5, Table 1. Thereafter the term I_aB/g can be calculated for the same intervals.

In the equation

$$S=MLQ+\frac{I_aB}{g}-R$$

all quantities are now known except S, the plantar-flexing torque exerted at both ankle joints by the postural activity in the posterior crural muscles of both lower limbs. The value of this factor can now be calculated at intervals of $\frac{1}{100}$ sec.

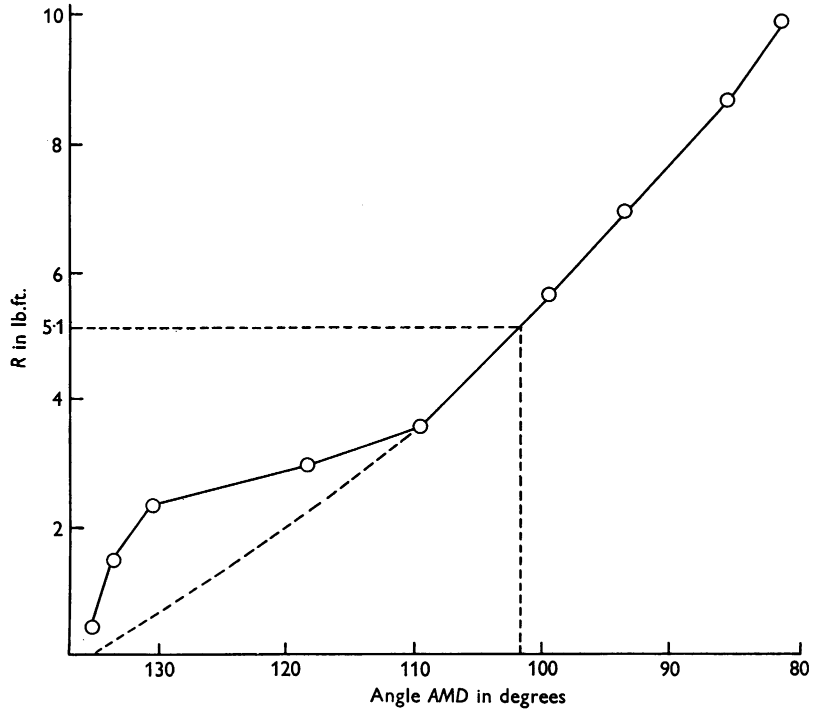
RESULTS

(1) *The moment of inertia of the body round the ankle axis* (I_a)

The general formula for the moment of inertia is Mr^2, where r is the radius of gyration which can be defined for the purpose of this investigation as the average distance of all particles in the body from the axis of rotation. It will therefore be apparent that the value of the moment of inertia is closely connected with the body build of the subject, being greater in the tall or heavy than in the short or light.

One example of the determination is given below. In this subject the mass (M) was 189 lb., the distance (D) from suspension to centre of gravity was 7·04 ft., the distance (L) from the centre of gravity to the ankle axis was 2·9 ft., and the time of oscillation (t) was 3 sec. Thus the moment of inertia about the suspension is

$$\begin{aligned} I &= \frac{t^2 MgD}{4\pi^2}, \\ &= \frac{9 \times 189 \times 32{\cdot}2 \times 7{\cdot}04}{4\pi^2}, \\ &= 9781 \text{ lb.ft.}^2. \end{aligned}$$



Text-fig. 9. The relationship of the passive tension in the tissues of the posterior crural regions of both limbs (R) to the angulation of the ankle joint ($\angle AMD$).

By the theorem of parallel axes (p. 547) the moment of inertia around the ankle axis is

