EVALUATION OF THE DEGREE OF DISABLEMENT OF PATIENTS SUFFERING FROM MULTIPLE SCLEROSIS AND SIMILAR CRIPPLING DISEASES IN TERMS OF MECHANICAL PARAMETERS AND THE EFFECT ON THESE PARAMETERS OF EXTERNAL MECHANICAL AIDS.

- by -

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SUMMARY

A study is made of the muscular strength of able-bodied people, as a basis for the comparison with the strength of disabled subjects. From the study, the ratio of the strengths of different muscle groups is found, and an empirical equation is derived, linking muscular strength with the general physique of the subject.

Two unenergised orthotic aids are then examined, and their effects on the performance of disabled subjects are assessed.

Finally, the muscular strength is expressed logarithmically in decibel units; enabling a more meaningful and direct comparison of the results obtained in the thesis, by means of a logarithmic scale, on which the muscular strengths are displayed visually.

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LIST OF DIAGRAMS

Nomenclature.

B	A function of the somatotype of the subject.
d	The length of the thigh link.
e	The length of the lower leg link.
h	The index of the function of the height or stature of the subject.
H	The height of the subject.
l_1, l_2, l_3	The distances of the centres of gravity of the upper body from the hip joint, of the thigh from the hip joint, and of the lower leg from the knee joint.
L_1	A backrest dimension on the chair aid shown in Figure (31).
L_{H2}, L_{H4}, L_{H5}	The perpendicular distance from hip joint to the reactions normal to the chair seat and normal and tangential to the footrest.
$L_{K4}, L_{K5},$	The perpendicular distances from the knee joint to the reactions normal and tangential to the footrest.
$M(5)$	The sum of the maximum muscular moments (as specified in Chapter 2) exerted at five body joints.
$M_N, M_H, M_K,$ M_S, M_E	The muscular moments exerted at the neck, hip, knee, shoulder, and elbow joints respectively.

r ------------- The co-ordinate of rotundity on the Sheldon Triangle.

R_H ------------ The force acting between the hands and the handrest.

R_4 ------------ The force acting between the feet and the footrest.

R_{HH}, R_{HV} ------ The horizontal and vertical components of R_H.

R_{4H}, R_{4V} ------ The horizontal and vertical components of R_4..

S ------------- The muscular moments expressed logarithmically in decibels.

t ------------- The co-ordinate of thinness on the Sheldon Triangle.

w ------------- The index of the function of body weight of the subject.

W ------------- The body weight.

W_T ------------ The body weight minus the weight of the feet.

W_1, W_2, W_3, W_4, W_5, W_6, W_7 ------ The weight of the component segments of the body, i.e. the upper body, the thighs, the lower legs, the feet, the upper arm, the forearm and the hand.

α_1 ...... The angle of the backrest of the chair.

α_2 ..... The angle of the seat of the chair.

$\beta_1, \beta_2, \beta_3, \beta_5, \beta_6$ ... The angles of the respective body links.

$\Delta_1 - \Delta_7$ .... The additional "non-equilibrium" force, acting between the chair mechanism and the respective body segment.

$\sigma_N, \sigma_H, \sigma_K, \sigma_S, \sigma_G$ ...... The standard deviations of the values of the muscular moments at the neck, hip, knee, shoulder and elbow joints.

$\Lambda_{FI}, \Lambda_{FII}, \Lambda_{FIII}$ .. Functions of the kinematics of the chair mechanism, used to calculate the moment of a force acting on the footrest.

Λ_{KR} .... A similar function, used for the kneerest .

$\Lambda_{HI}, \Lambda_{HII}$ ..... Similar functions, used for the handrest.

θ ..... The angles of the forces

Ω .... The seat hinge torque.

Chapter 1
Introduction

The work carried out in this thesis is based on relatively accurate measurement of data, relevant to the performance of able-bodied and disabled subjects in carrying out certain physical tasks; and the effects of unenergised orthotic aids on their performance. With the accumulation of enough data, certain trends and correlations emerge, from which information on the use, the design, and the effectiveness of these aids is derived. As a spin-off from the research on this limited application to two orthotic aids, there is an accumulation of more general data and techniques which can and, it is hoped, will be expanded, refined, and used, not only in assessing the performance of other aids, but in other fields altogether.

The thesis is split into four sections. Firstly there is an analysis of muscular strength, based on measurements taken on able-bodied subjects, of their strength in five major muscle groups. From these measurements, the ratio of the strengths of the muscle groups is found; and an equation is synthesized, relating the muscular strength to the physique of the subject.

In the second section, the effect is studied of an orthotic aid to help restore normal function to the arms of partially paralysed subjects. The effect is studied of the aid, as used therepeutically for hemiplegics. Here the aid gives the arm mobility, enabling a patient to exercise on his own. Also, the effect of the aid is studied on the ability of a subject to lift an object up to his mouth. The latter movement is regarded as being the most important function of the arm.

The third section is a study of rising, and sitting down in a chair; and the effectiveness of an aid to rising and sitting down is examined. In this section, an analysis is carried out firstly of unaided rising, and secondly of rising in the chair aid; and the two strength requirements are compared.

The final section is devoted to a new method of expression of muscular strengths: as a logarithmic function. This is in a similar manner to the expression of the loudness of a sound in the field of accoustics. The muscular strength scale is graded in decibels, and has the advantage over other quantitative ways of expressing muscular strength in that it magnifies the spectrum of the lower range of strength values, enabling them to be expressed in greater detail; and minifies the upper range of strength values, allowing them to be expressed in a better perspective.

The first and last sections of the thesis in particular are promising for wider application, and it is hoped that further work will be continued along these lines.

Chapter 2
Muscle Strength Testing.

2.1.1. Voluntary muscular strength will depend, by its very nature, not only on physical parameters of the subject, which may be measured, but also on unpredictable psychological factors which will control the degree of effort the subject puts into a voluntary muscle contraction.

In the following chapter, an account is given of an attempt to measure objectively the strengths of different muscle groups of a number of subjects by minimising as far as possible variations in the measurements due to psychological factors, and by treating the measurements statistically. Provided that there is not too great a scatter in these results, it should be possible to use them, for example, to deduce a figure for the percentage residual strength of a partially paralysed subject by measuring the strength of a muscle group, and comparing it with the expected strength of a healthy subject of the same physique.

In measuring the isometric force applied in a given direction by a subject at a point along one of the body links, the force will probably be dependent on a number of factors:

1. The physique of the subject.
2. The geometry of the body link and of the muscle group applying the force.
3. The position along the link at which the force is applied.
4. The ratio of the strength of this muscle group to the general strength of the body.
5. The age of the subject, his state of health, and the presence of any muscular fatigue.
6. The method of measurement of the applied force.
7. Any psychological effects.

A series of tests was carried out on able-bodied subjects, simply to measure the maximum force which they were capable of exerting at points on various body links. The tests were carried out using a fixed procedure, to minimise psychological effects, and to standardize the method of measurement, and the direction and manner of application of the force; and this includes the geometry of the body link under test, and the support given to other parts of the body affected by the force.

As other influences have been minimised, the magnitude of the force exerted will now be a function of the physical parameters of the subject, i.e. the height, weight, build, and age of the subject. Having tested a large number of subjects, the results, providing that they are fairly repeatable, may be expected to give two useful end products:-

1. The ratio of the strengths of different muscle groups may be found.
2. An equation may be synthesised by which the strength of a subject may be determined from his age and physique.

Using these, it should be possible to predict in any subject the expected strength of a given muscle group. The residual strength of partially paralysed subjects may be found by comparing directly his expected strength with his measured strength.

The results of these tests will be in statistical form; given by an average figure with a standard deviation. The success of this technique will depend on whether this deviation is too large to allow the residual strength to be quoted with an accuracy greater than the subjective measurement techniques that are used at present.

2.1.2. Outline Of The Tests.

It is not practicable to measure the strength of every muscle group in the body. Five major muscle groups were tested. They were those causing:-

a. Neck flexion
b. Hip extension
c. Knee extension
d. Shoulder horizontal adduction
e. Elbow extension

In each case, the force exerted by the muscle group was measured vertically or horizontally at a point along the link, perpendicular to the axis of the link, and in the direction of the attempted movement. All the movements tested were simple link rotations about a fixed joint, with the exception of neck flexion; and so for each test the distance of the force from the joint centre was measured, and the results were expressed as the moment of the force about the joint. Neck flexion is a more complex motion, involving flexion of the cervical region of the spine and forward rotation of the skull. The link length was simplified here to the vertical distance from the upper surface of the shoulders, at a fixed width, to the point of application of the force. As most of this force is supplied by the sternocleidomastoid muscles, which flex the cervical vertebral column, the upper surface of the shoulders is a good approximation to the centre of rotation. Neck flexion was chosen to be tested in view of the fact that the muscles of the neck are usually affected less severely than the other muscle groups by paralysing diseases.

2.1.3. Position Of The Subject In The Tests.

A standard position was fixed for each body link during the tests. The subject was seated in an upright posture, with the thigh link horizontal and the leg link vertical. Shoulder horizontal adduction was carried out with the whole arm held straight out in frontof the shoulders, and level with the shoulder joint. Elbow extension was measured with the upper arm vertical, and the forearm horizontal, its axis in the sagittal plane. The force in each case was measured perpendicular to the link axis, at a point near the end of the link. This will keep the magnitude of the force to a minimum, and reduce any discomfort caused by its application.

The subject was strapped into the chair during the tests to support his body as he was exerting a large force at one point. The chair itself comprised a short horizontal seat, supporting the buttocks, and a vertical backrest reaching to the shoulders.

A lap strap held the pelvis firmly in the seat, while not impeding thigh movements, and a chest strap held the trunk against the backrest.

The strap from the hip-extension dynamometer also serves as a constraint for the thigh during knee extension. (A large horizontal force at the bottom of the leg would require hip flexion forces to maintain equilibrium. This effectively is supplied by the thigh constraint). During elbow extension, the upper arm is strapped to the back of the chair, to relieve the shoulder of the need to apply extension forces; and during shoulder horizontal adduction, a symmetrical reaction to the forces set up is provided simply by applying a similar force with the other arm. These arrangements can be seen in Figures (1) to (5).

2.1.4. The Testing Apparatus.

The apparatus itself is a rigid box-like structure of Dexion, approximately cubic, of side seven feet wide (Figure (1-5)). In the centre, attached rigidly to the rest of the framework, is the chair. This has a short seat to allow free movement of the thigh, and long legs so that the feet of the subject clear the ground. On the outside framework are dynamometer attachment points. The dynamometers used are circular spring balances. These have a relatively low extension rate, and thus the muscular contraction will be close to isometric. The dynamometers are attached to the frame by a universal joint, and between the dynamometer and the body link is a flexible strap. This arrangement eliminates lateral forces on the dynamometer. Different sizes of subject are allowed for by adjusting the length of the straps, and by attaching them to the limb of the subject at a point along its length perpendicularly opposite the dynamometer attachment point.

2.1.5. The Force Measurement.

Two methods were considered for force application: either by resistance of the muscle group to a force acting on the limb, or by application of a force by the muscle group. The former would be carried out by applying a steadily increasing force to the end of the body link, the subject resisting any motion and thus applying an equal and opposite force. When the subject is unable to resist any further, the maximum force is recorded automatically. This method has its disadvantages; firstly in that the measuring apparatus will be complicated. Secondly we are more interested in the active force which can be applied, rather than the passive resistive force; so it was decided to allow the subject simply to apply the force himself.

Another important variable will affect the force applied - the psychological parameter. It was found during trial tests that merely by altering the tone of voice, or the wording used in asking the subject to apply a force, or even by allowing him to see the face of the dynamometer so that he knows how much force he is applying, it is possible to double the maximum force exerted. Throughout the tests, it was necessary to adopt the same tone of voice and wording in instructing the subject; this presented a particularly great difficulty when the understanding of the instructions by the subject seemed inadequate.

2.1.6. Analysis Of The Population Tested.

The subjects who volunteered for the tests were of two main groups: students at the University of Manchester Institute of Science and Technology, the majority of whom were aged between eighteen and twenty-five; and technicians from the Institute, who were mainly middle-aged. Twenty-eight males and one female were tested. But apart from these obvious weaknesses in the sample, there was a fair distribution of weaknesses and strengths among the subjects tested.

2.1.7. The Testing Procedure

Firstly a record of the weight, height, age, and somatotype was taken; whether the subject was left or right handed, and left or right footed. (If he were left handed, tests were carried out on his left arm, etc.). His profession or occupation and any physical pursuits he followed were also recorded. His somatotype was assessed on the Sheldon Triangle, which splits physique into three trends:- muscular, rotund, and thin. As shown in Figure (7), five increments were used between each extreme of build, so a subject classed as R_1T_5 will be extremely rotund, whereas one classed as R_4T_4 will be of average build with a slight tendency towards muscularity.

Having completed the record, the subject was strapped into the chair, and the neck flexion strap was placed around his forehead. The vertical distance from the strap to the shoulders was then measured with the device shown in Figure (1). This device incorporates two prongs, nine inches apart, which are placed on the shoulders. A pointer is located to the centre of the forehead strap; a scale along which the pointer slides indicates the vertical distance of the shoulders to the forehead strap, and a plumbline ensures that the device is vertical. The subject was then asked to push forward as hard as possible against the strap. This he was asked to repeat until a fairly consistant maximum force value was obtained. The maximum force was judged to be that force which could be maintained for approximately two seconds i.e. a steady force and not a jerk.

Measurements were taken, as shown in Figures (1) to (5), for the other muscle groups. Link lengths were measured from the centre of the strap to the joint centre. These were located from landmarks in the bone. Those used were:-

Neck Flexion :- Upper surface of the shoulders at a width of nine inches.

Hip Joint :- A point at the tip of the femoral trochanter 0.4 inches anterior to the most laterally projecting part of the femoral trochanter.

Knee Joint :- Mid-point of a line between the centres of the posterior convexities of the femoral condyles.

Glenohumeral Joint:- Mid-region of the palpable bony mass of the head and tuberosities of the humerus.

Elbow Joint :- The lowest palpable point of the medial epicondyle of the humerus..

For measurements of vertically applied forces i.e. hip extension and elbow extension, the subject was asked to relax the limb prior to the force application in order to measure the force at the dynamometer due to the weight of the limb. This force was subtracted from the force applied. All measurements were taken in pounds force and inches, and the joint moment for each muscle group was calculated simply by multiplication of force applied at the point along the link by the distance of the point of application of the force from the appropriate joint centre.

2.1.8. The Strength Ratios of the Muscle Groups

The moments exerted by the five muscle groups were summed to give an indication of the total strength of the subject, and the ratios of the strengths of the muscle groups were obtained by dividing the moment of each individual muscle group by the total strength. The figures obtained from the tests are given in the appendix (section (1)). The results are shown graphically in Figure (9) in the form of a histogram, each point representing an individual muscle group ratio. Thirty-two sets of results are plotted; those being the twenty-nine subjects tested, plus a repeat of the tests on the first three subjects.

It was found that each set of muscle group ratios is similar to a normal distribution about the arithmetic mean. The three weaker muscle groups (neck flexion, shoulder horizontal adduction, and elbow extension) have smaller mean deviations and ranges, but greater percentage deviations and ranges. The arithmetic means, mean and percentage deviations and other data, are given in section (1) of the appendix.

The intention of these tests is to utilize these figures for predicting the strengths of other muscle groups and multiplying by the strength ratio. This will give a figure for the residual strength of the muscle group, with a standard deviation which may also be calculated. The usefulness of these figures will depend on the magnitude of this deviation.

In the appendix (section 1) an analysis is carried out to find the magnitude of the standard deviation in the percentage residual strength of the knee in applying an extension moment. The strength of the knee is compared with that of the hip, elbow, shoulder, and neck, assuming these latter to be unaffected by any paralysis.

It is found that from these measurements the residual strength at the knee, expressed as a value R% can be given to an accuracy of $\pm$ 0.27R i.e. for a subject with a residual strength measured by this technique as 20%, this figure may be quoted as 20 $\pm$ 6%.

2.2.1 The Strength Equation

We have data for 29 subjects on their general physical strength, as given by thesum of the strengths of five muscle groups. We also have a record of the physical parameters for each subject i.e. the body height, weight, build, and age. It is reasonable to assume that these parameters will affect the physical strength to a greater or lesser extent, and that given the conditions of the tests - that all the subjects are able-bodied, and that psychological factors are minimised; we can assume further that it will be possible to synthesize an equation giving the strength of the subject as a function of these four parameters.

2.2.2. Synthesis of the strength equation

We can write an equation linking the strength with the physical parameters in the form:-

$$\frac{M(5)i}{\overline{M(5)}} = f_1\left[\frac{Wi}{\overline{W}}\right] \cdot f_2\left[\frac{Hi}{\overline{H}}\right] \cdot f_3\left[\frac{Bi}{\overline{B}}\right] \cdot f_4\left[\frac{Yi}{\overline{Y}}\right]$$

where $M(5)i$ is the strength of the subject in the five specified muscle groups.

$\overline{M(5)}$ is the average strength of the population.

$f_1\left[\frac{Wi}{\overline{W}}\right]$ is a function of the body weight of the subject divided by the average body weight of the population.

$f_2\left[\frac{Hi}{\overline{H}}\right]$ is a function of the height of the subject divided by the average height of the population.

$f_3\left[\frac{Bi}{\overline{B}}\right]$ is a function of the build of the subject divided by the average build of the population.

and $f_4\left[\frac{Yi}{\overline{Y}}\right]$ is a function of the age of the subject divided bythe average age of the population.

The equation will be empirical, and will be synthesized by giving each function a basic form, then finding the value of the function and its associated constants which give the closest fit to the experimental data. Unfortunately there will be a fairly large indigenous error present, due simply to the psychology of the subjects i.e. how much effort they put into exerting the forces. However, the larger the number of subjects tested, the better will be the average of the results, and the more accurately will the parameters of the strength equation be calculated.

Let us look firstly at the body weight function $f_1\left[\frac{Wi}{\overline{W}}\right]$. We must synthesize an expression which is likely to express the effect of the body weight on the muscular strength; and then solve the numerical constants of the expression from the data.

One factor determining muscular strength is the number of muscle fibres in each muscle: the greater the number, the greater is the strength of the muscle. Also, the greater the number, the greater is the weight of the muscle. This implies that as the muscle weight increases, so will the muscular strength; and that as a large proportion of the body consists of muscle, it seems probable that the muscular strength is proportional to some power of body weight, at least in health.

$$\text{i.e.} \quad \frac{M(5)i}{\overline{M}(5)} \propto \left[\frac{Wi}{\overline{W}}\right]^{w}$$

where w is the body weight index.

It is not clear however, at first sight, which way the body weight, height, build, and age will affect the muscular strength. If one were to corelate the strength of the subjects with any one of these parameters, one would probably be able to predict, for example, that the strength would be directly proportional to the body weight. However, as we are taking into account all the parameters simultaneously, this need not be the case. At the moment this is not important, for even if the muscular strength decreases as the body weight increases, the value of w will become negative, and the expression $\left[\frac{Wi}{\overline{W}}\right]^{w}$ will still be adequate.

A similar form of function may be used for the effect of the height of the subject on the strength:

$$\text{i.e.} \quad \frac{M(5)i}{\overline{M}(5)} \propto \left[\frac{Hi}{\overline{H}}\right]^{h}$$

where h is the height index.

The next function to be synthesized is the build of the subject. As mentioned, each subject is assessed on a Sheldon Triangle (Figure 7), and is given two numbers to represent his build; the first showing his tendency towards rotundity, and the second showing his tendency towards thinness. The two numbers together will show his tendency towards muscularity. The first number will be between 1 and 5: 1 showing the greatest tendency towards rotundity, and 5 showing the least. Representing this number by r, we can say that his rotundity is proportional to $(5 - r)$. Similarly, representing the second number by t, his thinness will be proportional to $(5 - t)$, and his muscularity by $(r + t - 6)$. The function representing build may be adequately represented by:

$$\frac{R(5 - r_i) + T(5 - t_i) + M(r_i + t_i - 6)}{R(5 - \bar{r}) + T(5 - \bar{t}) + M(\bar{r} + \bar{t} - 6)}$$

where R, T, and M are constants which will be found from the data. It will be expected that M will have the largest value - subjects with the highest value of $(r + t - 6)$ having the highest proportion of muscle in the body. R and T will be lower. Subjects with a higher value of $(5 - r)$ will have a high proportion of fatty tissue in the body; while those with a high value of $(5 - t)$ will have a high proportion of bone and other organs.

Finally, a function expressing the effect of age on strength must be synthesized. Figure (10) shows the form the function is expected to take. We can expect the strength to be small at birth, increasing quickly to a steady maximum, and dropping off at old age. The function might be represented by two portions of a sine curve, the first lying between the age of 0 and the age of maximum strength Ym, and the second between the maximum strength age, Ym, and 100 (100 being taken as an arbitrary upper age limit).

The function is written $\sin^y\left(\frac{Ym - Yi}{Ym}\right)$ for values of Y less than Ym, and $\sin^y\left(\frac{100 - Yi}{100 - Ym}\right)$ for values of Y greater than Ym. The function is given an index, y; the higher is the value of y, the sharper will be the maximum peak; while for low values of y, the plateau of maximum strength will extend over a large number of years.

The full equation now becomes:-

$$\frac{M(5)i}{\overline{M(5)}} = \left(\frac{Wi}{\bar{W}}\right)^w \cdot \left(\frac{Hi}{\bar{H}}\right)^h \frac{R(5 - r_i) + T(5 - t_i) + M(r_i + t_i - 6)}{R(5 - \bar{r}) + T(5 - \bar{t}) + M(\bar{r} + \bar{t} - 6)}$$

$$\times \left(\frac{\sin(F(Yi))}{\sin(F(\bar{Y}))}\right)^y \text{ where } F(Yi) = \frac{Ym - Yi}{Ym} \text{ when } Yi < Ym$$

$$\text{and } = \frac{100 - Yi}{100 - Ym} \text{ when } Yi > Ym$$

and $F(\bar{\bar{Y}})$ is the same function with $\bar{Y}$ substituted for Yi.

The equation is solved by computer. The technique used, and the full computer programme are given in the appendix (Section 2), together with a sample of the results obtained.

2.2.3. Analysis of the Population Tested

The ideal population to test would be a cross-section of humanity, with proportional representation of all races, sexes, ages, and occupations. Unfortunately, most of the subjects tested were male Caucasians, most of whom were between the ages of 18 and 25, and were students by occupation. This imbalance in the population makes it impossible to include a sex factor into the equation; and also the age factor is unreliable, as it is based on too few subjects at either end of the age spectrum.

The population includes, however, a good range and distribution of the height, weight, and build of the subjects; so the full effect of these parameters on the strength equation constants will be manifest.

2.2.4. The Strength Equation In Its Tentative Form

The equation, found from the data of the 29 subjects tested was as follows:-

$$M(5) = W^{-0.2}.H^{1.5}(4.44r + 4.60t - 11.75)$$

where M(5) is the strength of the five muscle groups specified in in-lbf.

W is the body weight in lbf.

H is the body height in inches.

and r and t are co-ordinates of the build, as given on a five point Sheldon Triangle.

The absence of the age function should be noted. The optimum value of the index of the function y, is found to be zero. This in effect means that the strength of those subjects tested is unaffected by their age. As mentioned previously, the range of ages tested is not good; and the absence of an age function will be true for this particular group tested, but is not necessarily true for humanity as a whole. Also there is no account of the sex of the subject. There is no reason, however, why the height, weight and build functions should not be a good representation of their effect on the strength of the subject, and the equation will be valid for calculating the expected strength of a subject.

Figure (11) shows the values of the strengths of the 29 subjects tested, as calculated from the strength equation, plotted against their measured strengths. The standard deviation of the calculated values from the measured values of the strength was found to be 15.5%. This error will probably be close to the indigenous error, mentioned previously, that will be present no matter how large a population is tested.

2.2.5. Application of the Strength Equation to the Estimation of Residual Strengths of Pathological Subjects.

In the appendix (Section 1), a technique is used for calculating the expected strength of an individual muscle group from the data obtained in the first part of the muscle strength tests. The strength equation may now be used to simplify this technique, and enable us to gauge the able-bodied strength of a muscle group of a subject to approximately the same accuracy of 27% standard deviation (i.e. the strengths of approximately 2/3 of all subjects will lie within 27% of the calculated value). The advantage will be that no actual measurements of the strength will be required, i.e. knowing the height, weight, and build (and also the age and sex) of a subject, the healthy strength of a muscle group may be calculated, and may then be compared directly with the residual strength of the muscle group, which may be simply measured.

This technique is also far better in the case of subjects partially paralysed in more than just one muscle group, as is so often the case. The estimation of the residual strength will be independent of any measurements on the rest of the body.

2.2.6 Suggested Further Work

It is possible that this equation and technique could be of some use in the fields of bioengineering, medicine, or physiotherepy. Before the results are used, however, it is suggested that the muscle tests are continued for a far larger number of subjects; and more important, for a broader range of types of subjects. This should be extended to cover all the major muscle groups, which would increase the usefulness of the technique. Should there be any demand for the use of this technique, the equation could even be expressed as a set of tables.

Finally, the magnitude of the indigenous error should be gauged, by repeatedly testing a few subjects, and observing how the results vary with each subject over a period of time.

~~Chapter 3~~

3.1.1. Tests On The Floating Arm Support.

The floating arm support (References 2, 5, 13) is a device designed to supply a vertical and near constant force to the forearm, while allowing at the same time free movement of the arm.

The device was designed for people suffering from multiple sclerosis and similar crippling diseases, whose arms are not strong enough to function normally, and who are either too weak to lift small objects, or cannot even support their own arm weight. The device is attached to a wheelchair, and is thus used in a seated position. The principle on which it works is to convert the potential evergy of the arm as it moves down into strain energy in a set of springs in the mechanism. Here the energy is stored until the arm is raised, whereupon the strain energy of the springs is returned to the arm, increasing its potential energy. In effect, the only work required to raise and lower the arm is a very small amount needed to overcome friction losses in the system. The effect of the device on the arm is to produce a feeling of floating.

The device as yet has had limited general use, and has received a mixed reception from the partially paralysed people who have so far used it. The feeling expressed most often is that the arm supports are good, but that the person himself would not want to use them (Reference 5). This may be partly due to psychological reasons, but it was also thought possible that this lack of acceptance was due to some biomechanical reason for which the aid does not in fact live up to its expectations.

A more promising use for the device is as an aid to physiotherapy in rehabilitating patients suffering from hemiplegia. In hemiplegia, half of the body, either the left or the right side, becomes almost totally paralysed. The degree of recovery and the speed of recovery are both affected by the amount of physiotherapy in the form of passive and active movements of the muscles which can be given to the patient. In the initial stages of recovery the patient is usually too weak to carry out significant movements of the upper limbs unaided; so a physiotherapist is required to move the weak arm for the patient. With the arm support the weight of the arm is supported, and the patient now requires very little strength to move the arm in three dimensional motion. The patient can thus be left on his own to exercise himself, saving the time of the physiotherapist, and increasing the time which he receives physiotherapeutic treatment. Initial trials with the floating arm support have proved to be very encouraging. What is now required is concrete evidence in the form of a controlled experiment to show whether therapy using the arm support does in fact increase the speed and degree of recovery as compared to orthodox therepeutic techniques.

Thus we wish to make objective measurements of the recovery of power in the arm using the aid therepeutically, and also to examine the effect of the aid on the biomechanical performance of the arm when in everyday use. To assess these, we would like to measure the position of the arm and the magnitude of the maximum vertical force applied by some point on the arm.

3.2.1. Assessing the Biomechanical Effect of the Arm Support on the Arm.

In everyday use, the arm support will be required to increase the mobility of the arm, increase the control which the user has over his arm, and increase the lifting power of the arm. Increased mobility of the arm will be required for increasing the range of the hand, increased control over the arm for better control of movements at the hand, and for increased lifting power as applied by the hand. The latter is regarded as the most important benefit brought about by the arm support, and so apparatus was designed primarily to measure this, but also to be capable of assessing the mobility and the degree of voluntary control. By far the most common movement carried out by the hand is in lifting objects from just above the thighs up to the mouth, as used for example when eating. This will be referred to as the "hand to mouth movement", and the path taken referred to as "the hand to mouth path".

3.3.1. The Arm Support Test Apparatus

The test apparatus should be capable of measuring the vertical isometric force applied by the hand, and at the same time be free to move in any direction so that the subject may select his own hand to mouth path.. To allow the hand free vertical movement and also to be able to measure the vertical force, it is necessary to be able to lock the measuring device in the vertical direction, and thus to measure the force at integral points along the hand to mouth path rather than continuously measuring this force.

At the beginning of the hand to mouth path, the hand is close to the top of the thighs. This makes it difficult to accomodate a dynamometer beneath the hand under test. To avoid this a compressive dynamometer was used, placed above the hand. As we wish to measure the isometric force at the hand, a deflection-free dynamometer is required. A piezo-electric dynamometer rather than a spring balance was therefore used. This also has other advantages. For example, it is possible to feed the results automatically on to a permanent record, which will introduce other interesting parameters into the results; the force will now be plotted against the time, and this will show whether any long term effects of the force on the time are present due to fatigue, and short term effects due to muscular tremor.

To allow movements of the hand in the x, y, and z direction, the mechanism shown in Figure (12) was constructed. This consists of two freely jointed links attached to a fixed support. This arrangement allows the point A to move over the whole horizontal range of motion required. A telescopic rod is attached to the free end of the links to allow vertical movements. The rod can be locked in a rigid position during actual measurement of the force, and unlocked to allow free movement of the hand. It is made from aluminium channel sections for lightness, and the locking device is a friction pad in each section, which may be operated by levers attached to the rods, or by remote control by brake handles connected to the rod by Bowdon cables. This is used when it is required that the tester should not affect or influence the hand to mouth path.

3.3.2. Measurement of the Hand Force

The dynamometer is fixed above the telescopic rod as it isnecessary to have as little movement as possible of the output wire. It is of a piezo-electric type; a static charge being generated proportional to the load applied between the end faces. A screened lead connects the dynamometer to a high impedance D.C. amplifier which amplifies the signal. The output from this is recorded on a U.V. recorder. As the output from the dynamometer is a minute D.C. charge of the order of a few micro-amps, the output lead from it is a heavily insulated, oil filled coaxial cable, and the impedance of the amplifier is very high, to minimise leakage of the charge. Even so there is some leakage, shown by a slow drift of the output reading. Prior to each force measurement, the dynamometer must be grounded. This eliminates any charge which has built up due to drift.

3.3.3. Measurement of the Hand Position

It is desirable to make the system of measurement of the hand co-ordinates as simple as possible, at the same time, however, bearing in mind the practical difficultie of measurement. We should like to locate the co-ordinate system to a point on the body of the subject under test. This would make the test independent of the apparatus, and it would produce slightly more meaningful results. However, the practical difficulties involved in fixing the co-ordinate system to the body are too great. The co-ordinates would need to be adjusted for each subject, and possibly even for each force measurement. A compromise is used in which the co-ordinate system is fixed relative to the chair, while as far as possible the position of the subject in the chair is also fixed. We cannot constrain the subject in any way, for example with belts, as this would probably affect his biomechanical performance. However, as the most important part of the body to fix to the reference co-ordinates is the head, a headrest is provided (shown in Figure(12)). This was adjusted for each subject to be comfortable in a relaxed sitting position. The co-ordinates were not fixed with the headrest as a reference, as again this would mean adjusting the apparatus for each subject. The headrest served, however, to locate the head in a consistent position during the testing. The actual reference co-ordinates were taken from the frame of the chair.

Measurement of the x, y, and z co-ordinates present a difficult problem. As the force output is being recorded on a U.V. recorder, it is desirable to record the co-ordinates of the hand on the same output. Due to difficulties involved in getting a direct electrical output of the hand position, the co-ordinates are indicated visually, and are transferred to the recorder by the tester, who dials the co-ordinates on a hand dial (shown in Figure (12)), which produces a permanent record on the U.V. output. Cartesian co-ordinates are used rather than polar co-ordinates, as they will be more meaningful during analysis of the results. An x - y co-ordinate scale was drawn out and was fixed to lie above the free links. A pointer attached to point A indicates the x and y co-ordinate readings (as shown in Figure (12)). The z co-ordinates are marked on a scale alongside the telescopic rod, and a pointer indicates on the scale, the z co-ordinate of the highest point of the hand. The scales are all marked in inches. Figure (13) shows the reference points of the co-ordinates. The x-axis lies across the seat, with x = 0 the centre line of the seat. The y-axis lies along the seat with y = 0 at the line of intersection of the wheelchair seat and the backrest; and the z-axis is vertical with the origin again at the line of intersection of the wheelchair seat with the backrest.

The co-ordinate hand dial circuit is shown in Figure (14). In the hand dial itself are the rheostat, the resistor, the battery and the switch. The face of the dial itself is marked out in the co-ordinate numbers. The tester moves a pointer round to the number to be read in, and then presses the switch. The rheostat is connected to the pointer, and thus a current, dependent in magnitude on the position of the pointer, runs through the circuit. This produces an output on the U.V. recorder.

Two perspex scales were made to simplify reading the U.V. recordings: one for measuring the magnitude of the force output. This was calibrated to read directly in pounds force. The other, for measuring the co-ordinates of the hand, was calibrated to read in inches. This speeded up analysis of the results considerably, enabling direct measurements of the force and co-ordinates of the hand position to be taken from the output trace.

3.3.4. Adaptations of the Apparatus for the Tests

In the first series of tests we wish to measure the maximum lifting force of the hands. Some sort of hand grip is required at the end of the telescopic rod for the subject to push up against. However, we do not want the results to be affected by the ability of the subject to grip, as the strength in his hand may be considerably reduced. To overcome this, a mushroom attachment (shown in Figure (12)) was fitted to the end of the telescopic rod, allowing the subject to hold the stem of the mushroom with his hand, and to apply a force to the underside of the dome of the mushroom. This also has the advantage of fixing the z co-ordinate of the hand with respect to the fixed frame of reference of the co-ordinate system: in the tests the z co-ordinate was measured to the underside of the dome of the mushroom. In later tests, the force was applied at a point on the arm. To allow this, the stem of the mushroom was removed and the force was applied to the dome of the mushroom.

In the second series of tests, the apparatus was to be used in two hospitals, transported weekly from one to the other, and erected and used by physiotherapists and doctors who, one must assume, will be ungifted in the use of the screwdriver, soldering iron, and spanner. The apparatus was thus made as light and as easy to dismantle and erect as possible; and only one connection was required to complete the electrical circuit on erecting the apparatus.

