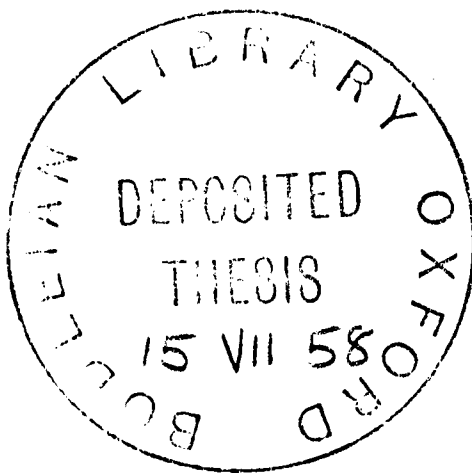


EXPERIMENTAL STUDIES ON POSTURE AND LOAD CARRIAGE IN MAN

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PREFACE

This thesis consists of four papers, in which an attempt has been made to study certain problems relating to the upright posture in Man. The thesis embodies the results of a study of the normal standing position, and how this is affected by the carriage of loads of varying weight. The first two papers relate to an experimental study of the muscular activity associated with maintaining the erect posture, and to the normal postural sway that occurs with standing. The other two papers are concerned with the manner in which the normal erect position is altered by load carriage on the back, and the effect this has on the postural activity. The common theme of the thesis has been to study some aspects of the muscular mechanisms controlling the upright posture in Man, using the 'intact organism' under as normal conditions as possible, and to examine the way in which this posture is affected by a common form of stress, using load carriage on the back for this purpose.

The experiments were conducted on a relatively small group of medical students, who, despite a fairly wide range of weight and height, must be considered a select and homogeneous group in relation to the population at large. It is believed, however, that a detailed analysis of the experiments on this small group, with proper statistical evaluation, has given results which are of general application.

"For station is properly no rest, but one kinde of motion, relating unto that which Physitians (from Galen) doe name extensive or tonicall....for in this position the muscles are sensibly extended, and labour to support the body, which permitted unto its proper gravity would suddenly subside and fall unto the earth..."

Sir Thomas Browne, 1646.

"Gravity is an unvarying standard factor in biological experience and plays its part in every movement and in every posture. Posture is essentially the counter-action of weight or in other words the active antagonism of gravity."

Derek Denny-Brown, 1928.

INTRODUCTION

By assuming the upright posture during the course of his evolutionary development, Man placed himself in a stance highly susceptible to the effects of gravity. It became essential to the continuance of this upright position that he developed an intrinsic mechanism whereby the ever-present effects of gravity could be counteracted. It has long been recognised that the chief means whereby the body resists the effect of gravity in maintaining the upright position is by the active contraction of certain muscles of the body, which have been termed the "anti-gravity" muscles. It has been believed that the necessary contraction of these postural muscles is an active and continuous process, while the body is in the erect posture. More than three hundred years ago, Sir Thomas Browne wrote about the muscles "labouring to support the body"; Denny-Brown, writing in more recent times, and representing the views of the Sherrington school of neurophysiology, described posture as being the "active antagonism of gravity." Sherrington himself described reflex tonus, which maintains an animal in the erect attitude, as static contraction (Sherrington, 1915).

An opposing point of view has developed in the past few years, which insists that the maintenance of the upright position in Man is not an active process, in the sense that it requires continuous contraction of the postural muscles. The views of this group can best be summarized by a quotation from Clemmesen (1951): "It is evident,

therefore, that we must reckon with the passive elastic tensions and forces in the muscles, and that this concept is much more correct than the now discarded concept of muscle tone." This school of thought believes that the effects of reflex postural tonus have been much over-estimated, and that the normal upright posture is chiefly dependent upon the mechanical arrangements of the skeleton (Kelton & Wright, 1949), and on the intrinsic elastic properties of the muscles (Ralston & Libet, 1953; Ralston, 1957). Such activity as may be found in the postural muscles is at most intermittent in occurrence and minimal in extent. They do not believe that the general level of muscle tone is due to the operation of the stretch reflex of Liddell and Sherrington (1924). This view has been developed more or less entirely on the evidence of the electromyograph, and furthermore, on evidence of a negative nature. It is claimed that no evidence of electrical activity can be detected in resting muscle, and that indeed there is very little activity in the postural or "anti-gravity" muscles. The evidence for this view has been summarized by Ralston and Libet (1953).

The present study was initiated with the thought of examining afresh the evidence for the existence of significant muscular activity associated with the upright posture. A study was made of the normal standing posture using primarily a technique which examined the mechanical end-result of muscular activity, rather than the electrical phenomena associated with contracting muscle, as in the electromyograph. The technique that was used, involving a strain gauge accelerometer mounted on a force analysis platform (see appendix)

is believed to give a much more representative picture of the overall muscular activity associated with standing than can be obtained by other methods, such as the electromyograph. The use of the E.M.G. has led certain workers to conclusions which are diametrically opposed to the findings of virtually the whole of the extensive experimental work on reflex posture, and it was therefore considered worthwhile to examine the subject afresh, using different techniques. Young adult males were employed as subjects for the experiments.

At the same time as the muscular activity associated with the upright stance was being examined, a study was also made of the normal postural sway. The static equilibrium of Man in the upright position is never perfect, and there is always a certain amount of postural sway, chiefly in an antero-posterior direction. This sway is the resultant of two opposing forces - the effect of gravity, tending to cause the body to fall to the ground, and the restraining effect of the "anti-gravity" muscles, which maintain the body in the upright position. The vertical projection of the centre of gravity of the body normally falls in front of the ankle joint axis (Morton, 1952), and there is a constant tendency for the body to fall forwards. The continuous interplay between gravity and the "anti-gravity" muscles results in what is known as postural sway. The logical place for measuring this postural sway is the centre of gravity of the body, and this is the procedure that has been adopted in the present experiments. A method of measuring the postural sway that takes place has been devised which is believed to be more accurate and more meaningful than previous methods. This involves a detailed planimetric

analysis of photographic records of the sway tracing, expressing the result as a mean deviation of the tracing from a baseline that had been fitted to a fixed length of the tracing.

Previous workers on postural sway have tended to assume that the body moves during postural sway in a rigid fashion, so that the line joining the axis of rotation of the ankle joints and the centre of gravity of the body can be represented as a straight line. This assumption, on which much work has been based (Smith, 1954, 1957), has never been experimentally examined to determine whether it is valid. To settle this point, a study was made of the sway at the waist, and this was compared with the sway as measured by the movement of the centre of gravity. If the body sways rigidly, the two results should be comparable, in that the mean angle subtended by the centre of gravity with the vertical should approximately equal the mean angle subtended by a fixed point at the level of the waist. This was not, however, found to be the case, and the significance of these results are discussed in the paper on postural sway.

The second half of the thesis deals with the effect of load carriage on the normal alignment of the body, and also its effect on the postural activity associated with standing. After studying the normal posture, it was decided that it would be profitable to examine the effect of load carriage on the body, to determine in what way the body adapted itself to a common form of "stress". The study was divided into two parts, in the first of which the effect on the normal body alignment of carrying weights of 12 and 24 kg. on the back was examined. In the second part, the effects on the normal

postural sway, and associated muscular activity, of carrying these weights were examined.

The obvious and immediate effect of carrying a heavy load on the back is to cause the body to lean forwards, and this observation is a matter of everyday experience. However, the precise manner in which the body leans forwards, and the effect of the position of the load on the back, has been but little studied. It has been claimed (Hellebrandt, Fries, Larsen & Kelso, 1944) that the effect of a load on the back is to cause the body to lean forwards "in toto" over the ankle joints. Using a photographic technique, and taking pictures of marked anatomical landmarks from the lateral side, it was found that there was in fact a highly significant realignment of the body segments, and that in fact the body did not lean forwards rigidly. It was found that the trunk altered its inclination according to the position of the load on the back, and as a result of this it was suggested that the vertical projection of the centre of gravity of the body plus the load remained relatively undisturbed.

Using the same subjects, the effects of load carriage on their postural activity was also investigated, and this study forms the subject of the fourth paper in this thesis. In addition to examining the effects of load carriage on sway, an attempt was made to assess the rate of working of these muscles controlling postural sway, using for this purpose an accelerometer mounted on a freely suspended platform. It was found that carrying a load on the back caused a significant increase in the postural sway of the subjects standing on the platform, and a significant increase in the output of the

accelerometer, which was interpreted to mean an increase in the rate of working of the postural muscles. Increasing the load caused an increase in the postural sway, and in the level of output of the accelerometer. These findings conflict with the work of Hellebrandt et al. (1944), who claimed that the addition of an Army pack on the back lessened the amount of postural sway.

The amount of postural sway is one of the most sensitive indices of the static equilibrium of the body, and it is of interest in this connexion that it was found that the position of a heavy load on the back did not significantly affect the level of postural sway. This finding does not support the views of Lippold & Naylor (1950), who, on the basis of electromyographic evidence, suggested that carrying a load high on the back increased body instability. If the body alters its alignment so as to keep the vertical projection of the centre of gravity constant, wherever the load is carried on the back, then it might be expected that the site of a load on the back would not significantly affect the rate of working of the postural muscles resisting the effects of gravity. In so far as the accelerometer output reflects the rate of working of the postural muscles, this conclusion is supported by the fact that the level of output did not significantly change when the position of a heavy load on the back was altered.

Certain general conclusions may justifiably be drawn from the experimental work described in this thesis. It is believed that the maintenance of the upright posture in Man is essentially a dynamic process, and it is considered fallacious to regard this

position as primarily a passive phenomenon. Gravity is a force which is always present, and is always threatening to upset Man's equilibrium. This equilibrium depends upon, and is largely controlled by, the continuous activity of the postural muscles. The part played by the ligaments and the passive elastic tension in the muscles is of a subsidiary nature. The body weight is supported against gravity chiefly through the active contraction of the muscles - the "Stütztonus" of Rademaker (1931). The point has been well put by Steindler (1955): "These rotatory components of gravity must be neutralized by active opposing muscle forces, if the equilibrium is to be established and the parts are to be maintained in their relative position." The continuous interplay between gravity and the restraining muscular contractions results in a so-called "static ataxia" which is normally remarkably small in amount. It is evidence of a finely-controlled balance that the centre of gravity of the body in the normal standing position does not sway over a distance of more than a few millimetres.

The reaction of the body to the stresses of load carriage is a good example of the powers of adaptation of the body. By keeping the vertical projection of the centre of gravity more or less constant, wherever a load may be positioned on the back, the body adapts itself to one position of the load as well as to another. As an example of how successful this adaptation can be is the finding that there was no significant change in the level of postural swaying when a heavy load was moved over a distance of 30 cms. on the back. The evidence would suggest that this adaptation is achieved, not by the body acting as a rigid column, either during sway or load carriage, but by the

pliant manner in which the body reacts to the stresses of gravity and load carriage.

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Paper 1THE POSTURAL MUSCULAR ACTIVITY ASSOCIATED WITH STANDING IN MAN

INTRODUCTION

The existence of significant activity in the postural muscles of Man when standing in the normal upright position remains an unsettled issue. It has often been demonstrated, since it was first shown by Adrian & Bronk (1929), that action potentials cannot be obtained from resting muscle using needle electrodes (Denny-Brown, 1949; Bigland & Lippold, 1954). It has further been claimed that little or no motor unit activity can be detected in the postural "anti-gravity" muscles of Man, standing in the normal upright stance (Hoefler, 1941; Weddell et al. 1944; Kelton & Wright, 1949; Floyd & Silver, 1950, 1955). These essentially negative electromyographic findings have been used to justify the claim that the upright posture in Man is primarily dependent upon the so-called "passive elastic tension" in the muscles, combined with the mechanical arrangements of the skeleton (Kelton & Wright, 1949). These latter authors suggested that muscle activity occurs only intermittently, and that osseous and ligamentous structures are more important in maintaining the upright posture.

The concept that the intrinsic elastic properties of the muscles are prepotent in maintaining normal standing, rather than active muscular contraction, has been developed apparently entirely

due to the fact that the electromyograph has failed to reveal more than minimal electrical activity in the postural muscles (Hoefer, 1941; Inman, Feinstein & Ralston, 1948; Clemmesen, 1951; Ralston & Libet, 1953). Because electrical activity is so little in evidence in the postural muscles during standing, as revealed by the E.M.G. technique, it has been claimed that postural attitudes in Man are not dependent on the stretch reflex, and that therefore the classical concept of muscle tone must be abandoned (Ralston & Libet, 1953; Ralston, 1957). It has been further observed that most of the classical work on posture has been done on laboratory animals, and so may not be relevant to Man (Clemmesen, 1951).

Jacobson (1943) denied that normal standing in Man was associated with the absence of action potentials, and he claimed to have demonstrated muscle activity quantitatively in terms of motor unit potentials, during normal standing. Göpfert (1952), using needle electrodes, demonstrated electrical activity in the muscles of relaxed subjects, in a reclining position. Göpfert believed that this activity was due to motor unit activity. Joseph and Nightingale (1952, 1955) demonstrated continuous, well-marked activity in the soleus muscles of an upright Man, and they found no evidence of the intermittent activity in the calf muscles described by some workers (Kelton & Wright, 1949; Smith, 1954). Portney and Morin (1956) found activity present in the gastrocnemii and hamstring muscles in the easy standing position more frequently than had been found by previous workers. They suggested that the discordant reports on the constancy of function of the postural muscles during standing could be

explained on the basis of minor displacement of the centre of gravity.

It is apparent that there is conflicting evidence from *as to the degree of muscular activity associated with* electromyographic studies ~~whether the upright posture in Man is associated with significant muscular activity.~~ The experiments described in the present study were designed to establish, using techniques other than electromyography, whether there is any evidence that the upright posture is associated with continuous skeletal muscular activity related to the maintenance of this posture.

METHODS

Experiments were conducted using two techniques. The level of the forces exerted at the feet during normal standing was measured, using an accelerometer. The second technique attempted to assess the comparative effects of standing and lying down on the production of ischaemic pain in the leg muscles. The experiments were conducted on twenty male medical student volunteers, aged 18 to 22. Their heights ranged from 170.6 to 192.3 cms., and their weights from 56.75 kg. to 84.75 kg. None of the subjects gave a history of any neuromuscular disorder, nor of any recent sporting injury.

The experiments were conducted using an accelerometer attached to a freely-suspended, optimally damped platform. The platform that was used for this purpose was actually a force analysis platform, the detailed construction of which has been described elsewhere (Whitney, 1958a; see also appendix). Only a

brief description will be given here of the relevant parts of the platform used in these experiments. The subjects stood on a rigid steel platform, 120 cms. square, which was suspended at each of its four corners by vertical lengths of piano wire. The other ends of the wires were attached to horizontal spring steel bars, which were firmly clamped by their roots to a steel chassis attached to the floor. Movement in the horizontal axes (side-to-side and front-to-back) was limited by horizontal piano wires running along the sides of the platform, attached at each corner to two vertical spring steel bars. The whole platform was optimally damped (approximately 60% critical) by oil-filled bellows, so that there was no perceptible movement of the platform when a subject was standing normally on it. The platform in fact felt completely stable to the subject. A more detailed description of the force analysis platform is given in the appendix. (1).

The accelerometer was attached to the platform so that its sensitive axis recorded acceleration in an antero-posterior direction (relative to the position of the subject on the platform) in the horizontal plane. The accelerometer employed a strain gauge transducer, and was manufactured by Statham Laboratories, Los Angeles, California (Model C-1-350). It was sensitive to acceleration over a range of plus or minus one G., and was connected up to a recording galvanometer on which continuous photographic records could be made. As the subject stands on the platform, he continuously sways to and fro, and through his feet exerts corresponding reactionary forces on the platform. The antero-posterior acceleration of the platform

in the horizontal plane, caused by these reactionary forces, is recorded by the accelerometer. The actual movement of the platform is very slight, as the tendency to displacement is resisted by the mechanical arrangements of the platform.

Mathematically, the second differential of displacement is acceleration, and the accelerometer may be considered to be doubly differentiating the horizontal displacement of the platform caused by the forces acting at the feet of the subject. The accelerometer output was invariably found to be oscillatory in nature, when a subject was standing on the platform. The frequency of the output was always around 10 c.p.s., and it is therefore believed that the accelerometer is detecting the oscillatory component of the muscular forces acting at the feet (Whitney, 1958b). The higher frequency force components are predominantly represented in the accelerometer record, as the low frequency force components (of a frequency substantially below 10 c.p.s.) are differentially attenuated by the accelerometer.

All subjects rested for at least five minutes before the start of an experiment. Light clothing was worn, and standardized plimsolls of the appropriate size were provided. Each subject was asked to adopt an "easy standing position", and to make no voluntary movements, such as turning the head or bending the knees. The heels of the subjects were placed 20 cms. apart, but they were allowed to spread their toes at will, in a manner such as they would normally adopt. Subjects faced away from the recording apparatus, and as far as possible, distractions were kept to a minimum.

On ten of the subjects, a second experimental technique was used on another occasion. This involved determining the effect of posture on the development of ischaemic pain in the legs. The sensations of the ten subjects were compared when they were standing in the upright posture with occlusive cuffs around both legs, just below the knees, and when they were in a recumbent position, with the leg circulation similarly occluded. The blood pressure cuffs (thigh type) used for the experiments were inflated from a reservoir to a pressure of between 210 and 220 millimetres of mercury. The duration of the experiment was twelve minutes in each case, and the effects of standing and lying down were compared on separate days. As a control, the subjects were also asked to stand normally for twelve minutes without leg cuffs. Subjects were asked to comment freely on any sensations they experienced, and as far as possible leading questions were avoided.

The analysis of the photographic records of the accelerometer output was performed by planimetry. The amplitude of the accelerometer tracing was measured over a fixed distance, and the mean height of the tracing derived from the area measured. Using a calibration factor, the output of the accelerometer could be obtained in centimetres per second squared. In ten of the subjects, the accelerometer output was measured on four separate occasions, each for a period of twenty seconds of normal standing. The records of the other ten subjects, obtained on two separate occasions, were measured for a period representing forty five seconds standing. The results of the seventy measurements obtained on twenty subjects are given in Table 1. The data was collected over a period of several months, with intervals of at least

several days (and often several weeks) between successive experiments on the same subject.

RESULTS

In all subjects, when standing normally in the "easy standing position", evidence was obtained of a continuous, well-marked output from the accelerometer. This output was oscillatory in nature, with a frequency of around 10 c.p.s. (Fig. 1). A continuous output was recorded from the accelerometer while the subject was standing on the platform, and there was no evidence of intermittent activity, in the sense that there was no period when activity was absent. There were, however, frequent bursts of output of a higher amplitude than the normal level.

In ten of the subjects, four measurements were made from the accelerometer records of each subject, which were obtained on four separate occasions. The records were all obtained over a period of three minutes standing in the normal standing position, and it was considered that by limiting the period of standing to three minutes, the effect of fatigue could be discounted. As will be seen in Table 1, in which is listed the four measurements obtained from each of ten subjects, the mean of the averages for the ten subjects is 0.764 cm/sec.^2 (S.E. 0.137 cm./sec.^2). This mean is close to the mean of the averages obtained from ten different subjects examined several months later than the group in Table 1, in which two separate measurements were made, each representing a period of standing of forty five seconds

TABLE 1

Table 1. Level of accelerometer output in cm./sec.² Normal standing in ten subjects, measured during four experiments, each on separate days. Output measured for 20 seconds standing.

Subject	Day 1	Day 2	Day 3	Day 4	Average
M.C.B.	0.78	0.41	0.67	0.46	0.580
J.C.E.H.	0.48	0.32	0.26	0.27	0.332
A.J.D.B.	0.67	0.60	1.76	0.25	0.820
D.F.J.A.	0.78	0.98	0.73	0.94	0.857
R.J.H.	0.42	0.37	0.45	0.30	0.385
J.H.O.	0.44	0.36	0.35	0.33	0.370
D.L.N.	0.78	0.79	2.11	2.52	1.550
K.J.T.W.	0.63	0.98	0.60	0.48	0.672
T.G.T.	2.56	1.60	0.76	0.99	1.477
C.J.T.B.	0.57	0.91	0.57	0.34	0.597
Average	0.811	0.732	0.826	0.688	0.764
Standard error	0.199	0.137	0.193	0.220	0.137
Range	0.42-2.56	0.32-1.60	0.26-2.11	0.25-2.52	0.332-1.550

(Table 2). The mean of the averages for this group was 0.684 cm./sec.^2 (S.E. 0.125 cm./sec.^2). If the twenty subjects are considered together then the mean of the total averages for the acceleration recorded is 0.624 cm./sec.^2 (S.E. 0.074 cm./sec.^2), with a total range of 0.25 to 2.56 cm./sec.^2 . There was therefore a factor of ten separating the extreme levels of acceleration recorded from all the subjects, and a factor of more than three separating the extremes recorded from any one subject. This great variability is of considerable interest when it is considered that the subjects were ostensibly adopting identical postures.

The oscillatory nature of the accelerometer output was always evident, and was always found to be within the range of 8-12 c.p.s., and usually 10 c.p.s. As the natural frequency of the recording apparatus is much higher than this frequency, it can be concluded that the acceleration recorded results from that component of the forces exerted which has an oscillatory frequency of around 10 c.p.s. In one or two subjects, it was found that occasionally the level of accelerometer output was so low that it was impossible to count accurately the frequency of the oscillatory output (Fig. 2a). The significance of the 10 c.p.s. nature of the accelerometer output will be considered in the discussion.

The experiments producing ischaemia of the legs were designed to determine whether the upright position affected the rate of onset of ischaemic pain in the legs. Subjects stood in a normal, comfortable standing position for approximately twelve minutes, while the circulation to their lower legs was cut off. They were asked to

TABLE 2

Table 2. Level of accelerometer output in cm./sec.² Normal standing in ten subjects, measured during two experiments on separate days. Output measured for 45 seconds standing.

Subject	Day 1	Day 2	Average
W.G.S.H.	0.75	0.83	0.790
J.M.S.W.	0.83	0.62	0.725
S.M.	0.37	0.29	0.330
D.N.J.	1.87	1.87	1.870
F.D.H.	0.54	1.46	1.000
J.G.R.	0.92	0.37	0.645
C.P.B.	1.21	0.67	0.940
J.S.E.	1.04	0.67	0.855
W.F.M.	0.71	1.12	0.915
W.G.B.	0.75	0.79	0.770
Average	0.899	0.869	0.884
Standard error	0.131	0.155	0.125
Range	0.37-1.87	0.29-1.87	0.330-1.870

comment on any sensations they felt, with especial reference to the lower limbs. Within a period of between two to four and a half minutes from inflating the cuffs, all subjects commented that they had specific sensations in their lower legs. Typical comments were that they felt "tired in the calves", the "front of my legs ached", and "the soles of my feet feel as if I'd been on a long run." The time to the onset of these or similar sensations is given in Table 3. By the end of twelve minutes, all subjects had developed feelings of discomfort in their calf muscles which varied from moderate to severe. Three subjects complained of discomfort in their hamstring muscles, as well as in their lower leg muscles. Two subjects reported that they could not have continued beyond the twelve minutes the experiment lasted, and a further subject fainted, and did not complete the experiment.