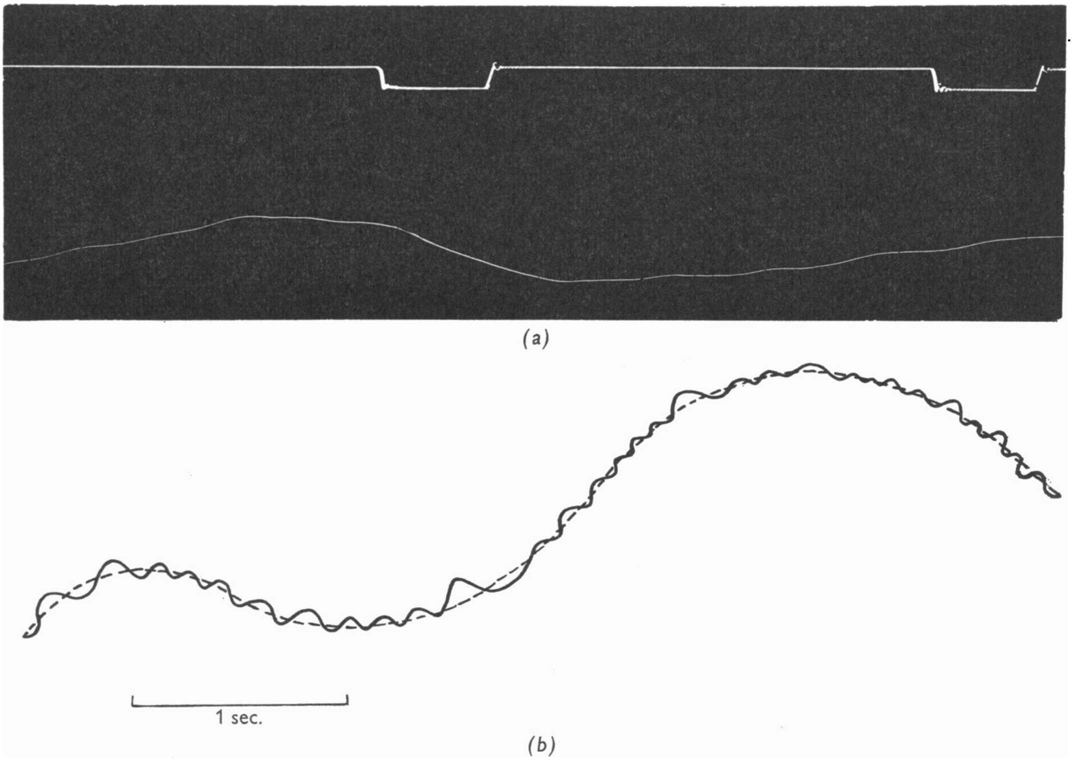
$$\begin{aligned} I_a &= I + M(L^2 - D^2), \\ &= 9781 + 189(2{\cdot}9^2 - 7{\cdot}04^2), \\ &= 1994 \text{ lb.ft.}^2. \end{aligned}$$

(2) *The passive tension in the tissues of the posterior crural regions of both limbs* (R)

The relationship of (R) to the angulation of the ankle joint ($\angle AMD$ in Text-fig. 3) varied very little in the subjects examined; that shown in Text-fig. 9 is a typical example. That part of the graph associated with dorsi-flexion (110–80°) is almost linear, whereas that part associated with plantar-flexion (110–135°) shows a distinct 'hump'. This deviation is the result of the approximation described on p. 549 by which the factor Wh is ignored in the equation

$$R/2 = K.DM.\ \sin \angle CDM - Wh.$$

It is apparent that the approximation does not appreciably affect the values of R for angulations of the ankle joint less than 110°. In the subject to which the data in Text-fig. 9 apply the single value of the angulation of the ankle joint during standing which was determined was 102°, and the corresponding value of (R) was therefore 5·1 lb.ft.



Text-fig. 10. (*a*) Enlarged segment of the kymogram in Text-fig. 5*a*. (*b*) The primary and secondary waves of the kymogram represented diagrammatically by the interrupted and solid lines respectively.

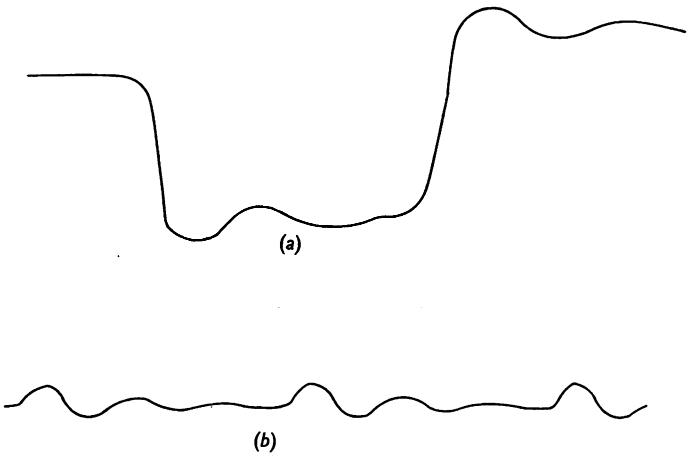
(3) *The values of the terms MLQ and* I_aB/g *at intervals of the order of* $\frac{1}{100}$ *sec.*