3.4.1. The Testing Procedure

The apparatus is erected with the telescopic rod vertical, and the x, y, co-ordinate canopy horizontal. The z co-ordinate scale is prefixed, but the x and y co-ordinates must be adjusted to the required position. A plumbline is provided for this purpose. It is first suspended from the point (x = 0, y = 0) on the scale to adjust x = 0 to the centre line of the wheelchair. A wooden lathe is then laid across the seat of the chair, and the plumbling is suspended firstly from the point B, and then from point C, whereby the scale can be adjusted, using markings on the lathe as a reference, to bring the x and y axes of the scale parallel to the chair frame, and to locate the line y = 0.

The subject is now seated in the chair, and is asked to sit in a normal, upright, relaxed posture. The headrest is adjusted for comfort in this position. The arm supports are then ~~adjusted~~ adjusted to supply 100% of the force required to support the arm at the mid-point of the hand to mouth path; the support force dropping off slightly as the arm is raised or lowered. A preliminary adjustment is made by weighing the arm on scales, and fixing the number and position of the springs from the arm support calibration graph (Figure (15)). The final adjustment is done by placing the subject's arm in the support, asking him to relax, and it can be seen if too much or too little force is supplied from whether the arm rises or falls. The test is then explained to the subject as follows:-

> "Imagine you are picking up an object, for example a cup of tea from a table in front of you, and are lifting it up to your mouth. Grip the handle of the mushroom, and starting from a position as if it were on a table, bring it up to your mouth. While you are doing this, I will stop you five times along the hand to mouth path, and will ask you to push up as hard as you can."

Where possible, the positions at which the force is measured are the starting position, with the hand on an imaginary table; the final position with the hand at the the mouth; and three evenly spaced intermediate positions. Just before the measurement of the force at each position, the subject is asked to release the mushroom and the dynamometer is grounded.

Two people are required to carry out the tests: one to instruct the subject, and to operate the brakes on the telescopic rod and the other to operate the U.V. recorder and amplifier, and also the co-ordinate hand dial. While the subject is applying the force he dials the x, y, and z co-ordinates, in that order.

Figure (16) shows a typical trace obtained during an actual test. One can see a straight portion of the curve where the dynamometer is being grounded. Moving from right to left, there follows the application of the force by the subject and the three lines representing the x, y, and z co-ordinates. Negative values of x (on the left side of the chair)are indicated by the tester by moving the pointer round the dial and back to zero before dialling in the x co-ordinate. This produces an inverted V on the trace. After the application of the force and the co-ordinate lines, there is a period of irregular forces before the next grounding while the subject moves his hand to a new position. This can be ignored. It is important at the beginning and end of each trace to record a baseline for the force and for the co-ordinates, so that they may be subsequently measured easily.

3.5.1. An Outline of the Purpose of Each Set of Tests and the Preliminary Findings

The first series of tests was carried out at the Frank Taylor Centre For The Disabled. This is a day centre for the disabled to which people of all ages and disabilities come. A total of sixteen subjects were tested here on the apparatus. The first nine subjects were tested for their ability to apply a force with the hand on the hand to mouth path. The last six subjects were also tested with the stem removed from the mushroom, applying a force on the underside of the dome of the mushroom with the wrist, the forearm being held in mid range of pronation-supination. Two subjects were also tested over the whole range of arm motion. The latter test was found to be too lengthy to carry out on every subject, so it was discontinued.

Figure (17) shows a typical trace obtained from these tests; one graph showing the hand to mouth path viewed from the side: the y co-ordinates being plotted against the z co-ordinates. Beneath that is shown the force applied by the subject plotted against the distance from the mouth. The first nine tests were carried out both with and without the arm supports. On subjects with two weak arms, the order of testing was:-

1. Left arm with arm support.
2. Right arm without arm support.
3. Left arm without arm support.
4. Right arm with arm support.

This order of testing would reveal any noticeable increase or decrease in the applied force during testing. In fact it can be seen that there is no significant difference in the force over the duration of the tests.

One immediate and unexpected observation that became apparent from the first set of tests, was that the arm support caused no consistent increase in the applied force. At first sight it would appear that if the subject were capable of applying unaided a force F_1 with his hand, and the arm support were supplying a force F_2, then the total force he could exert using the arm support would be $F_1 + F_2$. It is not valid to quote numerical comparisons between individual tests because of the large variation in individual results. It can be seen, however, that the force applied by the subject using the arm support is not consistently greater than the force applied unaided.

The only notable exception to this is with very weak subjects who are totally unable to raise their arm without the arm support, and are thus unable to supply a vertical force unaided, but were capable of exerting a force when the arm support was used. There were at first thought to be two explanations possible for these findings. The first involves the muscles of the shoulder joint, which are thought to be critical in applying a force at the hand; the hypothesis being that the force exerted by the hand is in general controlled by the strength of the muscles at the shoulder joint and elbow, and that the muscles relieved by the arm support will be different to those used to apply a force at the hand. The biomechanics of this will be examined in detail later in the chapter.

The second explanation could be that a limit is imposed on the force applied by the hand due to comparative weakness in wrist abduction. This is to say that although the arm is capable of applying more force; were it to do so, the wrist would simply adduct, and no extra force would be applied at the hand. If this were so, then in measuring the force exerted at a point on the end of the forearm link, just behind the wrist joint, we should find that the arm support causes a substantial increase in the force.

A second series of tests was carried out to test the latter theory. Measurements of the force were now taken as follows:-

1. Measurement at the hand with the arm support.
2. Measurement at the wrist with the arm support.
3. Measurement at the wrist without the arm support.
4. Measurement at the hand without the arm support.

For subjects with two weak arms, tests were also alternated between the arms. Six subjects were tested in this manner. The results taken, are analysed in the appendix (Section 3).

3.6.1. Conclusions Drawn From The Results Of The Two Sets Of Arm Support Tests.

1. The arm support does not significantly increase the maximum vertical force which can be applied by the hand, in subjects strong enough to support the weight of their own arm, in fact it would appear to reduce it slightly.
2. The maximum force applied at the wrist is significantly lower than the force applied at the hand; indicating that the strength of the wrist is not the limiting factor in

application of the force.

3. Force application appears to be a minimum when the hand is close to the mouth. This factor could be considered in the adjustment of the spring characteristics to raise the position of the maximum support force.

A statistical analysis of the results of the arm support tests which led to these conclusions is given in the appendix (section 3).

3.7.1. An Explanation For The Results.

In applying a vertical force at the hand on the hand to mouth path, with or without using the floating arm support, complicated muscular anatomy and biomechanics are involved. A large number of muscles are associated with the shoulder girdle and arm, all of which may affect the force at the hand to a greater or lesser extent. Section 3.7.2. deals with the function of the muscles in the use of the arm. In section 3.7.3. the biomechanics of the arm in applying a force at the hand is analysed, with the arm support in use, and by two methods without the arm support. From this final analysis, an explanation emerges for the limitations of the arm support in increasing the force at the hand.

3.7.2. A Reference To The Functional Anatomy of the Upper Limb.

This is based on a summary of electromyographical studies of the upper limb over the past two decades (Reference 1); and is given in the appendix (section 4).

3.7.3. The Biomechanics of the Arm With And Without the Arm Support in Use.

Figure (19) shows a plan view of a subject using the arm support, his hand midway along the "hand to mouth path". The arm support exerts a single vertical force R_{AS} at a point on the forearm, usually at about one third its length from the elbow joint. An imaginary object being lifted to the mouth (in the analysis this is assumed to be a British Standard cup of tea of weight 1 lbf) exerts a downwards vertical force at the hand. Figure (19) shows the arrangement diagrammatically.

A value for the weight of each link is now calculated using the figures given in the appendix (section 10), and the centres of gravity of the arm, forearm, and hand links are marked on the diagram as W_5, W_6, and W_7, respectively. The moment about the shoulder joint of each link is now calculated, and a moment vector polygon is constructed as shown. The resultant moment about the shoulder joint is found to lie almost exactly on the same line as the moment exerted by the arm support force about the shoulder, M_{RA}, and to be equivalent to a force of 4.8 lbf acting at the point of support of the arm support. We can assume thus that with the arm support in use, and adjusted correctly, the weight of the whole arm will be supported, and may be ignored. The only other force acting here will be the force at the hand. This is assumed to be of magnitude 1 lbf, and the moment produced by this at the shoulder joint, M_{RH}, is marked on the diagram.

To apply this force, it is necessary to exert a moment at the shoulder of 14.8 in-lbf. in shoulder external rotation, and of 4.5 in-lbf. in shoulder flexion/adduction.

With this configuration of the arm, but without the arm support, moments at the shoulder due to the arm segment weights will be acting, in addition to the moment due to the force at the hand, giving the moment vector polygon shown. Now it will be necessary to apply 31.5 in-lbf. external rotation moment, and 47.0 in-lbf. flexion/adduction moment to exert this force at the hand. This is far larger than the moment required with the arm support in use.

In observing people drinking tea, particularly disabled people, it can be seen that they do not, in general, pick up a cup of tea with the forearm nearly horizontal, as is necessary when using the arm support. The arm will usually be held as in Figure (20), with the upper arm nearly vertical, and the elbow held against the side of the body. A moment vector polygon is constructed, as before, to find the resultant moment at the shoulder when a vertical force of 1 lbf is applied at the hand. As can be seen, a shoulder external rotational moment of

14.3 in-lbf., and a shoulder flexion moment of 40.6 in-lbf. is required. In pathological cases, where the arm is weak, the need to exert the flexion moment can be overcome by holding the arm against the body. The frictional force between the arm and the side of the body will now supply some, or even all of this moment. The rest of the moment at the shoulder, that of adduction, is comparable in magnitude with that required in external rotation when using the arm support. In addition, there will be an elbow flexion moment acting of approximately 28 in-lbf. This may affect the ability of the subject to apply a force when muscles controlling the elbow joint are severely affected. However, as the average subject is stronger in elbow flexion, as is required when the elbow is held against theside of the body, than in shoulder external rotation, which is required when the arm support is in use; this moment at the elbow will probably not limit the force applied at the hand to any great extent. This explains the findings of the tests: that the arm support does not increase the vertical force that can be applied at the hand.

At first, the findings of the second part of the tests were completely inexplicable: that the force exerted at the wrist is significantly lower than that exerted at the hand. It would appear from simple kinematics that this force should be either equal to, or slightly larger than the force exerted at the hand. The explanation for this phenomenon (having caused the author several sleepless nights) now becomes obvious; for it is for the same reason that the arm support does not increase the force at the hand - one could even say it verifies this explanation.

In applying a force at the wrist, the stem of the mushroom, used as a handgrip, is removed; and the force is applied to the underside of the dome of the mushroom. To do this, it is necessary to hold the forearm horizontal, or to suffer the extreme discomfort of applying a force to the edge of the mushroom. This will explain the figures calculated in section (3) of the appendix: that in using the arm support, 22% more force is exerted at the hand than at the wrist, and that without using the arm support, where we are comparing the most natural with the most artificial arm configuration, 40% more force will be exerted at the hand than at the wrist.

3.7.4. Conclusions.

The floating arm support is an excellent device for giving support to the free arm; and as such is especially suitable for exercising the arm in the treatment of certain pathological conditions such as hemiplegia, where the arm is not capable of supporting its own weight against gravity. It can also be used in picking up light objects and for giving mobility to subjects with weak arms. However, for partially paralysed subjects who are able to raise their arm against gravity, the arm support tends to impede rather than help them, as it forces the arm into an unnatural position for lifting objects.

More promising from this point of view is the "close fitting" arm support, at present being developed (Reference (14), which does not restrict the user to keeping his forearm nearly horizontal; and allows a complete range of movement, not only of the hand, as with the floating arm support, but of the whole arm.

3.8.1. Tests to Gauge The Effectiveness of the Floating Arm Support As Used Therepeutically in The Treatment of Hemiplegia

The floating arm support has been used for some time at Oldham and Distric General Hospital, where its effect on the recovery of patients suffering from hemiplegia has been very promising. Tests were organized to produce objective measurements on the rate of improvement of the patients. Co-operating in the tests were Clatterbridge Hospital, who also adopted the arm support for therepeutic use. It was decided to test approximately fifty patients at the two hospitals. About one half at each hospital would act as a control group, who would receive orthodox therepeutic treatment; and the other half would receive treatment on the floating arm support. The patients would be channeled into control or test group on admission to the hospital, based on the month of birth. Each patient is tested fortnightly on the arm support apparatus, to measure his strength in applying a vertical force at the forearm, while his hand traces the "hand to mouth" path. The test apparatus is taken from one hospital to the other each week, and the tests are carried out by physiotherepists, and occupational therepists at each hospital during the week.

The arm support is useful in therepy while the patient has insufficient strength to raise the affected arm unaided. As soon as he recovers sufficient power to do this, the arm support is no longer required. We are therefore interested in measuring, not the force applied at the hand by the patient, but at a point on the forearm at which, by application of a single vertical force, the whole arm may be supported. This point is also the point at which the arm support acts. The force is thus measured on the forearm above the arm support pan pivot, and this force is assumed to be a direct measurement of the ability of the patient to raise his arm.

As with preceding tests, the arm support is adjusted for each patient, to support the arm at the mid point of the "hand to mouth" path, and to exert a maximum force here. The same adjustment parameters are used over the whole period of the tests for each patient. There will be a slight drop off in the arm support force to either side of this mid range position; but as the primary object of the tests is to measure the progress of the patient, a consistent arm support adjustment will enable us to compare the fortnightly measurements directly.

Testing of the patient requires a fair degree of comprehension and co-operation on the part of the patient. He is also required to maintain an erect sitting posture during the test. Involvement of the parietal lobe of hemiplegic patients often produces characteristic syndromes (Reference 7), dependent on whether the dominant or subordinate brain hemisphere is affected. On admission of each patient, a series of tests is carried out to assess any parietal lobe involvement; and only if the patient is free of any involvement will he take part in the tests.

3.8.2. Results Of The Tests So Far.

A number of patients have been tested at each hospital. Unfortunately, for various reasons, a disappointingly small number (four) were tested over a long enough period to produce any meaningful results.

The results of these four are shown in Figure (21). Subjects A, B, C, and D are in the control group, using orthodox physiotherepeutic techniques. It is too early at present to comment further on the results.

3.8.3. Recommendations For The Continuation Of The Arm Support Tests

Based on experience of the progress of the tests so far, the author wishes to make the following recommendations as to the future of the tests.

1. The experiment should, if possible, be extended to other hospitals who are willing to participate, to increase the sample available.
2. The experiment should be limited to in-patients.
3. One set of testing apparatus should be lent to each hospital.
4. The measurement of the position of the hand should be excluded from the tests, or greatly simplified.
5. The use of the floating arm support is not required during the tests, and the apparatus should be redesigned to exclude it.

Chapter 4
Unaided Rising

4.1.1. Unaided rising, like most biomechanical movements, is an individualistic function, varying from person to person. It will be affected by the chair in which one is sitting, the floor surface, and even more subjective variables such as the surroundings in which one is sitting, the reason for rising, and one's state of mind and body. To cite two extremes; one would use a different mode of rising from relaxing in an armchair than one would use in rising from a train seat in the rush hour.

Before we analyse unaided rising, we must therefore examine for what purpose the analysis is carried out, and thus what mode is likely to be employed. The analysis is to be carried out for disabled people and it is assumed that either the effective strength of the muscles will be reduced due to loss of power in the muscles or due to stiffness in the muscles or joints; or that excess pressures in the joints and stresses in the muscles will cause pain, which will inhibit rising.

To standardize the degree of disability, we will take the most pessimistic case of unaided rising - one in which the subject has just enough total residual strength to enable him to rise; and with any less he would not be able to do so. This will be a critical point as it may mean the difference between being "independent ambulatory" and "confined to a wheelchair"..

The purpose of the analysis is to find in quantitative terms the strength requirements for rising with the help of the chair aid. We will then be able to find exactly how much help the aid is giving, and assess its suitability for different disabilities.

4.1.2. The Biomechanics of Unaided Rising.

In general, normal unaided rising will take place in three stages. Starting in a normal sitting position, the subject will move into a position suitable for rising; usually moving the feet back, to transfer the line of action of the reaction at the floor closer to the centre of gravity of the body, and will place the hands on the armrest, if there is one on the chair. He will then rotate the upper body forward, keeping the thighs in contact with the seat. Finally, applying a force to the armrest, and if possible using the momentum of his upper body, he will extend the knees until the leg and thigh are vertical. While the thigh is rotating, the hands lose contact with the armrest, and the hip extends to bring the upper body back into a vertical position.

Essentially, rising from a seated position is a rotation of the thigh link from the horizontal to the vertical. The other body links, except perhaps the forearm and hand, will be at the same angle at the beginning and end of the cycle; in the seated and standing position. However, to achieve this thigh rotation, especially the initial rotation when the thigh is raised from the seat, movements of the leg and upper body and arm forces must be introduced to achieve and maintain equilibrium.

In normal rising dynamic effects are present, especially in the upper body. It has been found, however, that dynamic effects will on the whole increase the force requirements of the subject in rising.* It will be assumed therefore that rising takes place slowly, and that the subject will be in "quasi-static" equilibrium throughout the rise. This is in fact borne out in practice when one observes that most disabled people with barely enough strength to rise unaided will stand up slowly. It is also assumed that maximum forces will occur at the point in the cycle when the thigh loses contact with the seat. That is to say that in the position of maximum forces, the thigh link is still horizontal, but the body is only supported by the feet and hands, or even by the feet alone.

As can be seen, the emphasis so far in the analysis has been on the muscular forces required to rise, rather than on the energy or the power requirements. The reason for this is that rising from a seated to a standing position is a fast process never taking more than a few seconds to complete. It is unlikely that in this short space

* Reference 15

of time muscular fatigue will affect the subject. Thus we can say that the most important requirement to be fulfilled on rising will be the force requirement.

In rising, a person will reach, at some point along the cycle, a position of greatest difficulty, or maximum force requirements. If he has enough muscular strength to carry on rising past this position, we may assume that he is able to complete the cycle. This position of greatest difficulty will be a function of several factors. Firstly there will be static forces due to gravity acting on the subject. Also in some subjects increased muscle tone, and pain or stiffness in the joints may alter the critical position. In addition to this the effective torque about a joint exerted by a muscle will vary over the range of movement of the joint. It will be assumed, however, for the analysis that the position in the rising cycle at which maximum forces are required is the position at which the thighs lose contact with the seat, and also that this is the position of maximum difficulty. If the subject can meet the muscle force requirements in this position, it will be assumed that he is able to reach a standing position, and then sit down again.

4.2.1. Unaided Rising Without Using The Arms.

The essence of this mode of rising is to transfer the centre of gravity of the body vertically above the floor reaction at the feet. The subject will do this by flexing the knee to move the feet (and thus the reaction at the feet) backwards. He will also flex the hip and spine, to rotate the upper body forward. This will cause the centre of gravity of the whole body to move forward. When the centre of gravity is vertically above the foot reaction, he will extend the knee and hip to bring himself to an erect position. The critical position of maximum forces is that at which the centre of gravity is vertically above the reaction at the feet and he has applied sufficient muscular forces in knee extension that the reaction between his thigh and the seat is zero.

If the feet are placed far back, the subject will require to lean forward less; so the knee extension moment will be large, and the hip extension moment small. If the feet are placed farther forward, more hip flexion, and hence a greater hip extension moment will be required. These two interrelated variables will control the mode of rising.

The body angles in the critical position will thus depend on the ratio of the moment at the hip to the moment at the knee as applied by the subject. In the able bodied subject, where the moments used are small compared to those available, the body angles will vary a great deal. However, in the disabled, where the maximum hip and knee moments will be substantially reduced, the actual mode of rising used will depend a great deal on the ratio of the maximum hip and knee moments, and will be much more pronounced. In the extreme case of a subject having only just sufficient muscular forces to rise, he will succeed in rising only in one well defined critical movement.

In the appendix (Section 5) an analysis is carried out of unaided rising without use of the arms; and numerical results are calculated for a fifty percentile subject. Figure (22) shows how the parameters β_1 the trunk angle, M_K the knee extension moment, and M_H the hip extension moment, will vary with the angle of the lower leg β_3 in the initial rising position. It can be seen that β_3 is plotted over a large range of values. In practice only the middle values of the range will occur. Large values of β_3 will only be used by limbo dancers.

4.3.1. Unaided rising With Use Of The Arms.

This is a far more realistic method of rising when applied to the physically handicapped than rising without the use of the arms. In the tests carried out on disabled people, they were found, without exception, to use the arms in attempting to rise, even though the arm muscles in some cases were reduced in strength due to their disability. It is difficult to predict the best mode of rising for any given subject. This will depend on the position the subject wishes to adopt as he is about to rise. Having selected this initial position, he is still able to vary the forces he applies; applying a greater force at the hands will relieve the force at the feet, and vice versa.

Analysis of unaided rising is carried out here primarily as a direct comparison with rising on the chair aid. The same subjects who participated in tests on the chair aid to rising are used in the analysis of unaided rising, enabling a direct comparison to be made between the two, and to thus gauge the effect of the chair aid on the muscular forces required for rising.

The basic analysis, however, is simple and is applicable to any mode of unaided rising, restrictions only being added later to produce numerical results. Figure (23) shows a seated subject about to rise into a standing position. The same conditions apply to this analysis as to the previous analysis (without using the arms); that the rising cycle is too short for fatigue to have any effect, and the critical factor distinguishing success from failure in rising will be whether the subject can supply enough muscular forces to rise. Furthermore, as was shown before, the position in the cycle at which the greatest forces are required is at the beginning of the cycle, just as the thigh has lost contact with the seat, but as it is still (assumed to be) horizontal.

The body is simply acted on by three external forces: the reaction at the hand R_H with horizontal and vertical components R_{HH} and R_{HV} respectively, and R_4, the reaction at the feet, R_{4H} and R_{4V} again the horizontal and vertical components. Also there are gravitational forces acting on each segment of the body, the resultant, W_T passes through the centre of gravity of the whole body. If we look at the three resultant forces R_H, R_4, and W_T, it can be seen that they form a very simple kinematic force system. It is however, an indeterminate system: the lines of action of the forces R_4 and R_H may cross anywhere on the line of action of W_T. To carry out an analysis of this system, one restriction must be imposed on it.

The human body, being basically lazy and well equipped to carry out tasks in the easiest possible manner, particularly when physically disabled or weakened, will automatically find the easiest mode of rising. In pathological rising, this will be the mode requiring the smallest muscular forces. The one restriction we can place on the rising analysis will be that it will thus be the mode requiring the least muscular forces.

The only difficulty now is to decide which muscular forces shall be taken into account and which shall be ignored. As a subject rises, a large number of muscles and muscle groups will take part,. However, only four muscle groups are considered to be critical in rising. Those are the muscle groups controlling:-

a) Extension of the hip joint.(or flexion)

b) Extension of the knee joint.(or flexion)

c) Adduction/Flexion of the shoulder joint.

d) Extension of the elbow joint.

As they are not considered critical in rising, and need not be used at all in applying a force, any muscular forces at the wrist joint and the ankle joint are ignored. If either of these are used to any extent in applying a force, the effect is simply to decrease the sum of the moments in the other joints, while the total muscular moment exerted remains approximately the same.

Another factor to take into consideration is the relative strength inherent in the muscle group, and the degree of paralysis present. The muscle groups associated with the hip joint, for example, are stronger than those of the elbow joint. To keep the analysis straightforward, however, the muscular moments at each of the four joints are simply summed; and the mode of rising giving the smallest sum is assumed to be that used by the subject.

Figure (23) shows a subject in the critical initial position of the rising cycle. Three equations may be obtained by resolving the forces vertically and horizontally, and by taking moments about a convenient point. Numerical values of the constants are measured from disabled subjects tested on the chair aid to rising. The subjects were asked to treat the chair aid as an ordinary chair, and to attempt to rise unassisted. None of the subjects were in fact able to rise, but each attempt was photographed, and the relevant data was measured from each photograph.

In order to select the optimum mode for each subject, the moment at each joint is divided by the ratio of the strength in the joint (as found in Chapter 2), and these figures are summed. The smallest value will give an indication of the best mode. This will ensure that the moment exerted at each joint is not out of proportion to the strength, inherent in the muscle group controlling the joint. In the appendix (Section 6), there is a full analysis of the kinematics of Figure (23). Also the computer program, and the results obtained from it, are given.

Chapter 5

The Chair Aid To Rising.

5.1.1. A device has been designed, and a protype built as an aid for people suffering from multiple sclerosis and similar crippling diseases, to rise from a seated position, and to sit down from a standing position. (Reference 11). It is of the form of a chair (Figure 24), having a seat, a backrest, a footrest, a knee constraint, and a handrest. All these parts move during the rising cycle, and are connected to each other in the form of a one degree of freedom linkage system. A moving point on the mechanism is connected to earth (i.e. to the base of the chair) by a set of springs in such a way that the total energy of the system is constant over the rising-sitting down cycle. The potential energy of the subject and chair mechanism is converted to strain energy in the springs as the subject sits down, and is converted back to potential energy as the subject rises. The subject needs only to supply a comparatively small amount of additional energy to overcome losses in the system, and will thus need a relatively small degree of strength to rise and sit down.

We wish to examine these losses, and if possible to minimise them. Also we wish to examine how much strength is required to use the aid, which muscle groups are involved, and thus to deduce the effects of the aid on the biomechanics of rising by comparing the strength requirements of rising unaided with rising on the chair aid. This will give for different subjects tested, the direct effect of the aid on rising, the suitability of the aid for different disabilities, and the effectiveness of using different muscle groups to supply the extra energy requirements.

It is expected that the chair will have different uses for different disabilities. In the case of progressively deteriorating paralyses, such as multiple sclerosis, it will enable the subject to rise and sit down independently long after he is too weak to do this unaided. For paralysing disabilities from which there is progressive recovery, such as hemiplegia, the aid will probably increase the speed and the degree of recovery, using it in a similar manner to the arm support; making more effective therapy possible at an earlier point of treatment of the disease. Lastly, for disabilities from which there is no effective recovery or deterioration, such as paraplegia, use of the aid will, in some cases, mean the difference between the subject being able to rise on his own, and being unable to move from the seated position.

To study the effect of this device, the prototype was improved and modified for safety in use with disabled subjects. Dynamometers and other measuring devices were added to monitor the forces acting between the chair and the subject. These forces are caused by voluntary muscular contractions in the subject. The magnitude of these contractions can be most easily expressed as a torque about each joint caused by the muscle groups affecting the joint. From the forces measured by the dynamometers, the torques about the body joints exerted by the muscle groups may be found; and the effectiveness of these forces in causing the chair and subject to rise may be assessed.

5.2.1. Modifications To The Chair Aid

A locking device is required to hold the chair in the seated position. It should be strong enough to hold the chair safely even if the subject were to stand on the footrest with the chair in the seated position; and it should be possible to release it only when it is safe to do so (i.e. when the subject is seated in the chair and is in equilibrium with the spring forces).

A simple arrangement was designed to fulfil these conditions (Figure (25)). Two interlocking, wedge-shaped components fit together to form the catch, one rigidly attached to the seat and the other hinged to the base of the chair, and able to rotate enough to disengage. As the seat descends, the sloping surface XX' of the wedges meet, and the lower wedge rotates until it engages in the locked position, pulled by the spring shown. The catch is released by pulling the hand lever which transmits a force to the hinged wedge through a Bowdon Cable. The force required to disengage the wedges is proportional to the normal force acting between them. This fact was used to design a safety device or "fuse" to prevent the catch disengaging when it is not safe to do so. A pretensioned spring was attached between the hand lever and the Bowdon Cable. If the subject is not seated in a safe position for rising and the operating lever is pulled, the force required to open the catch will be greater than the pretension in the spring and the spring will extend leaving the catch locked.

Belts were attached to the seat and the backrest to strap those subjects into the chair who felt unsafe. However, there is no real need for the straps as the subject is always held securely in the chair by the kneerest; but on occasion they provided reassurance to the subject, and so were sometimes used. Paradoxically it was found necessary during tests on stronger subjects to use the lap strap to prevent them from lifting from the seat.

One other weakness in the design was corrected - this being the knee constraint catch, which was found on occasion to burst open. A simple safety catch was added here to eliminate this danger. In proper use, the chair was now considered to be free of any forseeable hazard.

5.3.1. Adaptation Of The Chair For Testing.

In the use of the chair aid to rising by the subject, we can assume that the chair is adjusted properly. The mechanism and subject will now be in equilibrium and the seat link torque at the seat hinge will be zero. Any attempt the subject makes to rise will set up additional forces between the subject and the chair mechanism, and between the mechanism and the chair base. This force system is shown in Figure (26). There may be additional forces acting on the backrest, seat, handrest, footrest, and kneerest, and also a torque at the seat hinge between the seat link and the base of the chair (ground). These forces are due to contact between the subject and the chair surfaces and arise from the exertion of muscular forces by the subject. To measure all the forces acting on the chair surfaces, their magnitudes, positions and directions, would require about thirty dynamometers. It would be necessary, because of their sheer numbers, to use electronic dynamometers - each requiring an amplifier and recorder. Also, major alterations to the chair would be necessary to accomodate this instrumentation. The work and cost involved in this is prohibitive, so the testing was simplified.

By measuring only the forces on the footrest and chair seat, a fairly accurate picture of the biomechanical force system can be drawn. Moments of the muscular forces about the knee and the hip joints, and under certain conditions the shoulder and elbow joints, can be calculated from the measurements of the seat and footrest forces. The geometry of the subject must also be measured for calculation of these muscular moments. In general, forces on the handrest and backrest can now be calculated by an analysis of the forces on the subject. Knowing all the forces acting on the chair mechanism, a value for the seat hinge moment can now be calculated. This was used in the preliminary tests as a check on the measurement apparatus, and the biomechanical theory on which the tests are based. If the calculated seat hinge moment were not equal to the value of the seat hinge moment obtained by direct measurement, this would indicate a fault, either in the measuring apparatus or in the method of analysis.

5.3.2. The Seat Dynamometer.

In measuring the resultant force on the seat, it is important to find also its position along the seat, or the distance from the seat hinge at which it acts. The dynamometer used should allow little deflection of the seat, but should give an easily visible reading of the force and its position. A hydraulic arrangement was used measuring the seat reaction, R_2, at two points on the chair seat (arranged as shown in Figure (27)). From these measurements both the magnitude and the position of the resultant force can be calculated.

The gauges are water-filled rubber tubing, sandwiched under a hinged flap on which the chair seat rests. The pressure in the tube is measured by a mercury manometer. Any force on the seat will be transferred to a hinged flap, which will increase the pressure in the rubber tubing. This increase will be registered by the manometer. Four such gauges are used as shown in Figure (27). One manometer monitors the front gauges, and the other, the rear. It was found in testing that they gave consistent readings over 55% of the seat width. This is adequate, even for the most eccentric sitter. Calibration of the seat gauges is given in the appendix (Section 7). Unfortunately the gauges are not free from dynamic errors. Application of a sudden force on the seat produces a vibrational response in the mercury manometers. Some damping is provided by friction of the liquid in the narrow bore tubing. Even so during testing great care was taken to ensure that only static loads were measured. To do this the subject was required for each measurement to apply a constant force for a few seconds to allow the mercury level in the manometers to become stationary. This will also ensure that it is the maximum constant force that is measured, and not the instantaneous peak force.

5.3.3. The Footrest Dynamometers.

Measurement of the forces on the footrest, R_4 and R_5, is comparatively simpler than measurement of the seat forces, as the position of the resultant can be located visually. It is assumed that the muscles controlling the ankle joint are relaxed and that the footrest reaction passes through the ankle joint. By projecting the resultant footrest reaction through the ankle onto the plane of the footrest, its position is located.

The force is measured in components parallel and perpendicular to the plane of the footrest. The arrangement for measurement is shown in Figure (28). For simplicity, bathroom scales are used to measure the perpendicular force component. The scales are fitted with a small wheel at each corner to roll on an aluminium track along the length of the footrest. The scales are constrained from moving freely along the tracks by a dynamometer measuring the tangential component of force. This, as shown, is a circular spring balance mounted vertically on the chair frame, and attached to the footrest scales by a nylon cord passing over a pulley. By this arrangement, only the force acting away from the chair can be measured. However, on rising the subject always applies a force in this direction, so the apparatus is adequate for these tests.

5.3.4. The Seat Hinge Torque Dynamometers.

The seat hinge torque measured by the moment about the seat hinge of a force acting on the seat link, is the torque exerted by the subject in applying forces on various parts of the mechanism. The balancing force acts on a rigid extension to the seat link, 24 inches long and at 45° to the line of the seat link, (shown in Figure (28)). Measurement of the seat hinge torque is required in carrying out two of the tests; one requiring static measurements of relatively high torques, and the other for measuring smaller torques as the chair is in motion. For the former, an arrangement is used, similar to measurement of the force R_5; a circular spring balance is mounted vertically on the chair frame (shown in Figure (28)). The scale is attached to the seat moment arm by a nylon cord which passes over a pulley. The length of the cord is adjusted with a block and tackle, so that the chair can be raised and lowered. In spite of the high forces required to adjust the chair, the pulley system allows this to be done quite easily by hand.

The other test carried out, in which the seat hinge moment is to be measured is referred to as the "deadweight" test. The subject sits relaxed in the chair, while the seat hinge moment, required to raise the subject and chair to a standing position, is measured continuously as a function of the seat angle. The cycle is completed by measuring the moment as the chair is lowered to the seated position. This measurement is to be continuous and the moments may be either positive or negative, although any dynamic forces present should be small. It is preferable in this case to measure and record the seat hinge moment and the seat angle electronically so that a permanent and continuous record can be taken of the readings.

A piezo-electric dynamometer is attached to the seat moment arm, at right angles to it. The output from the dynamometer passes through a charge amplifier and is recorded as the Y-coordinates on an X-Y plotter. A handle is attached to the dynamometer. The tester is required to apply a force to the handle necessary to raise and lower the chair and subject slowly. Readings of the seat angle are fed into the X-Y plotter as the X-coordinate. A simple circuit is used for this: the seat angle itself being measured by a goniometer, attached between the chair base and the seat link. The response of the plotter is adjustable, and the X-output is calibrated to give a pen deflection of 18cms between the seated and erect positions of the chair, i.e. for a 90° seat link rotation. The output of the goniometer circuit is a linear function of the seat angle; so that intermediate angles of the seat may be marked at regular intervals along the X-axis. The dynamometer circuit is adjusted to give a seat hinge moment output of 48 inch-lbs per cm. pen deflection along the Y-axis, and is capable of measuring positive or negative values of the moment.

A small amount of drift is present in the dynamometer: approximately 0.5 to 1.5 cms. over the whole cycle. This is taken into account in analysing the results. The dynamometer is grounded before each cycle to earth any charges built up in the circuit.