On another day, the subjects were asked to stand for twelve minutes in the normal standing position, with no impedance to their leg circulation. At the end of the experiment, six of the subjects had no comment to make, and four of them reported very slight aches in their calves or soles of feet. In none of the subjects did the sensations after twelve minutes of normal standing amount to even mild discomfort (Table 3). Following this experiment, occlusive cuffs were placed around their legs, in the same position as previously. On this occasion, however, the subjects adopted a supine position. The experiment again lasted for twelve minutes, and the cuffs were inflated to the same pressure as before. Most subjects spontaneously

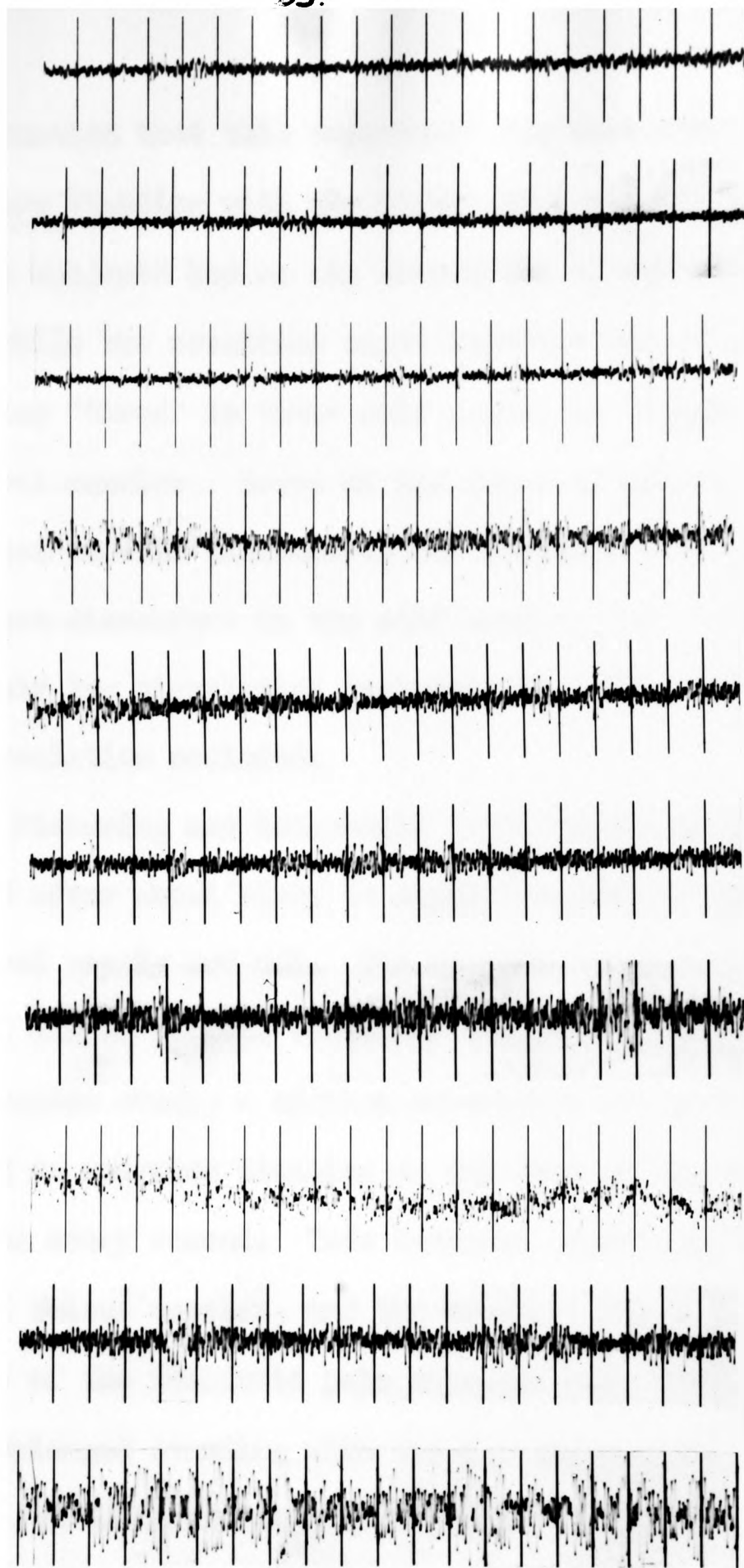


FIGURE 1

Fig. 1. Level of accelerometer output in ten subjects during normal standing. Vertical lines represent one second intervals. Note individual variation between subjects, and oscillatory nature of records.

Calibration mark represents 1 cm./sec.²

Sensitivity of recording unchanged throughout.

offered the information that this experiment was much more comfortable than when they were standing with the circulation cut off to their legs. Two of the subjects had no leg discomfort of any kind throughout the experiment, while the remaining eight reported sensations which varied from feeling "tired" in their calf region to "slight aching" in the anterior crural muscles. Seven of the subjects agreed that the sensations amounted to mild discomfort; all subjects were emphatic that there was much more discomfort in the calf muscles when they were standing with their leg circulation occluded, than when they were lying down with the circulation occluded.

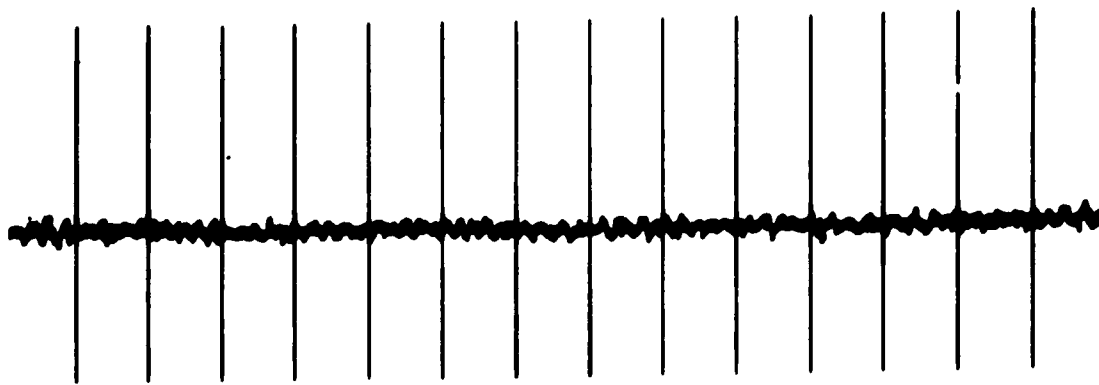
Lewis, Pickering and Rothschild (1931) found that intolerable pain was produced after about sixty to eighty seconds of exercise in man with the blood supply cut off. The exercise performed by their subjects involved almost maximal voluntary gripping movements with the hands. In the present study, a similar experiment was performed on the lower legs, using as exercise standing up and down on the toes at the rate of about once every second. This involved exercising the gastrocnemius and soleus muscles, and the subjects could then compare the type and site of the ischaemic pain produced with that which occurred after prolonged standing with the leg circulation cut off. In most subjects, intolerable pain in the calf muscles was produced after about sixty to eighty seconds of exercise. Like Lewis et al. (1931), we found that the pain in the calf muscles vanished completely within a few seconds of restoring the blood supply to the muscles. Some subjects were able to exercise for a short time beyond eighty seconds before the pain became intolerable. All subjects commented

TABLE 3

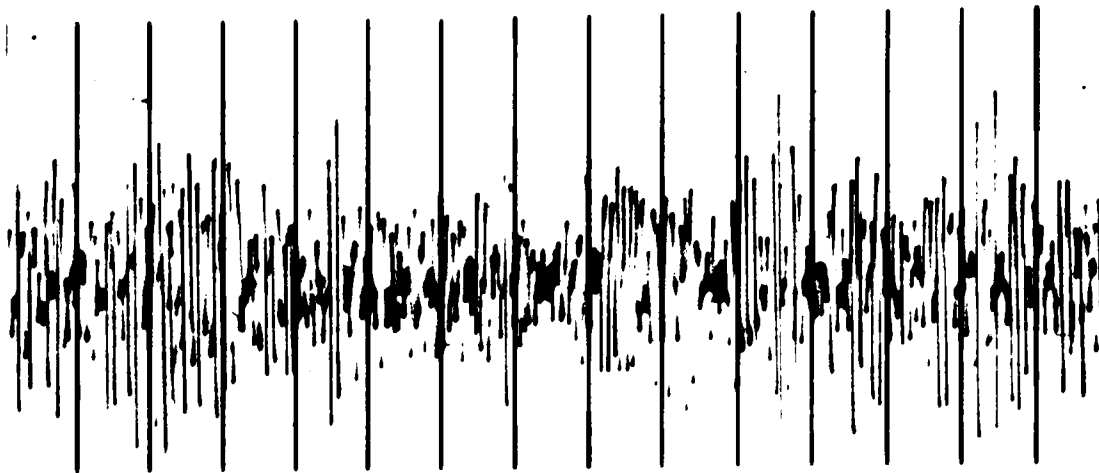
Table 3. Table showing the effects of occluding the leg circulation when standing normally and when lying down, in ten subjects. 12 mins. normal standing (A), 12 mins. supine with occluded leg circulation (B), and 12 mins. standing with occluded leg circulation (C).

Subject	Degree of discomfort in legs at end of experiment + = mild discomfort ++ = moderate discomfort +++ = severe discomfort			Time in minutes (to nearest $\frac{1}{2}$ minute) to first report of sensations in legs related to the muscles (see text)		
	A	B	C	A	B	C
M.C.B.	-	-	++	-	-	3
D.L.N.	-	+	++	10	5 $\frac{1}{2}$	4
R.J.H.	-	-	++	-	5 $\frac{1}{2}$	3
C.J.T.B.	-	-	++	-	6 $\frac{1}{2}$	3
J.H.O.	-	+	+++	7	4	3
J.C.E.H.	-	+	+++	-	5 $\frac{1}{2}$	2 $\frac{1}{2}$
T.G.T.	-	+	++	-	8	4 $\frac{1}{2}$
K.J.T.W.	-	+	+++	-	4	2 $\frac{1}{2}$
A.J.D.B.	-	+	++	12	3	2
D.F.J.A.	-	+	+++	12	9	3
Average					5.7	3.1
Range					5.5-9	2-4.5

* subject fainted before completion of experiment



(a)



(b)

(c)

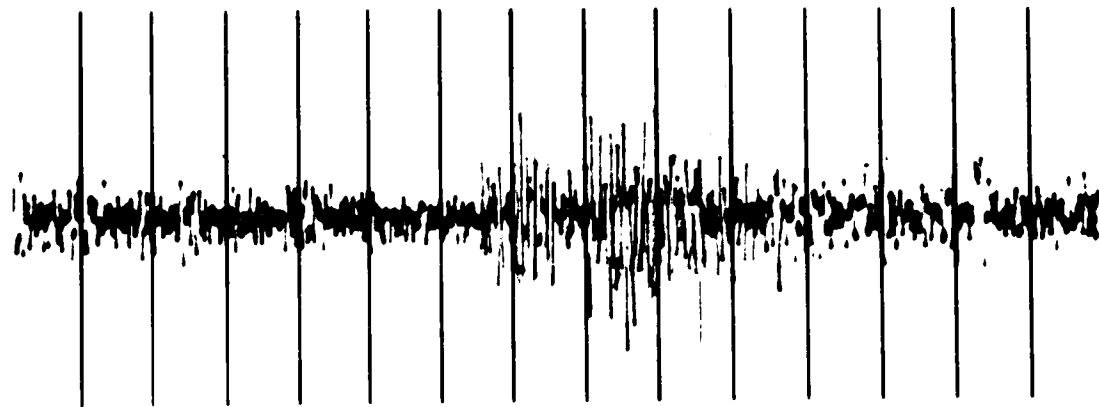
FIGURE 2

Fig. 2. Level of accelerometer output in two subjects, showing extreme ranges recorded. Vertical lines represent one second intervals. Calibration mark equals 1 cm./sec.² Sensitivity unchanged throughout.

(a) Subject with the lowest level of accelerometer output recorded.

(b) & (c) Same subject on different days.

that the discomfort they experienced at the end of twelve minutes standing with occluded leg circulation was a lesser version of the acute discomfort produced by exercising for approximately one minute, in the manner described. The site and type of pain was essentially similar in both experiments.

It is, of course, difficult and hazardous to compare the sensations of different subjects, and it is appreciated that "aching" and "discomfort" are vague and indefinite terms. Nevertheless, it seems reasonable to conclude that there is some factor which makes prolonged standing with arrested leg circulation a discomforting and even painful experience. The sharp contrast that was found between the sensations associated with standing and lying down during peripheral ischaemia suggests that the erect posture is associated with some factor that is operative to a much lesser degree, in the supine position. It is believed that this factor is the active contraction of the posterior crural muscles, especially soleus, in relation to the maintenance of the normal standing position.

Subjects were also asked to stand on the platform after they had been standing for ten minutes with the circulation to their legs cut off. The amplitude of the accelerometer output recorded under these conditions was found to be increased above the average level that was recorded during normal standing, with unimpeded circulation. It was found that, on average, the amplitude of the accelerometer output during ischaemic conditions was more than double that obtained during normal standing (1.96 cm./sec.^2 as compared to 0.85 cm./sec.^2).

DISCUSSION

The so-called "tremor" associated with skeletal muscular activity, with a frequency around 10 c.p.s., was first described by Schäfer (1886), and confirmed by Herringham (1890). It is this muscular tremor which, it is believed, the accelerometer is recording in the present experiments. Tremor is found to be associated with muscular activity (skeletal type) throughout the body, and as the tremor remains constant in frequency at whatever strength the muscle is being used (Redfearn, 1956; Whitney, 1958a), it is apparent that the tremor must be related to some fundamental property of skeletal muscular activity. It has been suggested that muscular tremor is due to some rhythmical propensity of the stretch reflex mechanism, and that the 10 c.p.s. activity represents a rhythmical oscillation in the muscle servo loops (Hammond, Merton & Sutton, 1956; Halliday & Redfearn, 1956; Lippold, Redfearn & Vucso, 1957).

Marshall and Walsh (1956) disagreed with the suggestion that tremor was due to oscillation in the feedback around the stretch reflex, and instead put forward the idea that tremor was due to the muscle acting as a kind of "low pass filter". They showed that muscles stimulated more frequently than fifteen times a second do not impart corresponding mechanical shocks of significant amplitude to the limb. Whitney (1958b) has now put forward the attractively simple suggestion that tremor is essentially a fundamental property of the muscle fibres, resulting from them having (in mammalian 'power' musculature) a twitch time of around 100 msec. He has demonstrated, on statistical grounds,

that given a twitch time of around 100 msec., it follows that a contracting muscle must exhibit a frequency of around 10 c.p.s.

According to recent experimental work, the nervous control of posture mainly employs the small motor fibre system, or gamma route, while rapid movements with minimum reaction time go through the alpha, or large motoneurone route (Eldred, Granit & Merton, 1953; see Fig. 3). They suggested that the advantage of the gamma system control of postural movements, which is the type of movement being considered here, is that during shortening the muscles still enjoy the self-regulating or servo properties of the stretch reflex. It was shown by Hunt (1951) that there is a constant tonic discharge to the intrafusal muscle fibres of the spindles, from the gamma efferent fibres, during postural reactions. Constant tonic activity in the gamma loop during normal posture would be expected to result in a diffuse, low-grade activity in the muscles, which is especially marked in those muscles subserving postural activity ("Postural Tonus"). The mechanical end-result of this continuous, low-grade activity in the postural muscles is believed to be what is being detected by the accelerometer used in the present experiments. It seems reasonable to conclude that the accelerometer output must be closely related to the level of postural reflex tonus in the muscles supporting the upright stance. If this is so, the considerable subject-to-subject variation in the level of accelerometer output (Tables 1 & 2) is understandable. Normal human muscle tonus is well recognised to exhibit marked fluctuations in level (Berkwitz, 1932), and this is presumably related, at least in part, to the general level of tonic gamma discharge. It has been

DIAGRAM OF THE STRETCH-REFLEX AND RELATED MECHANISMS.

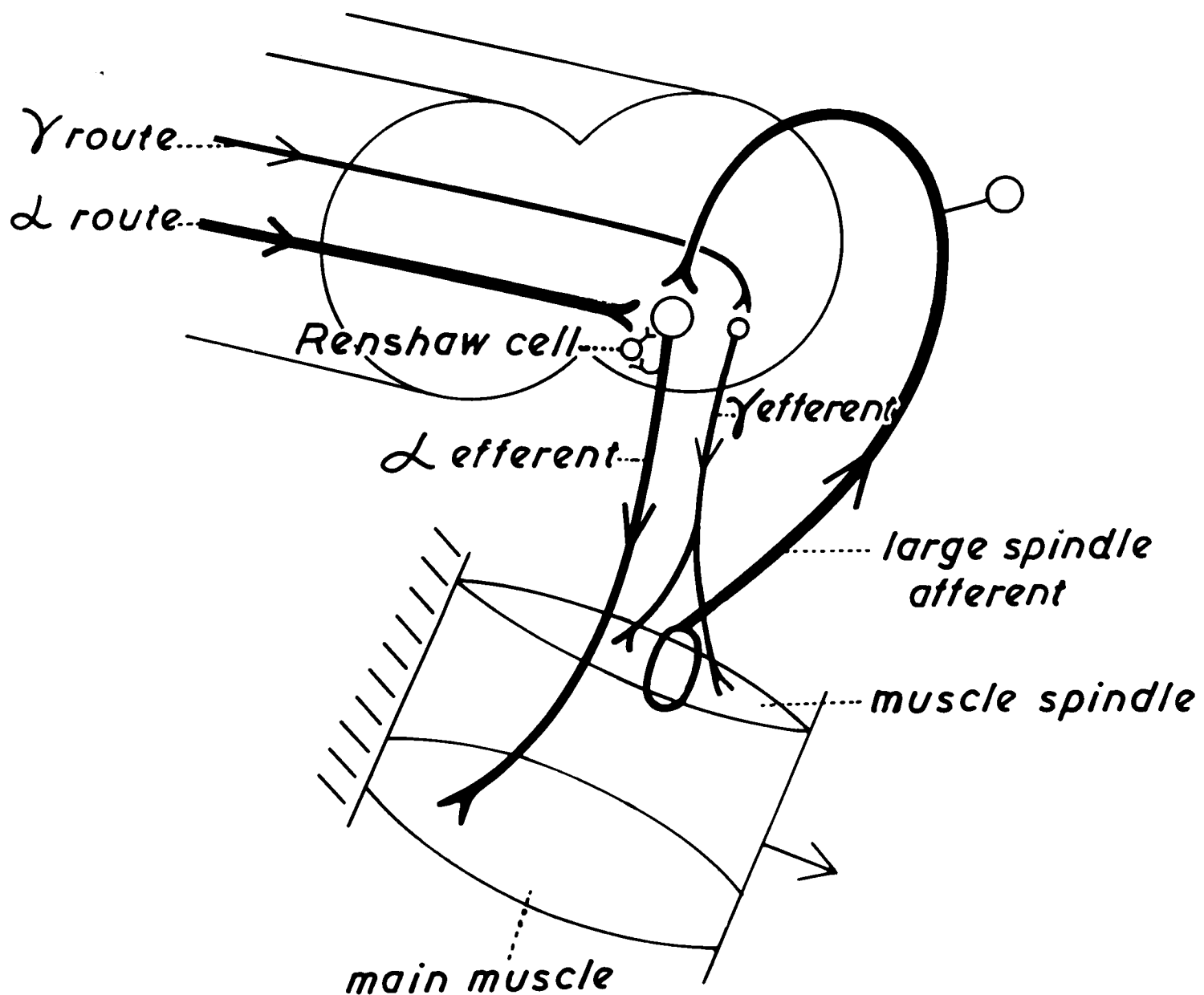


FIGURE 3

Fig. 3. The muscle spindle lies in parallel with the main muscle, and a gamma motor efferent nerve supplies the contractile poles of the spindle and can thus alter the 'bias' on the spindle sensory ending. The muscle contracts due to impulses either from higher centres exciting the alpha motoneurons direct or from impulses in the gamma efferents, which activate the muscle indirectly via the stretch-reflex. (after Hammond, et al., Brit. Med. Bull. 12, 214, 1956)

shown that this tonic discharge can be facilitated or inhibited by higher levels of nervous control (Granit & Kaada, 1952), and it is the changing level of this higher control which probably explains much of the individual variability that has been found to characterize the accelerometer output during normal standing.

It has been suggested by Whitney (1958b), that normal contracting skeletal muscle has a continuous oscillatory component, the period of which is the activation cycle (twitch time) of the individual muscle fibre. According to this theory, therefore, the oscillatory nature of the accelerometer output results from it recording the normal 10 c.p.s. oscillatory component of active muscle. The amplitude of the oscillatory component, according to Whitney, is chiefly dependent on the variation in the degree of synchronization between the muscle fibres. Those subjects with a high level of accelerometer output when standing normally would, on this theory, achieve this chiefly by increasing the degree of synchronization of the muscle fibres in muscles supporting the upright stance. There is no reason to suppose that this increase in accelerometer amplitude is primarily related to a change in the pattern of postural sway of the subjects. Some subjects, for example, were found to produce accelerometer output levels of more than three times that produced on previous occasions, with apparently identical standing postures (Table 1).

An experimental observation was made which tends to support Whitney's theory of muscle action. It was found that the level of accelerometer output could be more than doubled on average by standing

for ten minutes with the leg circulation cut off. According to Whitney (1958b), this higher level of output must result largely from an increase in the synchronisation of the active muscle fibres. Some factor associated with ischaemia must be providing the stimulus for greater synchronisation. It was shown experimentally by Matthews (1933) that cutting off the blood supply to a muscle spindle greatly increased the resting afferent discharge of the spindle. It could well be that a marked increase in the afferent discharge to the spinal cord from the spindles provides a powerful physiological synchronising factor to the motoneurone pools, stimulating them to an increased discharge. Interpreted along these lines, the increase that was found in the level of accelerometer output during ischaemia becomes explicable, and indeed it is predictable from Whitney's theory of muscle action.

Smith (1957), in a detailed study of the forces acting at the human ankle joint during standing showed that the upright posture is maintained by active and passive forces tending to cause plantar-flexion at the ankle joint. He demonstrated that the active force is by far the larger, and is due chiefly to the contraction of the gastrocnemius and soleus muscles. He estimated the magnitude of the force to vary over a range of 80 - 160 pounds, and to fluctuate in extent at a rate of about 10 c.p.s. Smith did not comment on the possible significance of the 10 c.p.s. activity that he found, although he demonstrated that active plantar-flexing forces were continuously present, to an extent that varied markedly. Both the work of Smith, and the experiments described in this paper, using different techniques, provide evidence for the existence of active, continuous

muscular forces with a frequency of approximately 10 c.p.s., being exerted in relation to the normal standing posture.

The experiments involving peripheral ischaemia were performed to determine whether the subjective changes associated with muscle ischaemia were influenced by the posture of the subject. If there is little or no active muscular contraction in the lower legs during normal standing in Man, and if standing were a purely passive act, then it should make little difference to ischaemic muscles whether the subject is standing or lying down. However, it was found that whereas ischaemia of the muscles produced at most a mild discomfort with the subject supine, a severe ache developed in the leg muscles when the subject was in the erect posture, amounting in some cases to frank ischaemic pain. The normal standing posture produced a diffuse, highly discomforting sensation in the calf muscles in all subjects, at the end of ten minutes ischaemia. Neither normal standing with unimpeded circulation, nor lying down with the circulation to the legs arrested produced a comparable effect. If the erect posture is associated with active muscular contraction in the lower legs, then occluding the circulation to these muscles would be expected to precipitate ischaemic pain much more rapidly than if the subject is lying down, and the muscles are inactive. Lewis et al. (1931), suggested that the pain of ischaemia was due to the accumulation of metabolites in the tissues ("Factor P"). Matthews (1933) gave an alternative explanation of the cause of ischaemic pain when he suggested that it was due to the central effects of a greatly increased afferent inflow from the muscle spindles to the spinal cord. This

suggestion is of particular interest in relation to the views recently expressed by Weddell (1957), in which he postulated that the sensation of pain is due essentially to "excessive afferent" inflow from the periphery to the central nervous system. The chief point in relation to the present study is that there would seem to be good evidence from the ischaemia experiments that the normal standing posture is associated with active muscular contraction, at least in the legs.

The evidence that forces from muscular activity are constantly present at the feet, the evidence that ischaemic pain can be produced quite readily in the calf muscles of a subject in the normal standing position, and finally, the finding of continuous electromyographic activity in the soleus muscles of the erect Man (Joseph, Nightingale & Williams, 1955; Naponiello, 1957), all opposes the suggestion that the muscular activity associated with standing is intermittent in character (Ralston & Libet, 1953), or the idea that muscular activity is of secondary importance compared with the osseous and ligamentous arrangements of the skeleton (Kelton & Wright, 1949). Normal standing, it is believed, is essentially dependent upon continuous postural muscular activity, to enable the upright position to be sustained against the effects of gravity.

Certain workers who have failed to find evidence of significant electrical activity in the postural muscles have been led into the position of denying the existence of active muscle tone, and have claimed that the classical idea of muscle tone must be abandoned (Clemmesen, 1951; Ralston, 1957; Nightingale, 1958). This position has resulted chiefly from assuming that significant muscular activity

cannot be present in the absence of detectable action potentials - which is a highly dubious assumption. A recent comment by Smith (1957) is germane to this point, when he says: "It is 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." On the other hand, the use of needle electrodes inevitably means that only a small portion of a muscle can be sampled, and here again, lack of detectable potentials need not necessarily mean a completely inactive muscle. The concept of passive elastic tension, denying the existence of active muscle tone, would seem quite contrary to the evidence deriving from a large body of recent research on neuromuscular physiology (Kuffler & Hunt, 1952; Granit, 1955). The classical concept of postural reflex tonus based upon the stretch^{reflex} of Liddell and Sherrington (1924) may be over-simplified in view of what is now known of the functions of the small motor fibre system. However, there seems no justification for abandoning this classical theory of muscle tonus, for the reasons such as Clemmesen (1951) has advocated; on the contrary, the fundamental ideas originating from the Sherrington school have been enlarged, and given increased meaning, by studies in the laboratories of Granit and Kuffler. The results of the present experiments would seem to accord with the views of this latter group on the mechanism of reflex posture.