The kymogram reproduced in Text-fig. 5 indicates the periodic sagittal displacements of the centre of gravity in the standing subject as a series of waves. However, when the kymogram is examined at a higher magnification (Text-fig. 10*a*) it becomes evident that superimposed on these large primary waves is a series of much smaller secondary waves so that the tracing follows the general form indicated diagrammatically in Text-fig. 10*b*.

Both primary and secondary waves are irregular in amplitude and frequency; the average frequency of the former is about 18/min. and that of the latter about 8/sec.

It is certain that the large primary waves of the kymograph tracing accurately represent corresponding displacements of the centre of gravity of the body, but before the smaller secondary waves can be regarded as arising in the same way, the possibility of artefact must be excluded.

Artefacts of such form might arise through vibration either in the kymograph drum or in the magnifying mechanism. Vibration in the drum is excluded by the absence of any ripples on the tracing when the magnifying arm is attached to a fixed point. Vibration in the magnifying mechanism might be evoked by either of two distinct factors. Thus, if there were a sudden large alteration in the angular velocity of the swaying body, this would be indicated by a correspondingly sudden displacement of the magnifying arm. The momentum of the magnifying arm might then be sufficient to cause a temporary elastic distortion of the whole mechanism and initiate a series of vibrations. In such circumstances the ripple would be restricted to those parts of the kymogram which immediately follow steep upward or downward deflexions, and would show a regular frequency and progressively diminishing amplitude as indicated diagrammatically in Text-fig. 11*a*.



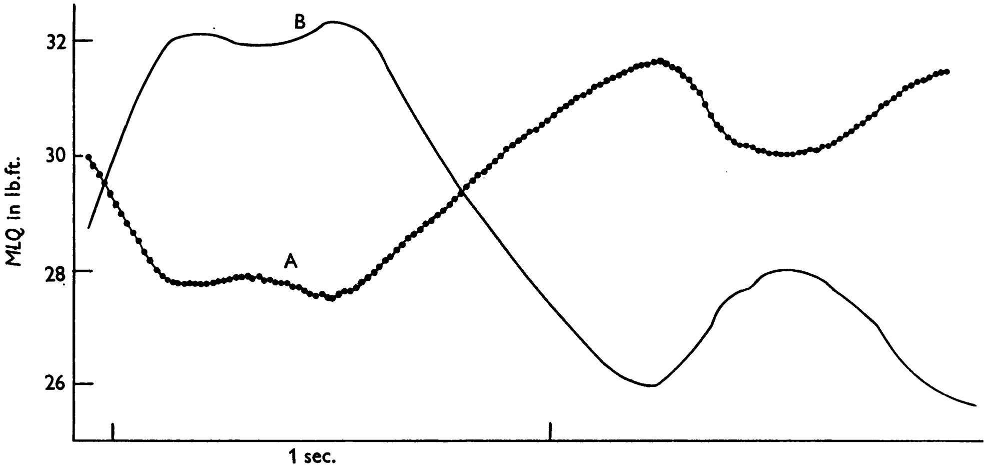
Text-fig. 11. The form of two types of vibration waves.

Secondly, vibration in the magnifying mechanism might be evoked by a continuously repeated series of sudden impulses on the centre of gravity of the body such as those imparted by the normal pulse. The ripple would then follow a pattern of repeated sequences of waves, each sequence having a uniform frequency and progressively diminishing amplitude as indicated in Text-fig. 11*b*.

Thus any artefact would be characterized by a certain regularity of pattern. The fact, therefore, that in the kymogram of body sway the secondary wave series follows no recognizable pattern, indicates, of itself, that these waves do not arise as artefacts. Moreover, this is confirmed by the fact that the form of the secondary wave series can be shown to be independent of the vibration time of the magnifying mechanism.