5.3.5. Photographic Recording Of The Tests.

It is important that recording of all the scale and gauge readings is carried out simultaneously, as well as measurement of the geometry of the subject and the chair mechanism. A photographic record of each test was taken. This gives an instantaneous and permanent record of all the readings. As we wish to measure the geometry of the chair and subject, it is necessary to photograph the apparatus from a distance to minimise angular distortions due to parallax. Using a camera fitted with a telephoto lens, the image of the apparatus filled the photograph at a distance of approximately 70 feet. Due to limitations of space, and the convenience of having the camera next to the apparatus, a large mirror was used to reflect the image of the chair. The camera was situated next to the chair, and a plane mirror measuring 5 feet x 4 feet was placed approximately 35 feet away. With this arrangement all the dynamometer readings are visible, with the exception of the footrest scale, as this faces upwards. It is necessary to use two small auxiliary mirrors to reflect the image of the scale into the camera. A camera timing of 1/8 second at F.6.5. was used throughout the tests, and two mercury vapour floodlights were used to illuminate the apparatus.

5.4.1. The Mechanics Of The Use Of The Chair.

The chair has now been modified and calibrated for testing. For the subsequent analysis of the mechanics of the subject, and also the mechanics of the interaction between the subject and the chair, tests were carried out with the help of healthy subjects. The subject will affect the mechanics of the chair in three ways:-

a) Applying gravitational forces to the chair.

b) Causing energy losses due to friction.

c) Applying an energy input to work the chair mechanism. This is in the form of muscular forces applied to the chair.

5.4.2. Analysis of the Mode of Application of Forces to Parts of the Chair.

The chair is a fairly complicated mechanism. If a force is applied to any point on the mechanism, e.g. at a point of contact between the chair and the subject, a complicated system of forces and torques will be set up in the links of the mechanism. We are interested in the torque generated about the seat hinge. A set of functions is now derived to express the seat hinge torque in terms of the magnitude and direction of a force applied at a point on the chair mechanism, and of the position of the mechanism in the rising cycle.

Expressed analytically, these functions are too complicated for repeated calculation. Each function is therefore plotted graphically, to enable the value of the seat hinge torque to be found easily; knowing the magnitude, direction, and position of the force, and the angle of the seat link. (The seat link angle α_2 is used as a measure of the position in the rising cycle of the mechanism.).

5.4.3. The Footrest Function.

Two forces act on the footrest, as shown in Figure (31). These are components of the force on the foot, normal and tangential to the footrest, Δ_4 and Δ_5. An expression for the forces transmitted through CJ and BI is firstly derived, and then an expression for the moment of the forces in these two links about the seat hinge A is derived. Eliminating between these two expressions gives an equation of the seat hinge torque as a function of the magnitude and position of the forces on the footrest, and of the mechanism link angles. These latter are functions of the seat angle α_2.

The equation expressing the seat hinge torque as a function of the forces on the footrest, is given by:-

$$\Omega = \Delta_5 \Lambda_{FI} + \Delta_4 (\Lambda_{FII} . JF + \Lambda_{FIII} . IF)$$

where a positive value of the seat hinge torque indicates a torque tending to cause the chair to rise. Where Δ_4 and Δ_5 are the normal and tangential forces on the footrest; Λ_{FI}, Λ_{FII}, and Λ_{FIII} are functions of the geometry of the chair mechanism, as given in Figure (30), and JF and IF are as marked on Figure (31).

5.4.4.. The Kneerest, Seat, Backrest, and Handrest Functions.

These expressions are all derived in a similar manner to the footrest function.

5.4.5. The Complete Seat Hinge Torque Function.

The moment about the seat hinge of the individual forces applied by the subject (due to his muscular exertion) on different parts of the chair, can now be summed to give the total effective moment. This will be given by:-

$$\Omega = \Delta_2(AU) - \Delta_1(d\sin(\alpha_2 - \theta_1) - L_1\sin(\theta_1 - \alpha_1)\frac{d\alpha_1}{d\alpha_2})$$
$$+ \Delta_3\Lambda_{KR}\sin(ZKL) + \Delta_4(\Lambda_{FII}(JF) + \Lambda_{FIII}(IF)) + \Delta_5\Lambda_{FI}$$
$$+ \Delta_7(\Lambda_{HI}\sin(XPQ) - \Lambda_{HII}\sin(XPR)) \quad \ldots\ldots\ldots\ 5.1.$$

where the forces, linear dimensions, and angles are as shown in Figure (31), and the calculated functions of the geometry of the linkage are given in Figure (30). These latter functions are characteristic of the chair, and will remain consistant, regardless of the test being performed on the chair.

5.5.1. Energy Losses in the Chair Aid.

Energy losses in the chair aid occuring when the subject rises or sits down in the chair, can take any one of four forms. Firstly there will be friction in the chair mechanism, probably in the joints connecting the links. Secondly, friction in the subject due to stiffness in the joints, muscle tone resistance, and also passive resistance of the body tissue. Thirdly there will be friction between the subject and the chair. This will be caused by variations in the physique of different subjects introducing a discrepancy between the motions of the mechanism and the subject. A slight sliding effect will then be introduced which will cause frictional resistance to the motion. The fourth energy loss will be due to the dynamic effects of initiating and stopping the motion. These are usually minimal as the subject, in general, rises and sits down slowly.

5.5.2. Friction in the Chair Mechanism.

It is difficult to measure friction in the chair mechanism experimentally; for to do so, the chair should be loaded as if it were occupied. This requires realistic loading not only of the seat, but also of the backrest, footrest, kneerest, and handrest. At the same time, the loading should not interfere in any other way with the free movement of the chair mechanism. The difficulties involved are too great to warrant carrying out the test; however, for some idea of the magnitude of the frictional forces involved, a simplified measurement can be made.. The chair springs are adjusted as for an average subject. A force is applied to the seat moment arm which will just raise the chair from the seated to a standing position. The magnitude of the force is measured by the piezo-electric dynamometer, and is recorded on the X-Y plotter. Figure (32) shows the seat hinge torque Ω as applied to the seat moment arm plotted as a function of the seat link angle α_2. The resulting curve is a narrow hysteresis loop which shows an energy loss just over 2% of the total energy supplied by the chair aid to the subject over the rising-sitting down cycle. It is expected that if the chair were loaded realistically, as if occupied by the subject, the energy loss would have been higher. Even so the losses are comparatively small; the same effective seat hinge torque can be applied by the subject merely by moving forward in the seat approximately one tenth of an inch.

5.5.3. Energy Losses in the Body.

In a healthy subject there will be little internal resistance to motion in the body. With certain disabilities, however, this resistance to motion may increase to a significant level. In chronic rheumatoid and osteo-arthritis the affected joints themselves deteriorate and this may cause a frictional torque opposing motion in the joint. Another cause of internal body resistance is due to stiffness in the muscles. This is often a side effect of paralysing disabilities such as multiple sclerosis and hemiplegia.

It was found while testing disabled subjects that increased muscle tone usually caused a resistance to limb movements greater in one direction than the other. For example, the knee extensors were found in general to exert a more powerful involuntary force than the flexors when affected by increased muscle tone. The effect of this is that the knee joint tends to extend and the leg straightens out. In sitting down in the chair, the knee joint is required to rotate from a fully extended to a flexed position. Extra work must be supplied by the subject to flex the knee. An indication of the magnitude of the additional work requirements can be seen in the "deadweight" tests, where the hysteresis curves for subjects with stiff muscles can be compared to those unaffected by stiffness, (Figure (33)). The tests are analysed in greater detail in the following chapter.

Muscular stiffness was found to vary a great deal during the tests, increasing after the affected muscle had been used to apply a large force, or when the subject was in a state of excitement or fear, and to decrease when the subject was calm. Because of this, the subject was allowed to relax in the chair for several minutes before any tests were carried out when any stiffness was present in the muscles.

5.5.4. Losses Due To Interference Between The Subject And Chair.

The chair was designed for a person of average height and build, with adjustments, where possible, to allow both large and small people to use it. The chair has two such adjustments: the spring adjustments to accomodate people of different weights and weight distributions; and the footrest height adjustment. The latter is necessary because the position on the knee at which the kneerest acts is critical to the kinematics of the chair-subject system. If the kneerest is not in proper adjustment, the subject will tend either to be pushed back along the seat or to slide forward during the motion. Either will require an additional work input from the subject to overcome the losses. This adjustment will allow for variations in the distance from the sole of the foot to the point on the patellar ligament on which the kneerest acts.

One other dimension is of importance: the length of the thigh. The position of the thigh on the seat is fixed during the motion by the kneerest. A subject with short thighs will sit farther forward on the seat than one with long thighs. Extremely long limbed subjects will tend to be "sandwiched" between the kneerest and backrest on rising, while extremely short limbed people will be unable to keep comfortably in contact with the backrest. To remedy this it is not possible simply to move the backrest backwards or forwards, as the kinematics of the mechanism will be altered. To alter the seat length it would be necessary to scale down each member of the backrest linkage. As this affected only a small proportion of subjects, no linkage adjustments were included.

5.5.5. The Deadweight Tests.

Tests were now carried out to measure the effect of the spring adjustment and the footrest adjustment on the kinematics of the chair aid and subject. The "deadweight" test apparatus is used. It is designed to plot the seat hinge moment applied by the tester against the angle of the chair seat link.

Figure (34) shows a typical trace obtained from this test. It takes the form of a complete cycle, from sitting to standing and back to sitting. Positive readings on the vertical axis show a moment applied to the seat link tending to make the chair rise; while negative readings show a moment tending to make the chair fall. The area enclosed by the curves is equal to the work lost in friction over the cycle. Ideally the curve should take the form of two superimposed straight lines along the x-axis. The purpose of these tests is to reach as closely as possible, this ideal by adjustment of the spring configuration and the footrest height.

A subject of average size was used as the "deadweight", and was instructed to sit comfortably in the chair, and to relax completely while the tests were being carried out. Figure (35) shows the spring arrangement. In the original chair design, the geometry of the springs was intended to be kept constant, adjustment being made by moving all the spring attachment points an equal distance along the spring adjustment arms. In testing a variety of people on the chair it became apparent that the required spring characteristics vary from person to person, and as such the spring configuration should be altered to suit each subject.

Figure (36) shows the effect of varying the configuration of spring B by moving the attachment point Z along the adjustment arm. This spring exerts a force over the first part of the rising motion. As the spring attachment point Z is shifted down the arm, the spring can be seen to exert a greater force over a larger part of the rising motion.

Figure (37) shows the effect on the characteristics of varying the configuration of the primary spring A by moving the attachment point Y down its adjustment arm while moving attachment point X up its adjustment arm. The seat hinge moment in the seated position will stay constant, while in the erect position it will increase. The reverse effect will occur if the spring attachment points are moved in the opposite direction. Using a combination of these adjustments, most spring requirements may be catered for. No quantitative measurements were made of the spring forces and the effects of spring adjustments on the seat hinge moment as it was felt that these were unnecessary, and would only complicate adjustment of the chair.

5.5.6. Summary Of The Spring Adjustment.

1. To increase the seat hinge moment supplied by the springs near the seated end of the cycle: move attachment point Z of spring B down the adjustment arm.
2. To increase the seat hinge moment supplied by the springs near the standing end of the cycle: move the attachment point Y of spring A down the adjustment arm.
3. To increase the seat hinge moment supplied by the springs over the middle of the rising cycle: move the attachment point X of the two springs down the adjustment arm.

The effects of these adjustments are in fact complex and have been simplified here to provide a general guide to adjustment of the chair springs.

In testing subjects on the chair, the technique used for the spring adjustments was to adjust the springs to be in equilibrium in the seated position, and then to take a deadweight trace. Figure (38) is a typical trace.. The first deadweight trace shows that far too much force is supplied at the seated position, and far too little at the standing position. The improvement in the characteristics can be seen as the springs are adjusted. Curve III is as close as could be approached to the ideal for this subject.

5.5.7. The Footrest Height Adjustment.

In curve III there is still a large amount of hysteresis and a pronounced jump in the curve between $\alpha_2 = 55^\circ$ and 70°. This is the point at which the kneerest contacts the knee. It is noticeable that on sitting down the kneerest loses contact approximately 5° later in the cycle. This would indicate that the subject slides forward in the seat slightly, due to incorrect adjustment of the footrest height. This sliding will cause a wastage of energy, shown by increased hysteresis in the deadweight curve.

Tests were now carried out to examine the effects of the footrest height adjustment on the energy losses, and to calibrate the footrest height scale to give minimum losses for different leg lengths of the subject. Deadweight tests were carried out on the same subject as used in the previous tests, using the spring adjustment found to give the best deadweight curve.

The footrest slides up and down in grooves. The zero on the scale is 16.5 inches vertically below the seat hinge. Starting with the footrest at -0.5 inches, which was uncomfortably high for the subject, it was lowered in increments to -2.6 inches, which was noticeably too low. A deadweight trace was taken at each footrest height. Figure (39) shows the traces obtained. These show the hysteresis losses to decrease significantly as the footrest is lowered, although the curves keep a very characteristic shape. In the last test, (curve IV), the subject experienced discomfort as the kneerest, which was intended to act on the patellar ligament was now acting on the kneecap. A kneerest height of -1.8 inches gave the least hysteresis while not causing discomfort to the subject. The total energy loss over the cycle amounts to 250 in-lbf., which is approximately 13.5% of the total energy required to rise unaided.

It is desirable to adjust the footrest before the subject gets into the chair. To do this a graph was drawn correlating the footrest height with a well defined dimension on the lower leg. The dimension used was the distance between the ground and the tuberosity of the tibia; this being easy to locate and close to the point at which the kneerest acts. Figure (40) of lower leg length against footrest height based on the measurement of several subjects.

5.5.8. Kinetic Energy Losses.

If the subject develops too much speed by the end of the cycle, kinetic energy will be lost in bringing the chair to rest. It is unrealistic to suppose, however, that if the chair were properly adjusted, a heavily disabled person would be able to develop any appreciable amount of kinetic energy. As this form of energy loss is voluntary and not inherent in the system, it is not necessary to carry out an analysis of it. During tests on disabled subjects it was seen that the chair was in fact used slowly, especially when, in the case of heavily disabled subjects, the energy losses were critical.

Chapter 6

Performance Of The Subject On The Chair Aid.

6.1.1. In the previous chapter, the mechanics of the chair aid with the subject seated passively in it, were examined. The following chapter is an analysis of the performance of the subject in using the chair aid, including the biomechanics involved, a series of tests carried out on disabled people, and a comparison with unaided rising, to determine objectively the effect of the aid on the biomechanics of rising.

6.2.1. Biomechanics Of The Use Of The Chair Aid.

Mechanically the chair aid can be regarded as a chain of rigid links, arranged as a one degree of freedom linkage system. The human subject may also be treated mechanically as a chain of rigid links. When the subject is seated passively in the chair, the chair and subject become one linkage system; the points of contact between the subject and chair being complex joints. During the rising cycle the effective position of these joints may change; for example, the position of the resultant force between the thigh and the chair seat may move towards the knee, as the subject exerts muscular forces, or as the chair rises and the geometry of the system changes.

As the subject applies muscular forces, this will have the effect of producing a torque about one or more of the body joints. For the chair-subject system, this will be an internally applied force which will set up internal forces through the rest of the linkage, and will be balanced either by an external force, such as a torque about the seat hinge, or will accelerate or retard the rising motion of the mechanism and subject. (As this motion has effectively one degree of freedom, a muscular force applied by the subject will cause the chair to either rise or fall). An analysis is now carried out of the effect on the seat hinge torque produced by the application of combinations of muscular forces.

There were found to be four basically different methods of causing the chair to rise from the seated position.

1. Extension of the knee joint, caused by contraction of the quadriceps ~~and~~ femoris. This will cause a redistribution of the forces on the footrest and seat; increasing the vertical force on the footrest, and reducing the force on the seat. As a result, a torque will be generated at the seat hinge, tending to make the chair rise.

2. Extension of the hip joint, causing contraction of the gluteus maximus and the hamstring muscles of the thigh; equilibrium of the upper body being maintained by pushing back against the backrest. The force on the backrest itself will tend to make the chair fall; however, the resultant torque at the seat hinge will be such as to cause the chair to rise.

3. Similar to the previous method, but forces maintaining equilibrium of the upper body are in this case applied at the handrest. This introduces the use of the muscles of the shoulder joint, and of the upper arm in flexing the elbow joint.

4. Upper body movements carried out by flexion of the hip joint and spine. This causes the reaction on the seat to be shifted forwards, producing a positive seat hinge torque.

For each mode, different sets of muscle groups are used. A subject attempting to rise will probably use a combination of these modes. To understand the effectiveness of each mode, to be able to calculate the moment of the muscle groups about each joint, and to be able to analyse which modes are being used, a biomechanical analysis is carried out of the internal and external forces acting on the subject as he applies muscular forces to rise in the chair.

We are interested in the strength required to carry out the different modes of rising, with a view to comparing these requirements with those of unaided rising. The same conditions will apply as with unaided rising i.e. it is the force input to the system, and not the work or energy input that we are really interested in. In the unaided rising analysis the arms play a critical role; and the muscular moment at the joints of the upper limbs are regarded as equally important as the muscular moments at the joints of the lower limbs. In using the chair aid, however, although the arms are invariably used in the free mode of rising, it will be shown that the use of the arms is not critical, and that much the same seat hinge torque is generated when the arms are not in use and other modes of rising are employed. The subject in using the chair will thus use some or all of the strength in his arms to rise, but it will not affect the seat hinge torque generated. We can thus say that the critical muscular moments will be required only at the hip and knee joints, and the total strength required will thus be equivalent to the sum of these two moments.

The body is assumed to be a two-dimensional chain of rigid links, freely jointed together. Muscle forces are assumed to produce pure torques at the joints, which are transferred to the chair mechanism, causing a seat hinge torque. A full analysis of the four modes of rising and the seat hinge torques produced is given in the appendix (Section 8)..

6.2.2. Preliminary Tests.

The first tests were now carried out on able bodied subjects to become familiarized with the measuring apparatus and technique, and to test the validity of the analysis. As mentioned in the previous chapter, there is provision on the chair for measurement of the seat hinge torque. This provides a check on the biomechanical theories that have been applied. If any of the modes show a consistent difference between the calculated and the measured readings of the seat hinge torques, there will be some false assumptions in the method of calculation for that mode.

Figure (41) shows a subject relaxed in the chair (A), and applying a knee extension moment (B), in attempting to rise. In order to use the equations given in the appendix (Section 8), it is first necessary to find the magnitude of the forces applied by the subject to the chair. This is simply done by subtracting the forces shown on the dynamometers as the subject is relaxed, from the forces shown as he attempts to rise.

In the appendix (Section 9), one set of results for each mode is calculated in full. Twenty seven results are calculated for two positions of the chair; in the sitting position, and at approximately one third of the way to the standing position. Eleven results were taken of the knee extension mode, five of hip extension with use of the backrest, five of hip extension with use of the handrest, and six of the trunk bending mode.

Figure (42) shows these results, with calculated values of the seat hinge torque plotted against the measured values, for different values of each mode of rising. As the calculated seat hinge torque should be equal to the measured seat hinge torque, all the points should lie on the line XX. As can be seen, the points are in fact scattered on either side of this line. There are no large, or visually consistent deviations of the points of any of the modes. It can be assumed that the scatter will be due to cumulative experimental errors, and that the theory on which the results are based is valid. These results show a standard deviation of 105.5 in-lbf., which is an error of 17.2%.

From the measurements it is also possible to calculate the effective moment about the body joints of the muscle groups involved in working the chair. Figure (43) shows the joint moments of the muscle groups plotted against the seat hinge moment. The muscular moment at the knee joint, and that at the hip joint are separated on the diagram. It can be seen how the knee extension mode requires a comparatively large knee extension moment, while upper body flexion is carried out using a large hip extension moment. Although there is a fairly large scatter in the results, the points at both $\alpha_2 = 90^\circ$ and $\alpha_2 = 60^\circ$ appear to lie on a straight line of estimated gradient approximately equal to one. This is to be expected when one examines the mechanics of the chair/subject system. In rising from the seated to the erect position, the angle of the links at the three joints at which critical torques are generated (the knee joint, the hip joint, and the seat hinge on the chair) all rotate through approximately 90°. The energy input (the product of the torque and the angle of rotation) which occurs at the knee and hip joint, will be approximately equal to the energy output at the seat hinge. The sum of the torques applied at the hip and knee joints will thus equal the torque generated at the seat hinge. In practice there is no energy output. The energy input is used to overcome energy losses within the system, (see sections 5.5.2 - 5.5.8.). But as we have expressed these losses in terms of the seat hinge torque, and we can now express the muscular moment input in the same manner, we are now able to express directly the muscular moments required to rise in the chair aid, i.e. those required to overcome the losses in the system.

6.3.1. The Chair Aid Tests On Disabled Subjects.

Since the construction of the prototype of the chair aid, six disabled subjects have tried it out. Observations made on these initial trials were completely subjective, but were extremely useful in enabling the design of the chair to be improved. Many of the small additions and improvements were the direct result of these trials. The main observations based on the initial trials were as follows:-

1. The chair aid works.

2. In addition to aiding subjects in rising and sitting down, it also holds them securely in an erect position. Several disabled subjects commented on this, as it was a new experience for them to be able to stand erect but relaxed.

3. The large amount of mechanical linkage and clutter attached to the chair makes getting into and leaving the chair a difficult and often hazardous process for both the subject and those who are helping.

The chair aid and measuring apparatus, as used in the preliminary tests, was now taken to the Frank Taylor Centre for the disabled. Here eight subjects were tested on the chair, all of whom were suffering from partial paralysis of the lower limbs, and some also from partial paralysis of other parts of the body. None of the subjects were able to rise unaided from a chair.

6.4.1. The Testing Procedure.

Before the testing begins, a short questionnaire is completed to record the disability of the subject, the presence of any stiffness in the limb joints, the weight of the subject, and some relevent link dimensions. The footrest height is then adjusted from the footrest graph, and the springs are adjusted approximately. The subject is now transferred to the chair aid. This is generally achieved as follows: with the chair in the standing position, the subject is pushed up a specially constructed ramp in his wheelchair. This brings his feet to the level of the footrest on the aid. With one helper on either side of the subject to provide support, he is now turned around and moved back against the erect chair aid. The kneerest is now closed and locked into position, and the subject is now fully supported in the chair.

A deadweight test (see section 5.5.5.) is carried out, firstly to check whether the footrest is adjusted properly, and to enable fine adjustment of the springs to be made. Tests are now carried out similar to the preliminary tests, in which the subject is asked to exert the maximum force in each of the four modes of rising. The results as before are recorded photographically. As a fifth mode, the subject is asked to attempt to rise in any manner he choses. This is called the free mode, and is usually a blend of the fixed modes that is most suitable to the subject in view of his disability. The chair is now locked in the seated position, and the subject is asked to attempt to rise as if he were seated in an ordinary chair. At the point of failure in the attempt to rise (i.e. the point at which a subject would normally lose contact with the seat, called in Chapter 4 the critical position), the subject is photographed. This photograph is used in the analysis of unaided rising, and will be compared later in the chapter directly with rising in the chair aid. Finally all restraints on the chair are removed, and the subject is asked to rise and sit down in the chair, while a record is made of the success or failure of the attempt, and of his comments.

6.5.1. Results Of The Chair Tests.

Figure (44) shows the results of the tests. For each subject the muscle forces in the hip and knee joints are shown both diagrammatically, and as plotted on a graph against the seat hinge torque generated. The deadweight test graph is also shown. As only eight subjects have been tested objectively, conclusions are difficult to draw from the results. We are able to examine the results as a whole, and draw conclusions as to the average, or general effect of the chair aid on the group, and we can also examine the results separately to explain individual measurements and observations. Unfortunately there are not enough results to divide the subjects into subgroups; for example into groups according to their disability. There follows, however, observations made on the results.

As estimation of the maximum seat hinge requirements in rising is made, based on the deadweight test graphs, and on the actual performance of the subject on the aid. Of the eight subjects tested, four (subjects IV, V, VI, and VIII) experienced no difficulty in rising and sitting down in the aid. The other four were all able to rise, but were unable to sit down. Of these two (subjects II and IV), it is estimated, would have been able both to rise and sit down had the chair been in better adjustment. Of the other two subjects; subject I, it should be noted, was a somewhat plump lady. The main restriction on her rising over the last half of the cycle seemed to be a "sandwiching" effect between the chair seat and the kneerest. The only other subjects on which anything similar to this phenomenon was observed was on tall, well built subjects, who tended to be "sandwiched" between the chair backrest and kneerest. The remedy in both cases would simply be to move the kneerest away from the seat.

Finally there is subject IV who is the only subject tested, for whom the chair is judged unsuitable. This subject is suffering from multiple sclerosis, and although he has a fair degree of residual strength, he has severe spasticity in the lower limbs. The knees tend to be held in a fully extended position, and a large external force is required to flex them. Perhaps a modified chair aid with restraints on the footrest which would force the knee to flex, would be good as a therepeutic aid; but as simply an aid to rising, this subject would probably find a simpler aid more effective; for example, one in which the seat only tips up. Using this he could keep the knee joint extended.

Of the free modes for rising selected by the subjects, on average most appear to have selected a more effective method of rising than any of the fixed modes. Figure (45) shows the muscular moments required for each mode, plotted against the seat hinge torque generated. It can be seen that both of the hip extension modes and the knee extension mode are of equal effectiveness; the upper body flexion mode is the most efficient of all; i.e. the seat hinge torque generated requires the smallest muscular forces of any mode. The free mode, on examination, fell into two distinct groups: four of the subjects used a form of upper body flexion in the free mode, using the handrest to provide support. This is a very effective method of rising in the aid, and is in fact imitating to a certain extent natural unaided rising.

The other four subjects used a mode that appeared to be a combination of knee extension, and hip extension with the handrest and backrest. This is more in keeping with the kinematics of the chair aid; and it was more the intention that the chair should be operated in this manner. This mode is recommended for subjects with weak arms, for if the subject leans forward without being able to support his upper body with his arms, he could lose control of the motion of the chair.

As a final test of the effectiveness of the chair aid, we can now compare the strength requirements of rising unaided with the strength requirements of rising in the chair aid. Before any numerical comparisons are made, the differences between the derivations of the figures for aided rising and for unaided rising should be listed.

1. The unaided rising figure is the sum of the muscular moments at the knee, hip, shoulder, and elbow joints; whereas in aided rising, the figure is the sum of the knee and hip joint moments only. The moments about the joints of the upper limbs are regarded as superfluous.

2. The unaided rising analysis is theoretical, taking no account of any internal body resistance in the subject. The analysis in using the chair aid is experimental, and many of the subjects tested had a large degree of spasticity or occasionally stiffness due to deterioration of the joints, which increases the strength requirements for rising.

3. In the unaided rising analysis, no account is taken of sitting down. In theory, sitting down is as simple as falling off a log, and requires no strength. But in pathological subjects this is often not the case, and sitting down presents almost as many problems as rising. Sitting down in the chair aid however, is an integral part of the cycle; and the chair can be adjusted to make rising easier and sitting down more difficult, or vice versa. In the analysis, the strength requirements for rising only, are considered; at the same time ensuring that the subject has sufficient strength to sit down in the chair aid (i.e. that the level of adjustment of the springs is not too great).

On the results sheet for each subject, the minimum moment required to rise for each subject is shown for unaided rising, and for rising in the chair aid. The ratio of the muscular moment requirements are seen to be fairly consistent.

We can now compare the average strength requirements in rising unaided, and in rising in the chair aid for the group as a whole. To make the results more meaningful, we can also calculate the able-bodied strength of the group from the strength equation in Chapter 2. We can also calculate from the tests the average residual strength of the group, and can thus find the percentage of the able-bodied strength required to rise unaided, and in the chair aid, and also the average residual strength of the group.

6.5.2. Able-bodied Strength Of The Group.

The average body weight, height, and build of the test group are now calculated ($W = 152$ lbf., $H = 65.2$ ins., $r = 3.75$, $t = 3.875$), and are substituted into the strength equation in Chapter 2 to find the average muscular strength of the group.

We have:-

$$M(5) = W^{-0.2}H^{1.5}(4.44r + 4.60t - 11.75)$$

giving $\underline{M(5) = 4380 \text{ in-lbf.}}$

This figure represents the average for the group of the sum of the maximum strengths at five specified joints. Using the muscle group strength ratios from Chapter 2, we can predict the healthy strength in the muscles applying moments in knee extension, hip extension, shoulder horizontal adduction, and elbow extension. These will be:-

M_H = 1820 in-lbf at each hip joint,

M_K = 1350 in-lbf at each knee joint,

M_S = 534 in-lbf at each shoulder joint,

and M_E = 407 in-lbf at each elbow joint.

6.5.3. Maximum Measured Strength Of Group.

In the chair aid tests, the subjects were asked to exert as great a force as possible in the different modes. We can thus use the largest calculated value of the muscular moment exerted at the hip and knee joint as the maximum strength of each subject at these joints; and can calculate the maximum, or residual strength of the group.

$$M_H = 486 \text{ in-lb}f \text{ at both hip joints,}$$

$$\text{and } M_K = 330 \text{ in-lb}f \text{ at both knee joints.}$$

This gives values for the average residual strength of the group as:-

<u>13.4%</u> in hip extension
and <u>12.2%</u> in knee extension.

6.5.4. <u>Unaided Rising Strength Requirements For Group.</u>

The average strength required to rise unaided is found from the appendix (Section 6). Unaided rising requires a combination of hip, knee, and elbow extension, and shoulder adduction/flexion. (The latter is assumed to be approximately equal in magnitude to the shoulder horizontal adduction, measured in the muscle strength tests).

The average for the group, of minimum strength requirements can be given as:-

$$M_H + M_K + M_S + M_E = 1571 \text{ in-lb}f \text{ for both limbs.}$$

The expected able-bodied strength in both arms and both legs is found from the calculation of the able-bodied strength in section 6.5.2., and from the muscle group ratios in Chapter 2; and is given by:-

$$M_H + M_K + M_S + M_E = (0.415 + 0.308 + 0.122 + 0.093) \times 4{,}380 \times 2$$

$$= \underline{8222 \text{ in-lb}f.}$$

This gives a value for the average percentage strength required for unaided rising as 19%.

6.5.5. <u>Strength Requirements For The Chair Aid.</u>

The average for the group of the maximum seat hinge torque, required in the rising cycle is 496 in-lb$f$. This is said to be numerically equal to the sum of the hip and knee moments required to rise in the chair aid.

$$\text{i.e.} \quad M_H + M_K = \underline{496 \text{ in-lb}f} \text{ in both limbs.}$$

The average for the group of the full strength at the knee and hip joints is <u>6340 in-lb$f$.</u> Thus the strength required to rise in the chair aid can be given as <u>7.8%</u> of the able-bodied strength.

6.5.6. Comparison Of The Strength Requirements.

The average residual strength of the group (12.8%) lies between the strength requirements for aided (7.8%) and unaided (19.1%) rising. The implications of these figures were seen to be true in practice in that none of the members of the group were able to rise unaided, but most could rise in the chair aid.

One surprising aspect, revealed by these figures, is the low percentage strength requirements for rising from a chair. This figure of 19.1% is not surprising however, when one considers that most healthy people are capable of rising fairly easily, using just one leg (although very few people will normally do this). The strength requirements of the chair aid (7.8% of the full strength) are approximately 40% of the strength requirements for unaided rising.

From the findings of the tests, it would seem desirable to have some adjustment of the seat length and its associated linkage, to cater for extremes of physique of subjects.

Chapter 7

7.1.1. The Logarithmic Strength Scale.

Physical strength, and the ability of a person to perform a task requiring physical strength, has been expressed in a number of different ways. In the field of sport, an athelete is either timed in his speed of performance of a task, or is measured in the distance he can throw or jump. As a test of pure strength he can lift weights; his strength being gauged as the heaviest weight he can raise above his head. This measurement is simple and practical, and measures the strength at the four main limb joints in knee, hip, and elbow extension, and in shoulder flexion/abduction. However, it does not differentiate between the muscle groups, and is only suitable for able-bodied people.

For disabled people, the main techniques used to express physical strength are as follows. Firstly there is a subjective notation, developed for use by physiotherepists and doctors to assess the strength of a patient in his muscle groups. The strength of the muscle group is placed in one of six catagories, and is graded from 0 to 5.

5 .... normal.

4 .... good, but with reduced strength.

3 .... fair.

2 .... able to lift limb against gravity.

1 .... Trace of movement in muscles.

0 .... No trace of muscular activity.

This scale is useful for measuring the progress of, or recovery from, paralysing diseases; but cannot be extended to measurement of strength in able-bodied people. Also, in being subjective, this technique is not accurate enough to be applied to the measurement of the effectiveness of an aid, or to measure the suitability of a subject for a particular aid.

Another scale, used mainly for multiple sclerotics, evaluates the disability by an overall function, based on the ability of the patient to carry out common activities of daily life. 78 activities are listed; a score of 1 is given for each success, and 0 for failure in each activity. The degree of severity in terms of the overall ability is loosely grouped into four catagories:-

1. Confined to bed.
2. Restricted to wheelchair.
3. Ambulatory with aids.
4. Independent ambulatory.

Lastly we have the technique used throughout this thesis so far, in which the strength is expressed as a torque about a body joint. This scale of measurement is suitable for both able-bodied, and disabled subjects, and is ideal for use in carrying out calculations; its main disadvantage however is its non-descriptiveness. A scale is required which magnifies low strength values and where the numbers used are of reasonable size at the upper end of the strength spectrum. Such a scale is the logarithmic, or decibel scale. After due consideration and trials, the best form of the scale was found to be based on the ratio of the muscular moment generated at the body joint, to an arbitrary base value of the muscular moment, taken as 1 in-lbf. It is desirable that, when used here, the decibel unit should be consistent with its use in other fields, where it is given as ten times the logarithem of the ratio of energy, work, or power levels. When such quantities as force, pressure, or moment are involved, the square of the ratio is used. To simplify this, the decibel unit is given as twenty times the logarithem of the ratio of such units as force, pressure, or moment.

We thus have the strength in decibels given by:-

$$S = 20 \log_{10} \frac{M}{M_o}$$

where M is the muscular moment exerted about a joint (or the sum about several joints); and $M_o = 1$ in-lbf.

$$\text{Thus } S = 20 \log_{10} M$$

where M is in in-lbf.

Figure (46) shows the muscular moment in in-lbf. plotted against the logarithmic strength in decibels.

We can now apply the logarithmic scale to the results obtained so far in this thesis.