SUMMARY

1. A study has been made of the skeletal muscular activity associated with maintaining the erect posture in Man. The extent and nature of this activity was assessed by the mechanical end-result it produced, using an accelerometer mounted on a freely-suspended and optimally-damped platform for this purpose.
2. Evidence was found of continuous, oscillatory forces which are exerted at the feet during the normal erect posture. The finding of this activity was discussed in relation to the views held by certain workers who have studied the erect posture by electromyography.
3. It was found that ischaemic pain was readily produced in the leg muscles when standing normally, whereas over a similar period of time, ischaemia produced virtually no pain if the subject was lying down. This finding was regarded as evidence that the erect posture is associated with muscular activity in the legs.
4. It was concluded that the erect posture in Man should be considered a dynamic state, associated with continuous muscular activity, especially in the lower limbs. This activity is necessary to actively counteract the effects of gravity. It is denied that normal standing in Man is dependent primarily on the mechanical arrangements of the skeleton, or the passive elastic tension in the postural muscles.

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Paper 2THE NORMAL POSTURAL SWAY IN MAN

INTRODUCTION

The normal postural sway of the human body in the upright position has been studied for many years. It has been demonstrated that in the erect posture the body is not motionless between periods of movement, but instead is continually swaying (Miles, 1922; Hellebrandt, 1938; Edwards, 1942; Hellebrandt & Franseen, 1943). Numerous factors have been shown to influence the extent of postural sway, such as the sex of the subjects (Orma, 1957), intoxication with alcohol (Goldberg, 1943), and the types of visual fields observed by the subjects (Wapner & Witkin, 1950). Several methods have been devised for measuring postural sway, the three commonest ones being the measurement of head sway using the Miles "ataxiometer" (Miles, 1922, 1950), or a modification of it; the measurement of shifts of the centre of gravity during postural sway (Hellebrandt & Braun, 1939); and the use of optical methods, using an illuminated point attached to the head (Goldberg, 1943; Orma, 1957).

A clear distinction has not always been made between the extent of postural sway as measured at the head, and sway as measured by displacement of the subject's centre of gravity. The relationship

or divergence of postural sway results as measured by these two methods has not been examined. It has been commonly assumed (e.g., Smith, 1957) that during postural sway a subject sways rigidly over the ankle joints, so that the body segment linking the axis of rotation of the ankle joints and the centre of gravity can be represented by a straight line. The justification for this assumption has not been experimentally established. The position of the feet is known to play a considerable part in determining the amount of postural sway occurring, and especially side-to-side sway (Fearing, 1924). This factor is often insufficiently stressed as a major cause of the variation in sway patterns recorded between different groups of subjects.

The present study was designed to determine whether the body sways rigidly over the ankle joints, and whether the sway as measured by the movement of the centre of gravity can be correlated with the sway as measured by an inclinometer mounted on a waistbelt. The shifts in the centre of gravity during normal standing were measured using a force analysis platform (Whitney, 1958), and the accuracy of the method was checked by a photographic technique.

METHODS

The displacement of the centre of gravity of the subjects was measured while they were standing in a normal, relaxed manner. The principle of the method used was established primarily by Hellebrandt (1938), and involves recording the change in the centre of pressure of the feet during postural sway, which in turn reflects the oscillation of the centre of gravity of the body. For the purpose of

this study, postural sway has been defined as the continuous oscillation of the centre of gravity of the body, which is occurring chiefly in an antero-posterior direction.

The force analysis platform that was used for determining the oscillations of the centre of gravity consists essentially of a freely-suspended and optimally damped steel platform, measuring 120 cms. square. The detailed construction of the platform has already been described (Whitney, 1958; see also appendix), and only the relevant part of the platform used in these experiments will be described here. The whole platform is suspended by piano wires at each of its four corners from horizontal spring steel bars, which are firmly clamped by their roots to a steel chassis attached to the floor. The platform is damped so that it felt quite rigid to a subject standing on it. The deflections of the steel bars at the corners of the platform are transduced to electrical changes by electrical resistance strain gauges mounted directly on the bars. The strain gauges constitute the arms of Wheatstone bridge circuits, and the electrical outputs of the bridge circuits were simultaneously recorded photographically by a multi-channel galvanometer.

The weight of the subject standing on the platform causes a downward deflection of the four supporting horizontal bars, the summed deflection of the bars being proportional to the subject's weight. As the subject sways forwards, the centre of pressure of the subject's weight also moves forwards, and therefore the forces acting on the front two bars increase. This increase is transduced electrically and summed in a separate circuit; from this result is subtracted the sum

of the deflection of the rear two bars. Initially, the platform is balanced electrically; as the subject sways, however, this balance is upset, and by taking moments the imbalance is converted into movement of the centre of pressure of the subject's feet. The result is recorded photographically on a multi-channel galvanometer, and a continuous record of swaying that takes place is obtained. By means of a similar arrangement, a record was obtained at the same time of the lateral movement of the centre of pressure (Fig. 5).

An accelerometer was employed as an inclinometer, attached to the rear of a belt which subjects wore around their waist (Fig. 1; see also appendix 2). The accelerometer (Model C-1-350, Statham Laboratories, Los Angeles), which was sensitive over a range of plus or minus one G., was mounted on the belt so that its sensitive axis recorded acceleration in the A.-P. horizontal plane, relative to the position of the subject. In the vertical position, the effect of gravity on the accelerometer output was nil, and it only recorded acceleration in the antero-posterior direction. However, as the accelerometer is inclined away from the vertical, as the subjects sway, the effect of gravity is such as to cause an increasing bias on the output, depending on the degree of tilt away from the vertical. As a result of the tilting, the accelerometer output has an undulating form (Fig. 2a), which is intimately related to the degree of postural sway of the subject. Essentially, the instrument acts as an inclinometer when it is used in this fashion. At the same time as the accelerometer is acting as an inclinometer, it is also recording the antero-posterior acceleration of the body, with the result that the



FIGURE 1

Fig. 1. View of accelerometer attached to the back of a belt around the subject's waist.

undulating sway record has a superimposed "tremor" of a frequency around 10 c.p.s. (Fig. 2a). The significance of this tremor, and its relation to skeletal muscular activity, has been discussed in a previous paper (Thomas, 1958a).

By equating moments about the mid-point of the platform, the movement of the centre of pressure of the feet can be converted into an equivalent change in the position of the centre of gravity of the subject. The only assumptions that this calculation involves are that the part of the body between the feet and the centre of gravity of the body acts as a rigid connecting link, and that the foot is not actively plantar-flexed during normal swaying in the upright position. The belief that the centre of pressure tracing accurately represented the movement of the centre of gravity of the subject was tested by a photographic technique. Subjects were asked to lean forwards as far as possible, keeping their feet flat on the platform, and taking care not to lose their balance. After leaning forwards for a few seconds, the subjects resumed their normal standing posture. The distance the centre of pressure tracing moved during this procedure was then measured on the record. Coincident with this procedure, a photograph was taken of the subject from his right side. The greater trochanter of the femur of the subject had been marked previously with a black marker, and the distance the marker was displaced when the subject leaned forwards was measured on an enlarged negative image. The details of the technique of photographic measurement used has been described elsewhere (Thomas, 1958b).

The position of the centre of gravity was determined for

for each of the subjects in the supine position. The subjects were placed on a stretcher, attached to the platform for this purpose, the weight and position of the stretcher having been previously balanced out. By equating moments, the position of the centre of gravity of the subject could be established. This figure was then used to estimate the height of the centre of gravity above the hip marker, in both the normal standing and leaning forwards positions. The distance the centre of gravity moved with leaning forwards, as determined by this method, was then compared with the distance the centre of pressure tracing moved forwards at the same time. If the centre of pressure tracing faithfully represents the movement of the centre of gravity, there should be close agreement between these two figures.

Detailed planimetric analysis was made of two sets of records. The antero-posterior body sway, which is defined here as the movement of the centre of gravity, was measured as a total range of movement over a period of time, and also as a mean deviation from a fitted baseline. The sway record was undulating in form (Fig. 2a), and a baseline was fitted by eye over a length of tracing equivalent to two and a half minutes of standing. The areas above and below the baseline were then measured by planimetry as a check that the baseline had been accurately fitted. It was found that, with practice, a baseline could be fitted by eye sufficiently accurately so that the areas above and below the baseline enclosed by the record showed very close agreement. Knowing the length of the baseline, the amount the tracing deviated from this baseline on average, over two and a half minutes, could therefore be estimated. To obtain a measurement of the gross amount

of swaying, the total range traversed by the tracing, above and below the baseline, was also measured for the same period of standing (Table 1). Knowing the weight of the subject, and by equating moments, the measurements obtained could be converted into the mean deviation from a fitted baseline and the total range of movement, respectively, of the centre of gravity of the subject. The body sway was also analysed from the accelerometer record, and here again a baseline was fitted by eye, and checked by planimetry (Fig. 2a). The areas above and below the baseline were measured by planimetry, and thus the mean angular deviation of the tracing from the baseline could be calculated. From this figure, the mean angular deviation from the vertical of the accelerometer, in radians, could be established (Table 2).

Ten male student volunteers, aged 19 to 22 years, were used as subjects. Their heights ranged from 172.7 to 192.1 cms., and their weights from 63.2 to 83.5 kgs. The subjects were free from any physical disability. They wore only light clothing, and standardized shoes of the appropriate size were provided (Dunlop plimsolls). During the experiments the subjects faced away from the recording apparatus, and were not informed of the nature of the experiment, beyond the minimum necessary for cooperation. An important factor was that the subjects were unaware that their postural sway was being measured. Considerable care was taken in positioning the feet of the subjects. The heels were placed along a line marked on the platform, at a distance of 20 cms. apart. Once the position of the heels had been fixed, the subjects were allowed to spread their feet at will. Most subjects adopted a position with the toes pointing outwards slightly.

Subjects stood in a normal easy standing position for three minutes, with their gaze directed in front of them, fixated on a wall some fifteen feet away. By limiting the standing period to three minutes, it was considered that the effects of fatigue on the normal standing posture could be discounted. Each subject was examined twice, usually at an interval of several days, and never twice on the same day.

RESULTS

The normal antero-posterior movement of the centre of gravity of the subjects was found to be on the average 2.78 cms. (S.D. $\pm$ 0.913 cms.) for the total range of movement of the centre of gravity, during a period of two and a half minutes standing. The average displacement of the centre of gravity, expressed as a mean deviation from a fitted baseline, was only 0.137 cms. (S.D. $\pm$ 0.068 cms.). It was found that the observed mean angular displacement of the accelerometer from the vertical during sway was significantly greater than the mean angular displacement calculated using as data the mean deviation of the centre of gravity (mean deviation of centre of gravity divided by height of centre of gravity). Very little lateral sway was observed (Fig. 5).

It will be seen in Table 1 that the postural sway expressed as a mean deviation from a fitted baseline gives a much lower figure than the postural sway expressed as a total range of movement. The mean deviation column refers to the mean amount by which the tracing departs from (above or below) a fitted baseline. There is in general good agreement between the relative values for sway, as determined by the two methods. However, it will be seen that the correlation is

TABLE 1

Subject	Height (cms.)	Weight (kg.)	Height of C. of G. (cms.)	Antero-posterior sway measured from oscillations of the centre of gravity (cms.)	
				MEAN DEVIATION	RANGE
M.C.B.	172.7	65.0	97.8	0.141	1.91
		65.9	97.5	0.120	1.62
J.C.E.N.	185.4	83.7	101.9	0.185	2.75
		83.4	101.4	0.142	2.37
A.J.D.B.	174.2	78.6	95.7	0.113	1.88
		79.1	96.1	0.134	1.91
D.F.J.A.	178.6	74.5	99.9	0.154	2.59
		74.1	99.9	0.358	4.31
R.J.H.	176.2	72.0	99.4	0.154	2.33
		74.0	99.0	0.125	1.74
J.H.L.	177.3	66.7	99.0	0.304	3.01
		66.2	99.0	0.184	2.20
D.L.N.	175.1	69.3	99.3	0.147	1.94
		70.2	99.2	0.190	2.50
H.J.T.W.	188.4	70.5	104.5	0.273	3.80
		77.8	104.5	0.317	4.13
T.L.T.	182.0	85.4	104.1	0.209	3.30
		82.7	104.8	0.196	3.15
G.J.T.B.	184.3	62.0	103.4	0.229	3.62
		63.6	102.6	0.195	3.40
Mean	180.97	73.71	100.92	0.187	2.70
Range	172.7- 192.0	60.0- 85.7	95.7- 104.8	0.113-0.358	1.52-4.31
Standard deviation				± 0.06	± 0.913

Table 1. Antero-posterior sway in 10 subjects, measured as a mean deviation and total range of movement of the centre of gravity.

not a very close one. The subject with the smallest amount of sway as defined by the total range of movement of the centre of gravity (subject M.C.B.) does not have the smallest amount of sway as defined by the mean deviation of the sway tracing from a baseline (subject A.J.D.B.). When comparing the postural sway of subjects, it would seem more reasonable to use a result which derives from the total amount of swaying that is occurring over the period of measurement (mean deviation from a fitted baseline), than to use the extreme range of movement recorded, which may only relate to a small part of the period of swaying that is being measured.

The amount of lateral sway that was observed is small (Fig. 5). The total range of the lateral sway, as measured over a period of two and a half minutes, was never found to exceed one centimetre movement of the centre of gravity, and it only occasionally exceeded a half a centimetre of movement. By far the greatest amount of sway took place in the antero-posterior plane, and much less occurred from side-to-side.

The observed values for the mean deviation of the inclinometer from the vertical are given in radians in Table 2. In Fig. 4, the distance d cms. represents the movement of the centre of gravity of the subject, expressed as the mean deviation from a baseline. If the segment of the body between the ankle joint and the centre of gravity of the body moved in a rigid manner, then the line OG' in Fig. 4 would be a straight line. In that event, the angle θ subtended by the inclinometer with the vertical would be equal to the angle COG' , and could be calculated by dividing the mean deviation of the centre of

TABLE 2

Table 2. Mean angular deviation from the vertical of the inclinometer, as calculated from the mean deviation of the centre of gravity, and as measured from the inclinometer tracing. Results are given in radians.

Subject	ESTIMATED VALUE (mean deviation of C.of G. over height of C.of G.)	OBSERVED VALUE (measured by planimetry from accel. tracing.)	DIFFERENCE (observed minus estimated)	PERCENTAGE DIFFERENCE (difference as % of est. value.)
M.C.D.	0.00144 0.00123	0.00256 0.00190	0.00112 0.00067	77.6 54.5
J.C.N.H.	0.00181 0.00140	0.00197 0.00194	0.00016 0.00054	8.8 38.5
A.S.D.B.	0.00118 0.00139	0.00256 0.00221	0.00140 0.00082	118.6 58.9
D.F.J.A.	0.00154 0.00358	0.00357 0.00310	0.00203 -0.00048	131.8 -13.4
R.J.W.	0.00155 0.00126	0.00281 0.00172	0.00127 0.00046	81.9 36.5
J.H.O.	0.00305 0.00186	0.00416 0.00364	0.00111 0.00175	36.4 94.1
D.E.N.	0.00124 0.00191	0.00230 0.00299	0.00111 0.00108	72.6 57.5
K.J.E.W.	0.00216 0.00304	0.00138	0.00172	54.7
T.G.T.	0.00199 0.00185	0.00301	0.00102	51.3
C.J.F.B.	0.00224 0.00190	0.00503 0.00334	0.00162 0.00154	73.3 61.1
Mean	0.00192	0.00292	0.00105	52.6
Range	0.00118-0.00358	0.00172-0.00438	0.00016-203	-13.4-131.8
Standard deviation	±0.000673	±0.00101		

gravity (d cms.) by the height of the centre of gravity of the subject (h cms.) This estimated value for θ has been calculated in radians, for each subject, and is given in Table 2. If the line OG' is straight, the actual position of the inclinometer does not affect the result, and the fact that the inclinometer is some centimetres (estimated to be approximately 10 cms. on the average) above the level of the centre of gravity is irrelevant in this connection.

The observed results for the mean angular deviation of the inclinometer from the vertical are considerably higher than the calculated values. The mean value for the calculated deviation (θ) was 0.00192 rad., as opposed to an observed value (θ^o) of 0.00292 rad. (1 radian = 57.3 degrees). The average for the ten subjects of the percentage differences by which the observed angular deviation exceeded the calculated deviation was 62.6%. The most likely explanation for this discrepancy between observed and calculated results is that the segment of the body between the ankle and the centre of gravity does not move rigidly, at least in the majority of subjects. Something analogous to the position illustrated diagrammatically in Fig. 4 must occur, where the line $OG'A'$ represents what is believed to happen. In other words, the body sways more like a pliant rather than a rigid rod, and the line OG' is a curved line rather than a straight one. This occurrence would explain why the observed angular deviation of the inclinometer (θ^o) is considerably larger on average than the estimated angular deviation (θ).

By measuring the position of the centre of gravity of the supine subject, and the anterior displacement of the hip marker which

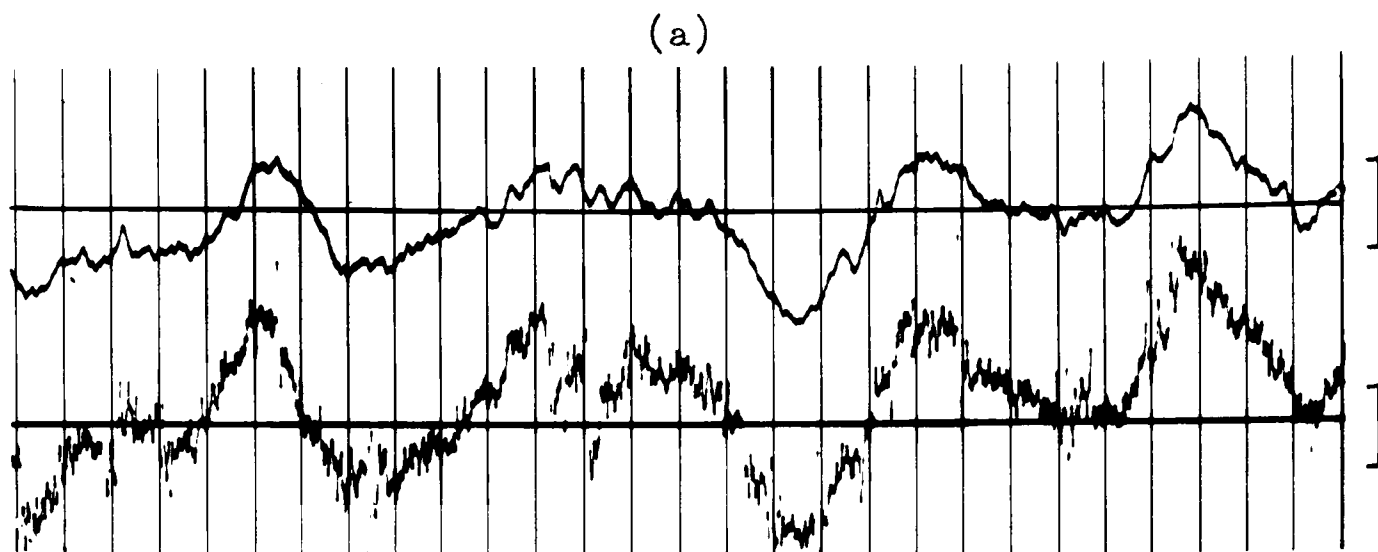


FIGURE 2

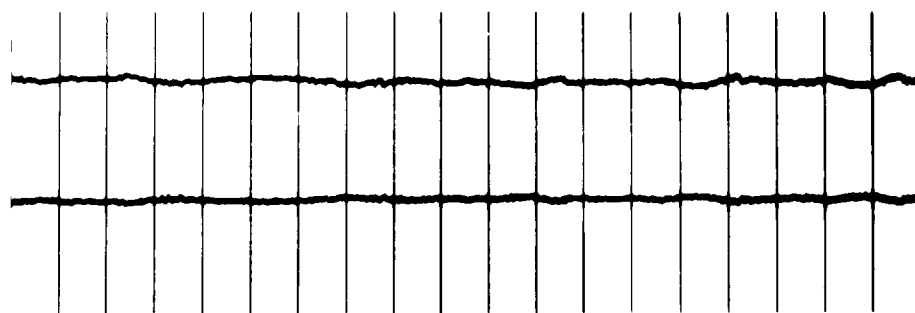


Fig. 2a. Record of antero-posterior movement of centre of gravity (upper tracing) and of output of an accelerometer mounted on waistbelt (lower tracing). Note that the gross movement of the accelerometer tracing reflects the movement of the centre of gravity, due to the accelerometer acting as an inclinometer. The fine tremor seen on the accelerometer tracing results from the antero-posterior acceleration of the body.

Vertical lines at 1 sec. intervals. Horizontal lines are baselines fitted for whole of the $2\frac{1}{2}$ min. record, and not only for the section illustrated (approx. $\frac{1}{2}$ min.). Calibration - upper tracing = 1 cm. movement of C. of G.; lower tracing = 0.01 radian, for gross movement of tracing.

Fig. 2b. Calibration record, showing output of platform (upper) before subject stands on it, and accelerometer output (lower) before attached to subject. Sensitivity unchanged between Figs. 2a and 2b.

occurred when the subject leaned forwards, it is possible to estimate the position of the centre of gravity in the subject leaning forwards. In Fig. 3 the line OAB represents the normal standing position, with B representing the position of the centre of gravity, A the position of the hip marker, and O the ankle joint. For the purpose of this calculation, all three points are considered to lie along a straight line, and to continue to lie along a straight line when the subject leans forward (line OCD). A'C' represents the horizontal displacement forwards of the centre of gravity. It follows, therefore, that B'D' can be deduced from the formula $-\frac{A'C' \times BB'}{AA'}$. The estimated horizontal displacements of the centre of gravity (B'D') were compared with the values recorded by the platform, for the ten subjects. The average deviation of the platform readings, irrespective of the sign of the readings, from the results estimated from the photographs, was 1.04 cms. (S.E. 0.192 cms.), which is significantly greater than zero at the 0.1% level. However, if the signs of the differences are taken into account, the difference between the two readings was only 0.04 cms. (S.E. 0.00511 cms.). This figure does not differ significantly from zero, suggesting that on the average the platform gives an accurate representation of the horizontal displacement of the centre of gravity, when the subjects lean forwards. This agreement can be regarded as very satisfactory in view of the unavoidable errors in calculating the comparison between the two sets of results.