It is considered, therefore, that both the primary and secondary waves indicate proportionate displacements of the centre of gravity during standing, and that both must consequently be taken into consideration in the calculation of the terms *MLQ* and I_aB/g.

The values of the term *MLQ* calculated at intervals of $\frac{1}{140}$ sec., and the segment of the kymogram in Text-fig. 5*a* from which they were assessed, are shown in Text-fig. 12. The value of the term shows a continual fluctuation which is synchronous with and proportional to the displacements of the kymogram of body sway; the fluctuations are thus almost entirely dependent on the primary kymogram waves and are influenced very little by the smaller secondary waves. During a stance of average duration, in a subject of average physique, the fluctuations occur within a range such as 26–34 lb.ft. The amplitude of the fluctuations varies considerably but has an average value of about 2 lb.ft.



Text-fig. 12. The line *A* indicates the values of *MLQ* calculated at intervals of $\frac{1}{140}$ sec.; the line *B* is the corresponding segment of the kymogram of body sway.

Text-fig. 13 shows the values of the term I_aB/g at intervals of $\frac{1}{84}$ sec. This factor undergoes a rapid and continuous fluctuation between negative and positive values at rapid and irregular intervals of about $\frac{1}{10}$ sec. Close comparison of this graph with the original kymogram shows that these fluctuations are synchronous with the secondary wave series, and are influenced very little by the form of the larger primary waves. During a stance of average duration, in a subject of average physique, the largest fluctuations are between about ± 8 lb.ft. whereas the average fluctuation occurs between ± 2·5 lb.ft.

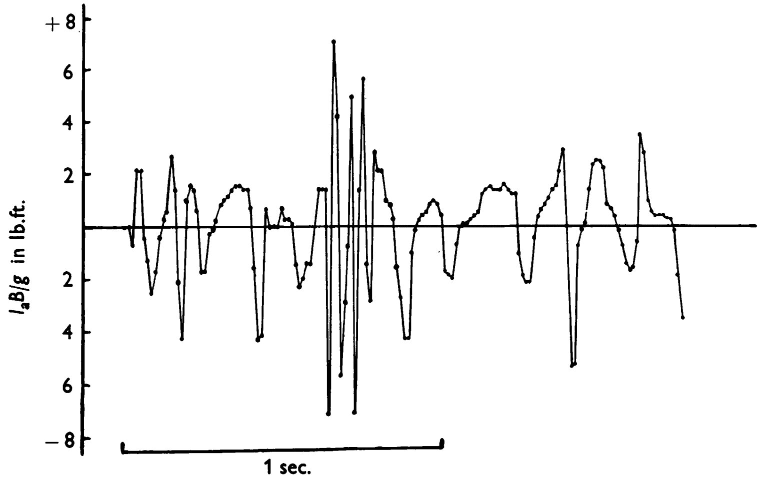
Thus of the two wave series in the kymogram (Text-fig. 10) the primary series are associated with small and comparatively gradual variations in the term *MLQ*, whereas the secondary series are associated with larger and more rapid changes in the term I_aB/g.

(4) *The plantar-flexing torque exerted at the ankle joints by postural activity in the posterior crural muscles of both limbs* (S)

The value of the active plantar-flexing torque operating at both ankle joints (S) is given by the formula

$$S = MLQ + \frac{I_aB}{g} - R \text{ lb.ft.}$$

Consideration of the practically constant value of R, and of the types of fluctuation which the values of MLQ and I_aB/g undergo, makes it clear that the value of S will fluctuate, in the manner of the term I_aB/g, about a mean which itself fluctuates in the manner of the term MLQ. It is apparent that because the factors M, L and I_a are dependent on the body type of the individual the value of S will vary with physique, being larger in tall or heavy subjects than in those who are small or light. However, in a young adult male of average physique it can be said that during a prolonged stance, as has been noted on p. 558, the value of MLQ varies in the range 26–34 lb.ft. and therefore since R is equal to about 5 lb.ft. and I_aB/g fluctuates between maximum and minimum limits of ± 8 lb.ft., the value of S will vary between maximum and minimum values of about 13 and 37 lb.ft. On the other hand, during any short period of stance of about 1 sec. the average fluctuation of MLQ is 2 lb.ft. and that of I_aB/g is $\pm 2{\cdot}5$ lb.ft. and therefore the average fluctuation of S is over a range of about 7 lb.ft. somewhere between the limits 13–37 lb.ft.