7.2.1. The Muscular Strength Of Able-bodied Subjects.

The strength equation, synthesized in Chapter 2 gives the muscular moment at five specific body link joints by:-

$$M(5) = W^{-0.2}H^{1.5}(4.44r + 4.60t - 11.75)$$

The logarithmic strength in decibels may now be given by:-

$$S = 30 \log_{10}H - 4 \log_{10}W + 20 \log_{10}(4.44r + 4.60t - 11.75)$$

In the muscle strength tests. the strengths recorded ranged from 67. 67.86 dB(2473 in-lbf.) up to 78.12 dB(8049 in-lbf.); a difference of 10.26 dB. This range is a measure, not of the numerical difference between the muscular moments, but of the ratio of the moments. The standard deviation is 1.26 dB.

We also have the ratio of the strength inherent in the five joints tested, given by average values for the group of:-

66.02 dB in hip extension.

63.44 dB in knee extension.

55.38 dB in shoulder horizontal abduction.

53.04 dB in elbow extension.

and 49.64 dB in neck flexion.

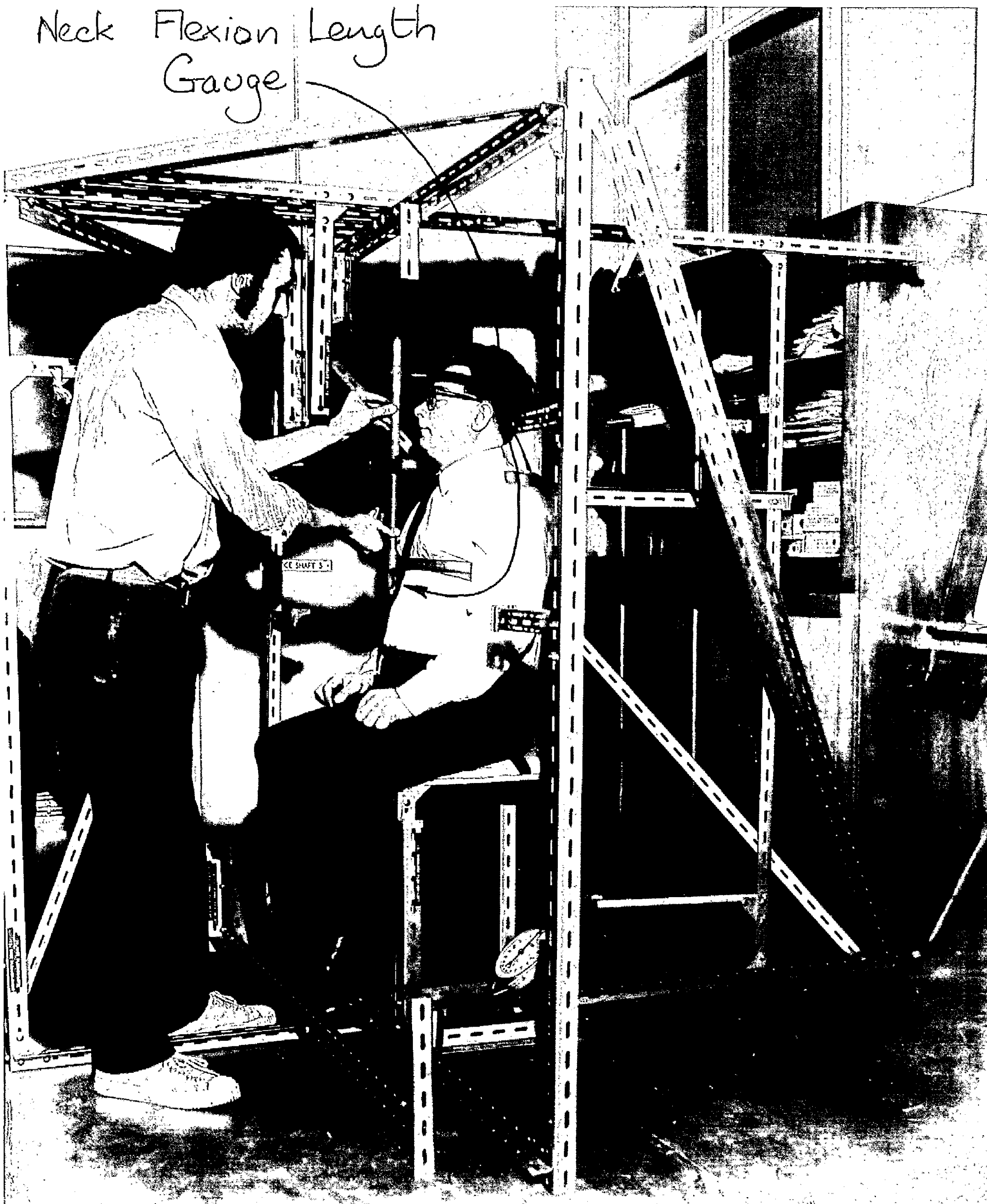
Also the average for the group of the total strength in the five joints is given by:-

$$S = 73.66 \text{ dB.}$$

We can now give the ratio of the strengths in each muscle group to the total strength. We have the ratio of the strength in hip extension to the total strength as 7.64 dB.

Also	the ratio in knee extension	= 10.22 dB
	the ratio in shoulder horizontal adduction	= 18.28 dB
	the ratio in elbow extension	= 20.62 dB
	the ratio in neck flexion	= 24.02 dB

FIGURE 1



MUSCLE STRENGTH TESTS

NECK FLEXION

7.3.1. Tests On The Floating Arm Support.

In the first part of the tests, the effect has been found of the arm support on the ability of the subject to hold the arm against gravity, and to apply a vertical force at the hand. The experimental results of the tests themselves are used to find statistically the effect of the arm support on the function of the partially paralysed arm; and the reasons for this effect. This is reported and explained in Chapter 3. Also in Chapter 3, the muscular moment at the shoulder is calculated, in applying a force of 1 lbf at the hand, with the arm support and unaided (Figures (19) and (20)).

We can now expand this to plot graphically the strength requirements at the shoulder in decibels, as the force at the hand is varied; and the effect of the arm support on these requirements. Figure (47) shows this. As can be seen, the arm support is only effective for applying a force of less than 1 lbf, and is most effective when used to support the arm only.

7.4.1. Tests On The Arm Support As Used Therepeutically.

The results as given in Figure (21) can now be translated into the decibel strength scale. In these tests, the force exerted by the patient is measured at the forearm in a number of positions as the hand travels along the "hand to mouth" path. An average is then taken of all these measurements. Tests are repeated fortnightly. The muscular moments required at the joints to exert this force (particularly at the shoulder joint) will be the product of the force and some distance, (for example, the distance from the point of application of the force, to the shoulder joint). However, the average of this distance will be constant for each fortnightly test. In expressing the strength requirement on the decibel strength scale, this unknown distance is irrelevant, as its effect will only be to shift all the points on the graph for each subject, an equal distance up the axis. As the choice of numbers on this axis is arbitrary, and as we are only interested in the gradient of a line drawn through the points, (i.e. the speed of recovery), we can express the force exerted logarithmically, and take measurements from this.

It is expected that in hemiplegia, the patient will increase in strength steadily over the initial stages of recovery, and that this

FIGURE 2

HIP EXTENSION

FIGURE 3

KNEE EXTENSION

FIGURE 4

SHOULDER HORIZONTAL ADDUCTION

FIGURE 5

ELBOW EXTENSION

increase will diminish as the patient nears complete recovery. These tests take place over the initial stages of recovery; and thus it will be assumed that the logarithmic strength will increase linearly. The gradient of a straight line through the points will give the rate of recovery in decibels, which will be the average ratio of the forces exerted in each test. Figure (57) shows the results for four patients taking part in the experiment. A straight line, estimated to be the best fit, is drawn through the points for each patient. The gradient of the line, or the rate of recovery of the patient, is shown on the diagram.

7.5.1. Unaided Rising, and Rising In The Chair Aid.

In Chapter 6, a fairly comprehensive analysis of these results are given. On the results sheet for each subject (Figure (44)), a scale is included showing the strength requirements in decibels for rising in the chair aid, for rising unaided, and for the maximum strength of the subject in knee extension plus hip extension. In fact it is not entirely valid to compare this maximum strength with the strength requirements for unaided rising, because the latter includes the strength at the shoulder and elbow joints. If the percentage residual strength of the arms of the subjects tested were equal to that of their legs, the total strength of the subjects, as used for comparison with unaided rising, would increase by approximately 2 dB.

Most of the subjects require approximately 10 dB more strength to rise unaided than in the chair aid; and in most cases, the strength of the subject lies between the two. Whereas both of the strength requirements (in aided and unaided) for rising will be fairly constant, the strength of the subject will probably be a function of time; depending on the nature of his disability. The usefulness of the aid will cover the time during which the strength of the subject, in moving up or down the scale, passes through this band between the strength requirements for aided and unaided rising.

7.6.1. Conclusion.

The author should like to conclude this thesis with a tasty diagram (Figure (48)), which shows all the major results in this thesis, presented on one logarithmic scale.

Figure 7

THE SHELDON TRIANGLE

for

somatatype classification

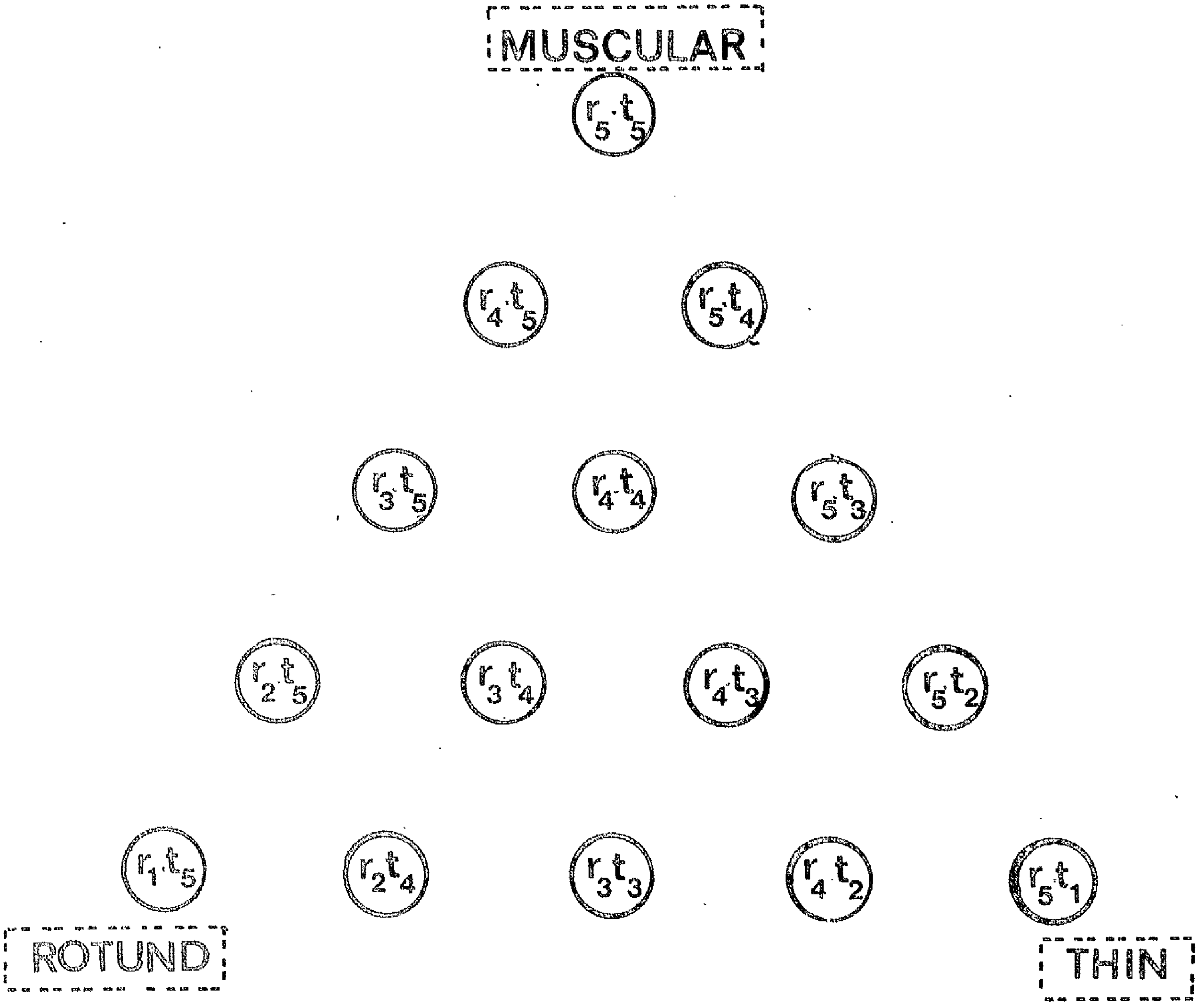


Figure 9 HISTOGRAM OF MUSCLE GROUP STRENGTH RATIOS

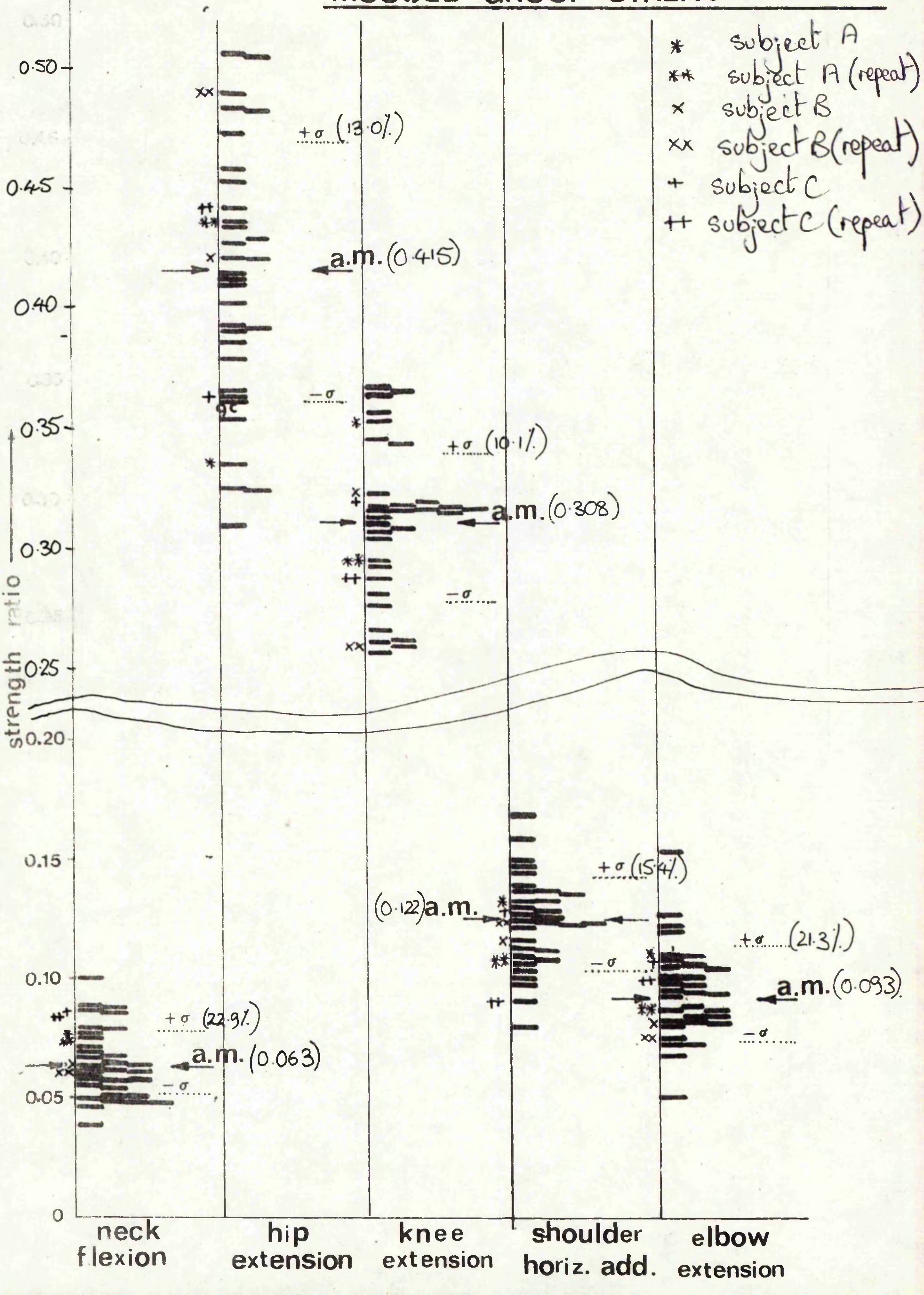


Figure 10

THE FORM OF THE AGE FUNCTION USED IN THE STRENGTH EQUATION

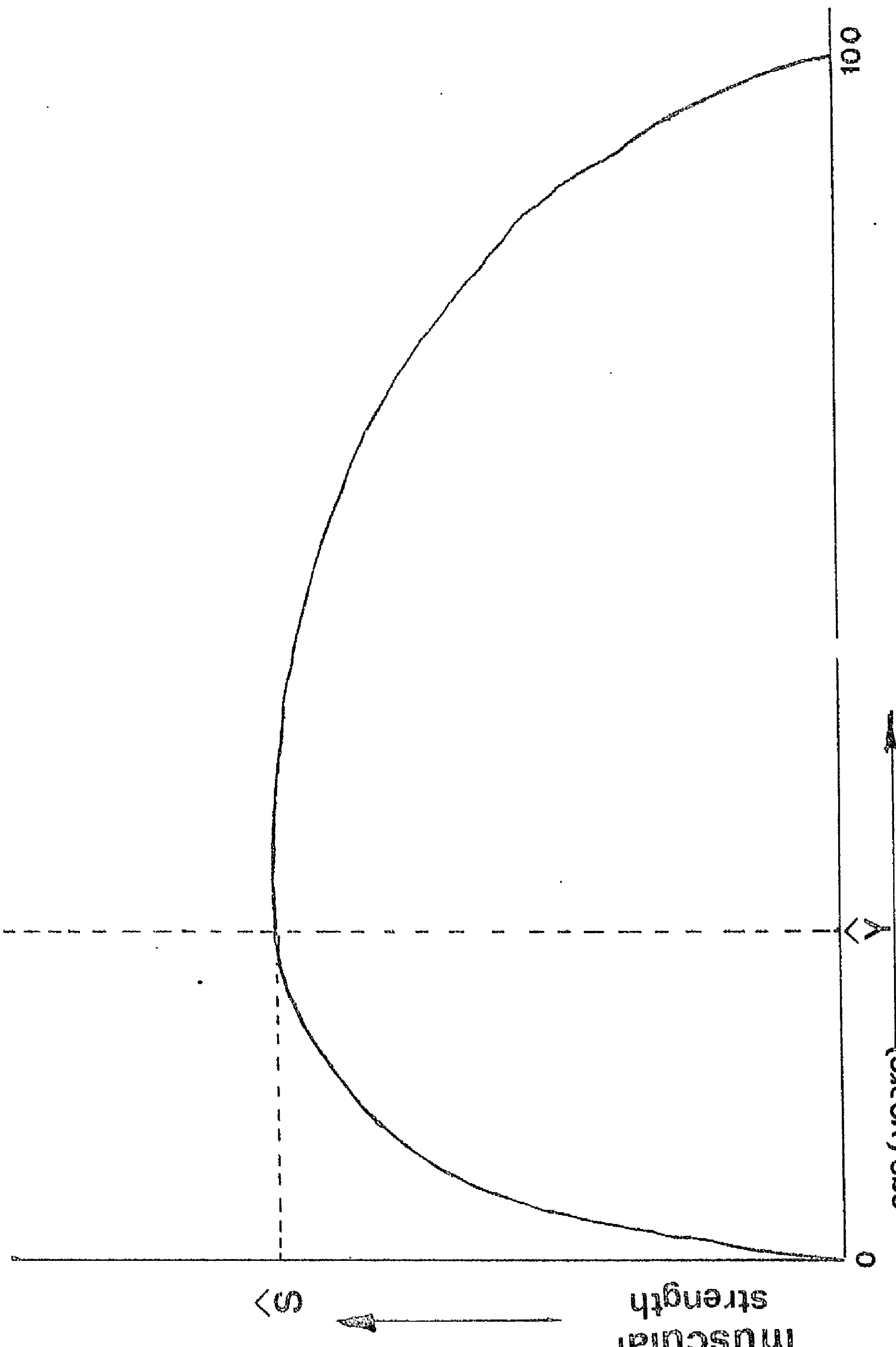


Figure 11

THE STRENGTH EQUATION

MEASURED v CALCULATED VALUES OF MUSCULAR STRENGTH

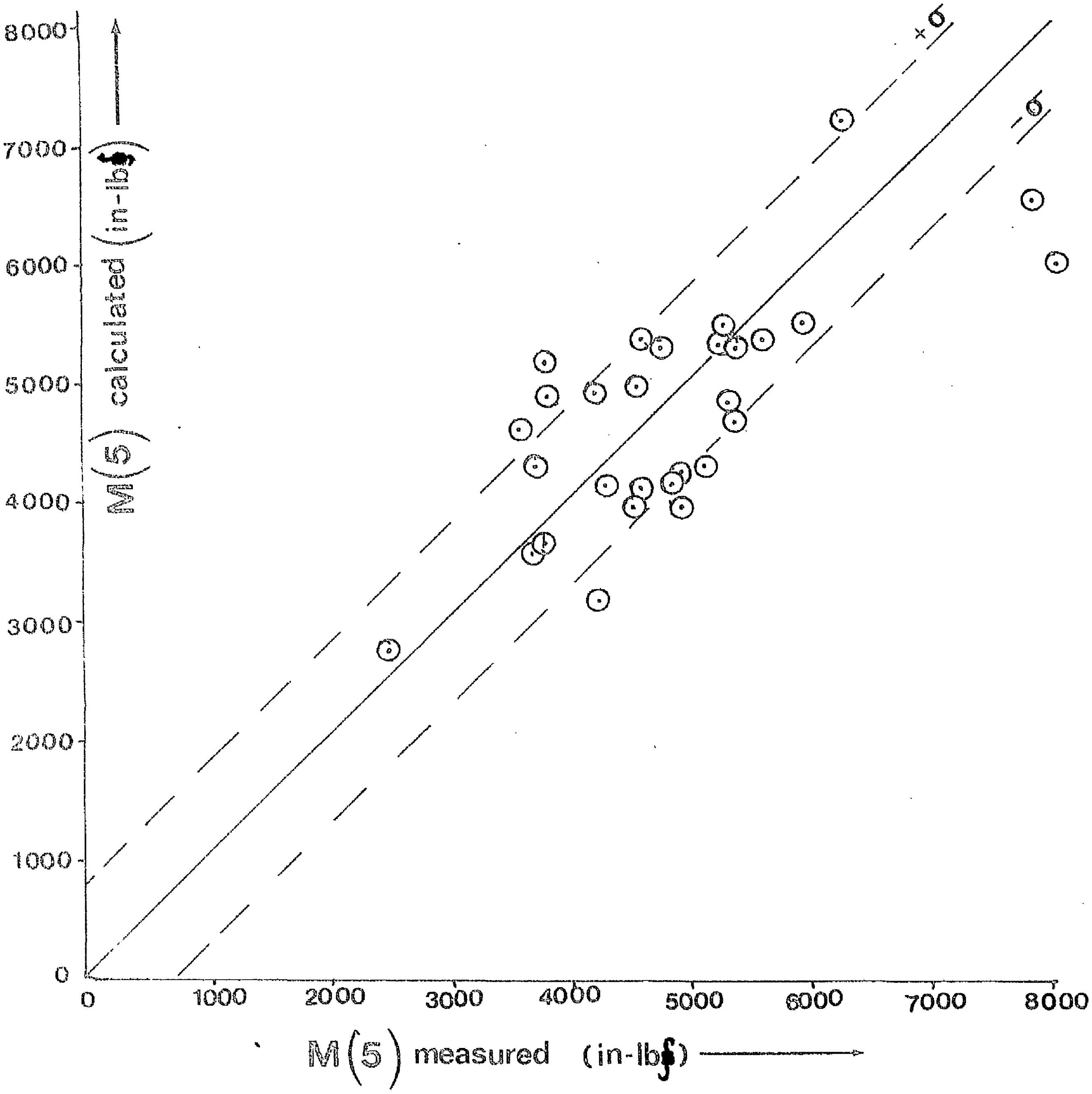
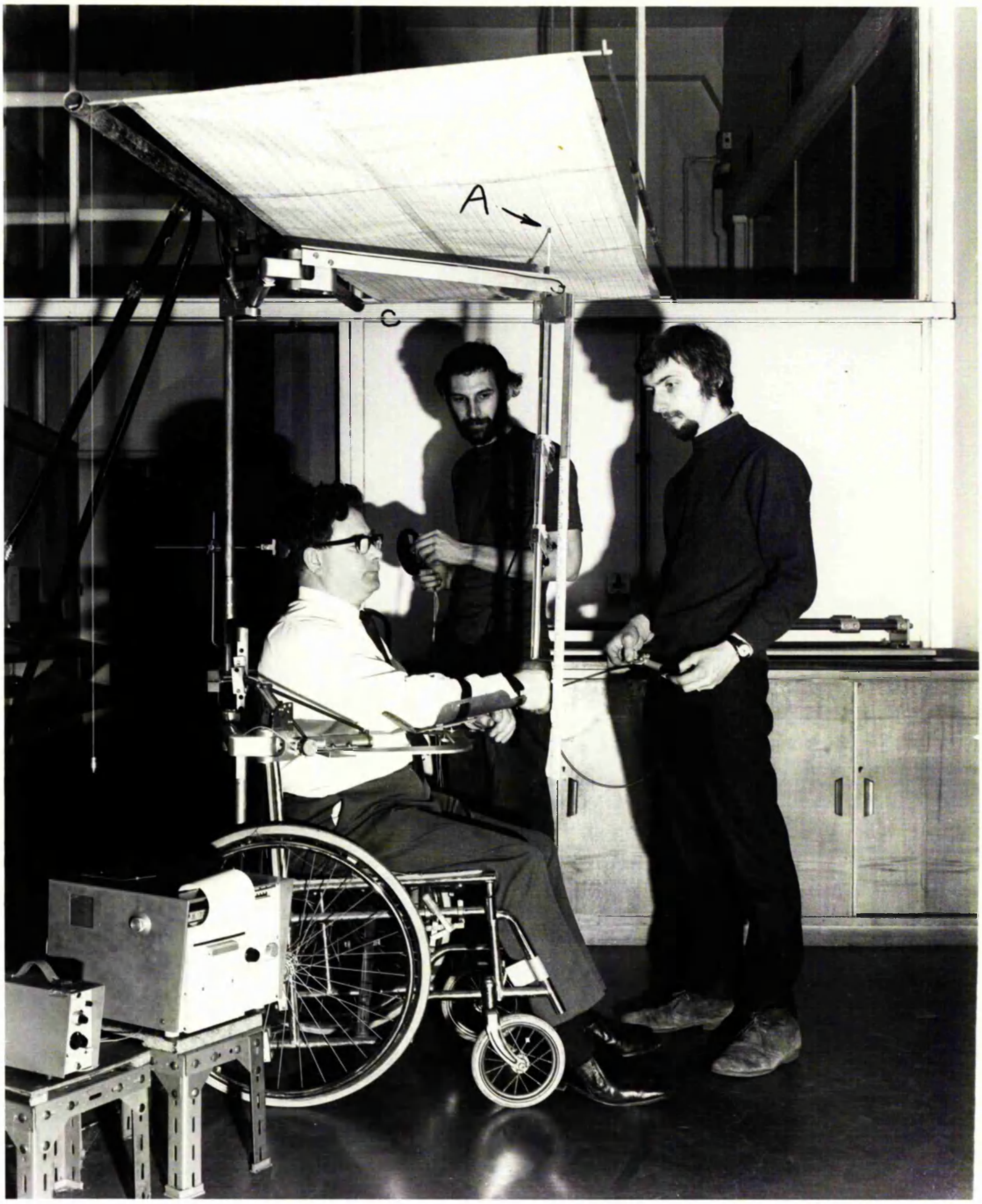


FIGURE 12



ARM SUPPORT TEST APPARATUS

THE COORDINATE SYSTEM ON THE CHAIR AS USED IN THE FLOATING ARM SUPPORT TESTS

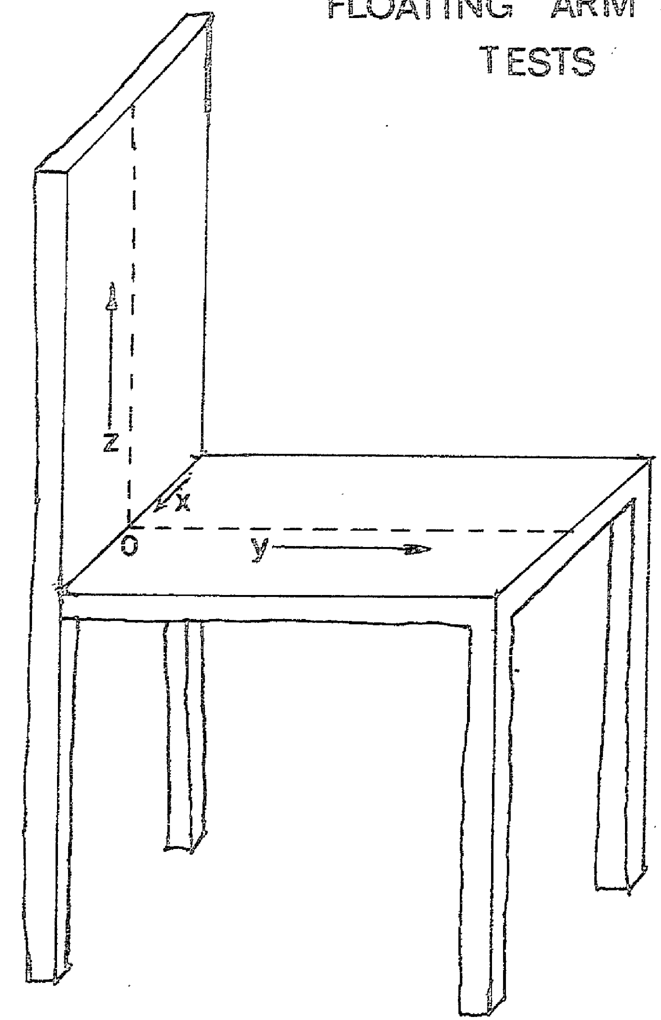


Figure 13

THE COORDINATE HAND DIAL CIRCUIT

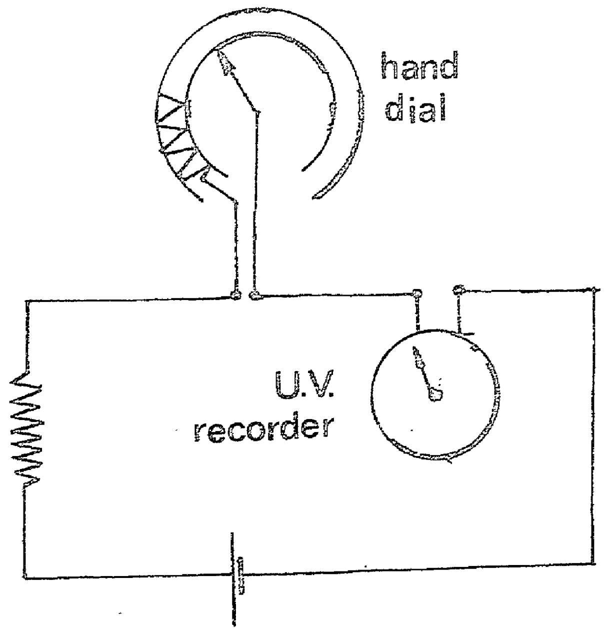
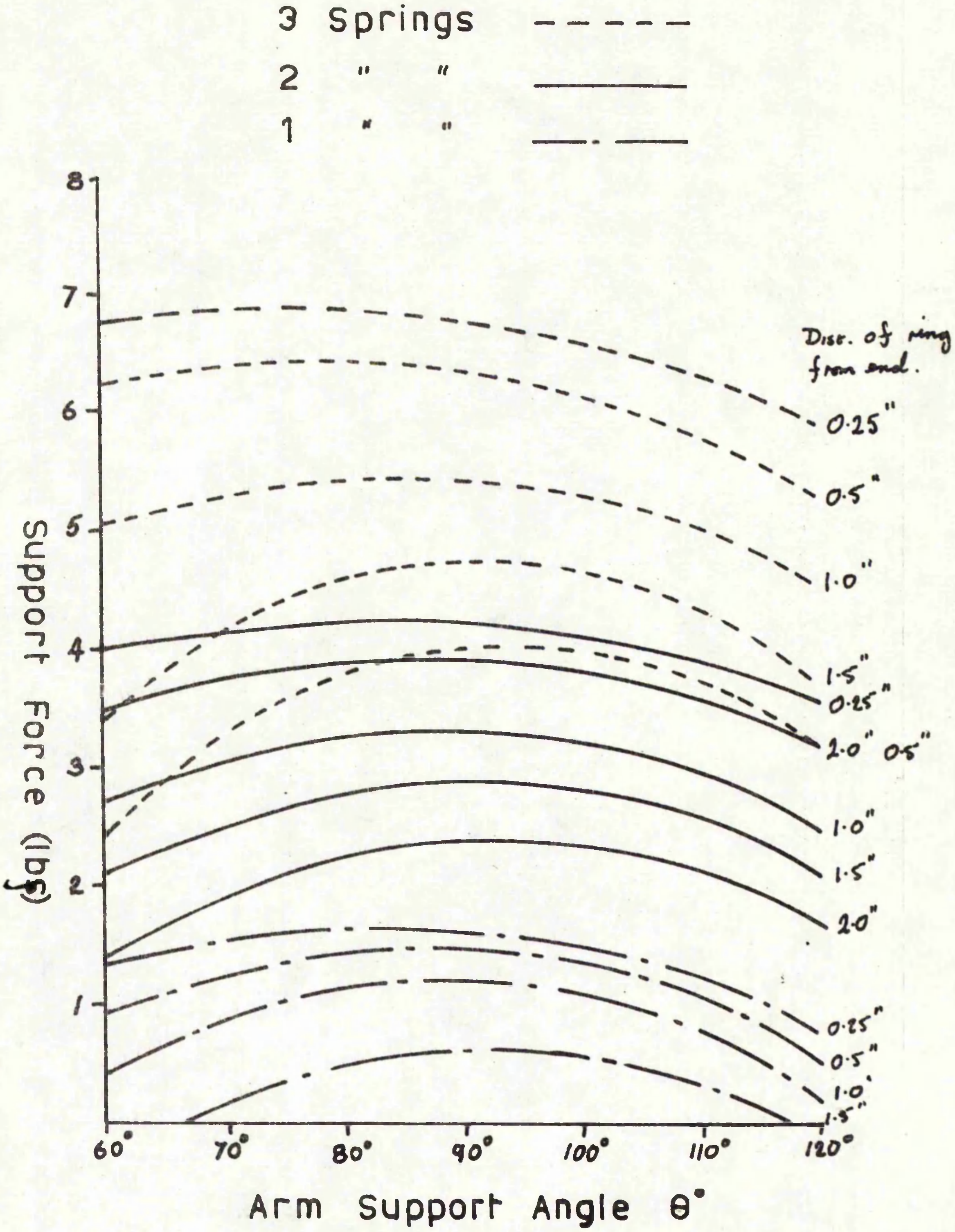


Figure 14



Support Force against Arm Support Angle.