It has been assumed that the lines OAB and OCD are straight lines, and as has been shown, this can only be considered an approximation. For the purpose of this calculation, the error from assuming

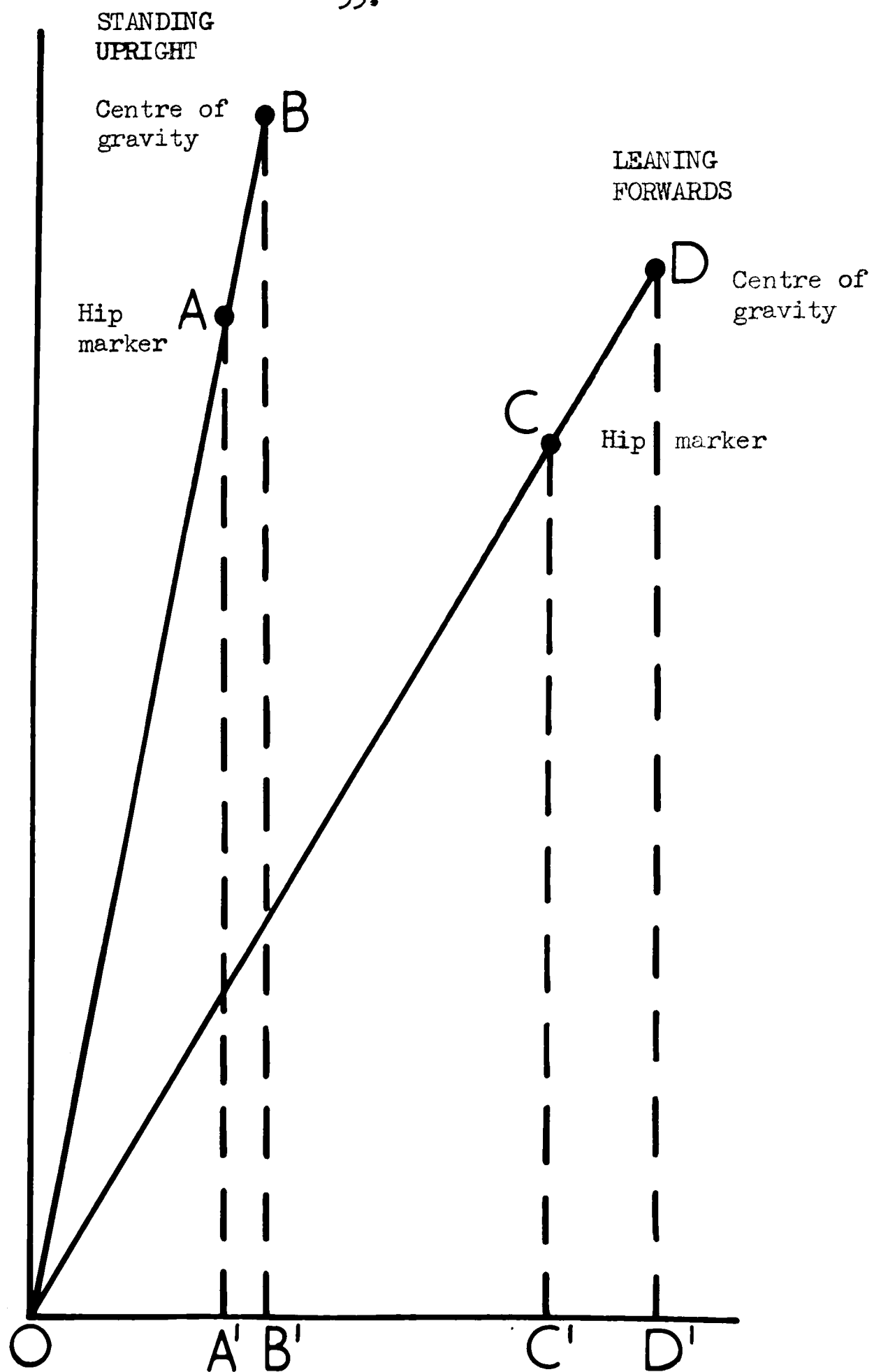


Fig. 3. Diagrammatic representation of the displacement of the centre of gravity when leaning forwards. Line OAB represents the normal standing position, with B as the centre of gravity and O as the ankle joint axis. Line OCD illustrates position of body on leaning forwards, with D representing the new position of the C. of G. For simplicity, body is drawn as a straight line (see text).

that the body moves rigidly is likely to be very small. Another potential source of error is that the photographic picture represents the position of the hip markers at only one instant, and cannot allow for the continuous movement caused by the normal postural sway. Such differences that are found between the readings obtained from the platform and the centre of gravity positions estimated by photography are probably due chiefly to these two sources of error. However, the results indicate that there is good evidence for believing that the centre of pressure tracing accurately represents the movement of the subject's centre of gravity.

DISCUSSION

An attempt has been made in the present study to express the normal postural sway as a movement of the centre of gravity of the body in the antero-posterior plane. The technique of presenting postural sway as a mean deviation from a fitted baseline, over a period of time, enables the degree of swaying to be measured with considerable accuracy. It is believed that this technique of measurement gives a more valid assessment of postural sway than is obtained by measuring the extreme limits of the sway. When the sway patterns of the ten subjects are compared in order of magnitude of the amount of sway, there is a general agreement between the results as measured by the two methods (mean deviation and total range), but the agreement is by no means complete. For example, subject C.J.T.B. rates as the second largest "swayer" measured by the total range of the sway record, but this subject drops to fourth place if his sway is measured as a mean

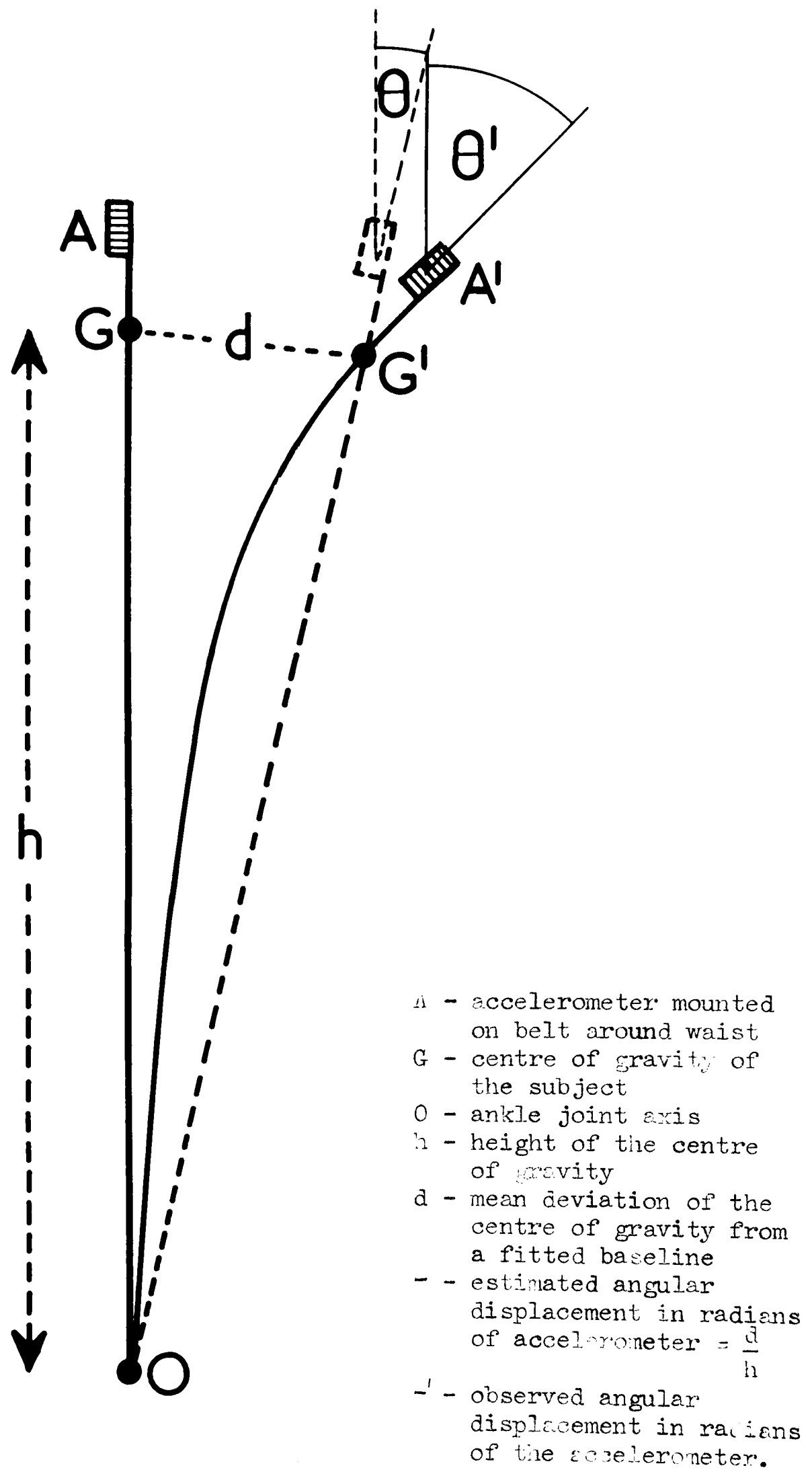


Fig. 4. Schematic illustration of the swaying of the body between the ankle joint axis and the centre of gravity of the body.

deviation. Similarly, subject R.J.H. rates as fifth in the order of magnitude of sway, measured as a total range, whereas if his sway pattern is recorded as a mean deviation he rises to third place. The advantage of expressing the sway as a mean deviation is that it takes account of the degree of swaying throughout the period of measurement, whereas expressing the sway as the extreme limits of the tracing, as Orma (1957) has done, may well fail to take into account the general level of swaying. A subject may make a sudden large movement, and then move very little for the rest of the period of measurement; only the large movement would be represented in the result if the sway was measured solely by determining the limits of sway.

The postural stance adopted during the measurement of postural sway is of considerable importance, particularly in relation to the degree of lateral sway. Very little lateral sway was observed in the present experiments, in which the subjects stood with their heels twenty centimetres apart. This agrees with the findings of Miles (1922), who found the position of maximum stability (in terms of postural sway) to exist when the feet were twenty centimetres apart. Orma (1957) studied his subjects when they were standing barefoot, with their feet together, and he found that lateral sway was almost as great as antero-posterior sway. In forty-five men, he found the antero-posterior sway to be 3.85 cms. (range 2.0 - 8.6 cms.) and the transverse sway to be 3.24 cms. (range 1.3 - 8.0 cms.) These are considerably higher figures than were found in the present study, where the antero-posterior sway, expressed as a total range of movement, was found to be 2.78 cms. (range 1.62 - 4.83 cms.), and the lateral sway

FIGURE 5

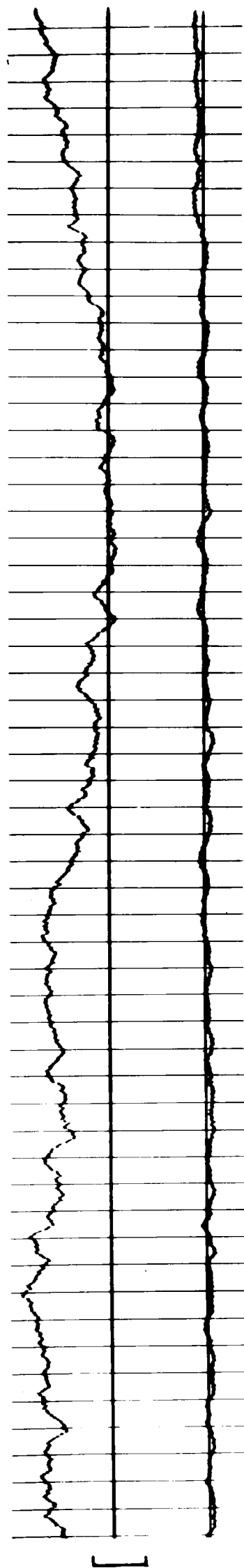


Fig. 5. Record showing amount of antero-posterior movement (upper tracing) and lateral movement (lower tracing) of the centre of gravity during normal standing. Note that during approximately 1 min. standing, there is little lateral movement of the centre of gravity.

Vertical lines represent 1 sec. intervals. Calibration mark equals 1 cm. movement of the centre of gravity. The two straight horizontal lines are made by unused galvanometers.

never exceeded 1 cm., and was usually much less. Orma (1957) was measuring sway at the head, which probably accounts for his higher figures for antero-posterior sway, but the relatively large amount of lateral sway he observed was almost certainly a function of his positioning of the feet. Most people are unaccustomed to standing without shoes, and do not generally stand with their feet together. It is readily understandable that under such circumstances the lateral sway should be increased, but it is a somewhat artificial situation, giving results that probably do not accurately reflect the normal postural sway. The need for carefully defining the position of the feet when comparing results of measuring postural sway is apparent.

Many measurements of the postural sway have been made using the movement of the head as the measure of sway (Miles, 1922; Goldberg, 1943). This suffers from the disadvantage that the sway of the body as a whole is not necessarily reflected accurately in the sway of the head. It seems more logical to measure postural sway as a movement of the centre of gravity of the body, rather than by using the movement of one part of the body. The method of using the centre of gravity oscillations as a measure of sway was introduced by Hellebrandt (1938). Hellebrandt, however, expressed her results as the area over which the centre of gravity oscillates in relation to the total base of support. This is expressed as a percentage of the area of support, which consists of the area bounded by the two feet, excluding the toes (Hellebrandt & Braun, 1939). The present study is believed to be the first detailed planimetric analysis made of the antero-posterior movement of the centre of gravity, and has the advantage in that it

does not introduce a further variable into the measurements - namely, the size of the subject's feet, which affects the area of the base of support.

Evidence has been presented to suggest that the tracing obtained from the movement of the centre of pressure of the feet parallels very closely the movement of the centre of gravity of the body. From both theoretical considerations of the characteristics of the force analysis platform (see appendix 1), and by direct photographic observation, it is concluded that the centre of pressure tracing gives an accurate representation of the movement of the subject's centre of gravity.

It has frequently been tacitly assumed that during the normal postural sway, the body sways "in toto" over the ankle joints, remaining rigid during the swaying (for example, see Smith, 1957). No experimental justification exists for this assumption, and in the present experiments it was found that the waist moved much further than would have been expected if the body acted as a rigid rod between the ankles and the waist. The angular deviation of the inclinometer from the vertical at the waist was on the average 62.6°, greater than than it would have been if the body segment linking the ankles and the waist swayed as a rigid column. The most likely explanation of this finding is that the part linking the waist and the ankle resembles a pliant column, and therefore the linkage is not rigid (Fig. 4). If this occurs for the segment linking the waist and the ankles, it seems likely that a similar process occurs between the waist and the head. If it is concluded that the whole body during postural sway moves in

a fashion analgous to a reed bending with the wind, swaying backwards and forwards in a gently undulating manner, it is possible to reconcile the large differences found between measuring the sway at the head and at the centre of gravity. If the body does sway in a pliant manner, it would be expected that the head would move relatively much farther than the centre of the body, and this agrees with what has been found experimentally.

Edwards (1942), summarizing previous work, noted that the degree of postural sway was influenced, *inter alia*, by opening and closing the eyes, the age and sex of the subjects, the extent and nature of the visual fields, and the state of attention of the subjects. The great majority of the work he quoted had been performed by measuring the sway at the head, and if this done, postural sway appears as a highly variable and easily influenced phenomenon. However, if the postural sway is examined as a continuous movement of the centre of gravity, and particularly as a mean deviation of the tracing from a fitted baseline, the relative constancy of the amount of body sway is more noteworthy. In twenty readings on ten subjects, the difference between the minimum and maximum readings obtained for the amount of sway (expressed as a mean deviation) was less than 3.5 millimetres. In one hundred and nine subjects, Hellebrandt and Braun (1939) found the area enclosed by the maximal shifts of the centre of gravity in the cardinal vertical orientation planes to be only 1.28% on the average, of the total base of support formed by the feet. There is, of course, no doubt that the sway can be much increased if the subjects' stability is reduced by making them stand unshod with their feet close

together, as Orma (1957) has done. As has been suggested, however, this is regarded as an artificial situation.

The results of the present study would support the view of Mellesbrunet (1950) that postural sway, if it be defined as the movement of the centre of gravity of the body, is relatively small in amount and constant in extent. The considerable attention that has been devoted, chiefly by psychologists, to the way in which postural sway, as measured at the head, can be influenced by numerous factors, has tended to obscure the basic fact that the position of the centre of gravity of the body is remarkably stable during normal standing. The oscillation of the centre of gravity that does occur should be regarded essentially as a constant response to the effects of gravity, resulting from the interplay of the flexing effects of gravity, and the ever-present restraining contractions of the extensor postural muscles.

SUMMARY

1. A study has been made of the postural sway occurring during normal standing in ten subjects. The procedure adopted involved measuring the continuous oscillations of the centre of gravity of the body. It is considered that measuring the movement of the centre of gravity is the most logical way to measure the postural sway that occurs during standing in the erect posture.
2. It was found that by far the greatest amount of sway occurred in an antero-posterior direction; with the foot position adopted (heels 20 cms. apart), very little lateral sway was observed.
3. An inclinometer was used mounted on a waistbelt, and an estimate was made of the angular displacement occurring at the waist during body sway. The angular displacement measured considerably exceeded the angular displacement estimated on the assumption that the body swayed rigidly between the ankle joint axis and the centre of gravity. It was concluded from this observation that, contrary to what has been commonly assumed, the body sways in a pliant manner over the ankle joint axis, and not as a rigid column.
4. If the postural sway is measured by detailed planimetry of the records, and the results expressed as a mean deviation of the movement of the centre of gravity from a fitted baseline, the relative constancy of the degree of body sway is noteworthy. In the subjects examined, the difference between the minimum and maximum antero-posterior sway recorded (expressed as a mean deviation of the movement of the centre of gravity) was less than 3.5 mm. It is concluded that

the body is remarkably successful in resisting the effects of gravity, and in maintaining the normal static equilibrium associated with the erect posture.

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Paper 3THE EFFECT OF LOAD CARRIAGE ON NORMAL STANDING IN MAN

INTRODUCTION

It is a commonplace observation that carrying a heavy load on the back causes the carrier to lean forwards, and that, in general, the heavier the load, the farther forward does the carrier lean. The addition of a load to the back causes the normal erect posture to be so modified that the vertical projection at the feet of the centre of gravity of the man plus the load is made to approximate in position to the centre of gravity of the unloaded man (Hellebrandt, Fries, Larsen & Kelso, 1944). Indeed, it has been suggested that the location of the centre of gravity is a "physiological constant" which it is impossible to disturb significantly in a standing man (Hellebrandt, 1950).

The manner in which the body alters its normal alignment to achieve stability under conditions of load carriage does not appear to have been examined in any detail. The present study is concerned with measuring the effect of load carriage and load position on the normal standing posture of young adult males, using a miniature camera photographic technique. The experiment was designed to find out in what way varying loads affect normal standing, and whether the same load has comparable effects on different subjects. The study was carried out on a relatively small group of subjects. The results were

sufficiently clear-cut, however, to suggest that the findings have general application for conditions of static loading.

METHODS

The experiments were performed on ten male medical students, aged 18 to 21 years. Their heights ranged from 170.6 cms. to 188.9 cms. and their weights from 59.75 kgs. to 84.75 kgs. Four bony landmarks, which were readily identifiable, were found and marked on all subjects. The four landmarks used were, 1. the lateral side of the greater tuberosity of the right humerus ("shoulder"), 2. the lateral side of the greater trochanter of the right femur ("hip"), 3. the knee joint immediately above the head of the right fibula ("knee"), and 4. the lateral malleolus of the right fibula ("ankle"). Small black adhesive markers were placed as accurately as possible on the landmarks. The subjects were then photographed from their right side against a white screen (Pl. 1).

The subjects stood on a rigidly constructed platform, with the contour of their heels placed posteriorly along a line marked on the platform. The heels of the subjects were 20 cms. apart, but they were allowed to spread their feet at will. The subjects were asked to adopt a comfortable standing posture and not to move the position of their feet. They were placed facing at right angles to the camera, and behind them in line with their right shoulder hung a vertical plumbline (Pl. 1). Alongside the plumbline was placed a metre calibration rule, which appeared in every picture taken. The camera was placed three and three quarter metres distant from the subject,

and exposures were made using an electronic flash, with a flash duration of approximately $1/2000$ th. of a second. The camera used was a Leica IIIIf, with a Summicron 5 cm. lens.

Each subject was photographed under the following conditions:-

1. normal standing, without load; 2. carrying an empty carrier; 3. carrying twelve kilograms attached to the carrier high on the back; 4. carrying twenty four kilograms as in 3.; and 5. carrying twenty four kilograms attached to the carrier low on the back. The load of 24 kg. was moved over a distance of thirty centimetres between the "high" and "low" positions (Pl. 1, Figs. 3 & 4). The carrier used for the experiments was adapted from a standard Bergen rucksack frame. Adjustable shoulder straps and waist belt were fitted, and on the back of the frame two vertical steel bars were welded, over which a platform carrying the weights could be moved. The weight of the carrier itself was four and a quarter kilograms, and additional weights of two, four and six kilograms were used. The weight of the load was concentrated into a relatively small mass, and the approximate centre of gravity of the load, in relation to the anatomical markers, is shown diagrammatically in Fig. 2. The shoulder straps of the carrier were adjusted individually for each subject, so that the waist-band supporting the load at the back fitted the natural waist of the subject.

For the experiments the subjects wore only trunks and vest. Subjects were photographed twice, but no subject came twice on the same day. It is considered that the heel positions on the platform did not vary over more than half a centimetre, from subject to subject. Subjects were photographed about half a minute after they had been

placed in position, for each of the successive load positions. The exact timing of the photographs varied slightly, and the release mechanism was operated surreptitiously. This, combined with the fact that the subject was not facing the camera, mitigated against the possibility of posing on the part of the subject.

The same procedure was then adopted with the subject carrying first twelve kilograms and then twenty four kilograms, without moving the position of his feet. The carrier was loaded from behind, and the subject did not have to move in any way. The weight-carrying platform on the carrier was used in only two positions - its upper and lower limits, which were thirty centimetres apart ("high" and "Low" positions). The centre of gravity of the load in the high position approximated to the level of the shoulder joint; in the low position, it approximated in position to the lower lumbar region of the subject (Plate 1, Figs. 3 and 4).