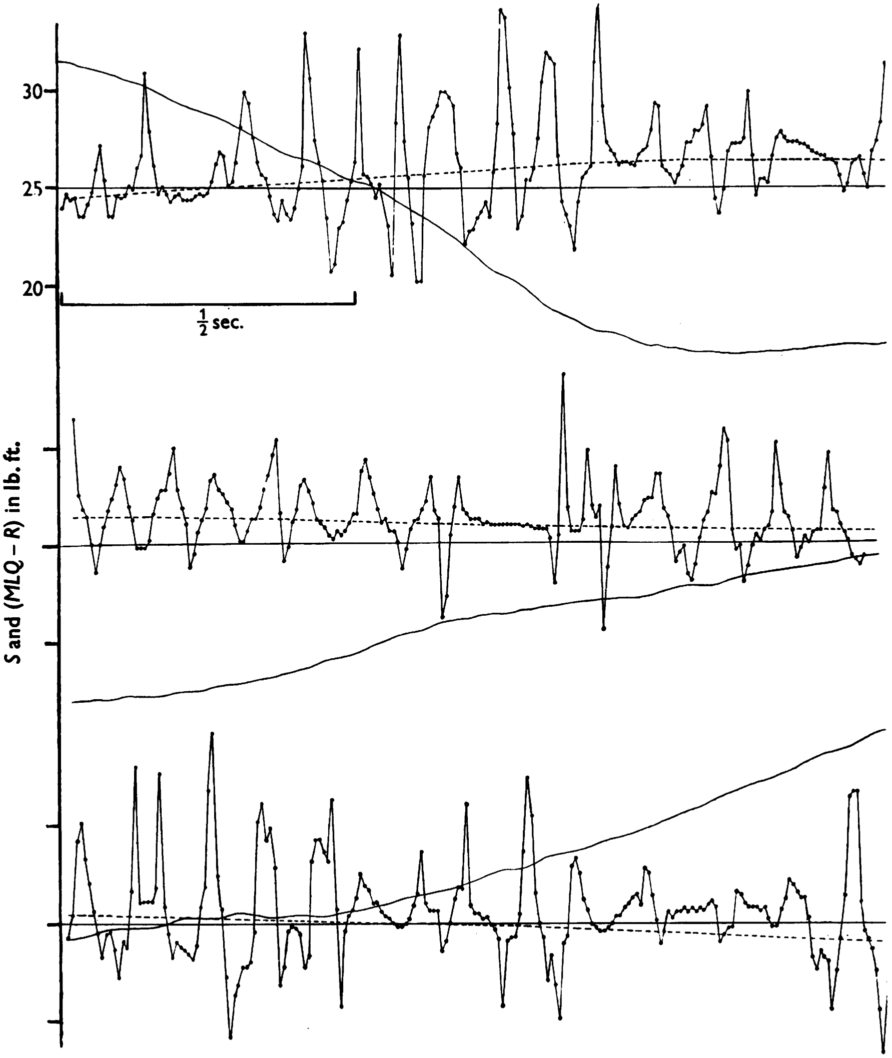


Text-fig. 13. The value of the term I_aB/g at intervals of $\frac{1}{84}$ sec.

Text-fig. 14 shows the values of S calculated at intervals of $\frac{1}{140}$ sec. over a period of about 4 sec. On the same graph variations of the factor ($MLQ - R$) over the same period are indicated by the interrupted line, and the corresponding part of the kymogram in arbitrary units is shown by the continuous line.

DISCUSSION

During standing the body oscillates antero-posteriorly over the foot, but although the centre of gravity consequently bears a variable relationship to the ankle axis it is always anterior to it. Thus body weight exerts at the ankle joint a dorsi-flexing torque which is constant in direction but variable in magnitude. The upright posture is achieved by a plantar-flexing torque which continually counters the effect of body weight, although it exactly equals and neutralizes it only at the anterior and posterior extremes of each oscillation of the body. Between these extremes the difference between the plantar-flexing and dorsi-flexing torques determines the



Text-fig. 14. The values of S at intervals of $\frac{1}{140}$ sec. in a subject of average physique. The term $(MLQ - R)$ is indicated by the interrupted line and the corresponding part of the kymogram of body sway by the continuous line.

direction and magnitude of the angular acceleration of the body around the ankle axis.