FIG 15

FIGURE 16

A TYPICAL ARM SUPPORT TRACE

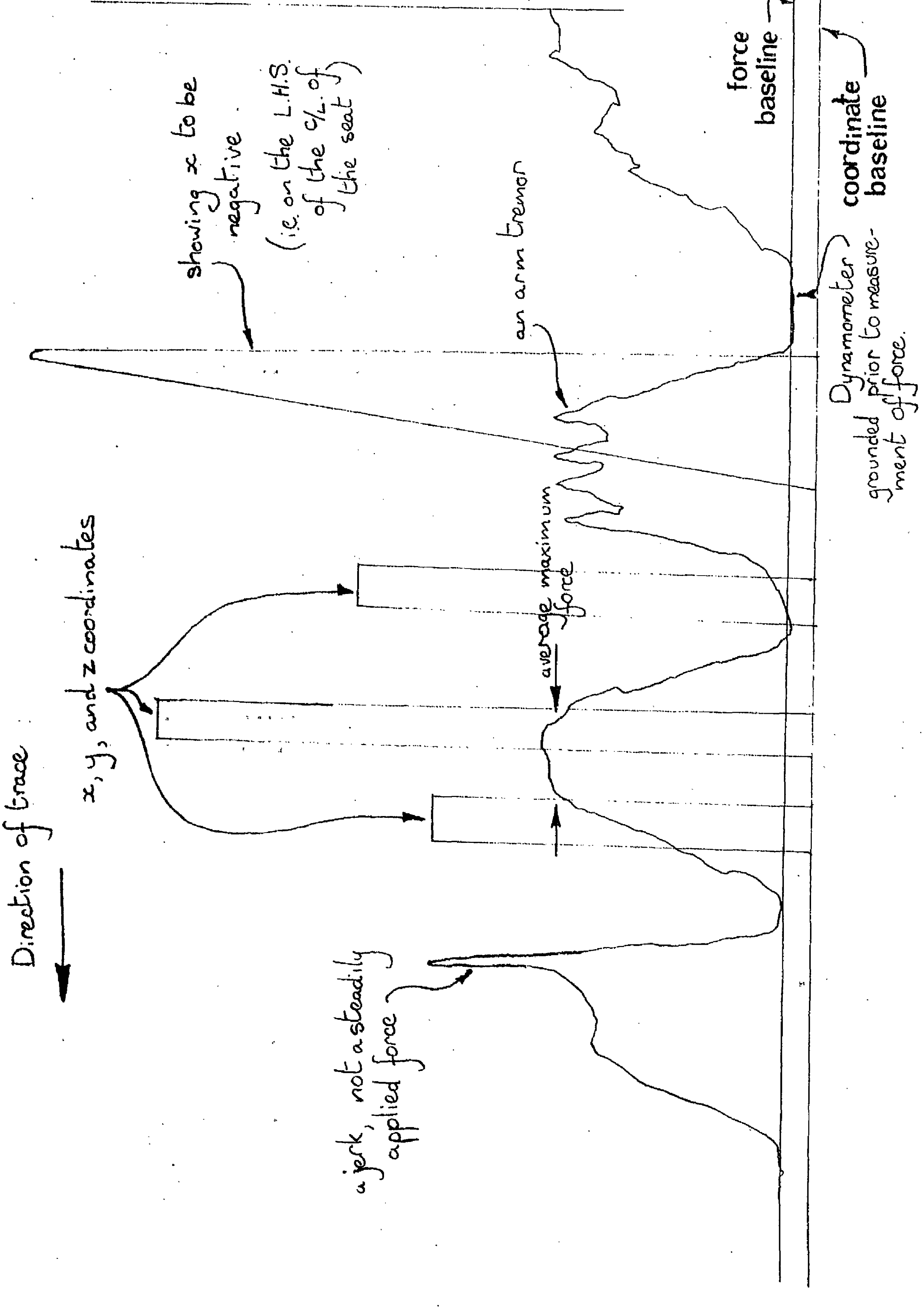
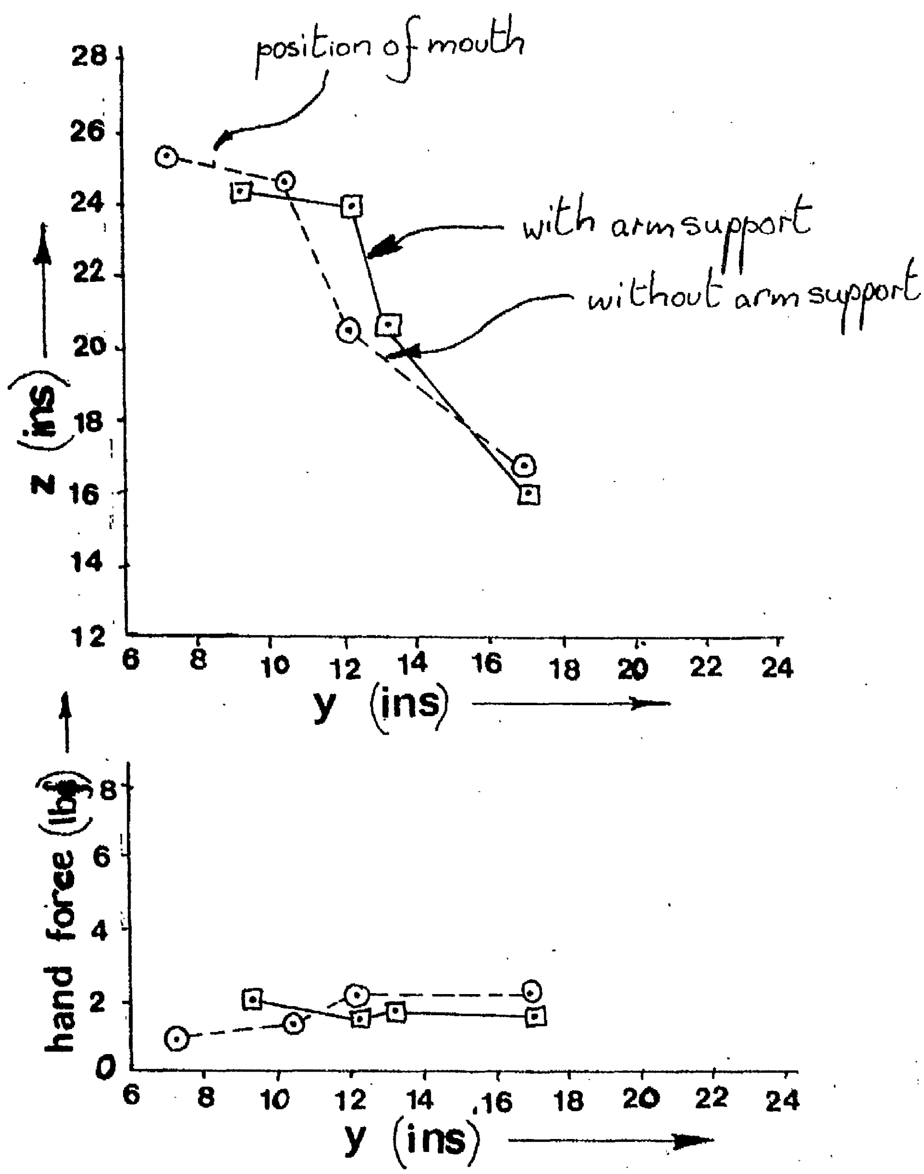


FIGURE 17

ARM SUPPORT TESTS:

A TYPICAL RESULTS GRAPH

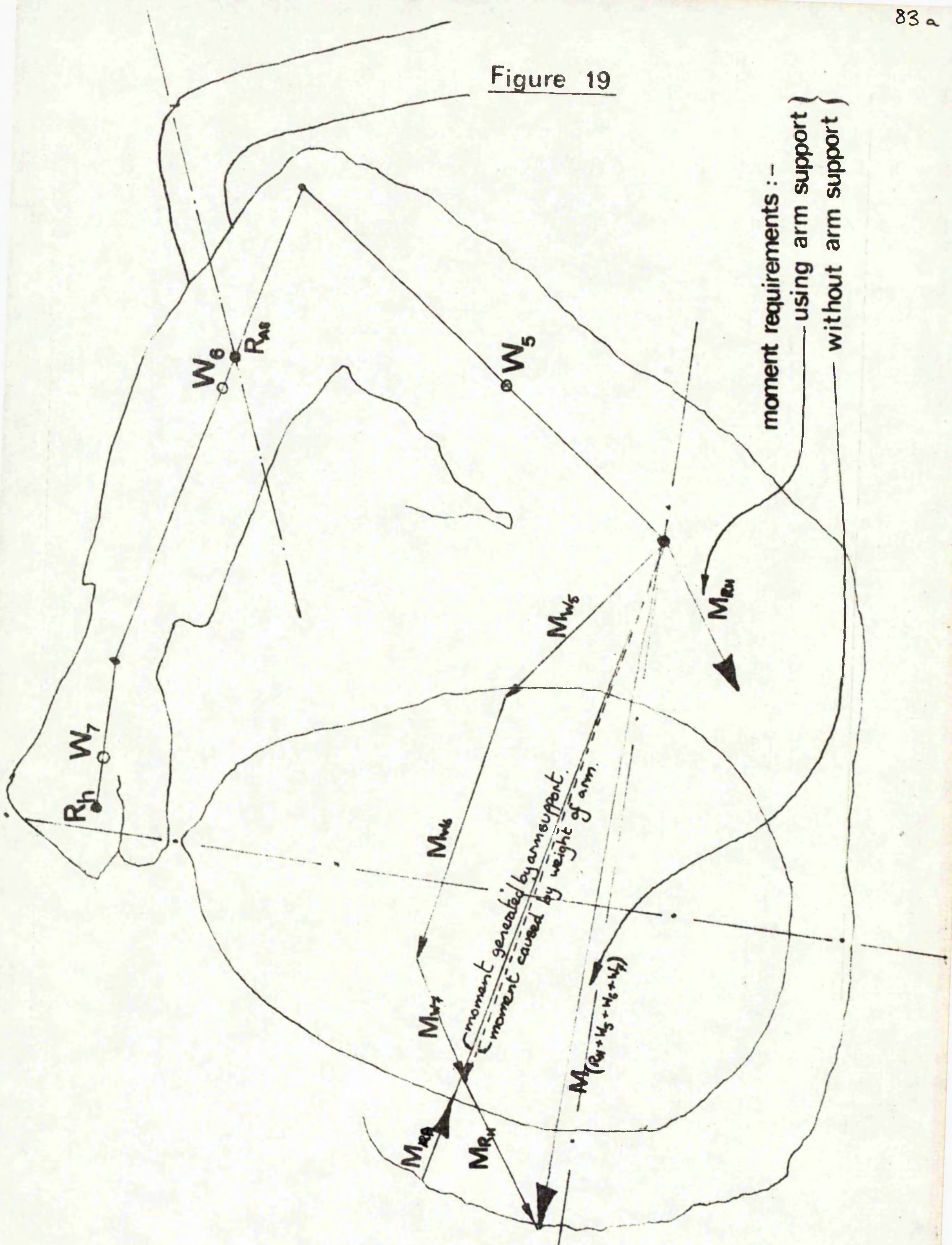


The X-coordinate is small (of average value less than 2 inches) compared to the Y and Z-coordinates (average value greater than 12 inches). The difference in the distance from the mouth, in taking the X dimension into account, will be approximately 0.1 inches. This is far smaller than the accuracy of measurement of the coordinates, and it was thus ignored.

FIGURE 19

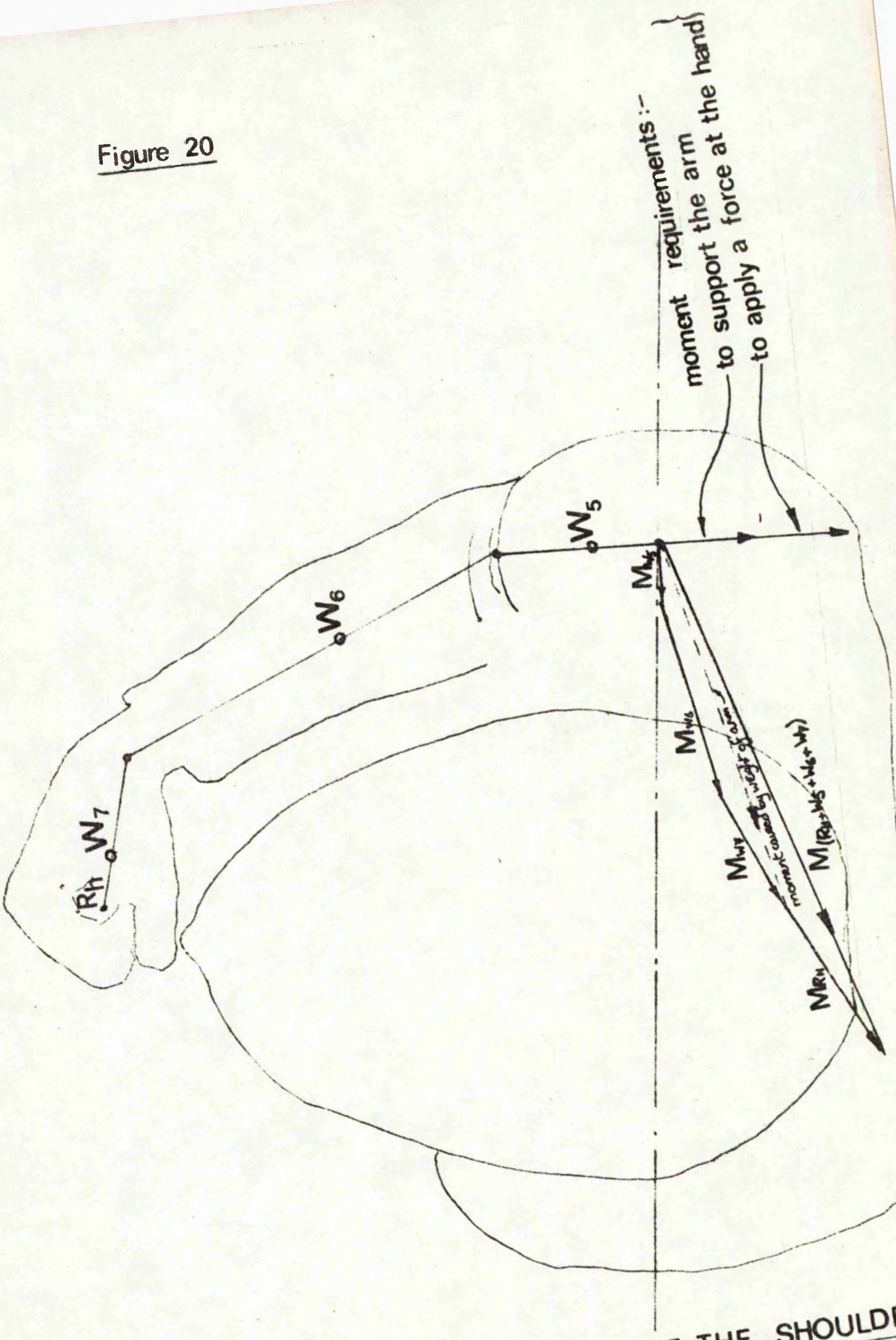
PLAN VIEW OF ARM SUPPORT IN USE

Figure 19



MUSCULAR MOMENTS REQUIRED AT THE SHOULDER JOINT IN USING THE FLOATING ARM SUPPORT

Figure 20



MUSCULAR MOMENTS REQUIRED AT THE SHOULDER JOINT UNAIDED

Figure 21

TESTS ON THE THEREPEUTIC EFFECT OF THE FLOATING ARM SUPPORT

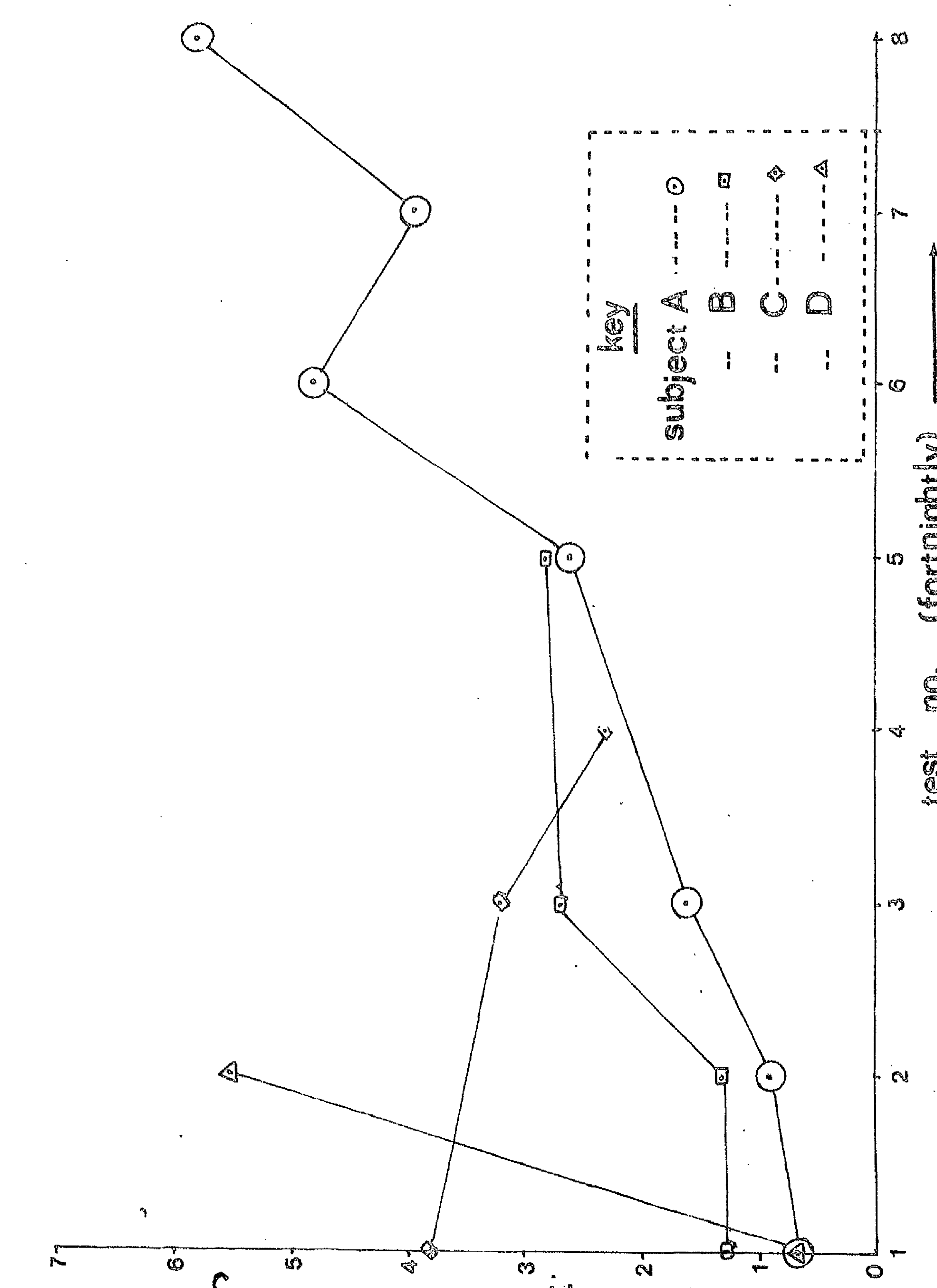


Figure 22

UNAIDED RISING WITHOUT USING THE ARMS

for a fifty percentile subject

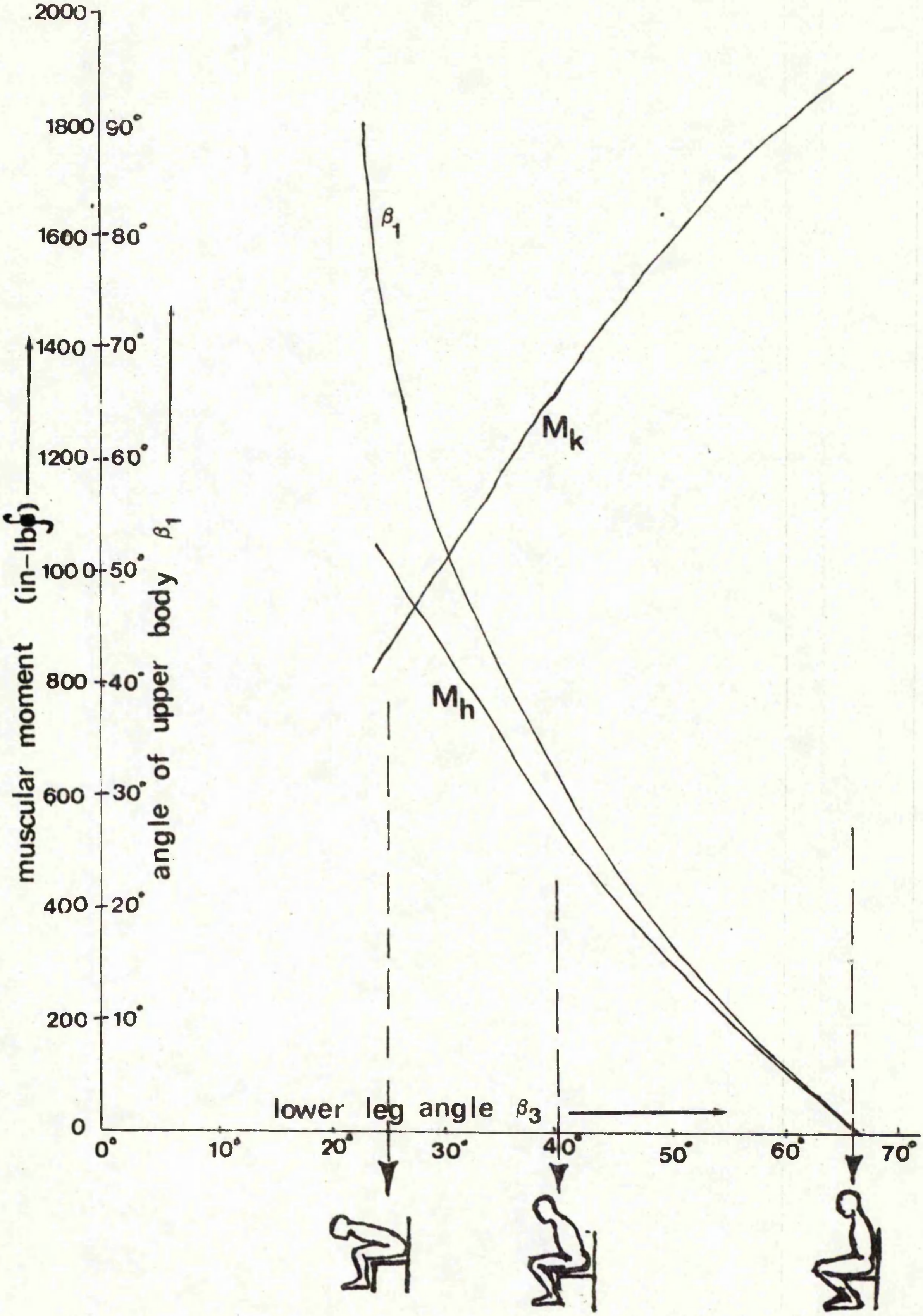
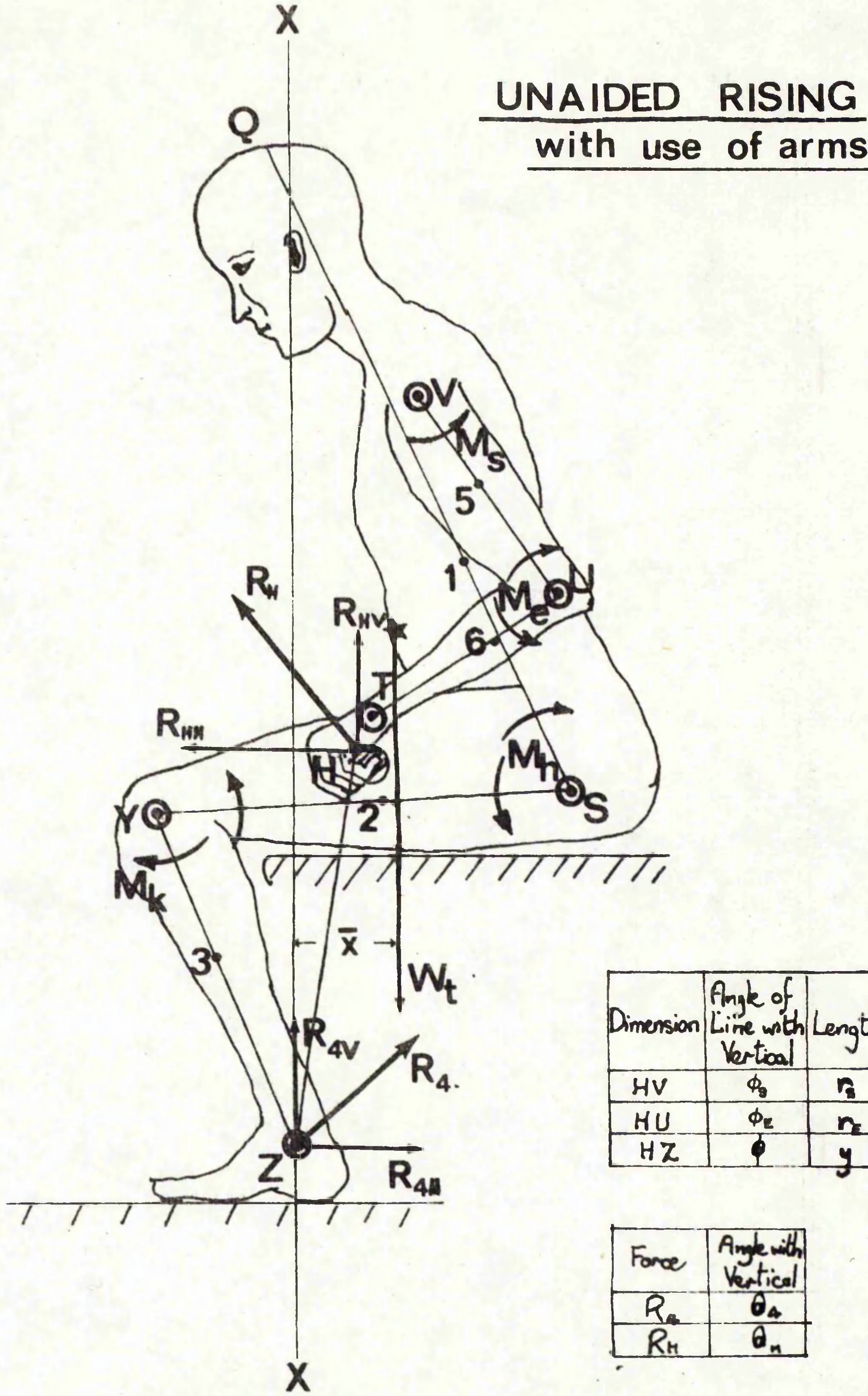


Figure 23

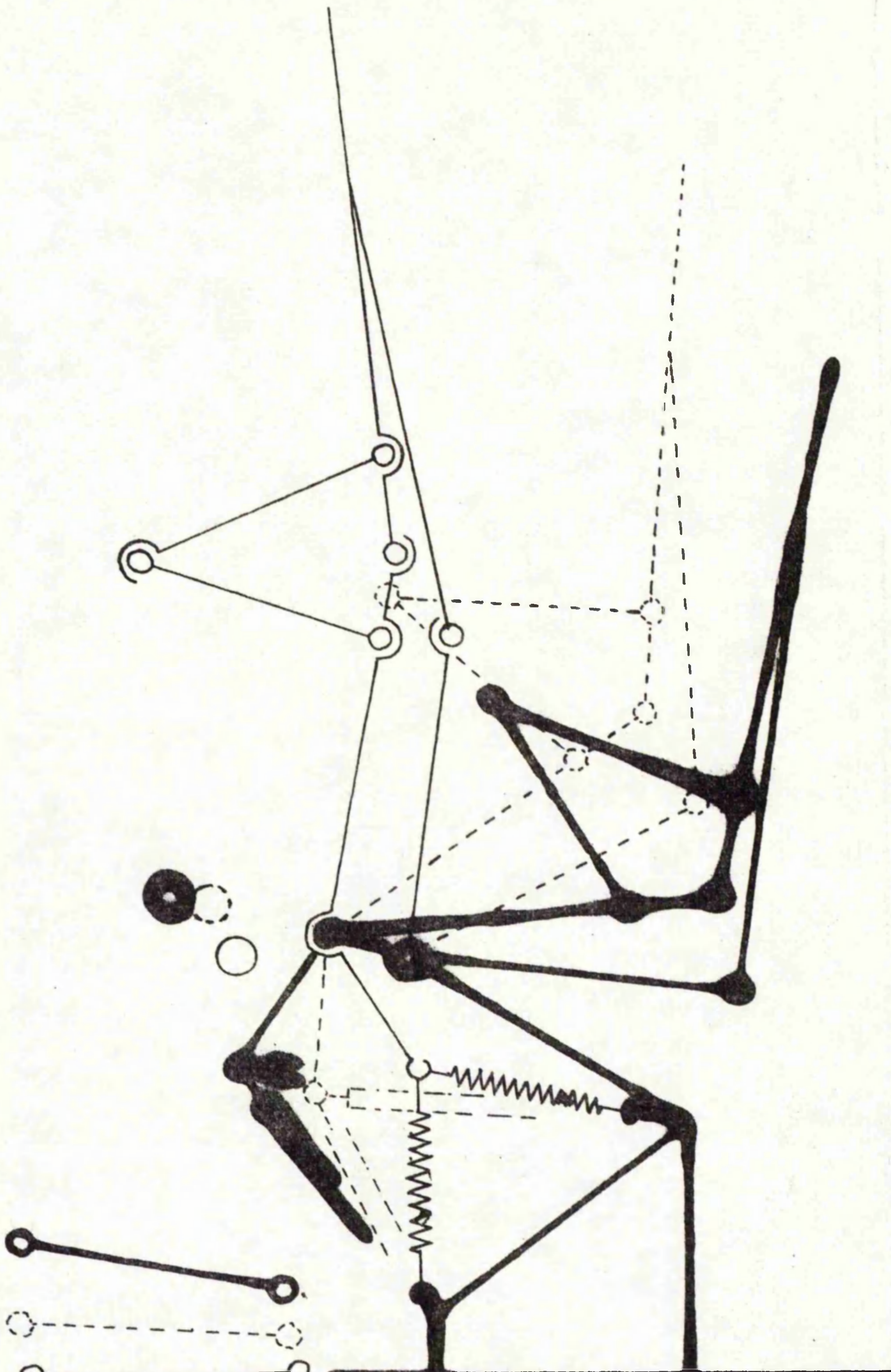


Dimension	Angle of Line with Vertical	Length
HV	ϕ_s	r_s
HU	ϕ_E	r_E
HZ	ϕ	y

Force	Angle with Vertical
R_4	θ_4
R_H	θ_H

Segment	Link	Angle of Link with Vertical	Weight of Segment	Link Length	Position of c.g.	Distance along Link of c.g.
Upper Body	QS	β_1	W_1	–	1	ℓ_1
Thigh	SY	β_2	W_2	d	2	ℓ_2
Leg	YZ	β_3	W_3	e	3	ℓ_3
Arm	VU	β_5	W_5	f	5	ℓ_5
Forearm	UT	β_6	W_6	–	6	ℓ_6
Hand	–	–	W_7	–	H	–