The film negatives were enlarged using a Leitz microfilm reader, and the size of the enlarged image was about one tenth of the actual size of the subject. Direct measurements were made on the enlarged image, which was projected on to a perfectly flat white screen. A ruled grid was used for measuring, one edge being lined up along the plumbline, and the other edge at right angles to the plumbline. The vertical height above the platform (using a line marking the position of the right foot on the platform as the baseline), and the horizontal distance from the plumbline could thus be measured for each of the four markers. The length of the metre rule was also measured on each negative, and this figure was used as the calibration factor. The

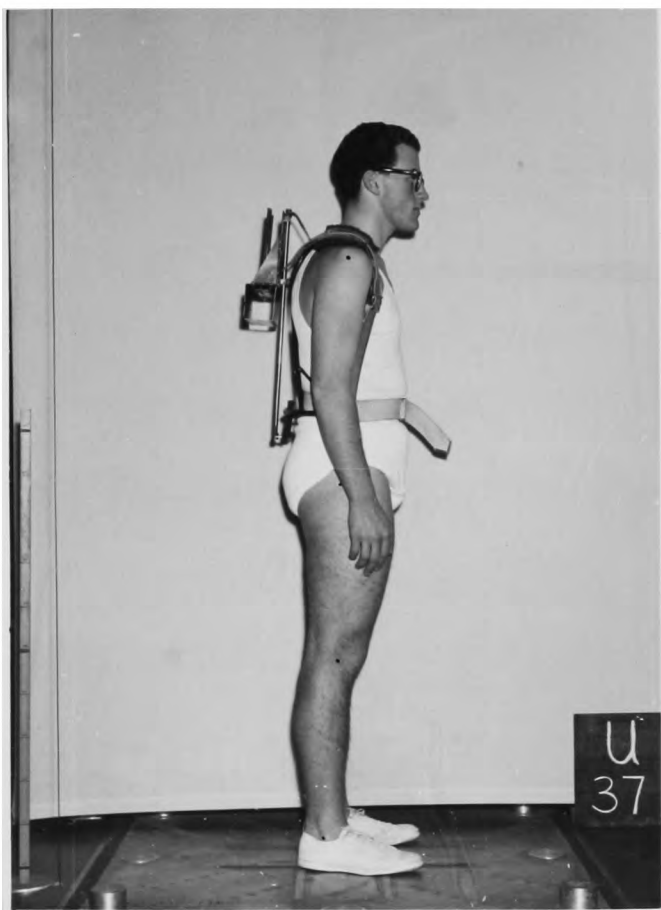


Fig. 1



Fig. 2

Fig. 3

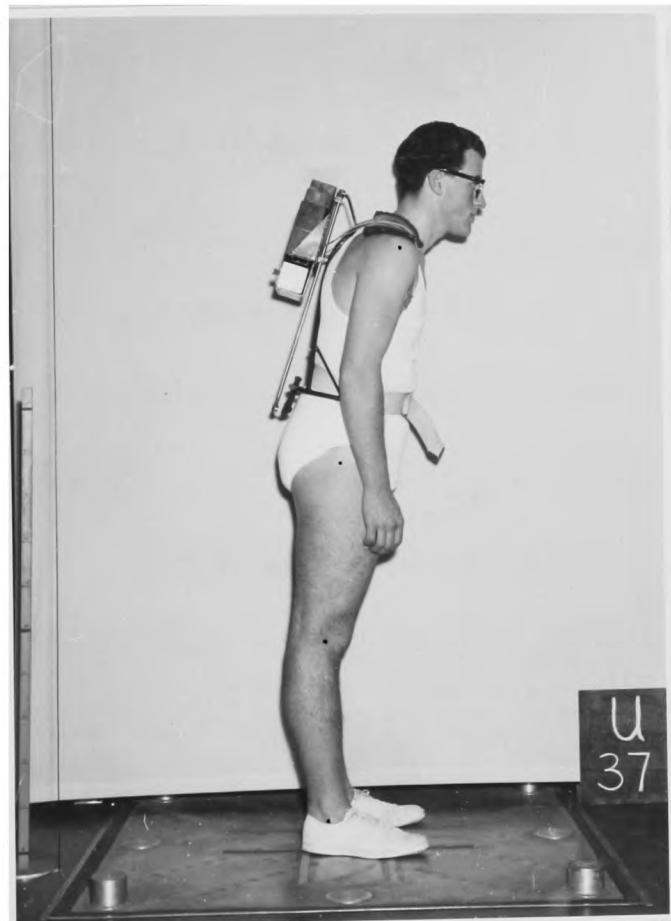


Fig. 4

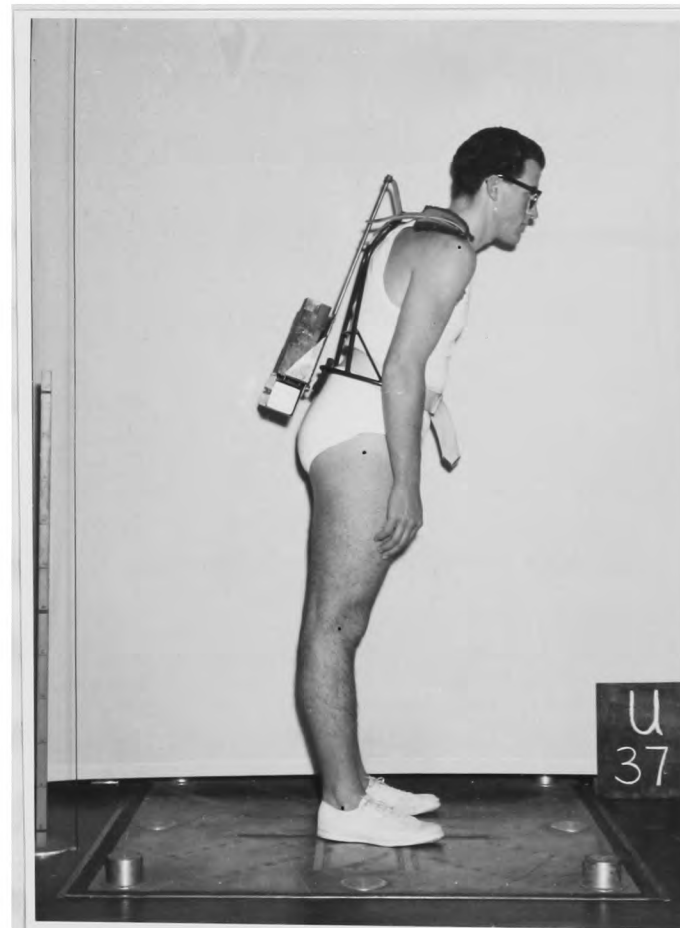
PLATE 1

Fig. 1. Carrying an empty carrier. Plumbline and metre rule are on left of picture. Fig. 2. Carrying 12 kg. (high). Fig. 3. Carrying 24 kg. (high). Fig. 4. Carrying 24 kg. (low).

position of the markers in relation to the two reference lines on the negative (plumbline and platform) was measured to an accuracy of half a millimetre on the enlarged image.

There are several potential sources of error in this technique, but the close similarity of the results for the two experiments (Table 1) suggests that these errors have been largely avoided, or do not significantly affect the results. The errors relating to the photographic distortion in measuring from thirty five millimetre negatives have been discussed by Tanner and Weiner (1949), who found distortions due to camera lens error, and differential distortion of the negative image during development to be immeasurably small. The chief error is likely to arise during printing of the negative, and this was avoided by measuring from the projected negative image. The error due to parallax was minimised by measuring the four markers from a plumbline in the same plane as the markers. As the metre rule appeared in all the pictures taken, all measurements were related to a calibration factor in the same sagittal plane as the markers.

Probably the main source of subject error arises from the fact that the subject never remains completely still. It is well recognised that the body is constantly swaying in an antero-posterior direction (Hellebrandt and Franseen, 1943), and it is theoretically possible that the pictures taken of the same subject under identical load conditions on different days may have been taken at the extreme limits of the forward and backward sway, respectively. The fact that the results show precisely the same trend for the second occasion on

which the subjects were photographed as they do for the first occasion, and that the actual figures are in general very similar, suggest that the difference due to the subjects having been photographically "frozen" in motion during varying parts of the postural sway cycle are small in relation to the effect of load carriage.

The analysis of the results was performed by comparing the positions of the four anatomical markers on each subject under varying load conditions. This was done for each subject on two occasions, so that there was a total of forty individual readings for each subject. Analysis of the data showed that there was no significant difference in the position of the ankle under the varying load conditions. The ankle marker was therefore taken as the zero position, and the positions of the other three markers were estimated in relation to a vertical line through the ankle marker. The results were compared by an analysis of variance, and the levels of significance determined by using a t-test.

RESULTS

The results showed a highly significant ($P < 0.001$) forward displacement of the shoulder with loading, and the same load carried in the low position caused a further displacement forward than when carried in the high position. The unloaded carrier, however, made no significant difference to the position of any of the markers. The position of the hip marker did not move significantly for the varying load conditions, but the position of the knee marker moved progressively backwards with increasing load carriage ($P < 0.001$). There is therefore

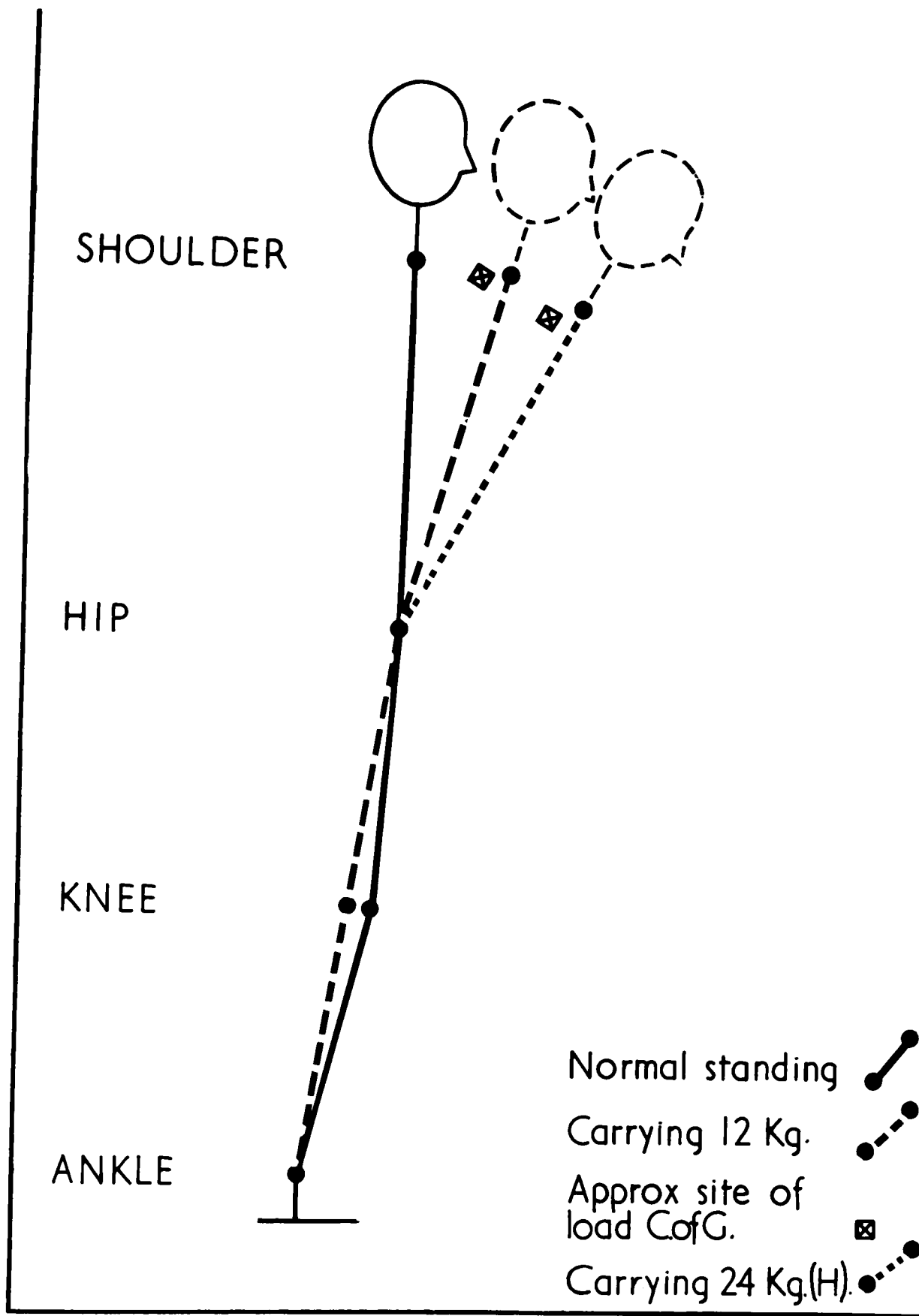


FIGURE 1

Fig. 1. Diagram showing disposition of body linkage under different loads.

a highly significant realignment of the body segments, as represented by the anatomical landmarks used in these experiments, with the carriage of loads above a certain minimum weight in the standing position.

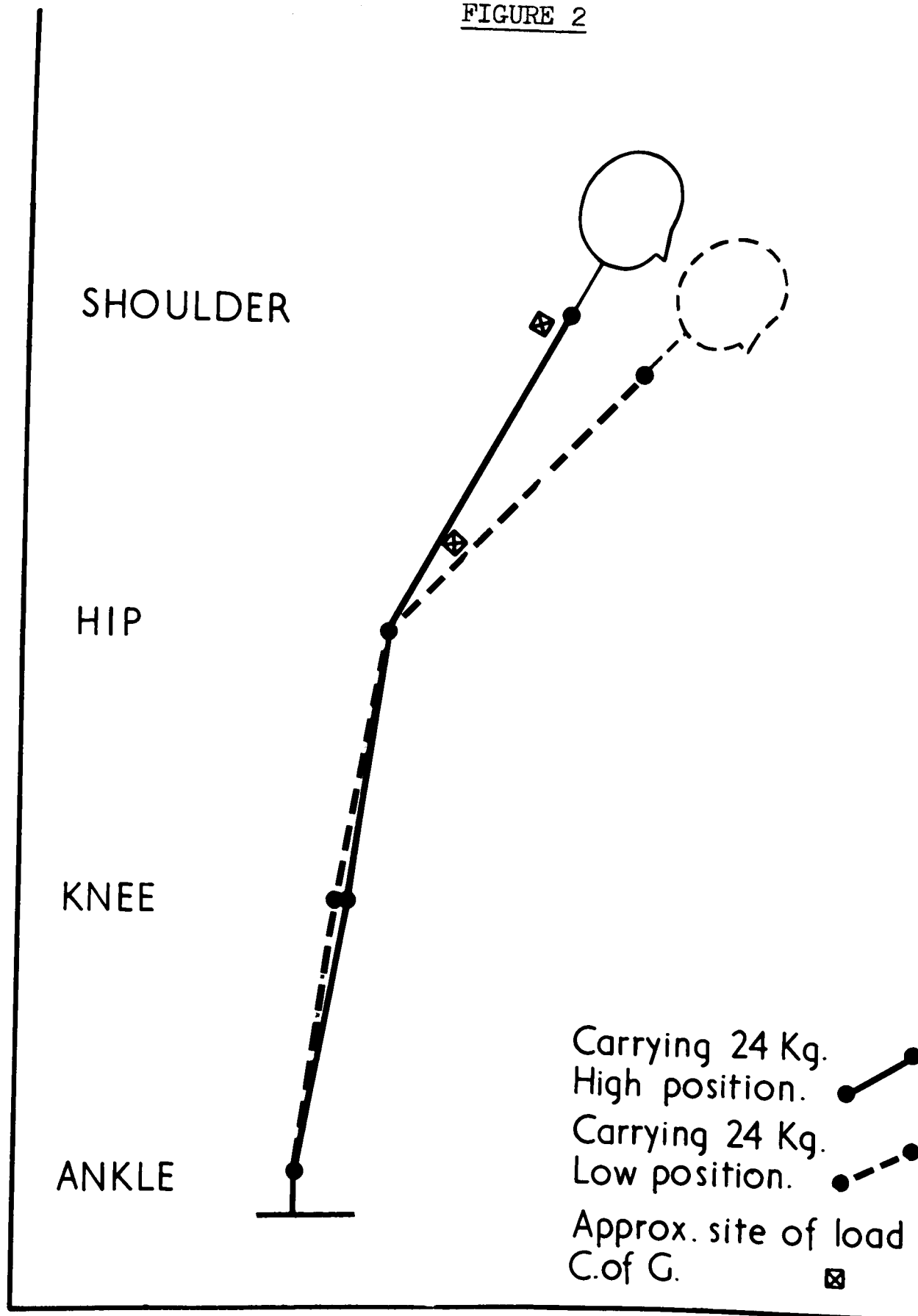
Figs. 1 and 2 illustrate diagrammatically the realignment that occurs.

The greatest amount of movement took place at the shoulder, as would have been expected. There were highly significant differences in position between normal standing and carrying the carrier plus twelve kilograms, between carrying twelve kilograms and twenty four kilograms, and between carrying twenty four kilograms in the low as compared to the high position ($P < 0.001$ in all cases). Thus the heavier the load, provided at least that it exceeded the weight of the carrier (4.25 kg.), the further forward did the subject lean, and for the same weight, the lower position on the back caused the shoulder to be displaced further forwards.

The position of the shoulder was also compared using the angular displacement from a vertical line through the hip marker, instead of the direct horizontal displacement of the shoulder with loading. This was done to see if the height of the subjects was a significant ^{factor} affecting the displacements. If the height of the subject was important in this respect, the effect would be most readily observable for the position of the shoulder. It was found, however, that the levels of significance between different effects of the varying load conditions were virtually the same as those obtained using the direct horizontal displacement.

The results for the position of the hip marker showed that the position of the marker did not significantly vary for any load

FIGURE 2



condition investigated. While the shoulder is moving forward with increasing load, the position of the hip remains relatively fixed. There is no statistically significant difference even between the position of the hip marker when the subject is carrying twenty four kilograms, and the position during normal standing without load. The position of the hip marker was also compared using the angular displacement from a vertical line through the ankle marker, instead of the direct horizontal displacement. As in the case of the shoulder, it was found that the angular displacement of the hip from the ankle gave results comparable to those obtained by direct measurement of the horizontal displacement. Measured by direct horizontal displacement, or by angular displacement from a vertical through the ankle, the position of the hip marker did not significantly vary with the varying load conditions.

The results for the knee marker showed clearly that there was an overall tendency for the knee position to move backwards with progressive loading ($P < 0.001$). Here, too, there was no significant difference between the subject standing normally without the carrier, and the subject with the unloaded carrier. But highly significant differences were found between the position of the knee in the subject standing normally (without carrier) and the position of the knee when the subject was carrying twelve kilograms, and when the subject was carrying twenty four kilograms (both positions). There was no significant difference in knee position between carrying twenty four kilograms in either the high or low position. In Table 1, the position of the ankle marker is considered as the zero position, and the

Table 1

Average displacement in cms. of shoulder, hip and knee markers vertically in front of the ankle marker. Data from ten subjects in two experiments.

	Normal	Empty carrier	12 kg. (high)	24 kg. (high)	24 kg. (low)	
<u>EXPERIMENT 1</u>						
Shoulder	6.77	8.30	13.41	19.14	24.08	
Hip	6.42	6.53	6.37	6.34	6.85	
Knee	4.94	4.58	3.72	3.33	3.46	
Ankle	0.00	0.00	0.00	0.00	0.00	
<u>EXPERIMENT 2</u>						
Shoulder	7.48	8.05	13.86	18.56	24.69	
Hip	6.51	6.42	5.48	5.33	6.26	
Knee	4.92	4.12	2.87	2.67	2.47	
Ankle	0.00	0.00	0.00	0.00	0.00	
<u>AVERAGE</u>						
Shoulder	7.12	8.18	13.63	18.85	24.39	<u>Sig.</u> P < 0.001
Hip	6.46	6.48	5.92	5.83	6.51	N.S.
Knee	4.93	4.35	3.30	3.00	2.97	P < 0.001
Ankle	0.00	0.00	0.00	0.00	0.00	N.S.

position of the other three markers is given as the distance they lie in front of the ankle in the vertical plane. The results represent the average displacement in centimetres for the ten subjects of the markers under the varying load conditions.

It was also found that the angle between the shoulder-hip line and the vertical, and the angle between the ankle-hip line and the vertical, showed no significant correlation with either the weight or the height of the subjects during load carriage. Therefore there was no evidence to suggest that the posture of the subjects when loaded was significantly related to their height or weight, at least under the conditions of these experiments.

Analysis of the data showed a significant difference between the average results for the position of the hip and knee for the two experiments. The average values for the position of the markers with each load are consistently lower for the second occasion on which the subjects were photographed. However, with the shoulder position there is no significant overall difference, four of the subjects showing a decrease in displacement in the second experiment, whereas six subjects showed an increase. These different effects are not significantly different, and so represent random fluctuations about the mean displacement. When individual subjects were compared, there was found to be a significant difference between the average displacements for the shoulder, hip and knee for the ten subjects, at the 0.1% level. There was also a significant interaction between loads and subjects, implying that the effects of the load differ significantly from subject to subject. However, with the shoulder

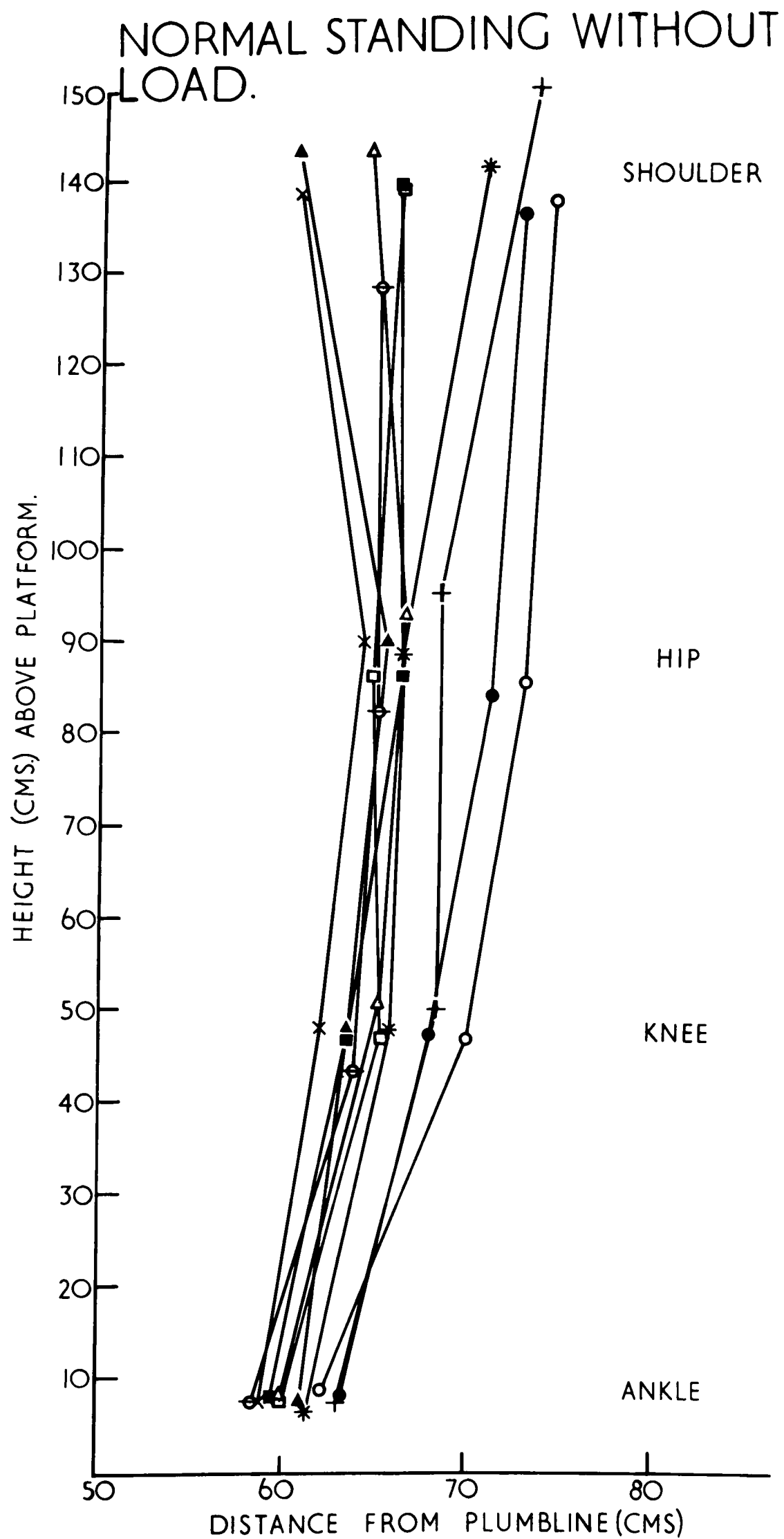


Fig. 3. Plot of positions of 4 landmarks on 10 subjects during normal

the trend is the same for each subject (displacement increasing with load), except in three cases in which the displacement with the carrier is less than normal. For the hip, there is no consistent trend, but there is also no significant difference between the loads.

DISCUSSION

The effect of load carriage on the normal standing posture has previously been investigated chiefly by Hellebrandt et al. (1944), and Hale et al. (1953). Hellebrandt et al. claimed that the body counterbalances the effect of load carriage by leaning forwards "in toto" over the ankle joints, without significant realignment of the rest of the body. They gave few details of the technique whereby they reached these conclusions, however, and it is difficult to evaluate their results. Hale et al. (1953), found that the carriage of a load low on the back tended to cause the body to lean further forward than a load carried high on the back. Their results were not statistically significant, however, and they did not comment upon the effect of load carriage on the alignment of the body segments.

It is apparent that the present data contradict the findings of Hellebrandt et al. (1944), and extend the findings of Hale et al. (1953). The results show that the body responds to the effects of load carriage on the back by altering the relative position of the body segments. The essential change takes place above the level of the hips, the trunk being displaced progressively forwards with increasing load. The plots illustrated in Figs. 3, 4 and 5 demonstrate the changes in marker positions for three of the five conditions

CARRIER PLUS 12 Kg. (HIGH POSITION).