The plantar-flexing torque is the sum of that exerted by the active contraction of the posterior crural muscles and that exerted by the passive tension in all the tissues of the posterior crural region ($T = S + R$).

In a young adult of average physique the passive torque in each lower limb ($R/2$) has an approximately constant value of about 2·5 lb.ft. On the other hand, the value of the active torque in each lower limb ($S/2$) fluctuates about ten times per sec. about a mean which itself fluctuates synchronously with the gross swaying of the body at a rate of about twenty times per min. Over a period of a few minutes it may vary within limits of about 6 and 18 lb.ft. while over a shorter period such as 1 sec. it usually fluctuates to an extent of 3 or 4 lb.ft. in the range 6–18 lb.ft. Thus despite the variable nature of the active torque it is always at least twice as great as the passive torque.

In a previous investigation (Smith, 1956) it was noted that the flexing torque which stabilizes the knee joint during symmetrical standing is similarly derived from active and passive sources, but in very different proportions, the active contribution being 30 % and the passive contribution 70 %. The smaller passive contribution at the ankle joint is dependent on a smaller passive tension in the relative extra-articular tissues. This in turn is due in part to the fact that, whereas in standing the knee joint is close to full extension, the ankle joint in similar circumstances is some 20° or so short of full dorsi-flexion. Another reason lies in the nature of the limiting articular mechanism at the two joints. That limiting extension at the knee joint operates over the terminal 10° or so of movement and consequently contributes to the passive force stabilizing the knee joint in standing. That limiting dorsi-flexion at the ankle joint, on the other hand, as can be shown in an osteo-ligamentous preparation of a recently amputated limb, operates over only the terminal 3° of movement and thus contributes nothing to stability during standing.

The exact location of the postural muscle activity which gives rise to the active plantar-flexing torque is not known with certainty, and consequently the torque cannot be accurately expressed in terms of muscle tension. However, despite the lack of certainty, there is some evidence (Smith, 1954) which suggests that a very large measure, if not all, of the activity is located in the triceps surae. If this distribution is accepted, it follows that the muscle tension is transmitted almost entirely through the tendo calcaneus, and that its value can be calculated by dividing the active plantar-flexing torque by the distance between the tendon and the ankle axis. The latter distance has been measured in fifteen subjects of average physique and has been found to vary from 0·08 to 0·12 ft. with an average value of about 0·1 ft. Thus the muscle tension developed in the triceps surae of each limb during standing in one of average physique varies between 60 and 180 lb. during a stance of average duration, and to an extent of about 35 lb. in that range over any period of a second or so.

This tension is considerably greater than that suggested by Joseph & Nightingale (1952). They suggested that in a subject of 140 lb. body weight the tissue tension, both active and passive, had a value of about 35 lb. However, their figure depended on the distance between the tendo calcaneus and the ankle axis being 8 cm., whereas, as noted above, the present investigation shows that the distance has an average value of 0·1 ft. or 3 cm.

The absence of fatigue in the triceps surae during standing, would seem to be associated with two factors. First, the muscle tension, high as it is, is only a small fraction of that commonly developed during normal activity. Thus Elftman (1939) has estimated that at a moderate rate of walking each triceps surae develops a tension at each step of about 600 lb. And secondly, the act of standing is essentially periodic in nature (Smith, 1953), periods of immobility, of an average duration of 30 sec., being separated by brief phases of movement in which the position of the body is shifted. Thus throughout standing the triceps is momentarily relieved of its burden of activity at frequent intervals.

The large amplitude of fluctuation in the value of the muscle tension in the triceps surae is also of interest. It might appear that fluctuations in muscle tension in the two limbs to an extent of 70 lb. at a frequency of about 10 cyc./sec. is incompatible with the very small variations in the position of the body which result. However, the displacement of a body does not depend only on the forces applied to it; it depends also on the duration of the force and on a third factor, which in the case of linear motion is the mass of the body and in the case of angular motion is the moment of inertia. Thus, even a large force opposing for a brief period the movement of a heavy flywheel would cause little change in its angular velocity because of the large moment of inertia of the flywheel and the short duration of the opposing force. Similarly, because of the large moment of inertia of the body about the ankle axis and the high frequency of the fluctuations of muscle tension, these large changes in the tension in the posterior crural muscles cause little displacement of the body as a whole.