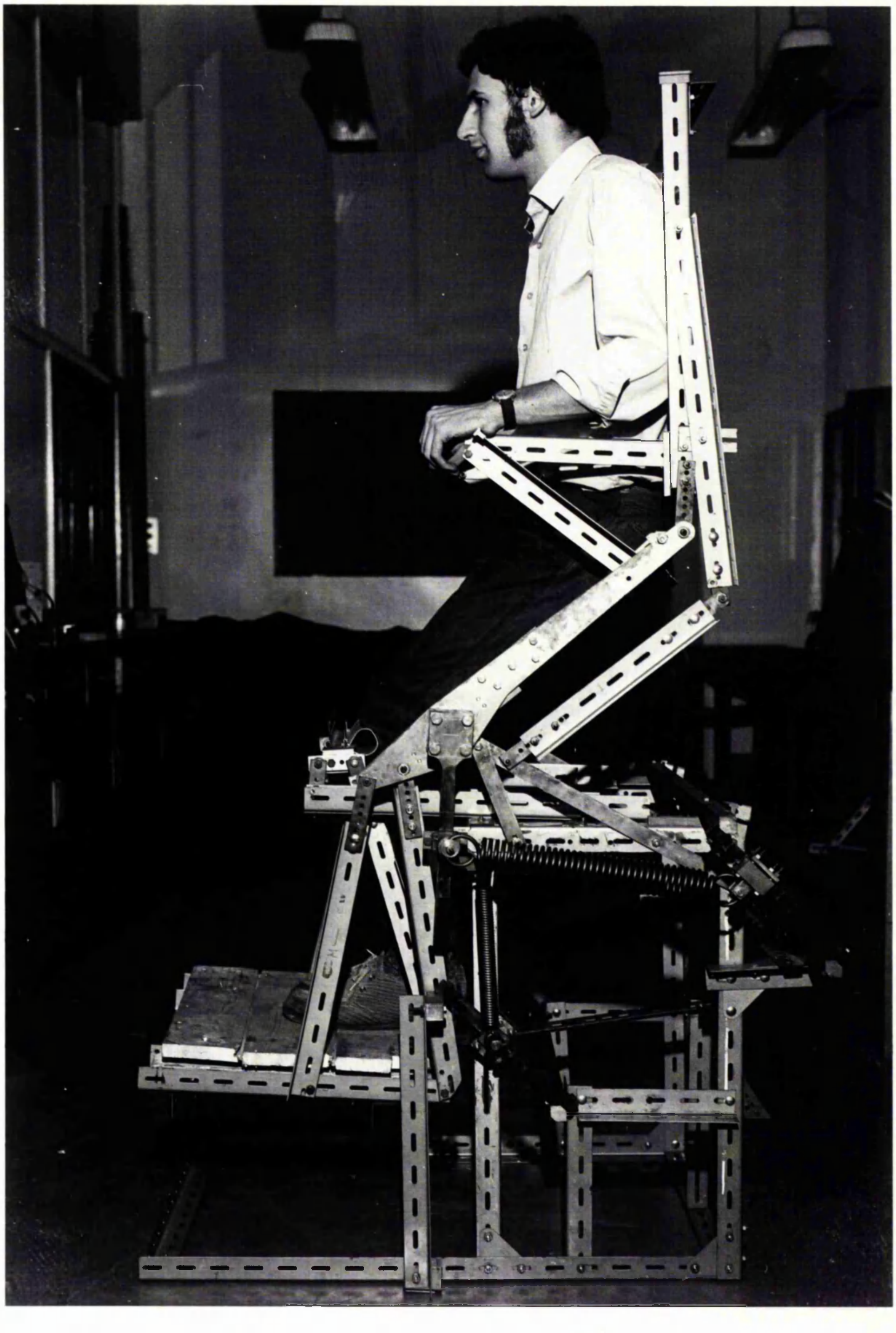
Figure 24



THE MECHANISM OF THE CHAIR AID TO RISING

shown in three positions in the rising cycle

FIGURE 24



THE CHAIR AID TO RISING

Figure 25

LOCKING ARRANGEMENT ON THE CHAIR AID

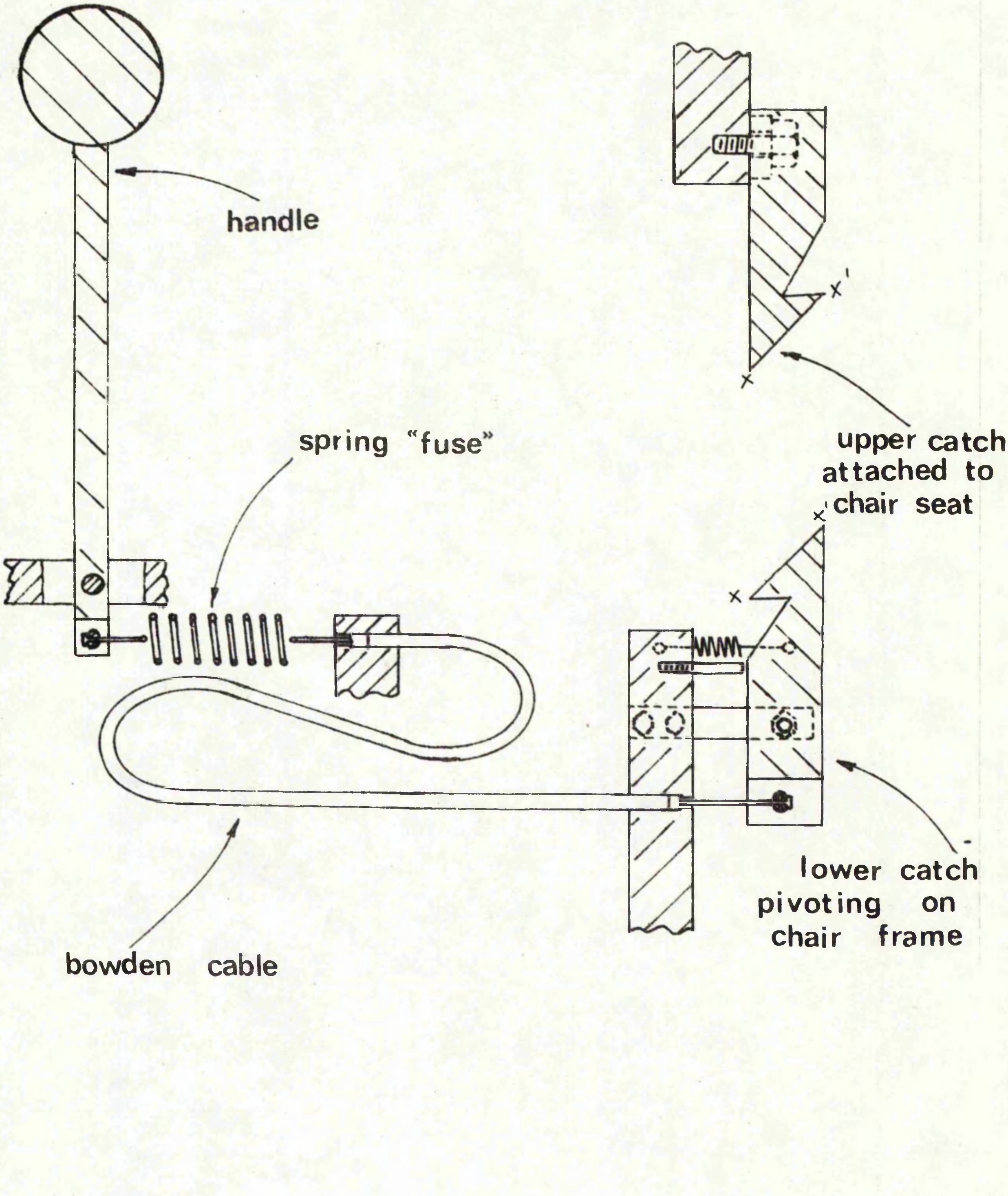


Figure 26

THE FORCE SYSTEM ACTING ON A SUBJECT SEATED IN THE CHAIR AID

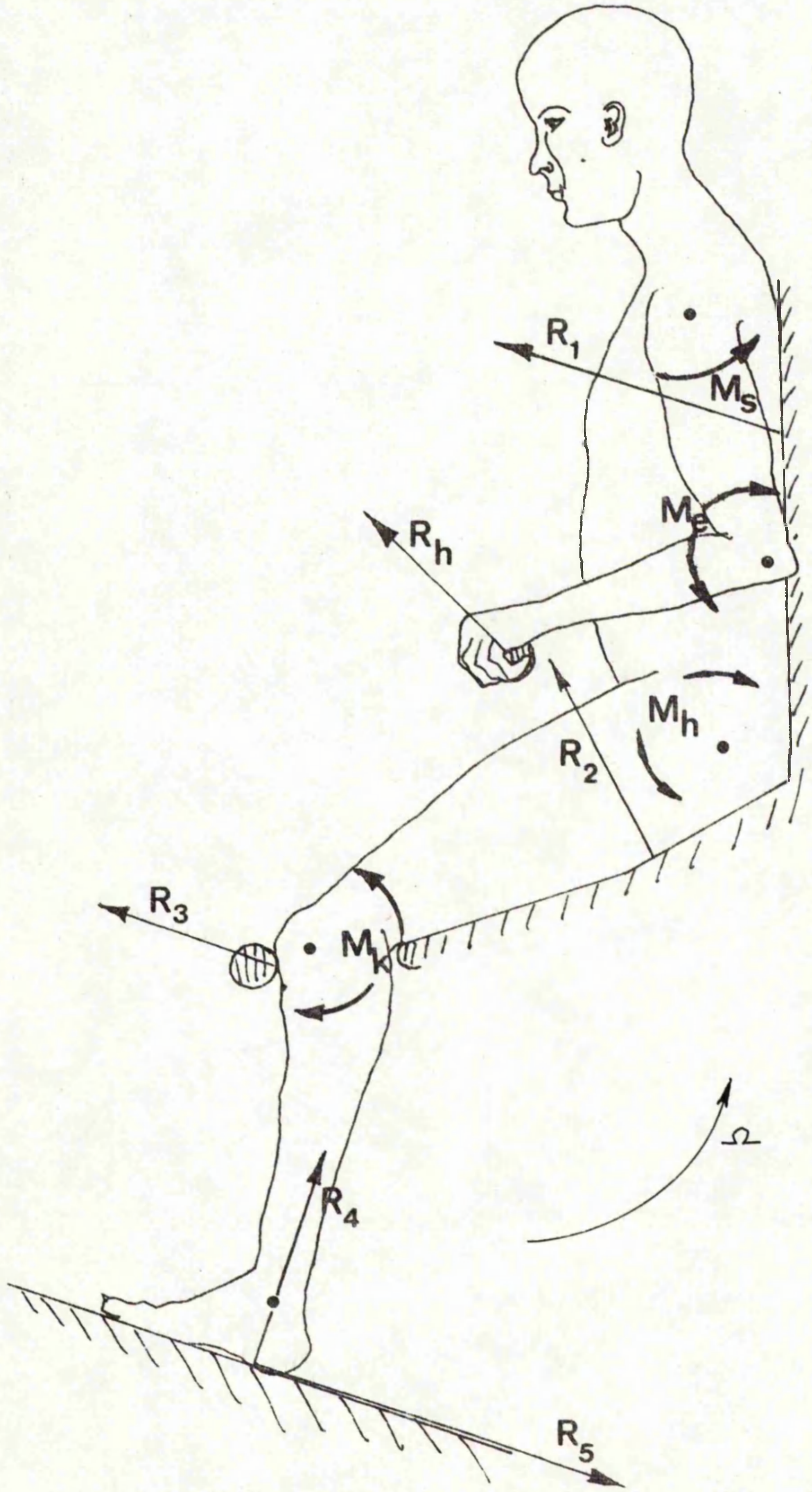
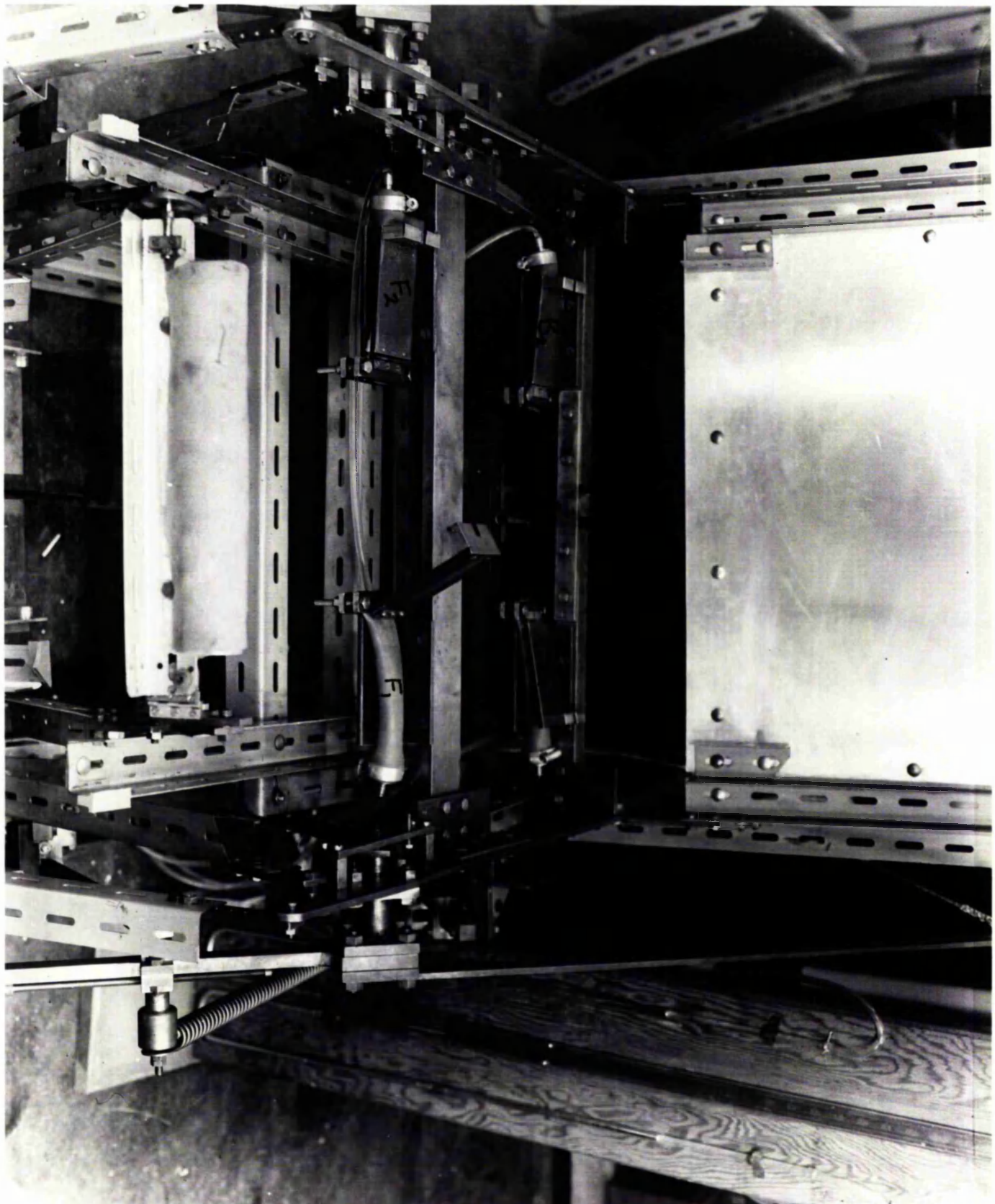


FIGURE 27



SEAT HYDRAULIC GAUGES

Figure 27

PLAN OF SEAT PRESSURE GAUGES

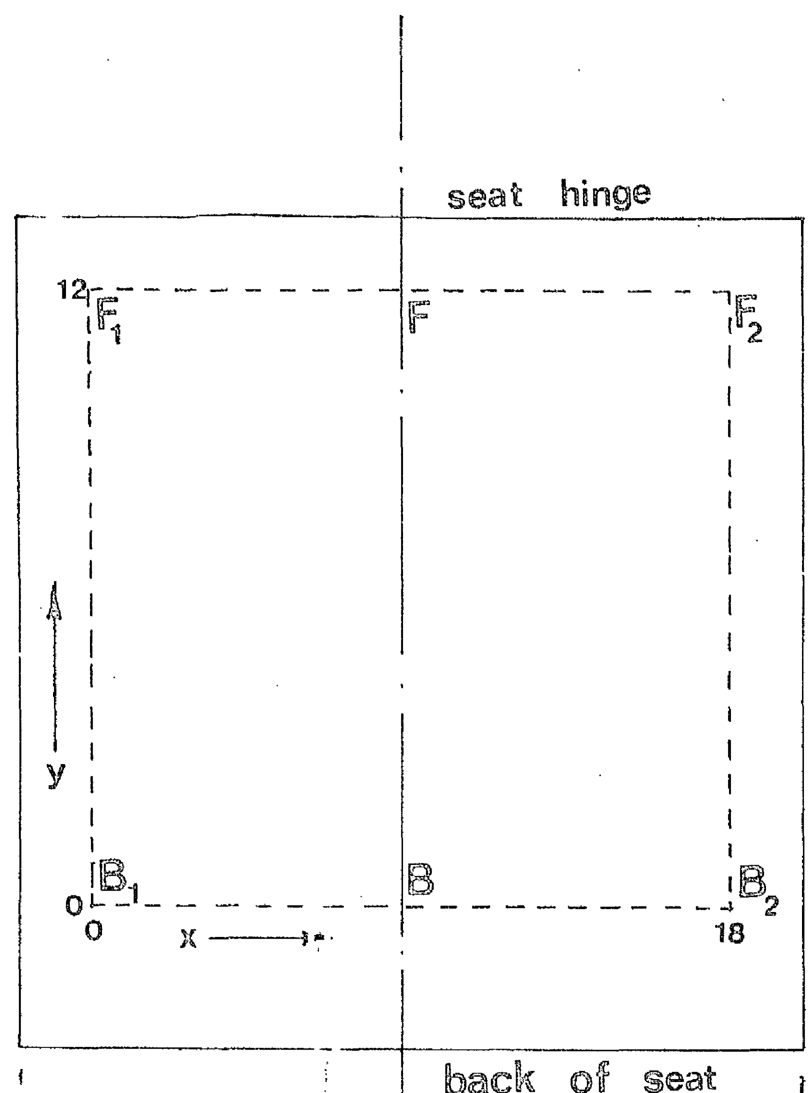


Figure 51

GAUGE RESPONSE ACROSS THE SEAT

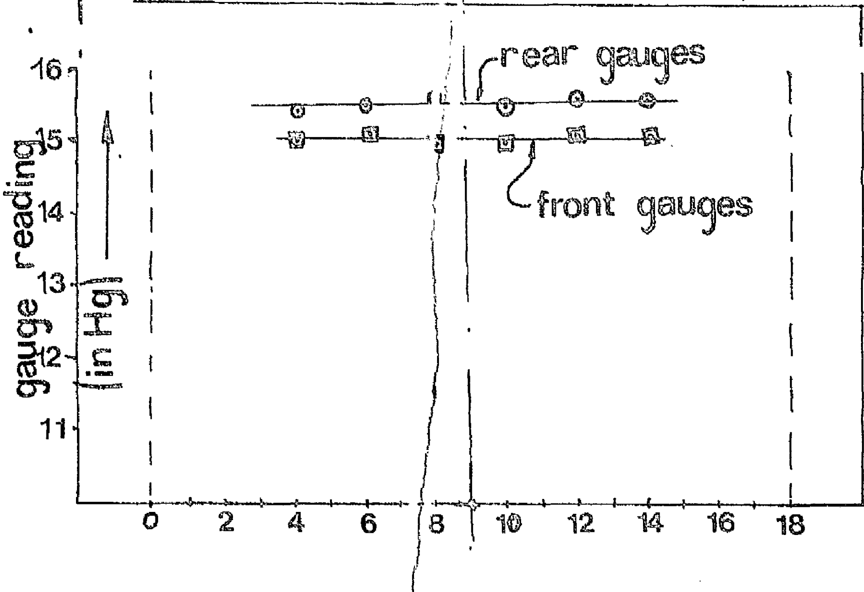


Figure 30

FUNCTIONS OF THE GEOMETRY OF THE CHAIR AID MECHANISM

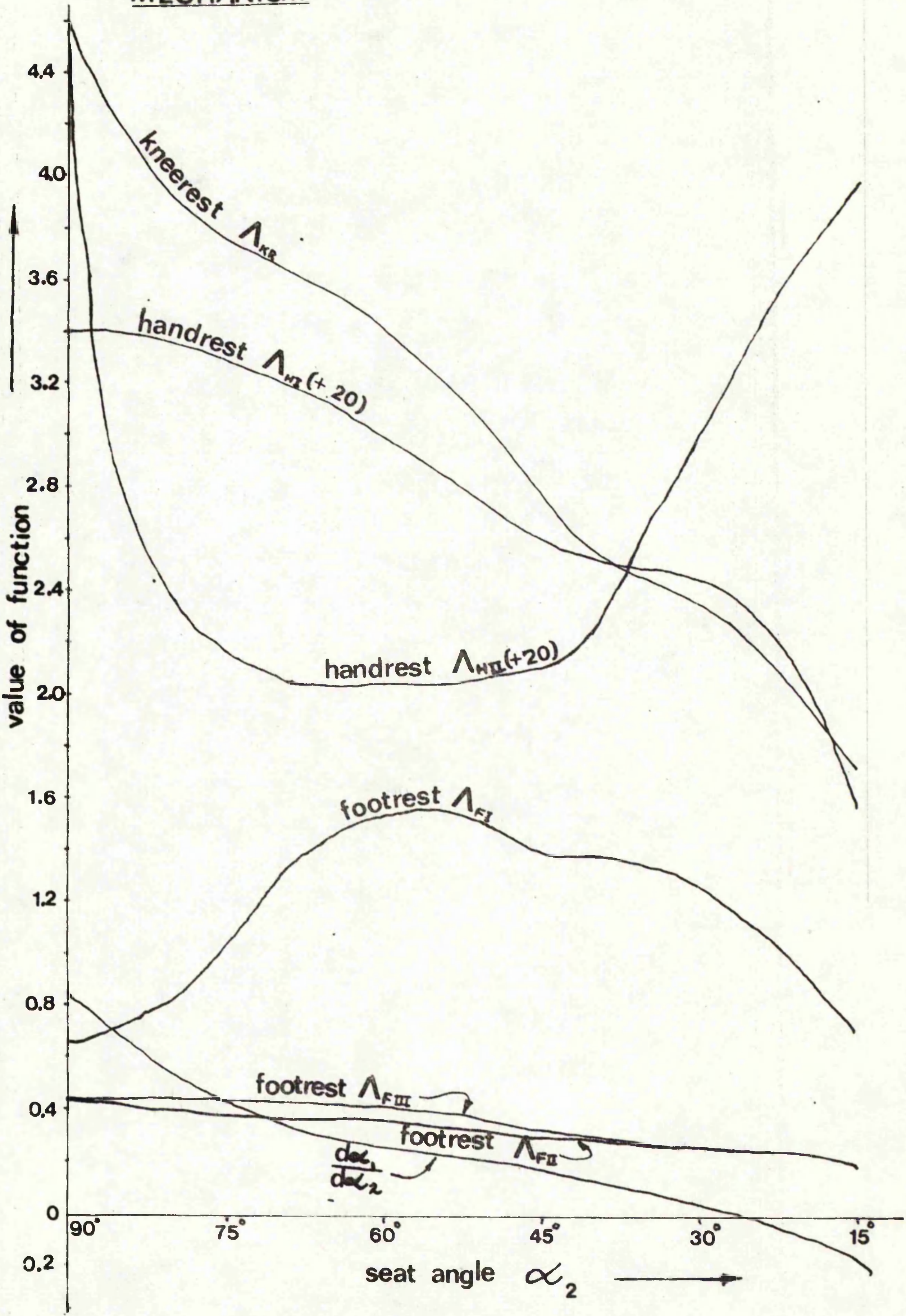


Figure 3D

LINK ANGLES OF THE CHAIR AID MECHANISM

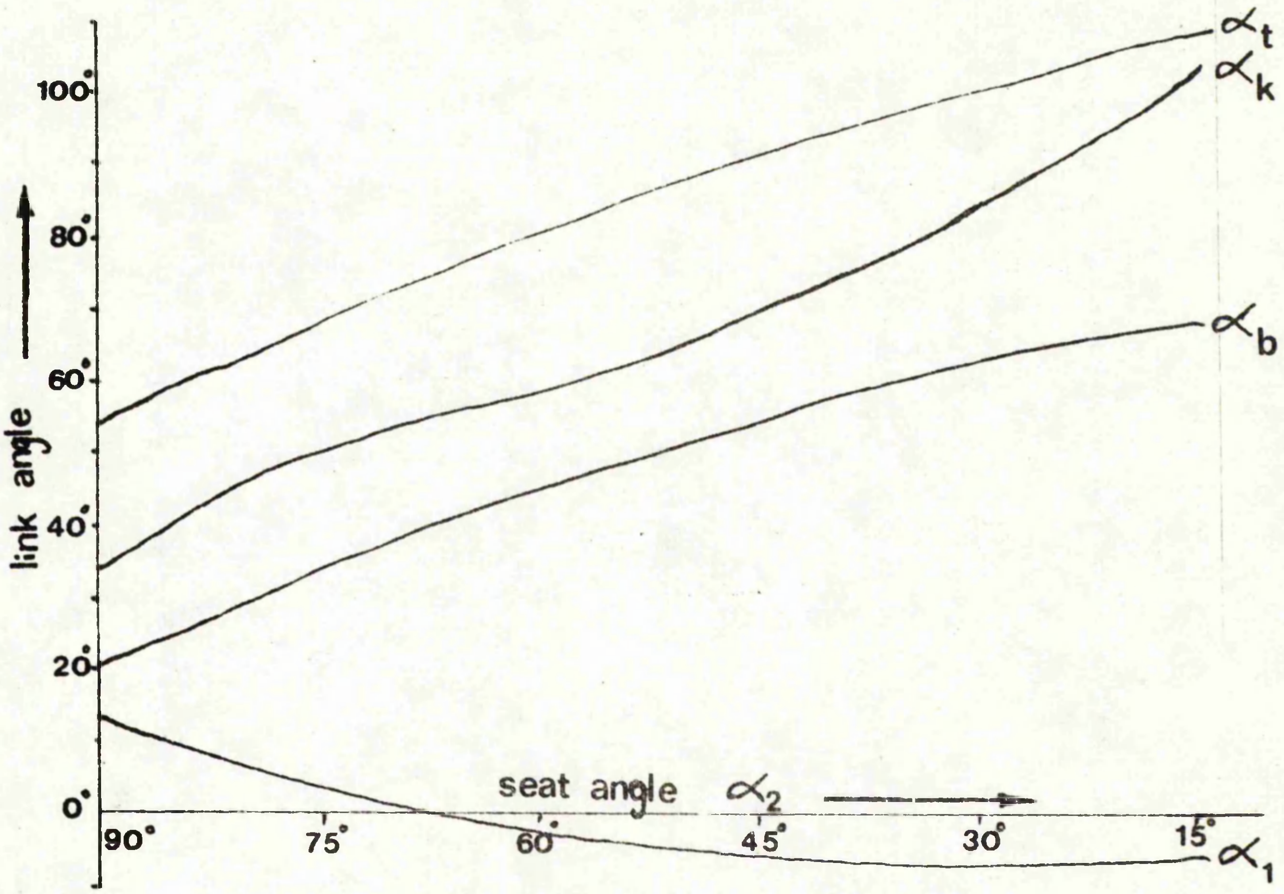
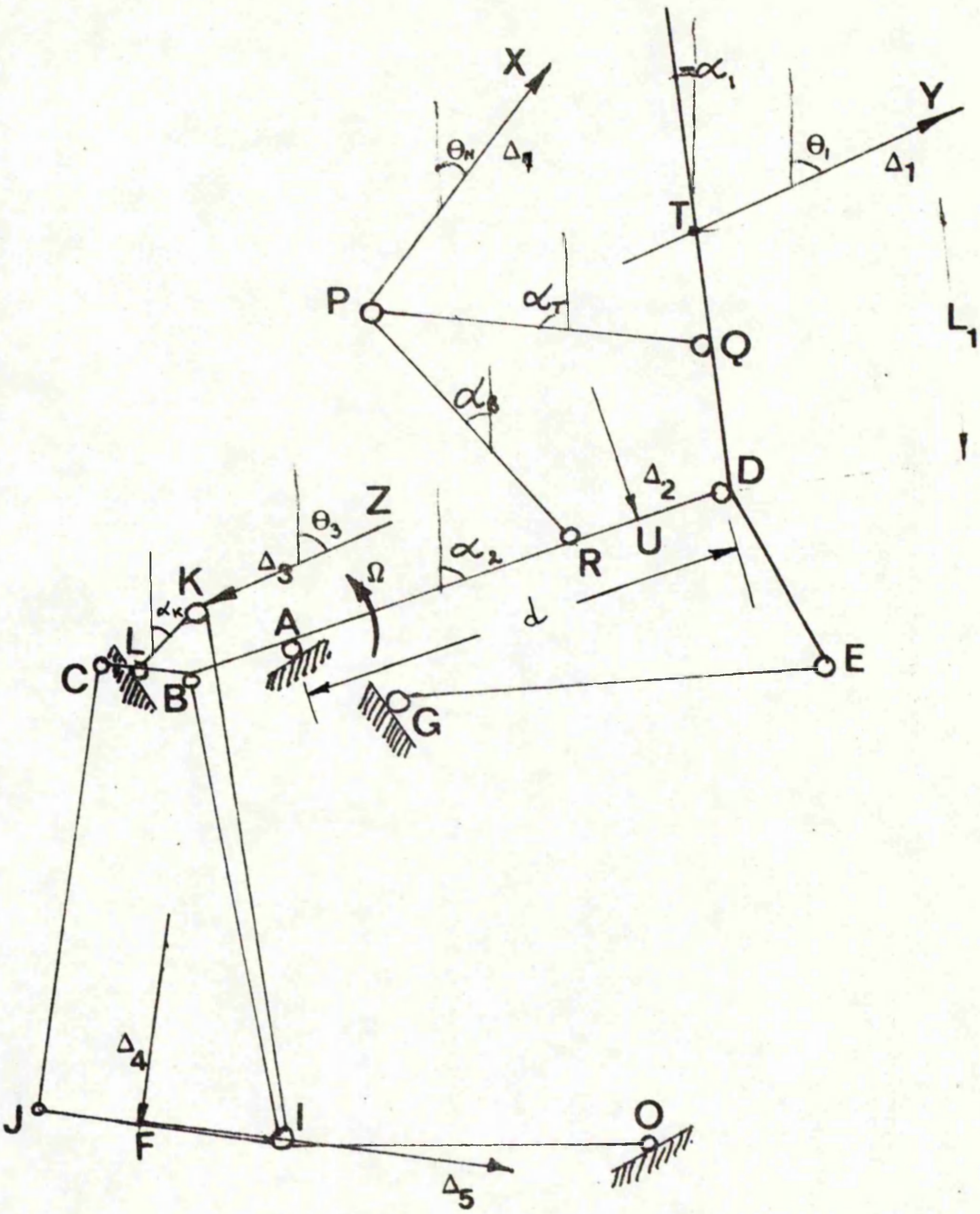