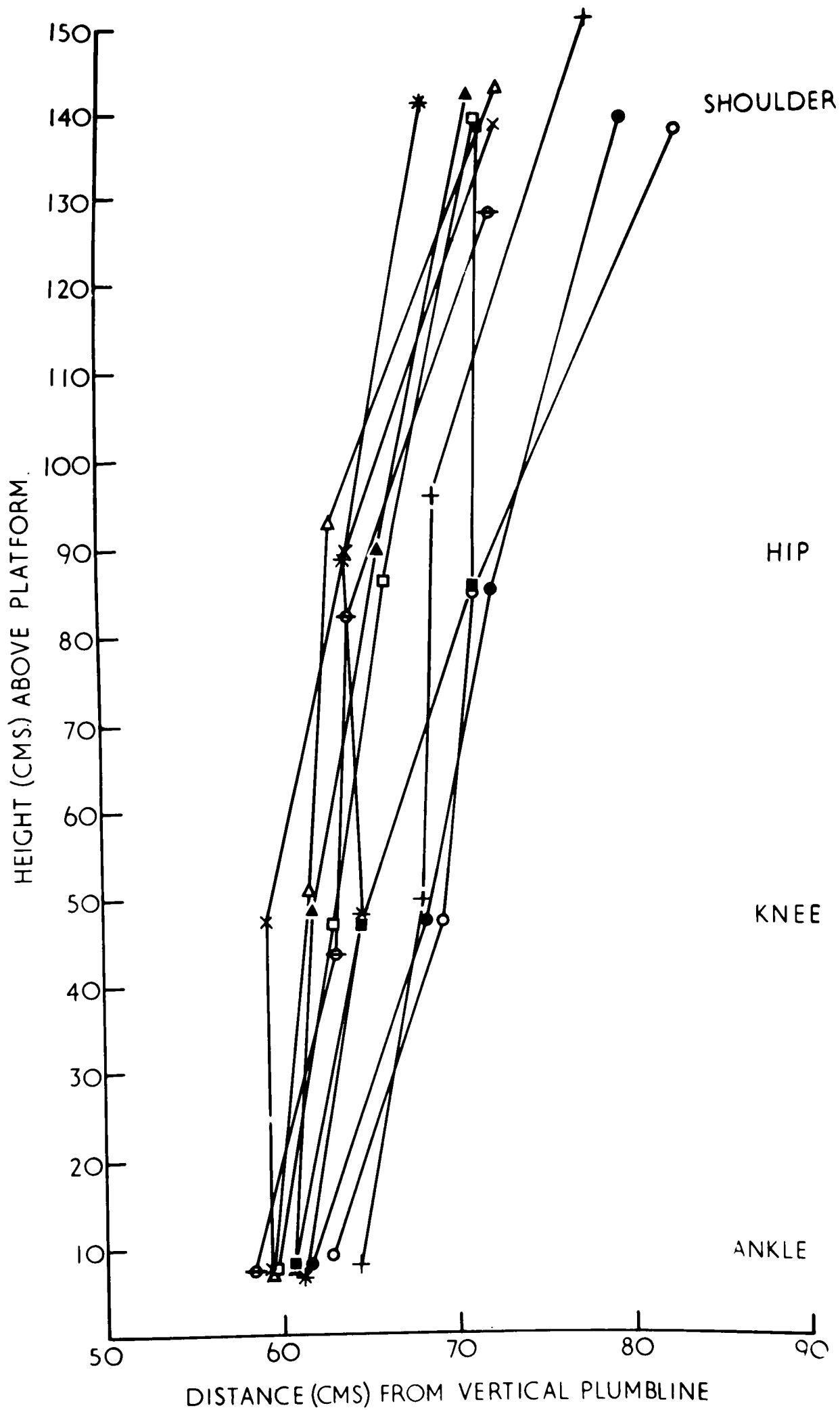


Fig. 4. Plot of positions of 4 landmarks on 10 subjects carrying 12 kg.

investigated. The plots were constructed from the results of the first experiment, and were drawn accurately to scale. They demonstrate clearly that the displacement caused by load carriage occurs between the hip and shoulder markers, and that for comparable loads the general pattern is very similar for all ten subjects. One subjects in Fig. 4 did not show any forward displacement of the shoulder with a 12 kg. load, as compared to the normal position. However, it will be noted in Fig. 5 that with the 24 kg. load the shoulder marker of this subject followed the same general trend as the other nine subjects. The scatter of the points for the positions of the various markers is due largely to variations in body stature, and to some extent to the fact that the horizontal spread of the points has been exaggerated for the sake of clarity.

The results show that, for a given load, the body will lean farther forward if the load is low on the back than if it is in a higher position. This suggests that the body treats the load on the back as essentially a problem in balancing. If the body keeps the vertical projection of the centre of gravity in a more or less constant position (Hallebrandt et al., 1944), it follows that lowering the position of a heavy weight on the back must cause the body to lean farther forwards to keep the combined centre of gravity of the body plus the weight in the same position. The fact that the trunk closely adapts its alignment in relation to the position and weight of the load it is carrying, as has been shown in the present experiments, accords with the view that the vertical projection of the centre of gravity varies within very small limits for any individual. Hale et

84.
CARRIER PLUS 24 Kg(HIGH).

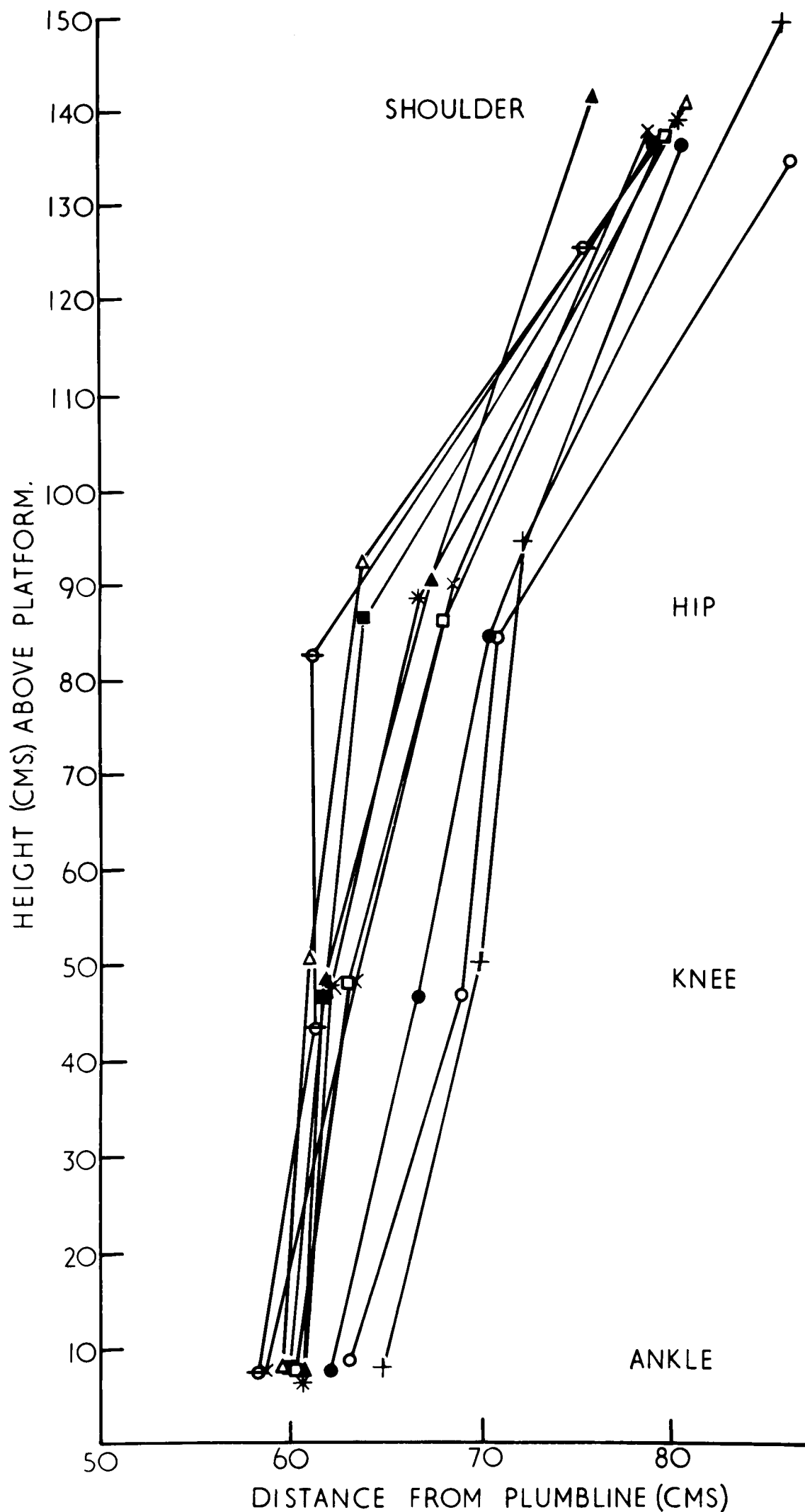


Fig. 5. Plot of positions of 4 landmarks on 10 subjects carrying 24 kg. high on the back.

al. (1953), also found that there was a general trend for a low pack to cause greater trunk inclination than a high pack, when carried on the back. Their results were not statistically significant, however, unlike the present findings.

The relative constancy of the hip marker under varying load conditions is probably due to its proximity to the centre of gravity of the body. The height of the centre of gravity is well established to vary between 55-56% of the body height (Cotton, 1931; Hellebrandt et al., 1938). Knowing the height of the subjects, an estimate could be made of the height of the centre of gravity for each of the subjects. It was found that a fairly short distance separated the centre of gravity from the horizontal projection through the hip markers - on the average some ten centimetres for the subjects used in the present experiments. If the position of the centre of gravity is a so-called "physiological constant" (Hellebrandt, 1950), then it might be expected that an anatomical landmark relatively near the centre of gravity would likewise show little or no variation in its position.

The evidence relating to the position of the knee joint indicates that heavy loading causes a certain amount of hyper-extension of the knee joint. Whether this is a purely passive phenomenon, or whether it is at least in part due to increased muscular activity in the legs cannot be decided on the basis of the present evidence. Smith (1956) has given a detailed analysis of the passive limiting mechanism which is operating at the knee joint in the normal standing position, and has emphasised that the adult knee joint is stabilized during standing by two complementary factors, namely the postural contraction of the

flexor muscles and the passive resistance of the tissues. It is reasonable to assume that during load carriage there is increased muscular activity helping to stabilize the joint, and this increased muscular activity, together with passive stretching of the tissues, causes hyper-extension of the knee joint.

Lippold and Naylor (1950) recorded electromyographic activity in the trunk muscles under two load conditions, and reported that when a load was carried high on the back, greater activity was recorded than when the load was carried around the pelvic girdle. They suggested this result was due to the higher combined centre of gravity when a load in the high back position is carried, and this results in greater instability. To maintain the body equilibrium under these less stable conditions, increased muscle activity is required. Lippold and Naylor's results cannot be directly compared with the results of the present experiment, as they investigated somewhat different load conditions. However, as Hale et al. (1953) pointed out, Lippold and Naylor did not refer to the possible effects of the inclination of the trunk varying with the position of the load. It has been shown that low back carriage causes increased inclination of the trunk, and this of itself could alter the level of muscular activity. It may be, for example, that while a load carried high on the back causes increased erector spinae activity, the increased inclination of the trunk caused by carrying a load low on the back results in increased activity of the hip extensors, the overall muscular activity remaining relatively constant. This suggestion is supported indirectly by the findings of Daniels et al. (1952, 1953) who found on the basis of energy expenditure

experiments no significant difference in the energy cost between carrying a load high or low on the back. Further indirect support for this view is gained from the work of Reid et al. (1955), and Reid and Draper (1957), who found, after extensive field trials, that in general there was no significant physiological difference between different types of Army packs. If the position and type of load carriage significantly affected the overall muscular activity, it would be expected that this would be demonstrable in physiological tests.

The present experiments, in which two more or less extreme positions of practical load carriage on the back were adopted, suggest that as far as the anti-gravity muscles are concerned, the relative position of the load is not of prime importance. Considering the body as a whole, a load on the back of, for example, 24 kg. requires the muscles resisting the effect of gravity to work as much for one position of the load as for another. It seems reasonable to suggest that for comparable loads the total muscular activity necessary to resist the collapsing effect of gravity will not vary significantly, wherever the load is placed on the back. It was shown by Hellebrandt et al. (1944), that the combined centre of gravity of the body plus the load remains in virtually the same vertical projection as the centre of gravity of the unloaded body, and the present data show that this is accomplished by alterations in the position of the trunk. It is suggested that the trunk acts as a counterbalance, altering its inclination according to the site of the load on the back, and in this manner the projection of the centre of gravity at the feet remains relatively constant.

SUMMARY

1. The effect of load carriage on the normal standing posture has been studied using a miniature camera photographic technique.
2. The displacement from the vertical of marked anatomical landmarks was measured on enlarged negative images, during normal standing and when loads of 12 kg. and 24 kg. were carried.
3. It was found that there was a highly significant ($P < 0.001$) forward displacement of the shoulder with load carriage, while there was no significant change in the position of the hip. There was a highly significant ($P < 0.001$) backwards displacement of the knee, while the position of the ankle was not significantly affected.
4. It was concluded that a highly significant realignment of the body position takes place during load carriage above a certain minimum weight.
5. A weight of 24 kg. was found to cause further forward displacement of the shoulder when it was carried low on the back than when it was carried high on the back ($P < 0.001$).
6. The suggestion is made that the trunk acts as a kind of counter-balance, altering its inclination according to the position of the load on the back so that the vertical projection of the centre of gravity remains relatively undisturbed.

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Paper 4THE EFFECT OF LOAD CARRIAGE ON POSTURAL ACTIVITY IN MAN

INTRODUCTION

The present study consists of an analysis of the effect of load carriage on the postural activity of standing. In a previous study (Thomas, 1958c), it was shown that load carriage above a certain minimum weight changes the normal alignment of the body in the standing position. It was suggested that the trunk acted as a kind of counter-balance for loads on the back, and adapted its alignment so as to preserve the normal vertical projection of the centre of gravity. The present experiments were designed to test the hypothesis that the position of a load on the back does not significantly affect the rate of working of the muscles that counteract the effects of gravity.

Postural sway is well known to be invariably associated with the upright position in Man, and the extent and nature of this sway has been frequently investigated (Hellebrandt & Franseen, 1943). The effect of load carriage on the normal postural sway has received little attention, however, and has been examined in this study using loads carried on the back. An attempt has also been made to assess the rate of working of the postural muscles that control sway, during the carriage of loads of varying weight and position. The experiments were conducted using a force analysis platform (Whitney, 1958a), to which

had been attached an accelerometer. Simultaneous records were made of the displacement of the centre of gravity of the subjects and the amount of horizontal movement imparted to the platform by the forces acting at the feet of the subject (Thomas, 1958a, 1958b). The object of the study was to determine in what way the normal postural activity was affected by load carriage, and to establish whether the amount of sway could be correlated with the output of the accelerometer. It is believed that the rate of working of the muscles controlling the body sway can be related to the amplitude of the accelerometer output.

METHODS

The force analysis platform used in these experiments has already been described (Whitney, 1958a; see also appendix 1), and its application for measuring the movement of the centre of gravity during the normal postural sway has also been described (Thomas, 1958b). Attached to the platform was an accelerometer (Statham Laboratories, Los Angeles, California, Model C-1-350), which recorded the displacement of the platform caused by the forces acting at the feet of the subjects. The application of an accelerometer for this purpose has been considered previously (Thomas, 1958a). The accelerometer is believed to be recording the oscillatory component of the forces imparted to the platform in the antero-posterior horizontal plane by the feet of the subjects as they swayed back and forth (Whitney, 1958b). The movement of the centre of gravity of the subject, and the output of the accelerometer, were recorded simultaneously on a multi-channel galvanometer.

The experiments were performed on ten medical student volunteers, aged 18 to 21 years. Their heights ranged from 170.6 cms. to 188.9 cms., and their weights from 59.75 kgs. to 84.75 kgs. The subjects were positioned on the platform so that their heel contours were constant along the same transverse axis, and at a fixed distance of twenty centimetres apart. They were allowed to spread their feet at will, once the position of the heels had been fixed. Standardized shoes of the appropriate size were provided, and the subjects wore only light clothing. Instructions were given to adopt an easy standing position, and to make no unnecessary movements. The experiments were conducted so that as far as possible, only one factor (the load carried) was altered throughout the experiment. Each subject was studied on two occasions, on separate days.

The loads were carried on a modified Bergen rucksack frame, as previously described (Thomas, 1958a). The subjects were studied, 1. in the normal standing position, 2. carrying an empty carrier (weight 4.25 kgs.), 3. carrying an added weight of 12 kgs. high on the back, 4. carrying an added weight of 24 kgs., as in 3, and finally 5. carrying an added weight of 24 kgs. in a position low on the back. The position of the 24 kgs. load was varied over a distance of thirty centimetres, the high position approximating the upper thoracic region and the low position the lower lumbar region of the subject. The analysis of the photographic records was performed by planimetry, and the results expressed as a mean deviation of the tracing from a fitted baseline (Thomas, 1958b). A fixed length of the record was measured for all subjects, during the five load conditions investigated, the length of

the record being the equivalent of forty five seconds standing (Plates 1 & 2). The sway tracings were also measured to determine the extreme range of movement of the tracing during the same forty five second period. The total range of movement of the tracing was measured for the five load conditions, and the results are given in Table 2. The results were compared by an analysis of variance, and the levels of significance determined by a t-test.

RESULTS

The analysis of the results shows that increasing the load carried causes significant increases in the amplitude of the accelerometer tracing, in the mean deviation from a fitted baseline of the centre of gravity tracing, and in the total movement of the centre of gravity, in an antero-posterior direction (Tables 1, 3 & 4). It was found that in all subjects there was a positive linear correlation between the amount of sway (expressed as a mean deviation from a baseline) and the amplitude of the accelerometer for the five load conditions investigated (Fig. 1).

It was found that there was a highly significant increase in the amplitude of the accelerometer tracing between standing normally and carrying 12 kg. ($P < 0.001$), and also between standing normally and carrying 24 kg., in the high or low position ($P < 0.001$). No significant differences were found between carrying an empty carrier and normal standing, nor between carrying 24 kg. in the high as compared to the low position (Table 4). The results for the sway patterns, both for total range and mean deviation, were closely

Table 1

Table 1. Mean deviation in millimetres of sway record above or below a fitted baseline during 45 secs. standing. Data from 10 subjects in two experiments, during 5 load conditions.

Subject	Normal standing	Empty carrier	12 kg. (high)	24 kg. (high)	24 kg. (low)
W.G.J.H.	2.21 2.48	1.86 1.95	2.60 2.74	3.83 3.98	2.39 4.33
J.M.S.W.	1.04 1.04	2.37 1.80	2.84 2.70	4.07 4.36	5.21 3.79
S.M.	2.27 1.00	2.27 2.00	3.19 4.91	3.82 4.00	5.36 2.09
D.N.J.	1.66 1.59	3.68 1.73	2.74 3.32	4.33 6.13	3.98 6.92
F.D.H.	3.10 2.79	3.57 3.80	4.42 2.25	5.20 4.19	7.22 2.33
J.G.R.	2.15 2.06	1.79 2.42	5.35 5.19	3.67 4.74	7.61 3.58
C.P.B.	2.26 2.26	3.67 2.36	4.43 3.20	4.99 5.37	5.94 4.80
J.S.E.	1.21 1.99	1.99 3.12	4.24 3.21	3.46 3.81	4.24 8.14
W.F.H.	0.97 1.17	1.07 2.04	1.46 2.24	3.60 3.31	2.53 3.31
W.G.B.	1.85 2.40	4.05 2.06	3.64 4.73	3.09 4.46	4.73 3.64
Mean	1.87	2.48	3.47	4.22	4.61
Standard deviation (means for each subject)	± 0.59	± 0.64	± 0.96	± 0.62	± 1.04

comparable to the accelerometer results. There was no significant difference between the results for normal standing and carrying an empty carrier, nor between carrying 24 kg. in either the high or low positions (Tables 1 & 3). However, highly significant increases ($P < 0.001$) occurred in the sway tracings between normal standing and carrying 12 kg., and between normal standing and carrying 24 kg. (high or low positions). There is therefore a general increase in the movement of the centre of gravity tracing, expressed either as a mean deviation or as a total range of movement, and there is also an increase in the level of acceleration recorded at the feet, with increases in the amount of load carried. However, the position of a load of the same weight on the back did not affect either the sway pattern or the acceleration recorded at the feet.

The degree of sway, expressed as a mean deviation from a fitted baseline, was correlated with the amplitude of the accelerometer tracing, and the correlation coefficients are given in Table 2. With ten subjects, the correlation coefficient is significant at the 5% level only when the coefficient is 0.85 or above. Five of the subjects reach this level of correlation, and further two subjects just fail to do so. One subject had a considerably lower level of correlation than the other subjects. The results may be considered to give in general a high level of correlation with the limited degrees of freedom allowed by the experiment. The overall association between the ten correlation coefficients was tested and found to be highly significant ($\chi^2 = 48.9$, and $P < 0.001$). Graphs plotting the accelerometer amplitude against the mean deviation of the sway tracing

Table 2

Table 2. Correlation coefficients between sway (mean deviation) and acceleration for ten subjects.

Subject	Correlation coefficient (sway and acceleration)	Significance
W.G.J.H.	0.89	$P < 0.05$
J.M.S.W.	0.996	$P < 0.001$
S.M.	0.80	$P < 0.1$
D.N.J.	0.96	$P < 0.01$
F.D.H.	0.82	$P < 0.1$
J.G.R.	0.75	N.S.
C.P.B.	0.42	N.S.
J.S.E.	0.89	$P < 0.05$
W.E.M.	0.88	$P < 0.05$
W.G.B.	0.76	N.S.

showed that the correlation was a linear one, although the inclination of the lines showed considerable subject to subject variation. A graph showing the linear correlations between the sway and the level of acceleration recorded, using the average results for the two experiments, is given in Fig. 1. The points on the graph have been omitted for clarity.

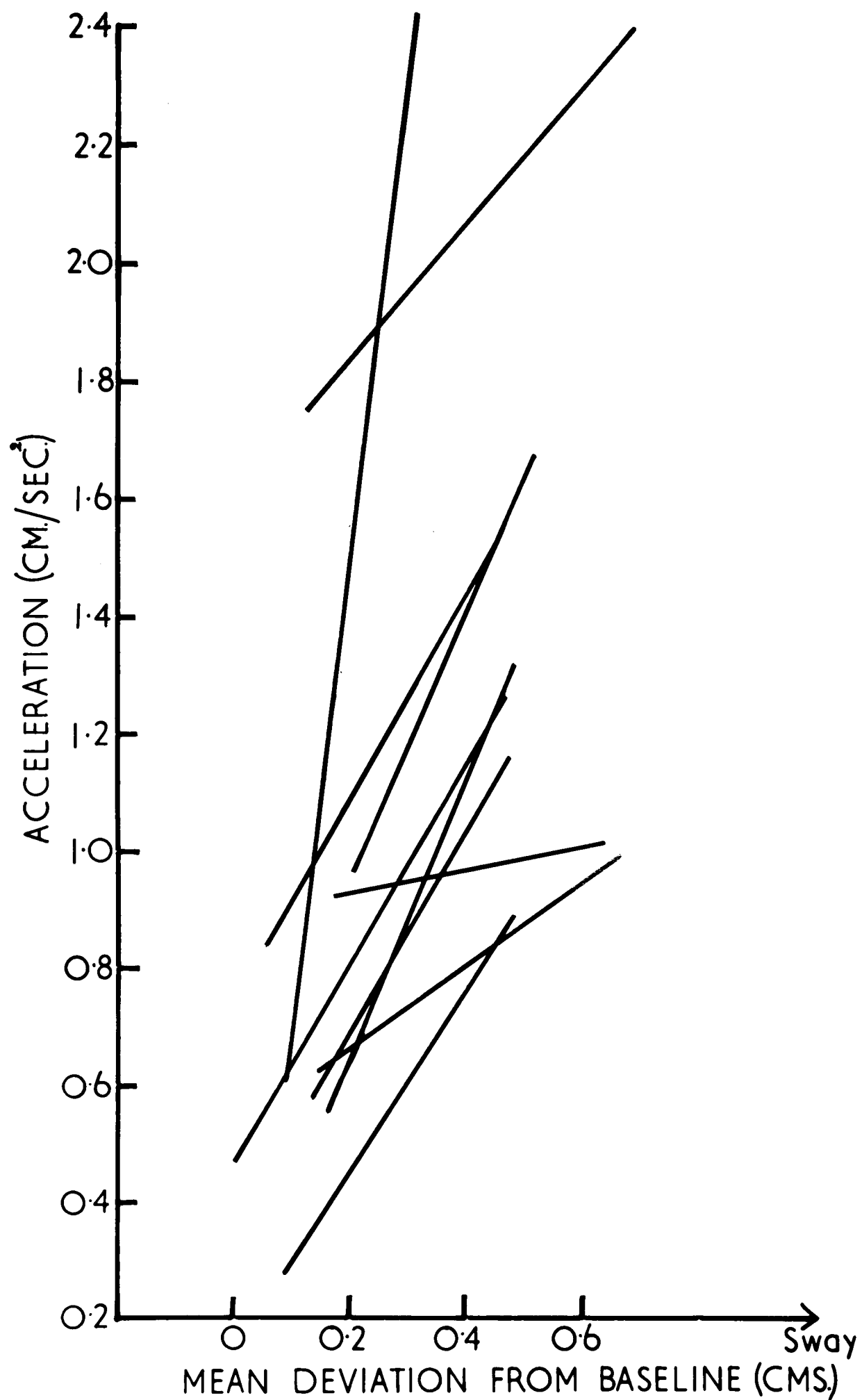


Fig. 1. Graph of sway plotted against acceleration in ten subjects. Points omitted for clarity. Note lines show similar inclination in 7 subjects.