As regards the cause of the fluctuations in muscle tension it seems apparent that they are the result of two factors, namely, an incomplete tetanus of the motor units involved and a measure of synchronization in the firing of these units. Such an explanation is fully in keeping with the observations of Denny-Brown & Nevin (1941) and Denny-Brown (1949), who noted that postural activity of muscle differs from that associated with voluntary movement in two respects. It increases by recruitment of units all firing at a low maximum rate of 5–10/sec. in contrast to the increase of willed effort both by recruitment and by an increase in the rate of firing of previously active units. Secondly, it is characterized by a periodicity in the intensity of the discharge of action potentials and consequently by a 'clonic' type of electromyogram; this is in contrast to the typically asynchronous discharge and continuous electromyogram of voluntary movement.

It is difficult to reconcile the results of this mechanical estimation of the tension resulting from postural activity in the posterior crural muscles during standing with those of the many electromyographic examinations published during recent years. These results have varied considerably. Ralston & Libet (1953), reporting the results of the biomechanics group at the University of California, maintain that no activity occurs in the calf muscles during standing, except during excessive swaying of the body. On the other hand, Kelton & Wright (1949), Floyd & Silver (1950) and Smith (1954) have all observed a mild degree of activity which was intermittent in nature and synchronous with the antero-posterior swaying of the body. And thirdly, there is a large body of opinion, of which the works of Hoefer (1941), Seyffarth (1940), Weddell, Feinstein & Pattle (1944), Joseph & Nightingale (1952) and Joseph, Nightingale & Williams (1955) are representative, which considers that activity is continuous and of more or less uniform intensity throughout standing.

Consideration of the variations in muscle tension during standing described above suggests that the appearance of an electromyogram might well vary with its duration. Thus an electromyogram of short duration might be expected to show a continuous discharge of practically uniform amplitude, whereas a longer recording should show a discharge which, although continuous, fluctuated considerably in amplitude. It therefore seems reasonable to infer that the recordings quoted above which show an intermittent activity indicate either an instrument of insufficient sensitivity, as Joseph & Nightingale (1952) have suggested, or an unconscious exaggeration of body sway on the part of the subject—an exaggeration which can readily occur if the subject suspects the purpose of the experiments. On the other hand, those recordings which show no significant variation in amplitude with body sway may simply be too short to show such a variation; variations in amplitude synchronous with body sway are very difficult to appreciate on the fluorescent screen and are only readily apparent when recordings of over about 10 sec. are examined.

Quite apart from the exact pattern of the muscular discharge in standing, the results of the present investigation seem to have a bearing of some importance on the general use of surface electromyography. In their examination of the posterior crural muscles in standing, Joseph *et al.* (1955), using a very sensitive and well screened instrument, obtained deflexions of the order of 40 μV. above the background disturbance. The present results show that these deflexions represent the electrical activity associated with a comparatively high muscle tension of the order of 120 lb. It therefore seems questionable whether surface electromyography in its present form is to be regarded as an adequate method for determining inactivity or assessing minor degrees of activity in muscle—purposes for which it has been frequently employed during the last few years.

SUMMARY

1. During standing gravity constantly tends to carry the body forwards around the axis of rotation of the ankle joint. This dorsi-flexing force is resisted, and the upright posture is maintained by active and passive forces tending to cause plantar-flexion at the ankle.

2. Methods are described by which these active and passive forces can be assessed.

3. The active force is the result of a postural contraction which is located in large measure in the triceps surae. The magnitude of the force fluctuates continually at a rate of about 10 cyc./sec. between high and low values in the range 80/160 lb.

4. The passive force is the result of tension in passive extra-articular tissues in the posterior crural region. The plantar-flexing torque which it exerts at the ankle joint remains approximately constant throughout stance and has a value of about 2·5 lb.ft.

5. The results of this investigation are correlated with those of the numerous electromyographic studies of the posterior crural muscles in standing which have been reported in recent years.

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