Figure 31



FORCES ACTING ON THE CHAIR AID MECHANISM IN GENERATING A SEAT HINGE TORQUE

Figure 32

HYSTERESIS LOOP OF EMPTY CHAIR

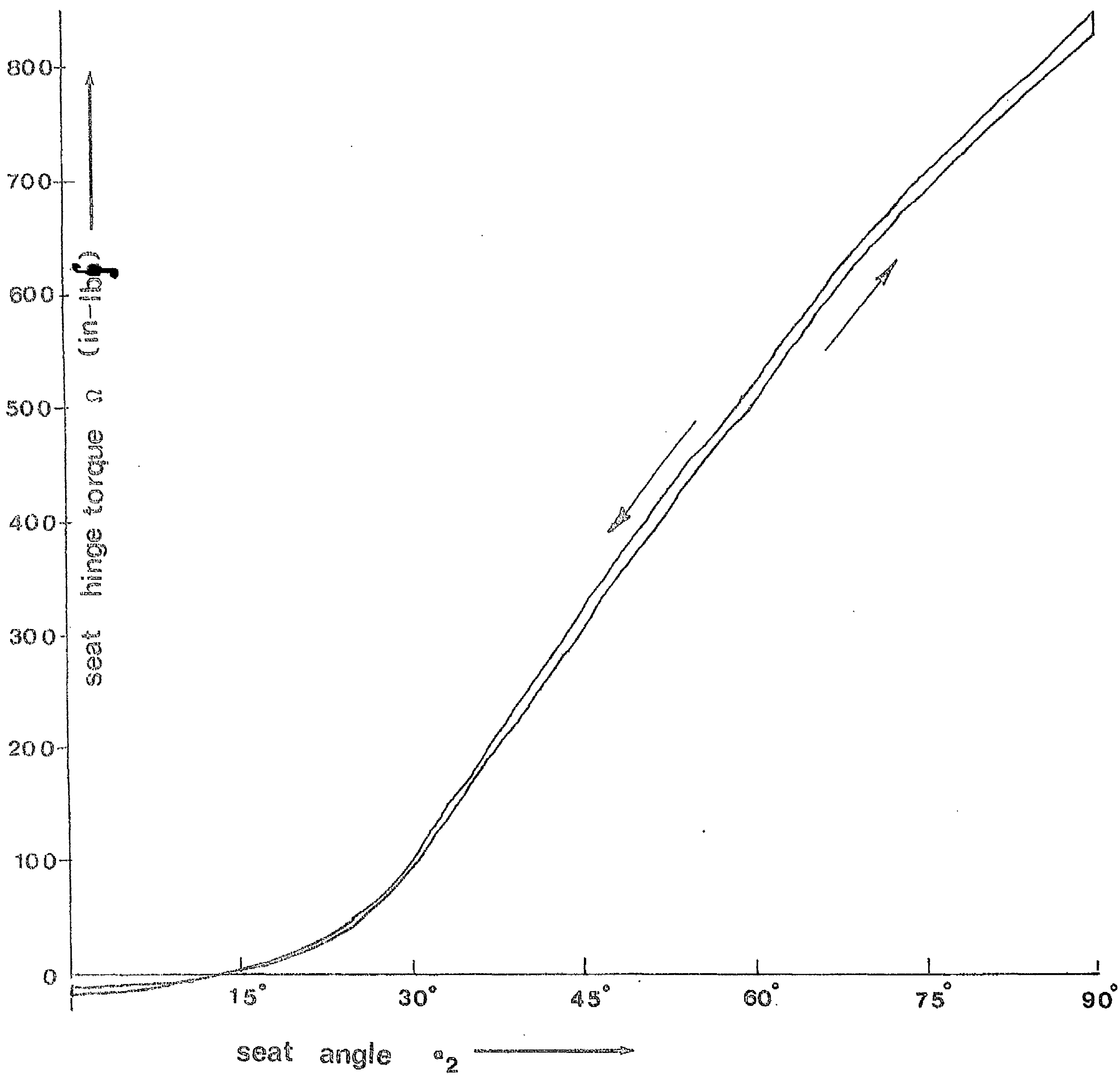


Figure 33

INTERNAL BODY FRICTION IN USING THE CHAIR AID

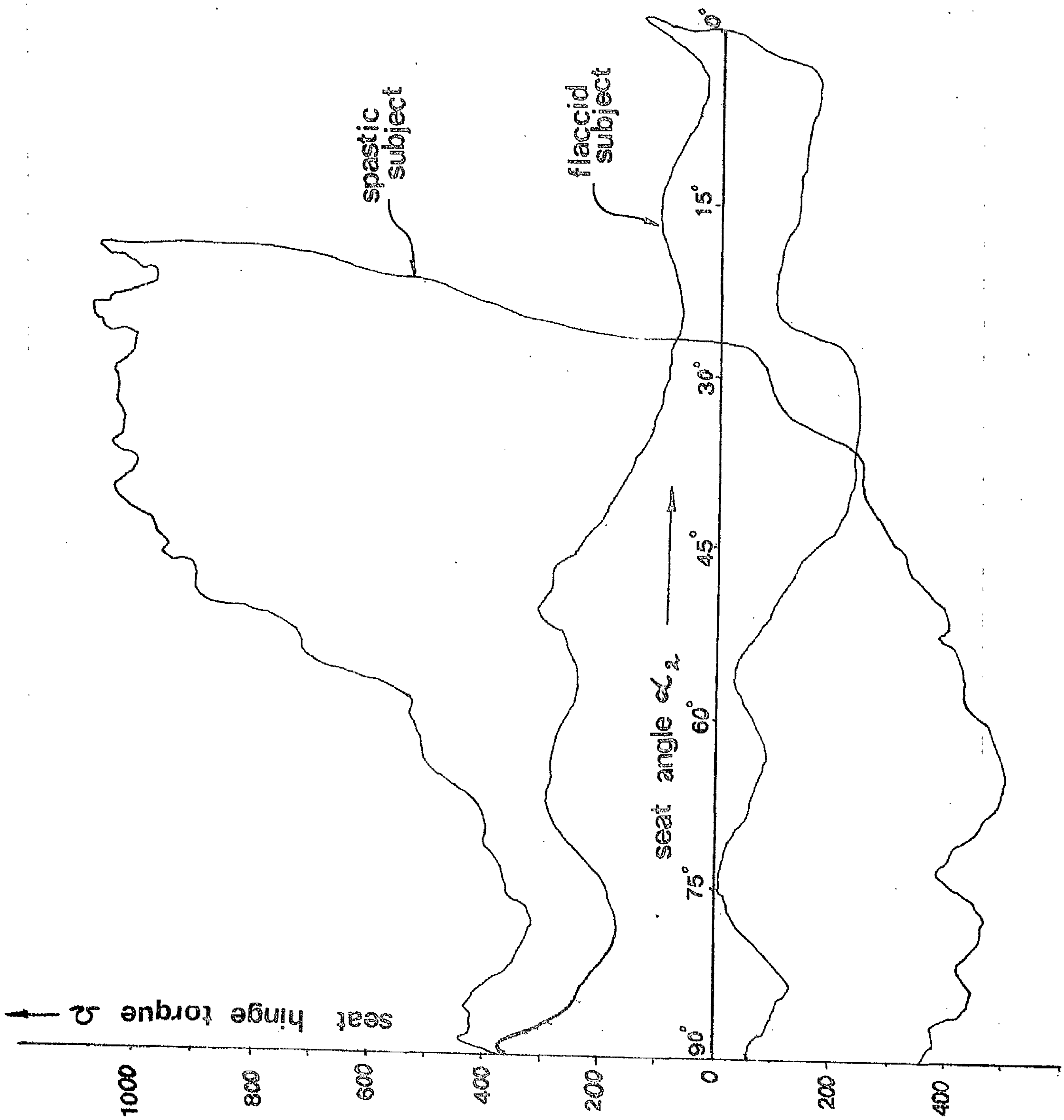


FIGURE 34

A DEADWEIGHT TEST CURVE

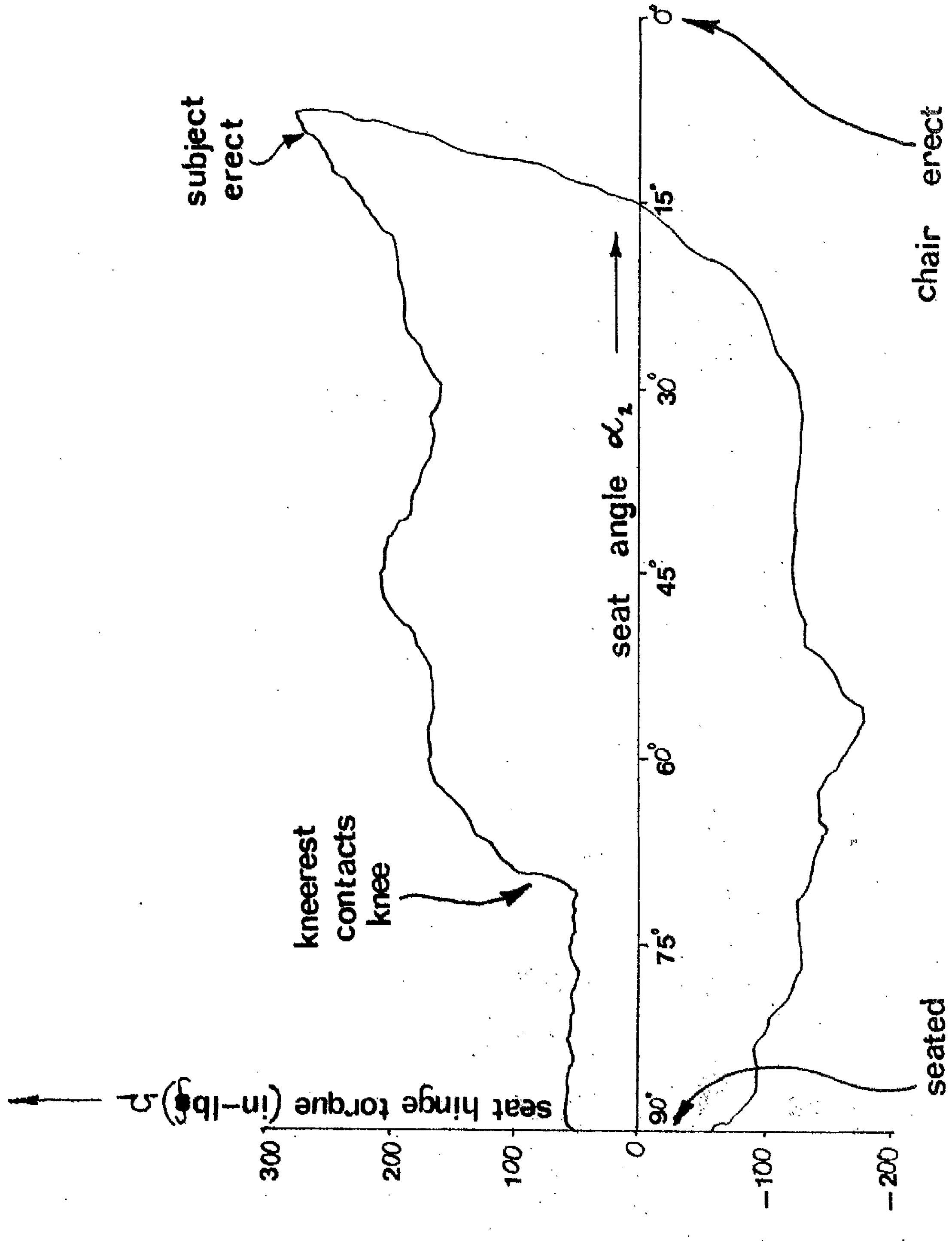
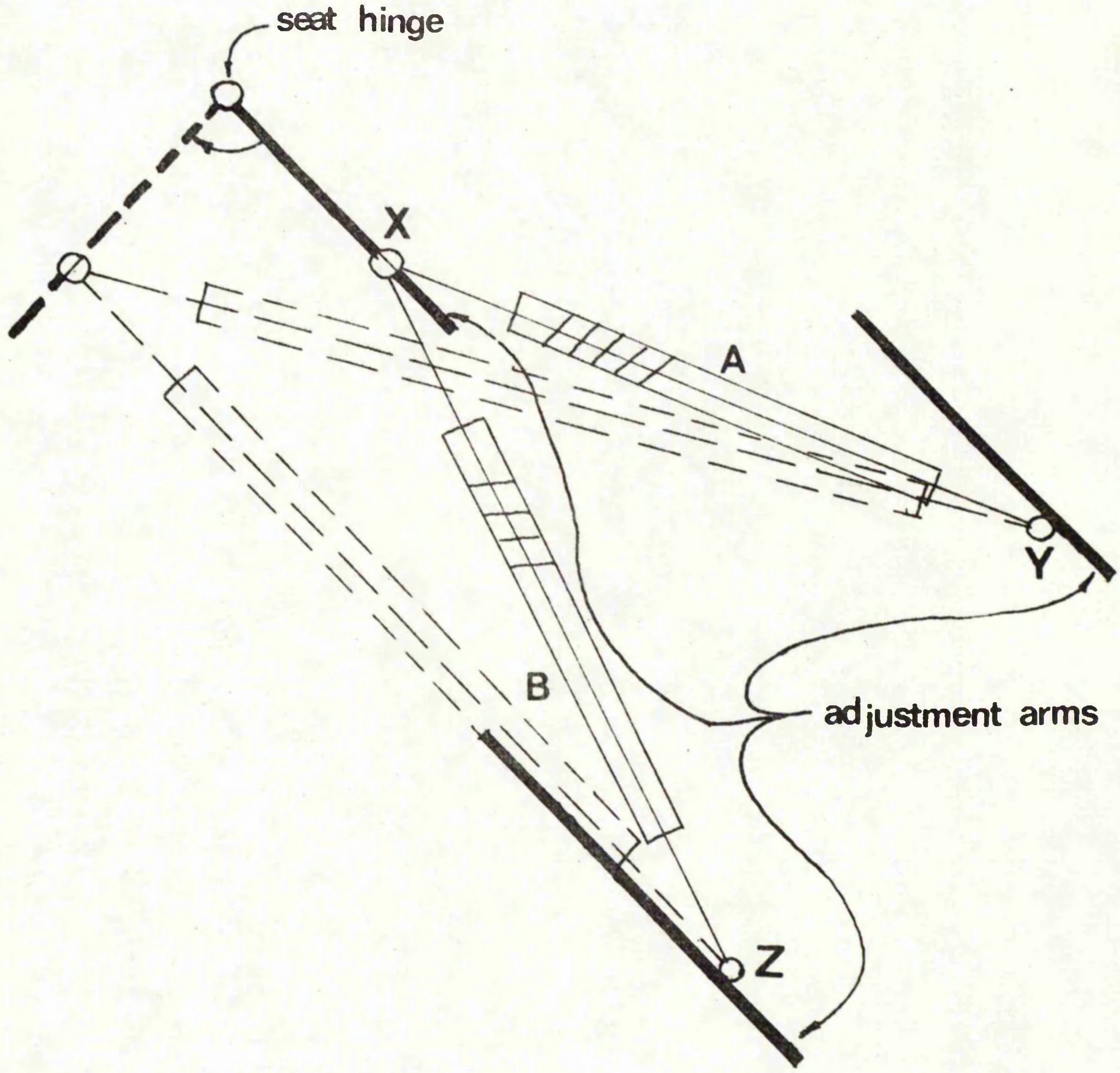


Figure 35



THE SPRING ARRANGEMENT ON THE CHAIR AID

FIGURE 36

ADJUSTMENT OF SECONDARY SPRING B

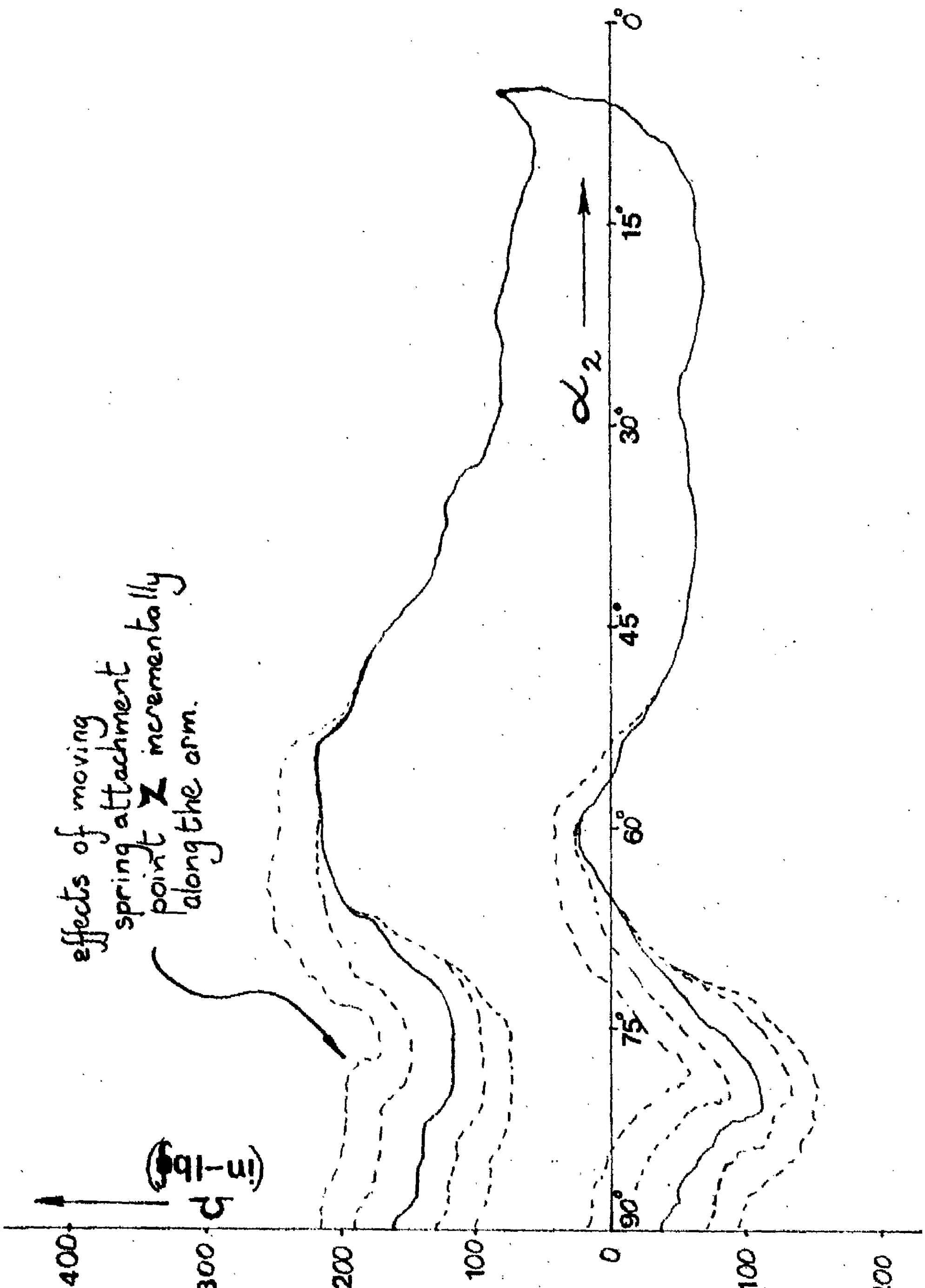


FIGURE 37

ADJUSTMENT OF PRIMARY SPRING A

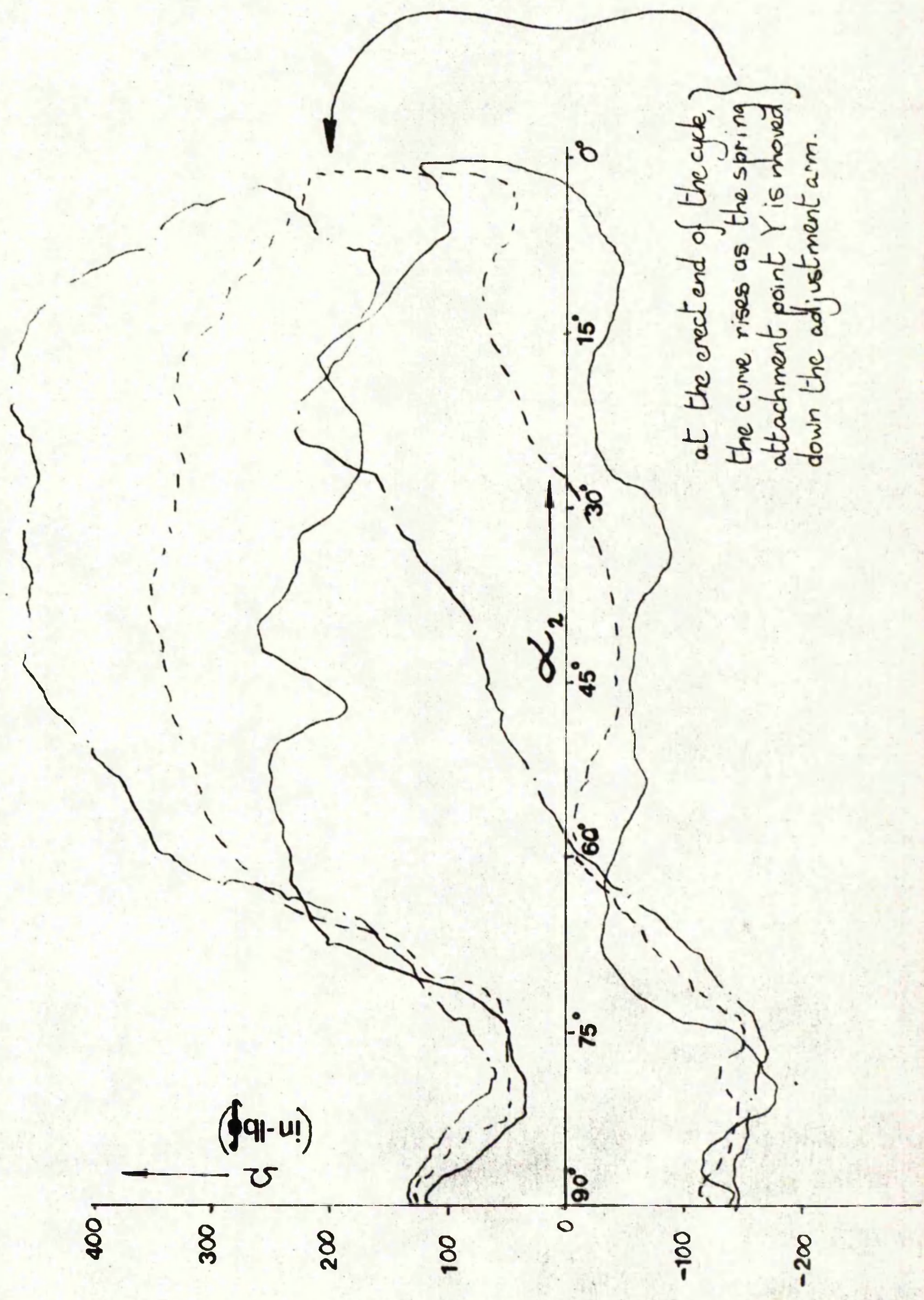


FIGURE 38

PROGRESSIVE ADJUSTMENT OF CHAIR DEADWEIGHT CHARACTERISTICS

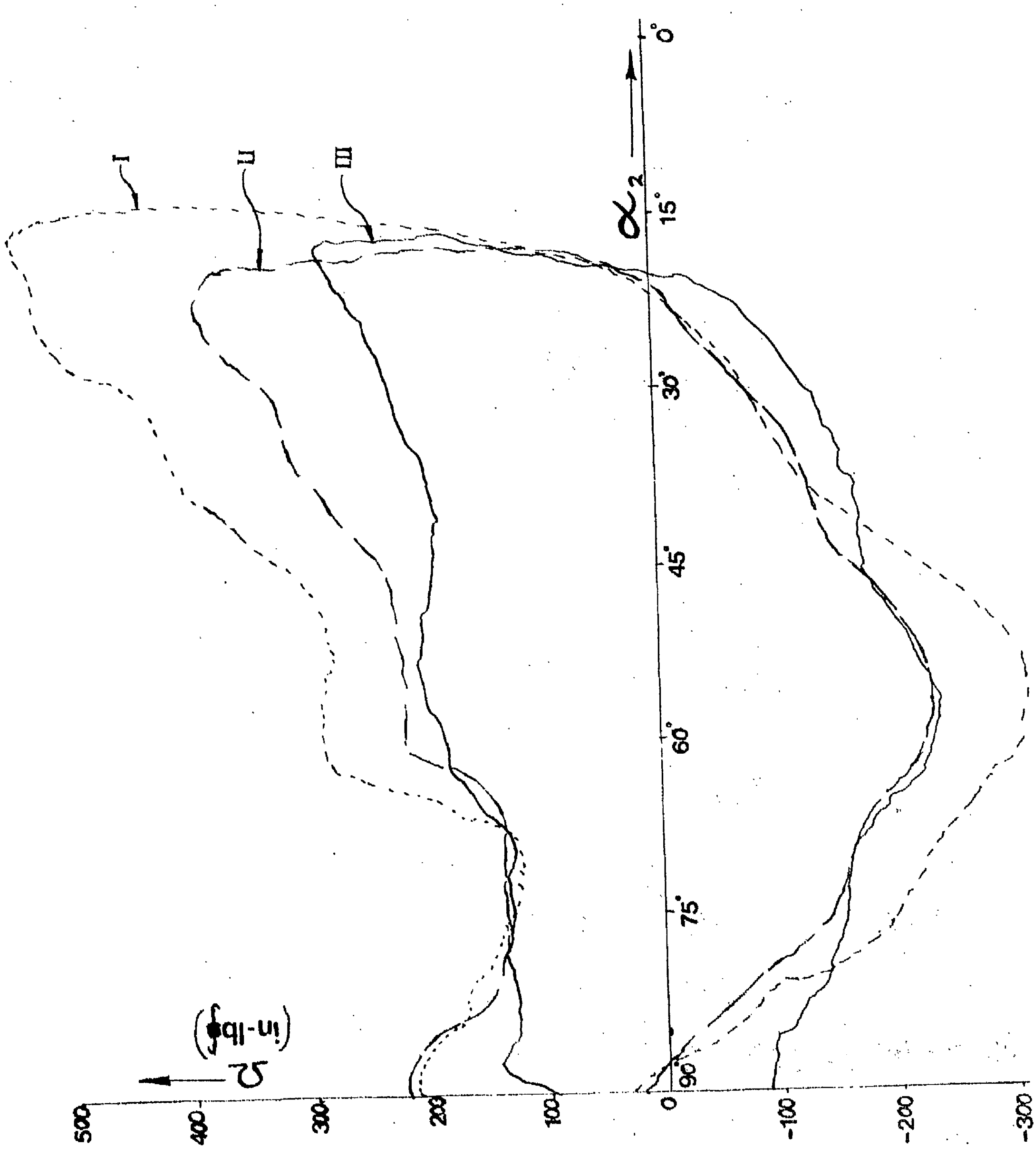
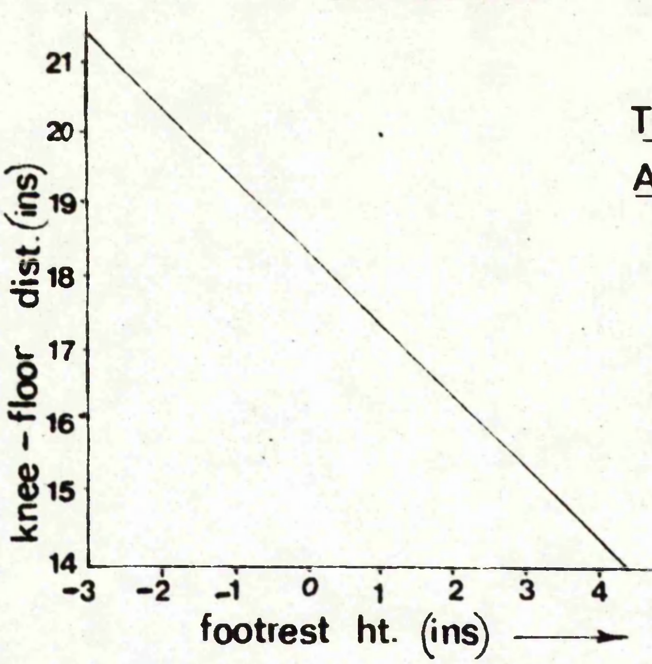