Table 3

Table 3. Total range of sway record in centimetres during 45 secs. standing. 10 subjects in two experiments under 5 load conditions.

Subject	Normal standing	Empty carrier	12 kg. (high)	24 kg. (high)	24 kg. (low)
W.G.S.H.	2.93 2.34	2.63 2.78	3.99 3.66	6.47 6.14	2.63 5.27
J.T.S.W.	1.88 2.04	2.66 2.35	3.76 3.43	4.39 5.33	5.96 4.70
S.M.	1.95 1.95	2.56 2.71	3.01 4.81	5.86 6.46	6.46 4.06
D.N.J.	2.50 1.91	3.10 2.39	3.46 3.58	6.20 6.32	5.01 6.08
F.D.H.	3.34 2.82	4.75 3.98	5.90 3.59	10.27 7.70	7.57 3.85
J.G.R.	3.55 2.81	2.66 2.07	5.36 5.03	7.40 6.22	10.95 3.85
C.P.B.	2.80 2.34	5.61 4.05	4.99 3.74	5.76 7.17	10.28 6.08
J.S.E.	1.58 2.29	2.72 3.87	4.01 4.73	5.16 5.01	4.73 11.03
W.F.M.	1.45 1.61	1.61 1.93	1.77 4.34	4.51 4.51	3.86 3.70
W.G.B.	3.18 2.27	5.22 3.29	5.67 4.54	4.54 4.54	4.88 5.56
Mean	2.38	3.15	4.17	5.996	5.82
Standard deviation (means for each subject)	± 0.54	± 1.01	± 0.71	± 1.35	± 1.53

DISCUSSION

The extent and variation of the normal postural sway has been studied for nearly a century (Vierordt, 1861; Orma, 1957), but the effect of load carriage on postural sway has been little studied. Little or no work has been done attempting to correlate directly the amount of postural sway with the rate of working of the muscles that control the swaying. It has been claimed that normal postural sway is reduced by the carrying of an Army pack on the back (Hellebrandt, Fries, Larsen & Kelso, 1944).

Postural sway has been analysed in this study both as the mean deviation of the centre of gravity from a fitted baseline, and as the total range of movement of the centre of gravity tracing, for a fixed period of time. In this experiment, only antero-posterior sway has been measured, as it is believed that the lateral component of sway is small and relatively insignificant in the foot position adopted in these experiments (Thomas, 1958b). The technique of accurately measuring the postural sway by planimetry has been discussed in a previous paper (Thomas, 1958b), and it is believed to give an accurate and meaningful figure for the degree of swaying that is occurring. While it has been found, under the conditions of this experiment, that the effect of load carriage on the sway pattern is such as to produce comparable results whether the sway is expressed as a total range of movement, or as a mean deviation, it is considered that results expressed as a mean deviation from a baseline are, on the whole, more representative and less liable to error.

The data obtained show that the average results for the

Table 4

Table 4. Amplitude of accelerometer tracing in cm./sec.² during 45 secs. standing under five load conditions. Ten subjects, investigated twice on separate days.

Subject	Normal standing	Empty carrier	12 kg. (high)	24 kg. (high)	24 kg. (low)
W.G.S.H.	0.75 0.83	0.50 0.75	0.94 0.75	0.15 0.96	0.79 0.83
J.M.S.W.	0.83 0.62	1.91 0.83	2.75 1.42	3.29 3.41	3.71 3.75
S.M.	0.37 0.29	0.50 0.54	0.54 0.58	0.87 0.83	0.79 0.75
D.N.J.	1.87 1.87	1.91 1.79	1.87 1.96	2.37 2.04	2.41 2.25
F.D.H.	0.54 1.46	1.33 1.33	1.67 1.25	1.79 1.37	1.83 1.25
J.G.R.	0.92 0.37	0.75 0.54	0.87 0.67	1.29 0.67	1.25 0.62
C.P.B.	1.21 0.67	1.21 0.62	1.25 0.79	0.96 0.96	1.12 0.83
J.S.E.	1.04 0.67	0.75 0.67	1.08 0.92	1.42 1.08	1.83 1.25
W.F.M.	0.71 1.12	0.83 1.04	1.04 1.33	1.29 1.45	1.21 1.08
W.G.B.	0.75 0.79	0.75 0.71	1.96 0.83	1.12 1.00	1.04 0.96
Mean	0.88	0.96	1.22	1.42	1.48
Standard deviation (means for each subject)	±0.39	±0.43	±0.49	±0.82	±0.92

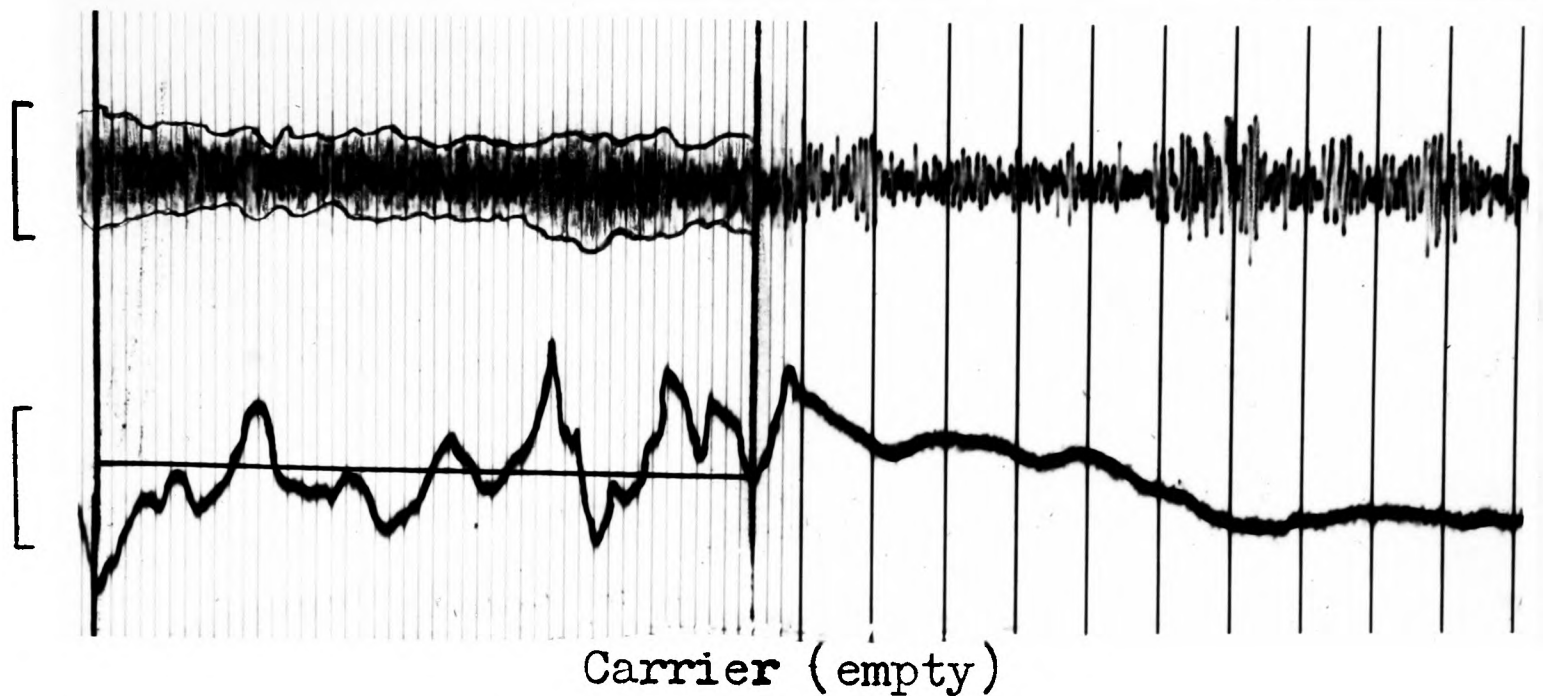


PLATE 1

Effect of load carriage on sway and accelerometer patterns

Accelerometer (upper) and centre of gravity (lower) tracings of subject carrying an empty carrier pack (weight 4.25 kg.). Recording at two paper speeds, the vertical lines representing 1 sec. intervals. Note envelope drawn in over accelerometer tracing and baseline fitted to centre of gravity tracing. Left half of tracings were portions analysed.

Calibration (upper) = 1 cm./sec.²
 (lower) = 1 cm. movement of C.of G.

movement of the centre of gravity during normal standing was 2.33 cms. (S.E. 0.067 cm.) for the total range of movement, and 1.87 mms. (S.E. 0.074 mm.) for the mean deviation from a fitted baseline. However, when 24 kg. is carried the amount of swaying is more than doubled, with an average of 6.00 cms. (S.E. 0.454 cm.), and an average mean deviation of 4.22 mms. (S.E. 0.133 mm.) Hellebrandt et al. (1944) claimed that the addition of an Army pack suppressed the level of normal body sway. The findings of the present experiments are at variance with their results, and it has been shown that there is a significant increase in the amount of swaying between carrying 12 kg. and 24 kg. ($P < 0.05$). It is difficult to believe that these divergent results can be wholly attributed to variations in the types of pack used in the two studies, and the only reasonable interpretation of the present findings would seem to be that postural sway, far from being suppressed by load carriage, is in fact significantly increased.

There is a high level of correlation between the sway results and the accelerometer results. Increasing the load carried causes the body to sway more, and this is associated with increased acceleration recorded at the feet. A linear relationship can be demonstrated between the degree of swaying (mean deviation from a baseline) and the ~~sway~~ acceleration recorded (Fig. 1), and if the ~~sway~~ acceleration is proportional to the rate of working of muscles controlling the body sway, then it follows that the degree of swaying is also proportional to the rate of working of the muscles. In other words, the more the body sways, the more must the rate of working of the controlling muscles increase. The fact that neither sway nor

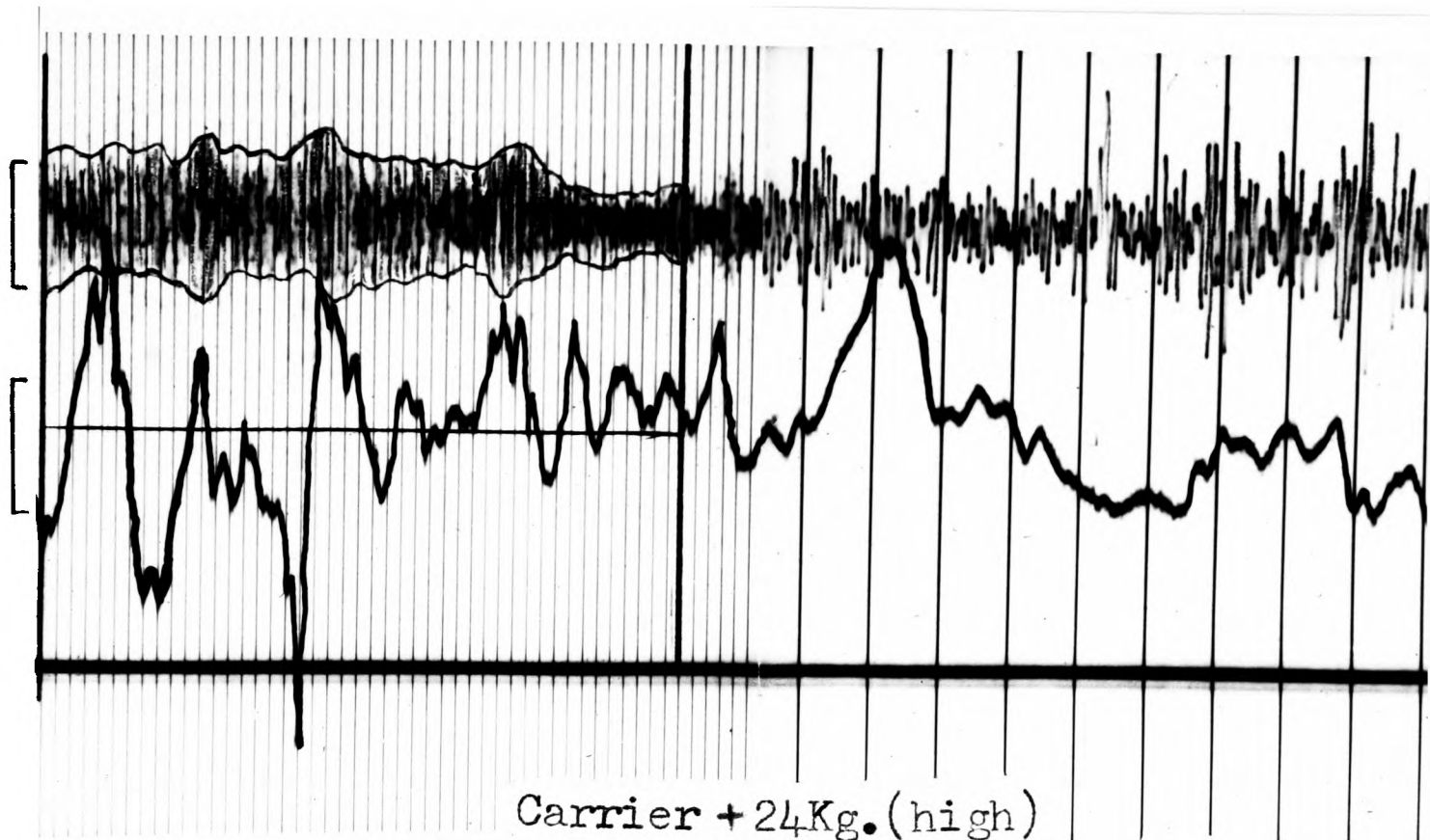


PLATE 2

Effect of load carriage on sway and accelerometer patterns

Same subject as in Fig. 1, but now carrying 24 kg. added to the carrier, in a position high on the back. Note the increase in the amplitude of the accelerometer output (upper) and also in the amount of movement of the centre of gravity tracing (lower).

Recording at two paper speeds, the vertical lines representing 1 sec. intervals. Envelope drawn in over accelerometer tracing and baseline fitted to centre of gravity tracing. Left halves of tracings were analysed by planimetry.

Calibration = 1 cm./sec.² (upper)
 = 1 cm. movement of the
 centre of gravity. (lower)

acceleration is significantly altered by moving a 24 kg. load over a distance of thirty centimetres on the back suggests that the general position of a load on the back does not affect the rate of working of those muscles counter-acting the effects of gravity, at least in so far as their activity is reflected in the accelerometer output. This suggestion fits in with the findings of Daniels et al. (1952), who found no significant difference in the energy cost between carrying the same load in a high or low position on the back.

The accelerometer results demonstrate very clearly an oscillatory activity at a frequency of around ten cycles per second (Plates 1 - 4). An oscillatory activity of this frequency has been recognised for many years to be an accompaniment of skeletal muscular activity, since Schäfer first described the phenomenon in 1886. This so-called muscular tremor has been much studied recently (Halliday & Redfearn, 1956; Marshall & Walsh, 1956; B.M.J., 1957; Whitney, 1958b), and a more detailed consideration of this ten cycle per second tremor in relation to muscular activity has been given in a previous paper (Thomas, 1958a). In the present context, the 10 c.p.s. nature of the accelerometer output, so clearly evident in the present experiments, can be regarded as further evidence of the muscular origin of the forces producing the acceleration that is being recorded. The general level of the accelerometer output has been shown to be directly related to the work done by those muscles supporting the load (output increasing with load), and it seems reasonable to conclude the level of the accelerometer tremor recorded represents the mechanical end-results of happenings in the postural muscles.

PLATE 3

Effect of carrying 12 kg. on sway and accelerometer patterns

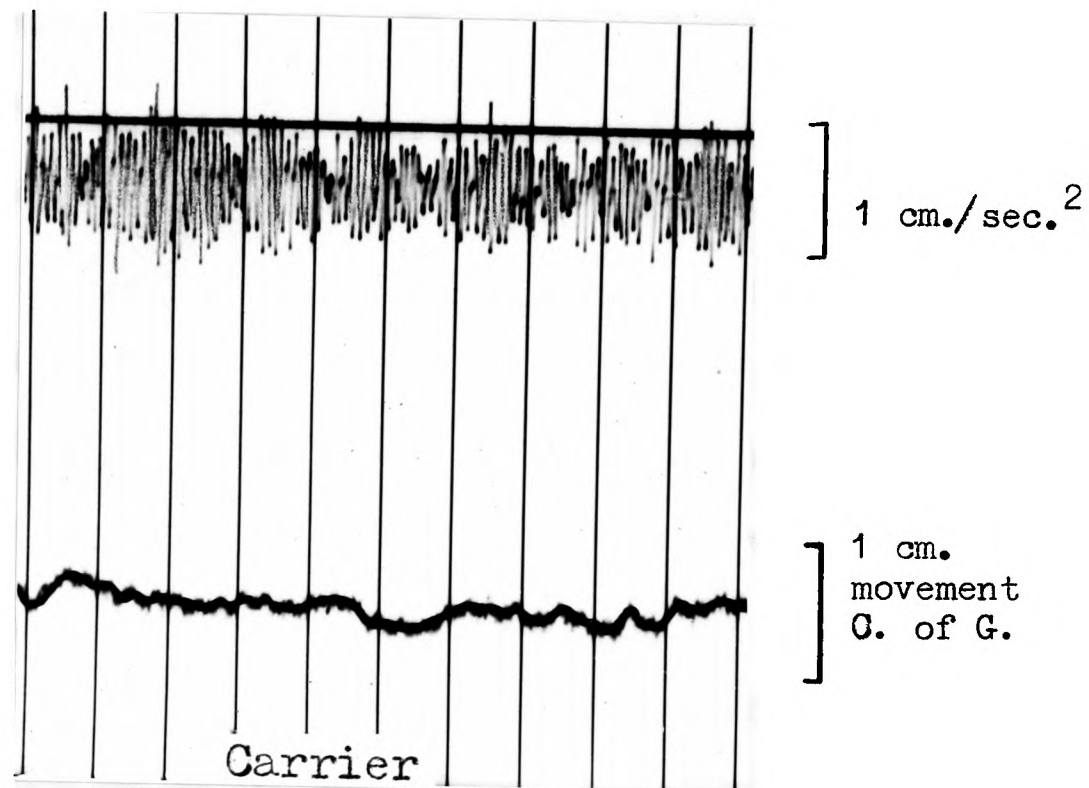


Fig. 1. Accelerometer (upper) and centre of gravity (lower) tracings of subject carrying an empty carrier pack (Wt.- 4.25 kg.)

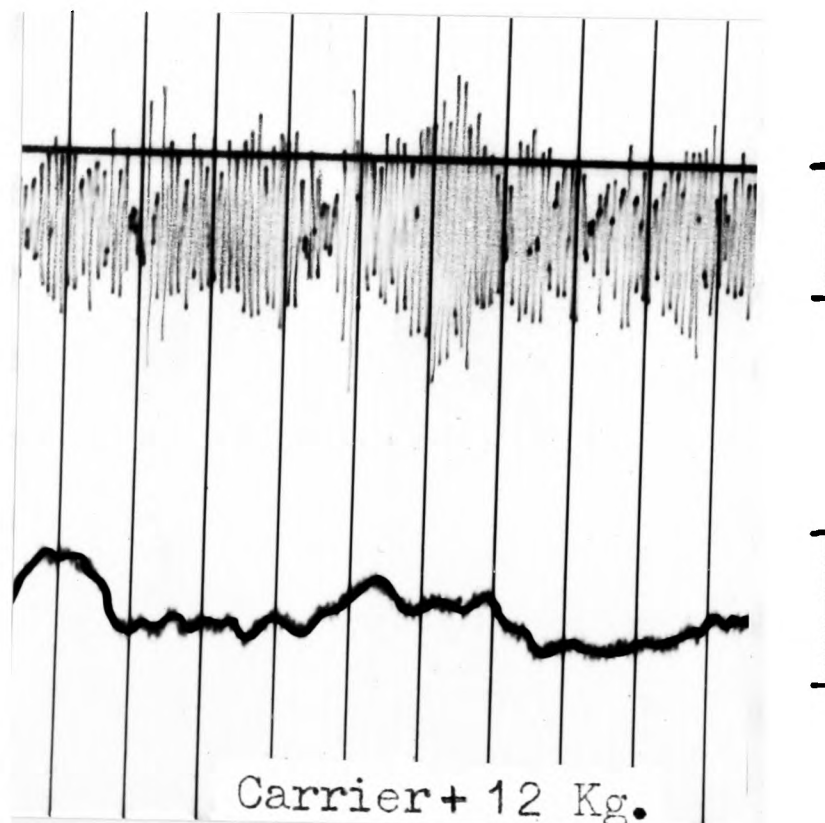


Fig. 2. Same subject now carrying 12 kg. added to pack in a position high on the back. Note increase in amplitude of accl. tracing, and increased movement of centre of gravity tracing.

The exact site of the muscles that control postural sway, and whose activity is being recorded on the accelerometer, cannot be determined by the present technique. It may be inferred however, that the erector spinae muscles, and the lower limb muscles generally (but particularly gastrocnemius and soleus), must be predominantly involved. There is good electromyographic evidence that the soleus muscles are continuously active during standing (Joseph & Nightingale, 1952; Joseph, Nightingale & Williams, 1955), and it appears likely that these muscles (among others) play a large part in controlling the swaying of the body.

The present data suggests that the position of a load on the back is unimportant in comparison with the weight of the load. No evidence was found that, for a load of the same weight, the position of the load on the back affected either the amount of postural sway, or the level of activity of the muscles supporting the load. At least in so far as static conditions are concerned, the important factor would seem to be the weight of the load, and the heavier the load, the more the body sways; the more the body sways, the more the muscles supporting the weight of the body plus the load have to work to maintain the body equilibrium.

These findings are of interest in view of the suggestion put forward (Thomas, 1958c) that the body reacts to load carriage by altering the normal alignment of the body so that the vertical projection of the centre of gravity to the feet remains undisturbed. As a result of this realignment, it was suggested, the anti-gravity muscles work at much the same rate for differing positions of a load

PLATE 4

Effect of carrying 24 kg. on sway and accelerometer patterns

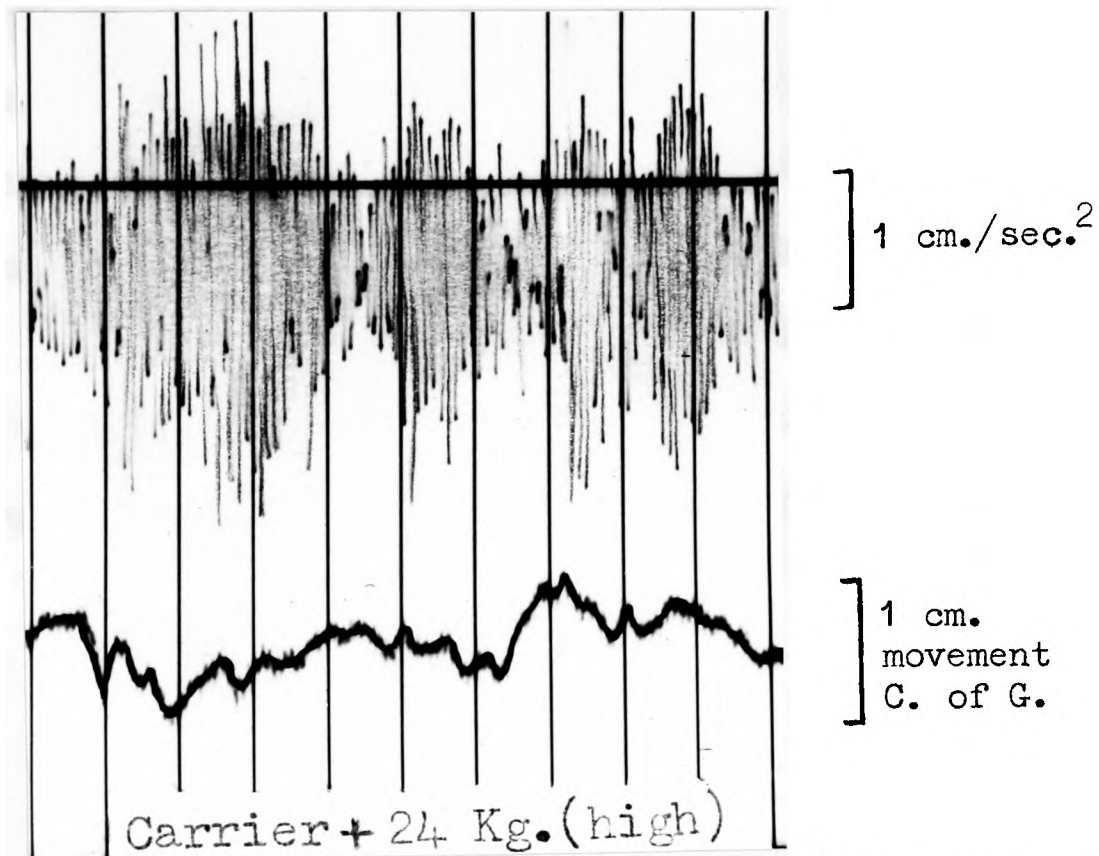


Fig. 1. Same subject as in Pl. 3, now carrying 24 kg. added to pack. Note further increase in both amplitude of accelerometer tracing and movement of centre of gravity tracing.

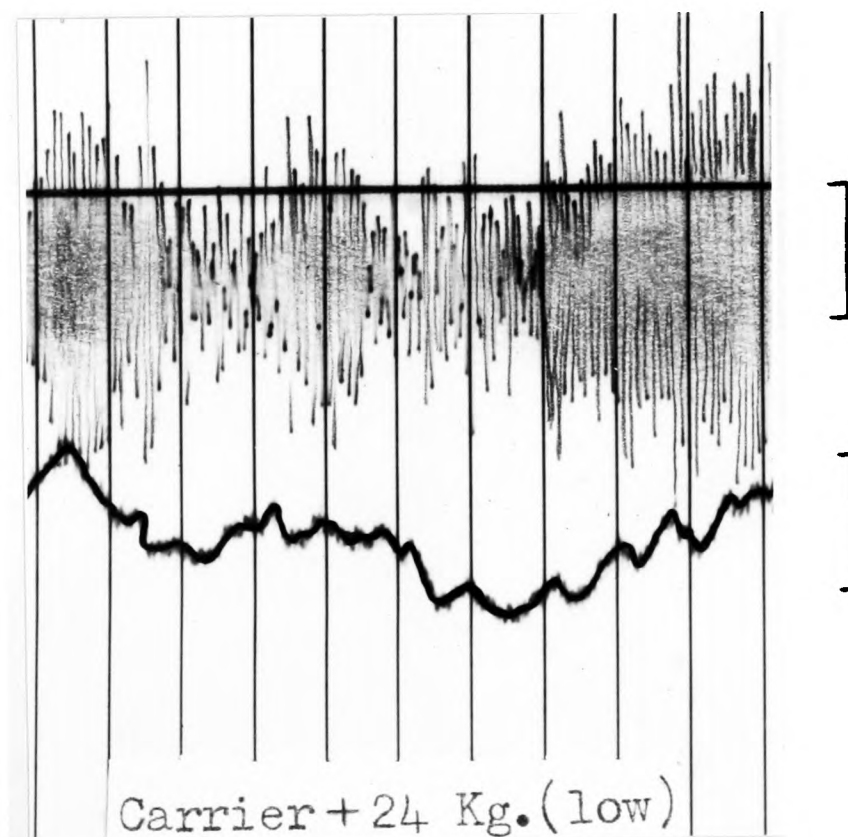


Fig. 2. As in Fig. 1, but with pack now moved to a position low on the back. Note there is no marked change from Fig. 1.

of the same weight. The general view may be put forward, therefore, that the same load on the back will not alter the level of the postural sway, or the rate of working of the postural muscles, wherever the load is situated on the back. The body simply adapts itself, by altering the inclination of the trunk, so that the projection of the combined centre of gravity remains constant. If the weight of the load is increased, however, the body sways more, and more work has to be done by the postural muscles to maintain the erect posture.

SUMMARY

1. The antero-posterior displacements of the centre of gravity of the body (postural sway) of ten subjects has been studied under varying load conditions.
2. The rate of working of the muscles controlling postural sway has been arbitrarily assessed on a mechanical basis, using an accelerometer.
3. It was found that carrying a load of 12 kg. in a position high on the back caused a significant increase in the amount of postural sway ($P < 0.001$). The addition of a further 12 kg. caused a further significant increase in the amount of postural sway and in the rate of working of the muscles ($P < 0.05$).
4. The movement of a 24 kg. load from a position high on the back to a position low on the back (distance 30 cms.) did not significantly affect either the amount of postural sway or the rate of working of the muscles.
5. It is concluded that the site of a load on the back does not affect the amount of work that is done by the muscles maintaining the upright posture. The body simply alters its normal alignment according to the position of the load, so as to keep the vertical projection of the centre of gravity at the feet in a relatively constant position.

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APPENDIX 1A Description of the Force Analysis PlatformGeneral arrangement

The apparatus described in this appendix was used for most of the work on which this thesis is based. The apparatus was designed and constructed solely by Dr. R.J. Whitney, of the Medical Research Council Unit for Research on Climate and Working Efficiency. A description of the platform has already been published (Whitney, 1958), but it was thought advisable to include a brief account of the platform in this thesis. The description of the platform given here relates to its basic construction, and to those parts of it which have been used in the experiments described.

The platform consists essentially of a rigid steel platform, 120 cms. square, which is suspended at each of its four corners by vertical lengths of piano wire. The other ends of the wires were attached to horizontally disposed spring steel bars, which were firmly clamped by their roots to a heavy steel chassis attached to the floor. Movement of the platform in the horizontal axes (side-to-side and front-to-back) was limited by horizontal piano wires running along the sides of the platform, and attached at each corner to two vertically disposed steel bars. The steel bars were carefully chosen to have identical deflection rates. Deflections of the bars at the corners of the platform were transduced to electrical changes by electrical resistance strain gauges mounted on the bars. The strain gauges constituted the

arms of Wheatstone bridge circuits, and the electrical outputs of the bridge circuits, when energised, were continuously recorded on a multichannel galvanometer. The weight of the subject causes a downward deflection of the four horizontal bars, the summed deflection of the bars being proportional to the weight of the subjects. As the subject sways forwards, the centre of pressure of the subject's weight also moves forwards, and therefore the forces acting on the front two horizontal bars increases relative to the rear bars. This increase is transduced electrically and summed in a separate bridge circuit; from this result is subtracted electronically the sum of the deflection of the rear two horizontal bars. Initially, the platform is electrically balanced; as the subject sways, the balance is upset, and this electrical resistance imbalance is proportional to the amount of movement of the centre of pressure of the subject's feet. The result is recorded photographically on a multi-channel galvanometer, and gives a record of the antero-posterior movement of the centre of pressure. An essentially similar arrangement is used for recording the side-to-side movement of the subject's centre of pressure. The whole platform is hydraulically damped, using oil-filled bellows, and the damping is optimal (about 60% critical). The platform is sufficiently rigid for a subject to feel completely stable when standing on it.

The horizontal displacement of the platform (antero-posterior in relation to the subject) was also recorded by an accelerometer, which was attached to one end of the platform. The accelerometer that was used employed a strain gauge transducer, and was sensitive to acceleration over a range of plus or minus one G. (Model C-1-350,

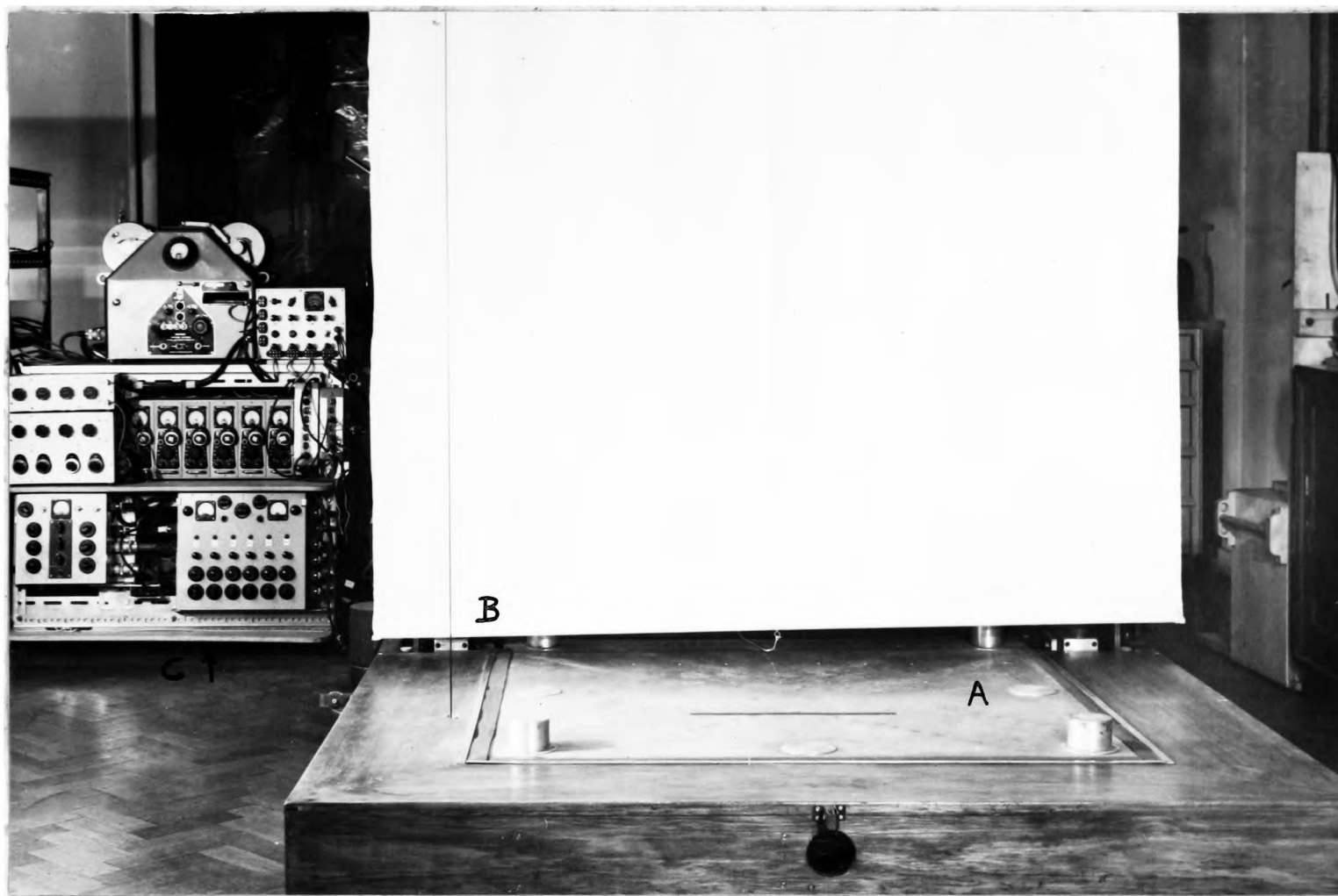


Figure 1. The Force Analysis Platform

In the foreground of the picture can be seen the platform (A), with its wooden surround. The white backcloth (B) behind the platform, and the vertical plumbline can also be seen. The recording apparatus can be seen on the left of the picture (C). Subjects faced to the right of the picture, when standing on the platform.

Statham Laboratories, Los Angeles, California). It was connected up to a multi-channel galvanometer, where a continuous record could be made of its output. As the subject sways to and fro, continuous forces are being exerted at his feet, and the accelerometer is recording the acceleration of the platform caused by the forces acting antero-posteriorly at the feet of the subject. The accelerometer can be considered as doubly differentiating the displacement of the platform.

Mechanical details of the platform.

The deflection rates of the platform, directly determined by dial gauges, are 0.0012 cm. per kg. for deflections in the vertical axis, and 0.0018 cm. per kg. for deflections in either of the two horizontal axes (side-to-side and front-to-back). Departure from linearity in the deflection rate is less than one per cent total discrepancy at a loading of 100 kg. It will be appreciated from these figures that a subject standing normally on the platform, even with a heavy load on his back, will not be aware of any movement at his feet. The inherent stiffness of the platform confers a moderately high natural frequency and a low response time on the platform. With a subject of normal weight on the platform, the response time is of the order of 0.02 sec.

The mechanical arrangements of the platform are such that a force, exerted in any part of the suspended platform and in any downwards direction in relation to the surface of the platform, is resolved along three known rectilinear axes and produces deflections

of some or all of the suspension bars, and the deflection of each individual bar depending on the magnitude and direction of the force, and on the position of the centre of the force within the surface of the platform. The strain gauges, which are mounted on the suspension bars, are connected electrically in sets, each set constituting the strain sensitive elements of an independent Wheatstone bridge. The electrical output of the bridge will be proportional to the co-ordinate or force required.

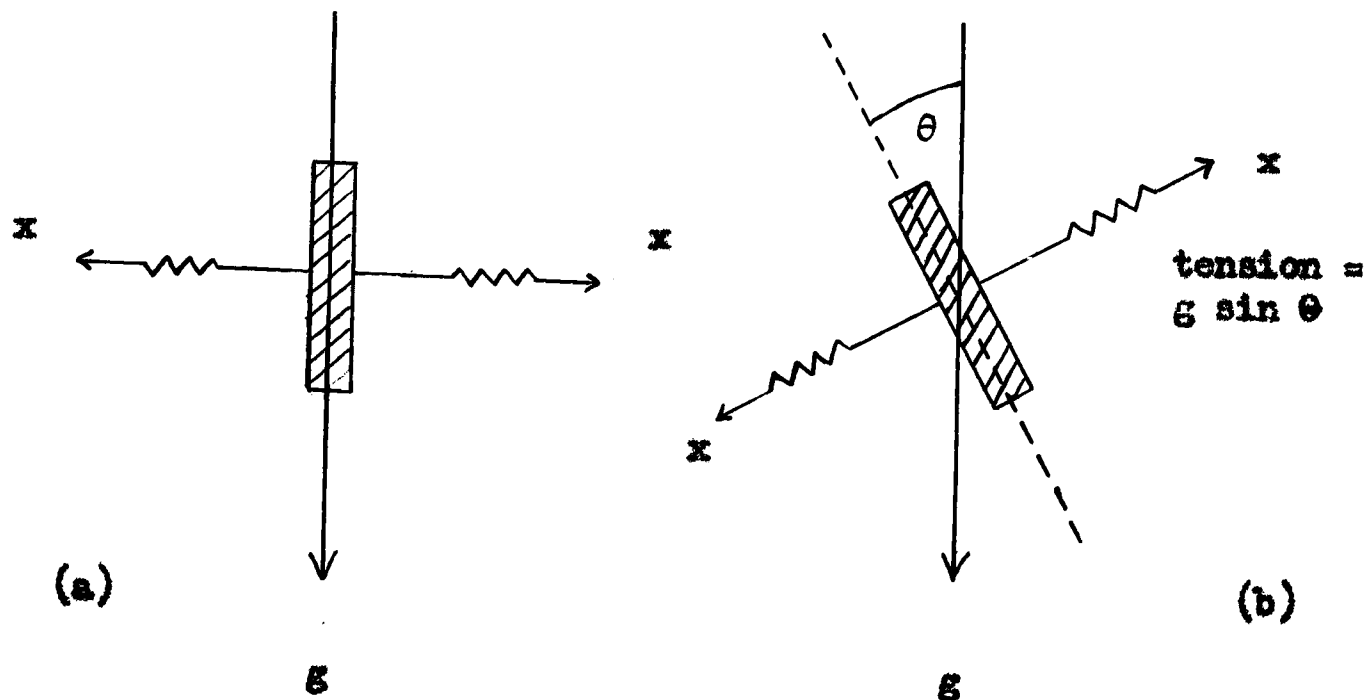
One bridge network (W-bridge) has an output which is proportional to the total force exerted normal to the platform surface, at any point within the surface, and this bridge is used for weighing the subjects. This bridge network sums the resistance changes at each corner of the platform due to the force resolved at each corner in the vertical axis, and therefore the output of the bridge will be proportional to the total vertical force exerted on the platform surface, irrespective of where the force due to the subject's weight is exerted within the surface of the platform. It was possible to weigh the subjects to the nearest 100 grams using this network.

Two separate bridge networks were used to determine the difference between the vertical forces exerted at the sides of the platform, in the horizontal axes. The output of these bridges, with a constant energizing voltage, is proportional to the difference between forces acting 100 cms. apart, due to a single force exerted at some point between the sides (either side-to-side or front-to-back). If the difference of the forces is divided by the sum of the forces (i.e. the single force), the dividend will be proportional to the

co-ordinate of the single force. This division is carried out electronically by deriving a voltage from the output of the weight bridge network (W-bridge) which is inversely proportional to the W-bridge output, and then using this derived voltage as the energizing voltage for the horizontal force bridges. As the subject sways, the differences between the vertical forces acting at the front and back of the platform will change with the swaying, as the centre of pressure of the feet moves backwards and forwards. A continuous record of this change therefore gives a picture of the antero-posterior movement of the subject's centre of pressure of the feet. By a similar process, a record can be obtained of the side-to-side movement of the centre of pressure of the feet. By equating moments, the movement of the centre of pressure can be converted into movement of the centre of gravity of the subject.

APPENDIX 2The use of an accelerometer as an inclinometer

In the experiments described in Paper 2, a strain gauge accelerometer, (Statham Laboratories, Model C-1-350), was mounted on

Fig. 1

a belt at waist level and used as an inclinometer (Fig. 1). When the sensitive axis of the accelerometer (xx) is horizontal, as in Fig. 1a, the effect of gravity is such as to exert no bias on the output of the accelerometer, and the accelerometer reproduces solely acceleration occurring along its sensitive axis. When, however, the accelerometer is tilted, as in Fig. 1b, the effect of gravity is such as to cause a bias on the output, depending in its extent to the degree of tilting that is occurring. The effect of the bias is such as to cause the output to have an undulating form, as the accelerometer is tilted back

and forth by the swaying of the body. By measuring this output, the mean angular deviation of the accelerometer can be determined (mean deviation of the gross movement of the record from a fitted baseline - see Fig. 2a, p.53).

Thus, in Fig. 1b, the acceleration recorded = $g \sin \theta$,
and therefore, $\sin \theta = \frac{\text{acceleration}}{g}$.

When the angle θ is small (as it is in this case), it follows that:

$$\theta = \frac{\text{acceleration}}{g} \text{ radians}$$

If the acceleration recorded is equal to 1 cm./sec.², then

$$\begin{aligned} \theta &= \frac{1}{g} \\ &= \frac{1}{981} \\ &= 0.00102 \text{ radians.} \end{aligned}$$

(1 radian is the equivalent of 57.3 degrees)

Therefore, 1 cm./sec.² is the equivalent of 0.00102 rad.

The accelerometer has a built-in calibration of 1.25% g., and at the sensitivity it was used in the present experiments, 1 mm. deflection on the record was equal to 1.1 cm./sec.²

$$\begin{aligned} \text{Therefore, } 0.01 \text{ radian} &= 1 \times \frac{0.01}{0.00102} \\ &= 9.8 \text{ cm./sec.}^2 \end{aligned}$$

Thus, if 1mm. deflection on the record equals 1.1 cm./sec.²,
then 9.8 cm./sec.² = $\frac{9.8}{1.1}$ mm. = 8.91 mm.

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ABSTRACT

A study has been made of the skeletal muscular activity associated with maintaining the erect posture in Man, using twenty medical students as experimental subjects. The extent and nature of this muscular activity was assessed by the mechanical end-result it produced, using for this purpose a strain gauge accelerometer mounted on a freely-suspended and optimally-damped platform. Evidence was found for the existence of continuous 10 c.p.s. oscillatory forces exerted at the feet during normal standing. The extent of these oscillatory forces showed considerable subject-to-subject, and day-to-day variation, but they were nevertheless always present. It is considered that these forces result from skeletal muscular activity. It was also found that ischaemic pain was readily produced in the leg muscles of subjects standing normally, whereas over a similar period of time, ischaemia produced virtually no painful sensations if the subject was lying down. This finding was regarded as evidence that the erect posture is associated with active muscular contractions in the legs. It was concluded from this study that the erect posture in Man is a dynamic state, associated with continuous muscular activity, especially in the lower limbs. Gravity is a constant force, and continuous activity by the postural muscles is required to counteract its effects. It is denied that normal standing in Man is dependent primarily on the mechanical arrangements of the skeleton, or the passive elastic tension in the postural muscles, as has been claimed. *such as Blemessen (1951) & Ralston & Libet (1956),* It is believed that some workers have tended to place too much

emphasis on the findings of electromyography of the postural muscles, and have been led into the false position of denying the existence of active muscle tone. The findings of the present experiments would suggest that there is good evidence that the upright posture in man is maintained by low-grade and continuous muscular activity in the anti-gravity muscles ("postural tonus").

A study was made of the postural sway occurring during normal standing in ten subjects. The procedure adopted involved measuring the continuous oscillations of the centre of gravity of the body. It is considered that measuring the movement of the centre of gravity is the most logical way to measure the postural sway that occurs during standing in the erect posture. It was found that by far the greatest amount of sway occurred in an antero-posterior direction; with the foot position adopted in these experiments (heels set 20 cms. apart), very little lateral sway was observed. To obtain an estimate of the angular displacement of the body occurring at the waist during postural sway, an inclinometer was mounted on a waistbelt. It was found that the angular displacement measured was considerably in excess of the angular displacement estimated on the assumption that the body swayed rigidly between the ankle joint axis and the centre of gravity. It was concluded from this observation that, contrary to what has been commonly assumed, the body must sway in a pliant manner over the ankle joint axis, and not as a rigid column. If the postural sway is measured by detailed planimetry of the records, and the results expressed as a mean deviation of the movement of the centre of gravity from a fitted baseline, the

relative constancy of the degree of body sway is noteworthy. In the subjects examined, the difference between the minimum and maximum antero-posterior sway recorded (expressed as a mean deviation of the movement of the centre of gravity) was less than 3.5 mms. It is concluded that the body is remarkably successful in resisting the effects of gravity, and in maintaining the normal static equilibrium associated with the erect posture.

The effect of load carriage on the normal standing posture was studied using a miniature camera photographic technique. The displacement from the vertical of four marked anatomical landmarks (right shoulder, hip, knee and ankle) was measured on enlarged negative images, during normal standing and when loads of 12 kg. and 24 kg. were carried. It was found that there was a highly significant forward displacement of the shoulder with load carriage, while there was no significant change in the position of the hip. There was a highly significant backwards displacement of the knee, while the position of the ankle was not significantly affected. A weight of 24 kg. was found to cause further forward displacement of the shoulder when it was carried low on the back than when it was carried high on the back. It was concluded that a highly significant realignment of the body position takes place during load carriage above a certain minimum weight, thus contradicting the findings of previous workers who had claimed that the body adapted itself to load carriage without significant realignment of the body segments. The suggestion was made that the trunk acts as a kind of counter-balance, altering its inclination according to the position of the

load on the back, so that the vertical projection of the combined centre of gravity to the feet remains relatively undisturbed.

The antero-posterior displacements of the centre of gravity of the body (postural sway) of ten subjects was studied under varying load conditions. The rate of working of the muscles controlling postural sway was arbitrarily assessed on a mechanical basis, using a strain gauge accelerometer mounted on a freely-suspended platform. It was found that carrying a load of 12 kg. in a position high on the back caused a significant increase in the amount of postural sway of the body, and also in the rate of working of the muscles controlling the sway. The addition of another 12 kg. caused further significant increases in both the sway and the rate of working of the muscles. The movement of a 24 kg. load from a position high on the back to a position low on the back, over a distance of 30 cm., did not significantly affect either the amount of postural^{sway}/recorded, or the rate of working of the postural muscles. It was concluded that the site of a load on the back did not affect the amount of work that was performed by the muscles maintaining the erect posture. The body simply altered its normal alignment according to the position of the load, so as to keep the vertical projection of the centre of gravity at the feet in a relatively constant position.

It is concluded that the maintenance of the upright posture in Man is a dynamic process, and should not be regarded as primarily a passive phenomenon. Gravity is always threatening to upset Man's postural equilibrium, and this equilibrium depends upon, and is largely controlled by, the continuous contractions of the postural muscles.

The interplay between gravity and the restraining muscular contractions produces the normal postural sway, which is usually remarkably small in amount. During load carriage, wherever a load may be positioned on the back, the body adapts itself to one position of the load as well as to another, and succeeds in keeping the vertical projection of the centre of gravity relatively constant. It is suggested that this successful adaptation is achieved by the pliant manner in which the body reacts to the stresses of gravity and load carriage.

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