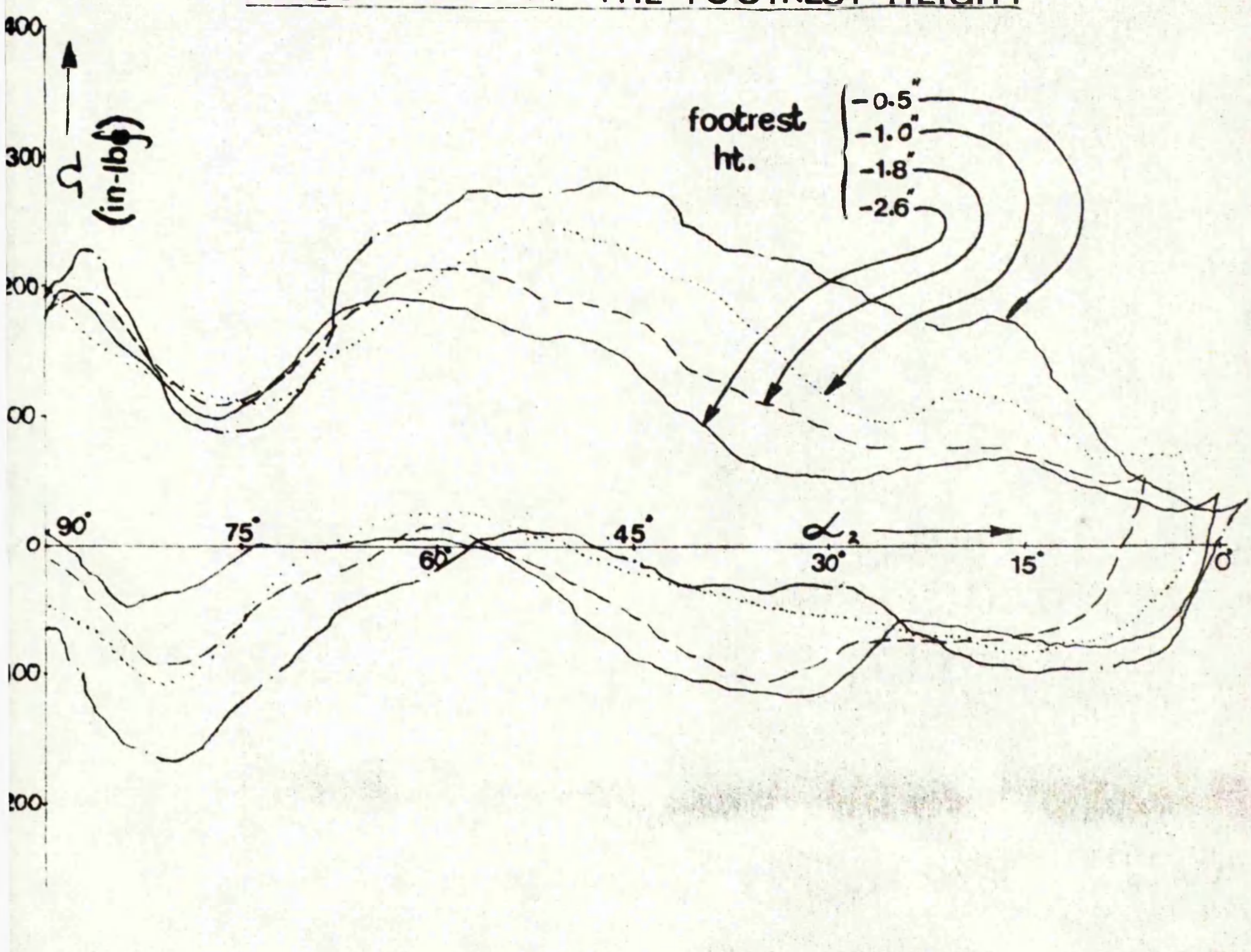
FIGURE 40



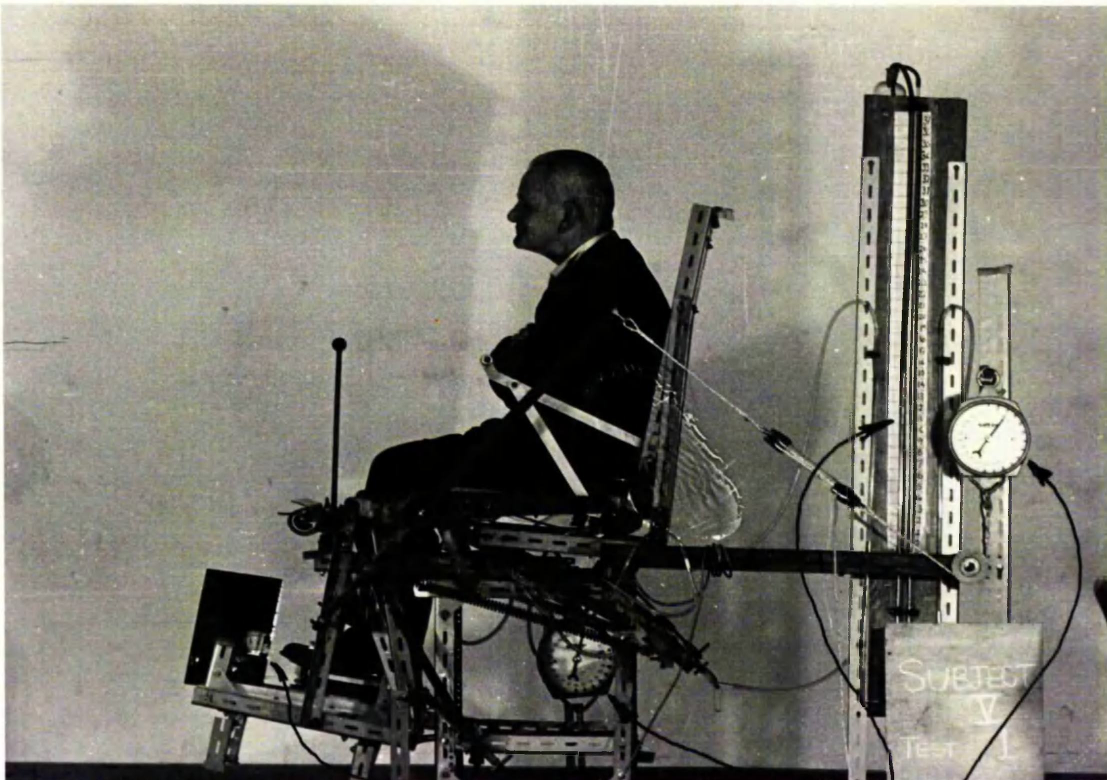
THE FOOTREST ADJUSTMENT SCALE

FIGURE 39

ADJUSTMENT OF THE FOOTREST HEIGHT



CHAIR AID TESTING — FIGURES 41 + 28



(A) RELAXED

Measurements of $\{R_4, R_5, R_2, \Omega\}$

(B) APPLYING A KNEE EXTENSION MOMENT

Figure 42

RESULTS OF PRELIMINARY CHAIR AID TESTS

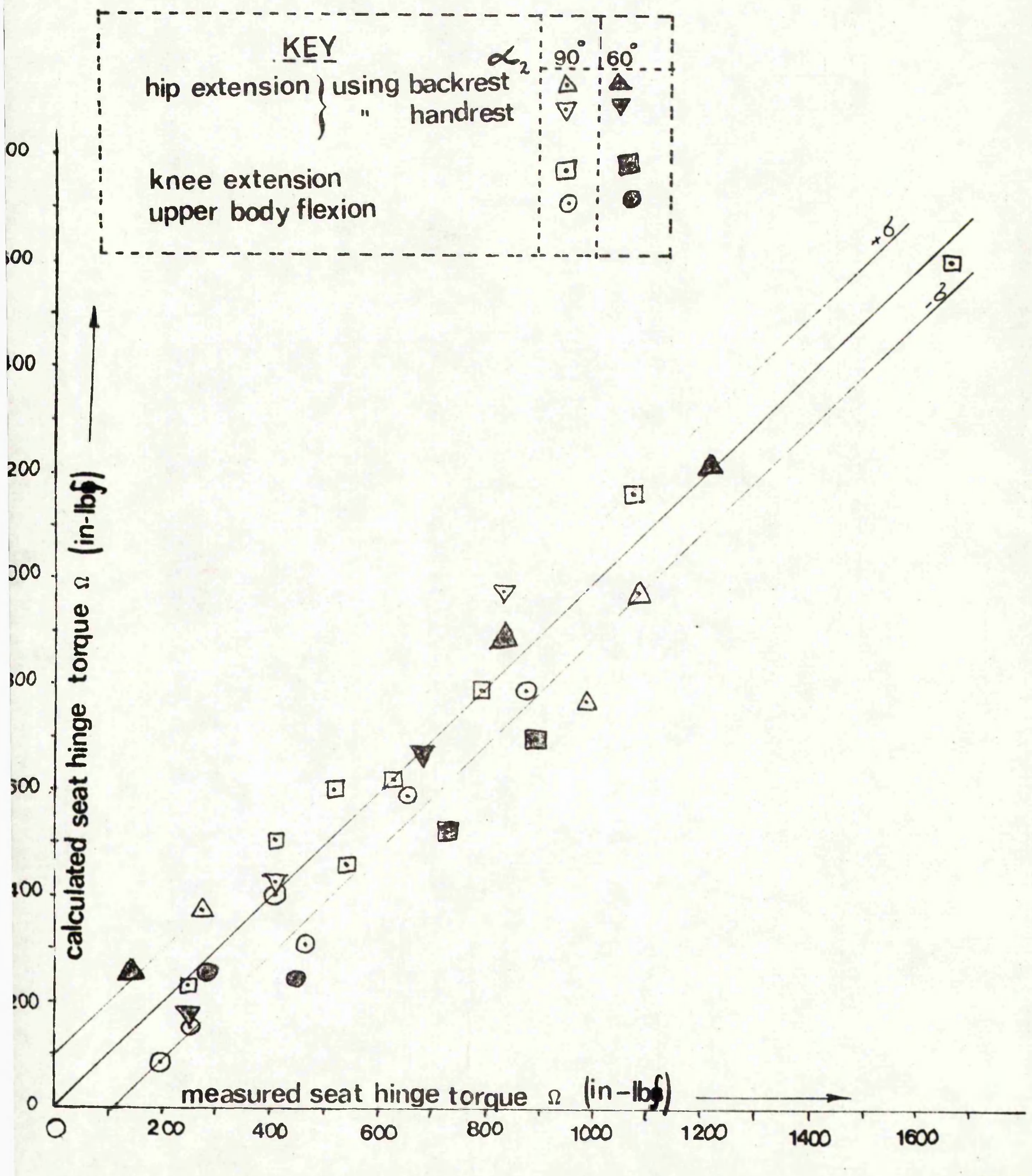


FIGURE 43

PRELIMINARY CHAIR AID TESTS: EFFECTIVENESS OF THE MODES OF RISING

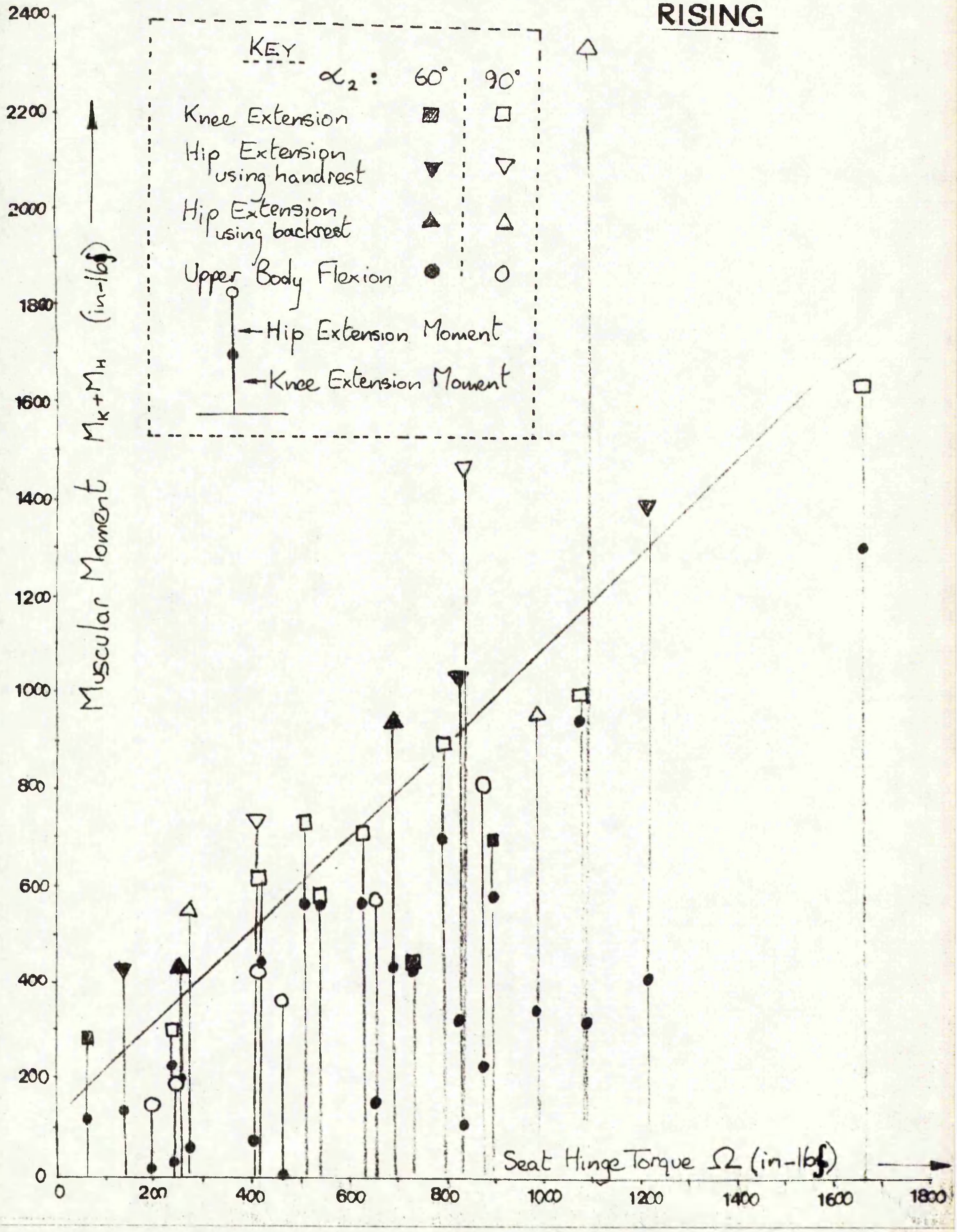


figure 44

Subject I

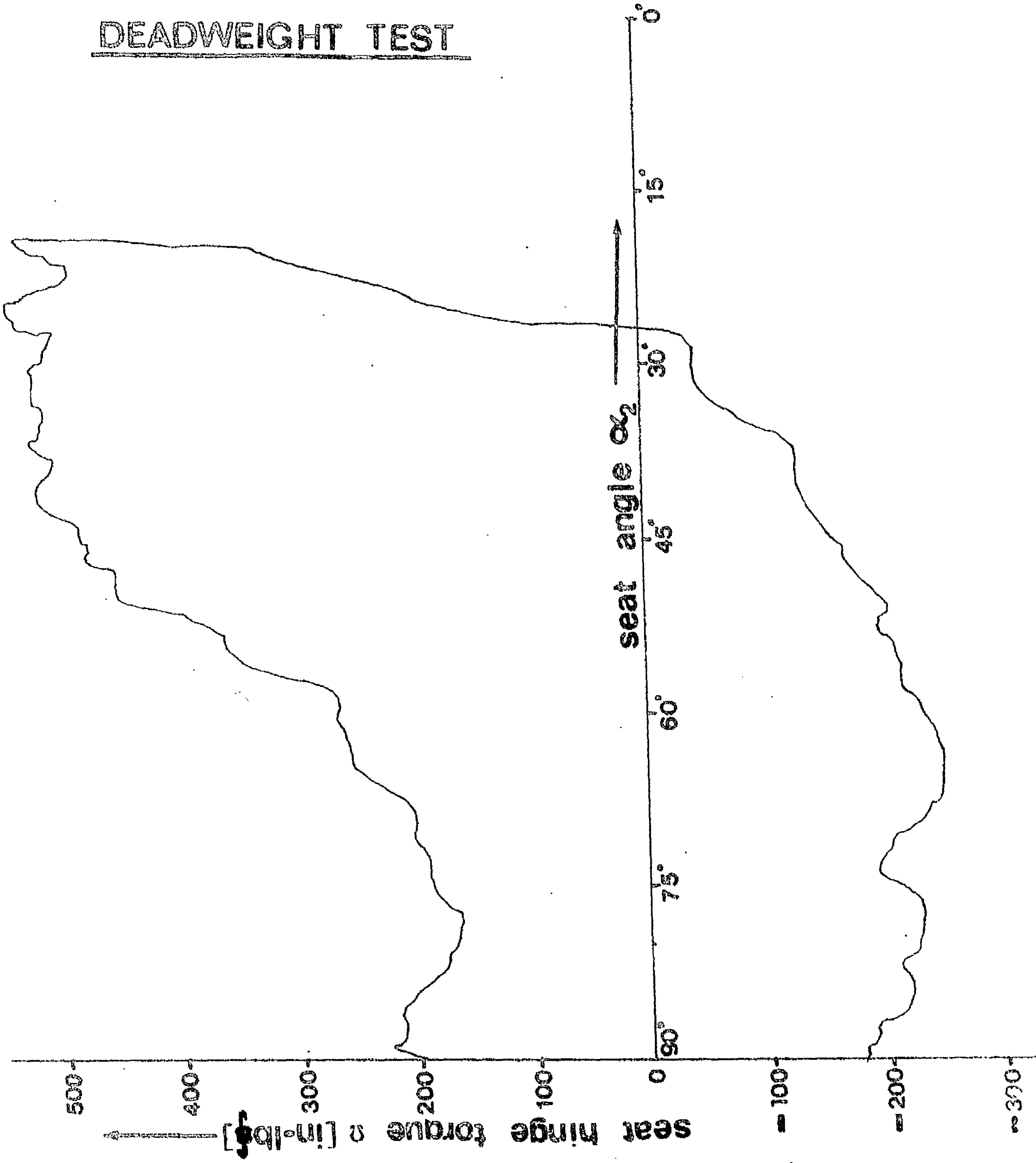
disability ------ Hemiplegia.

physique ------ Small, very plump.

sex --- Female **age** --- 50-55

effectiveness of chair aid ---- Able to rise fairly easily, unable to sit down.

DEADWEIGHT TEST



CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject I

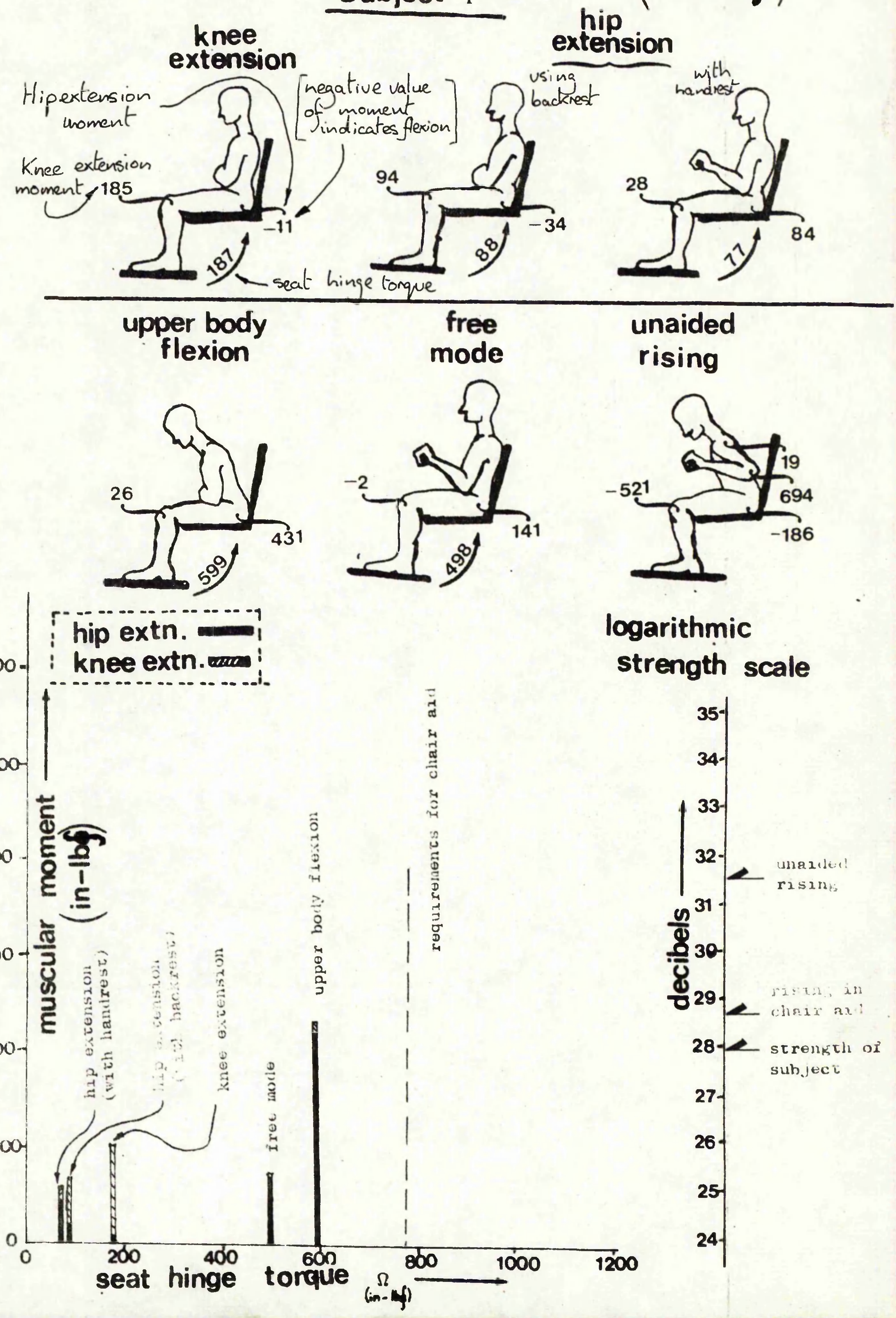


figure 44

Subject II

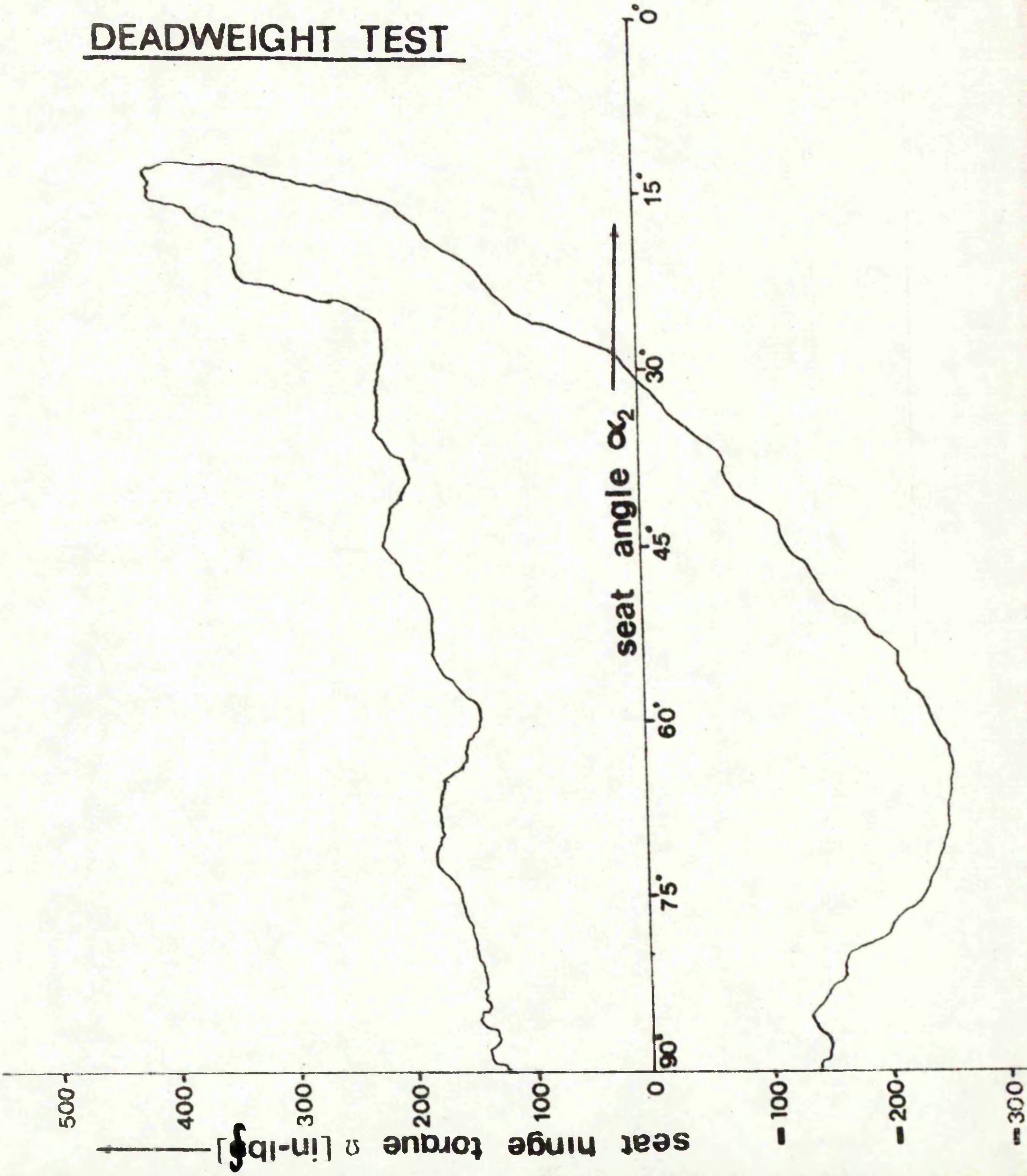
disability ----- Multiple Sclerosis.
and a painful arthritic hip joint.

physique ----- Small and wiry.

sex --- Male **age** --- 40-45

effectiveness of chair aid ---- Rose easily. Sat down with greater difficulty. Aid does not relieve painful hip.

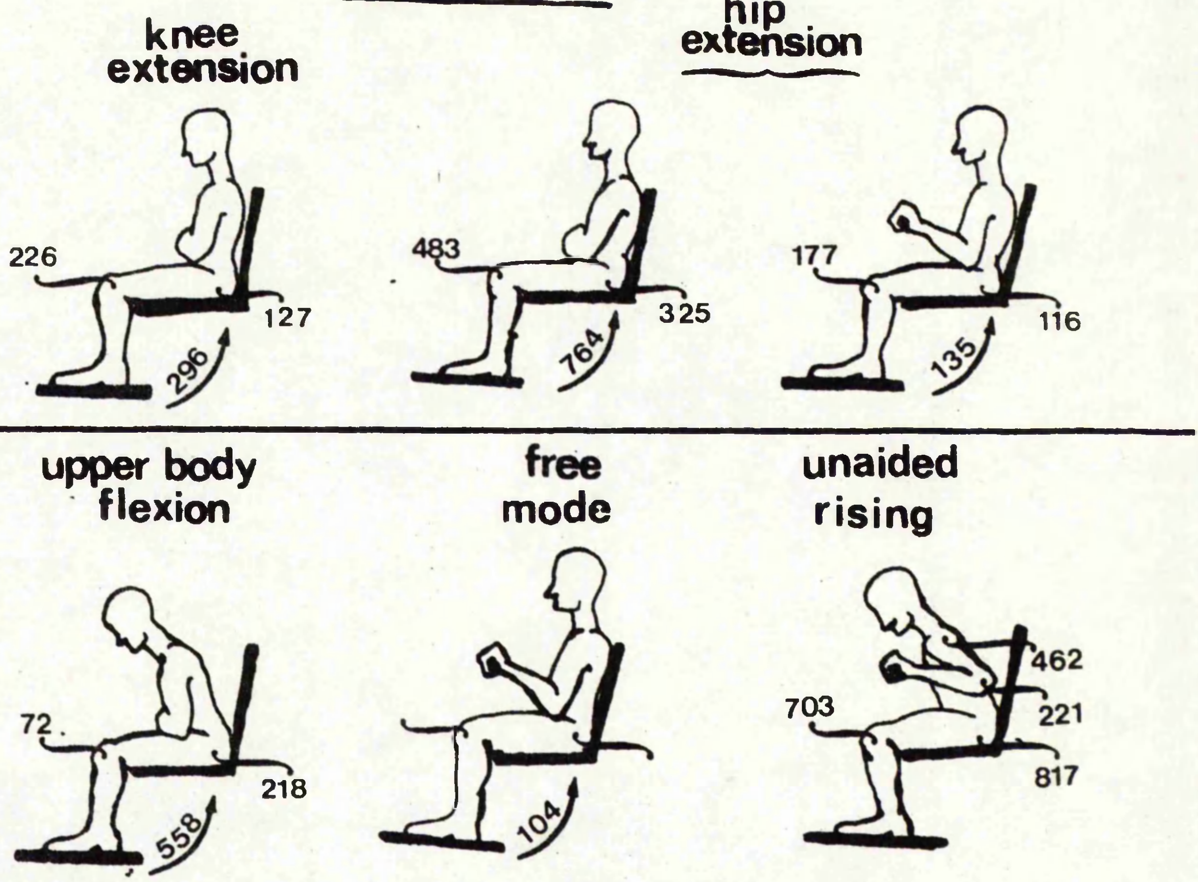
DEADWEIGHT TEST

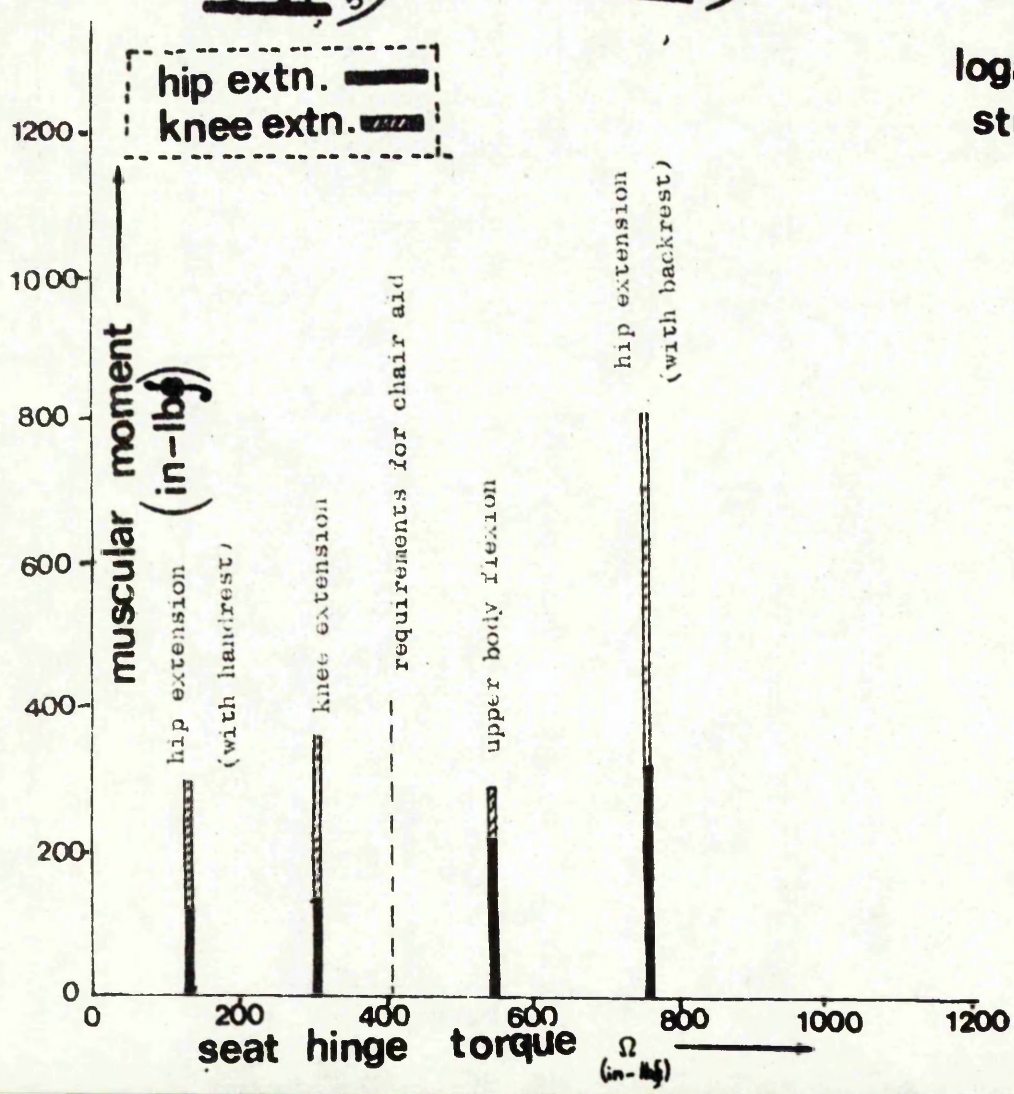


CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject II





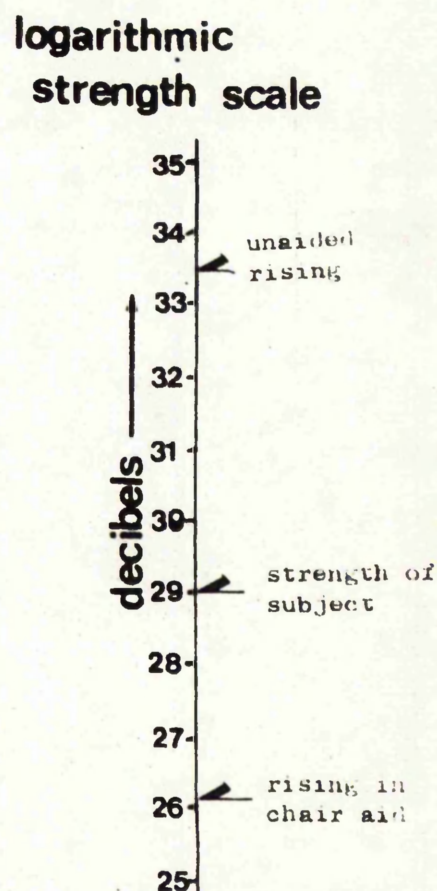


figure 44

Subject III

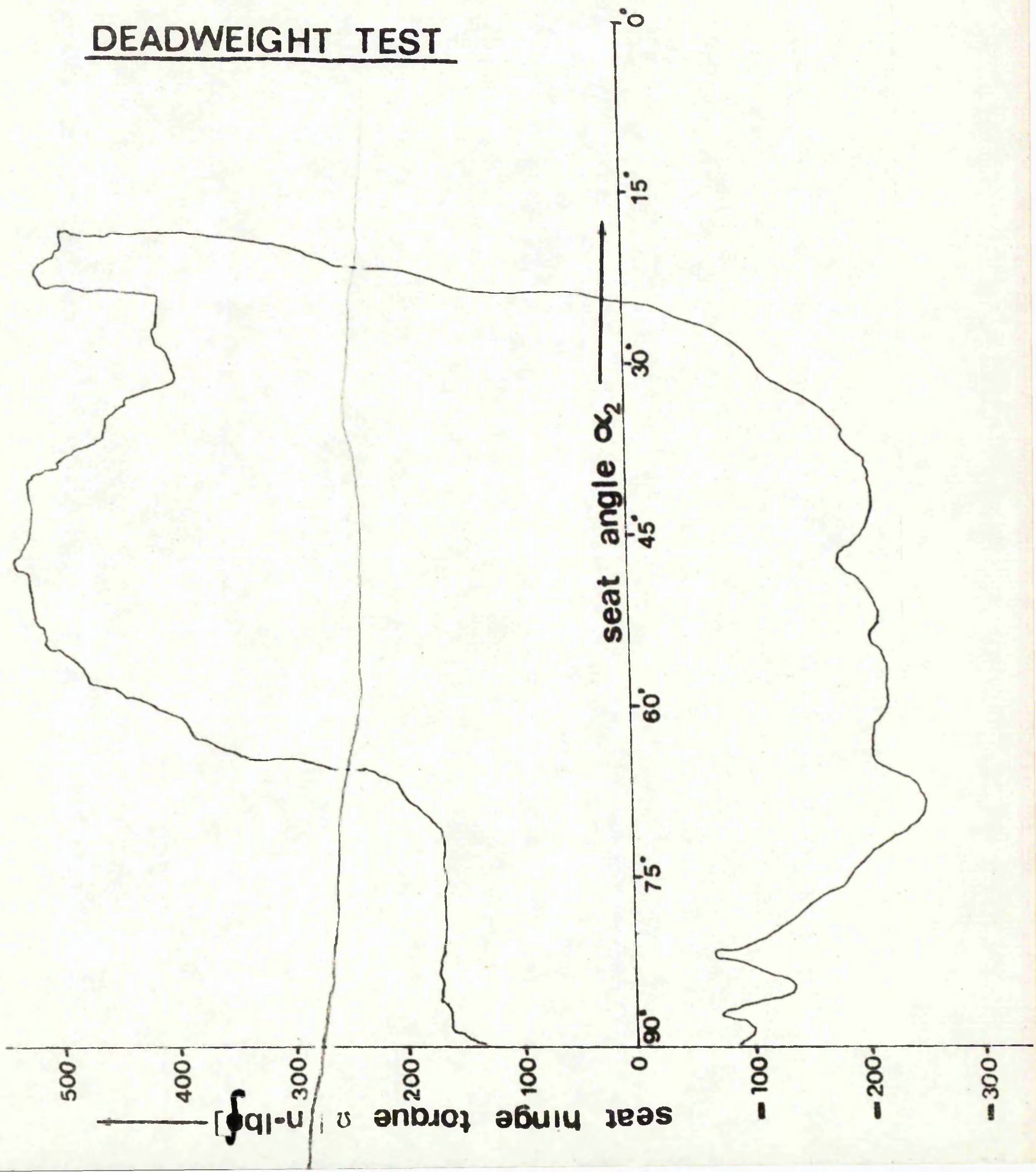
disability ----- Multiple Sclerosis
Stiffness in knee and hip joints.

physique ----- Medium height, medium build.

sex --- Female **age** --- 45-50

effectiveness of chair aid ---- Found aid very effective. Commented on how good it felt to be able to relieve the pressure of sitting down.

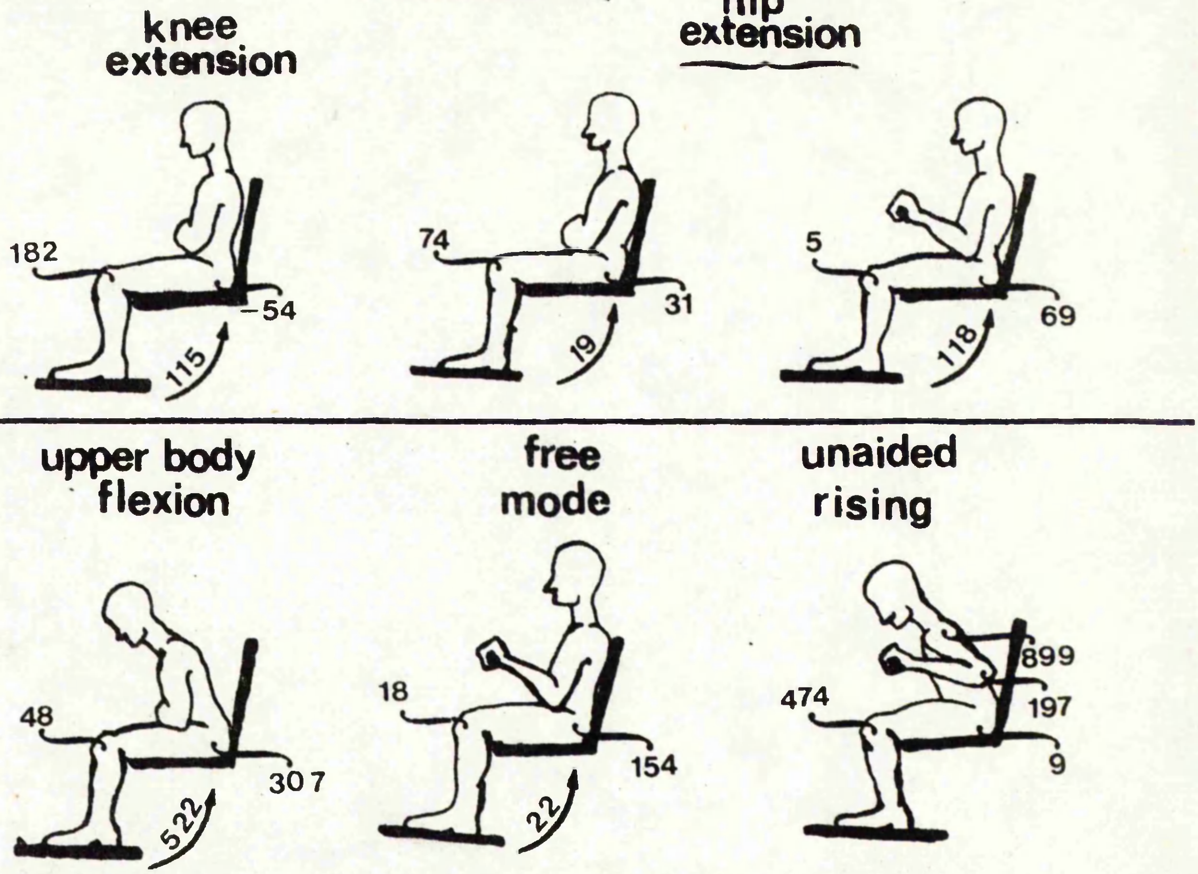
DEADWEIGHT TEST

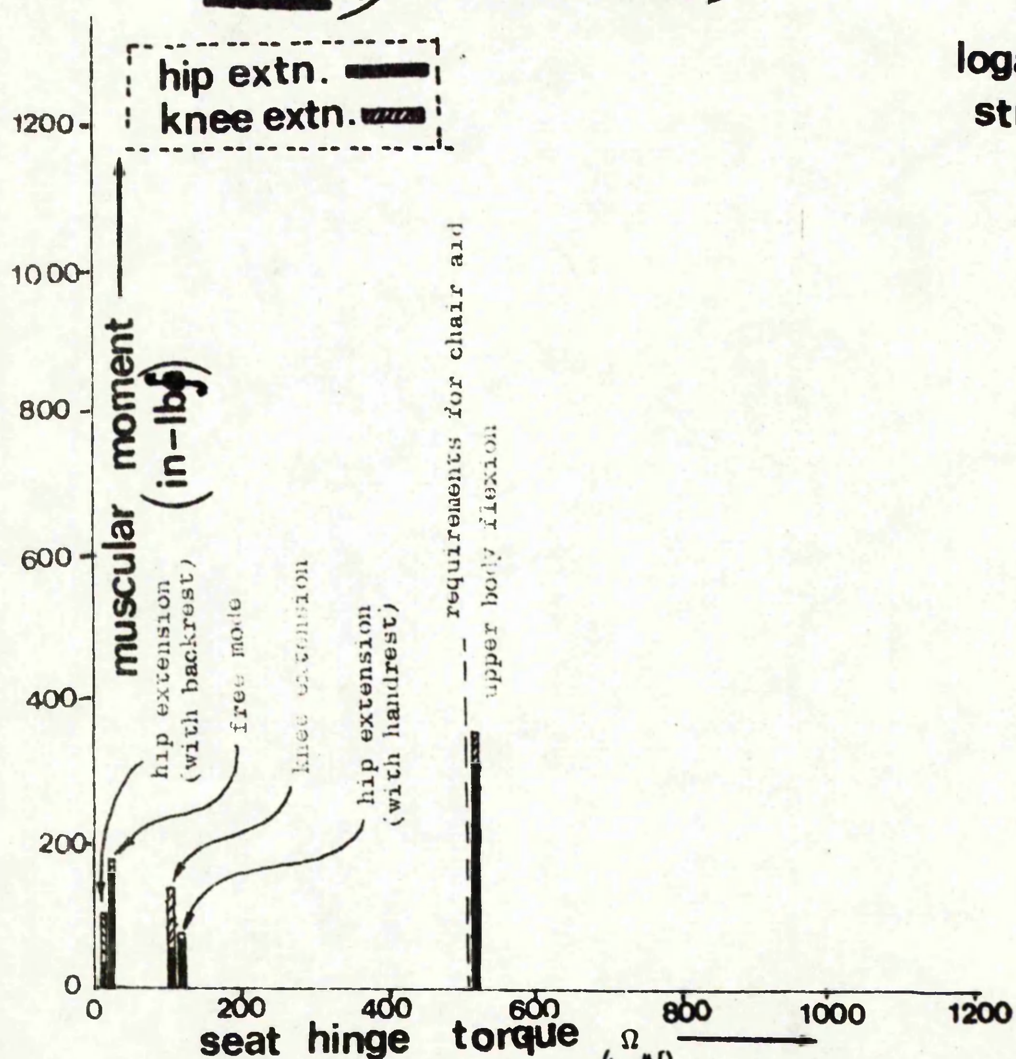


CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject III





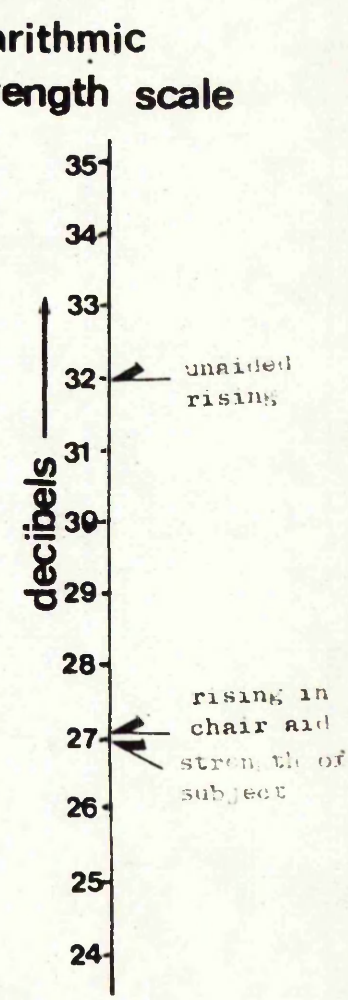


Figure 44

Subject IV

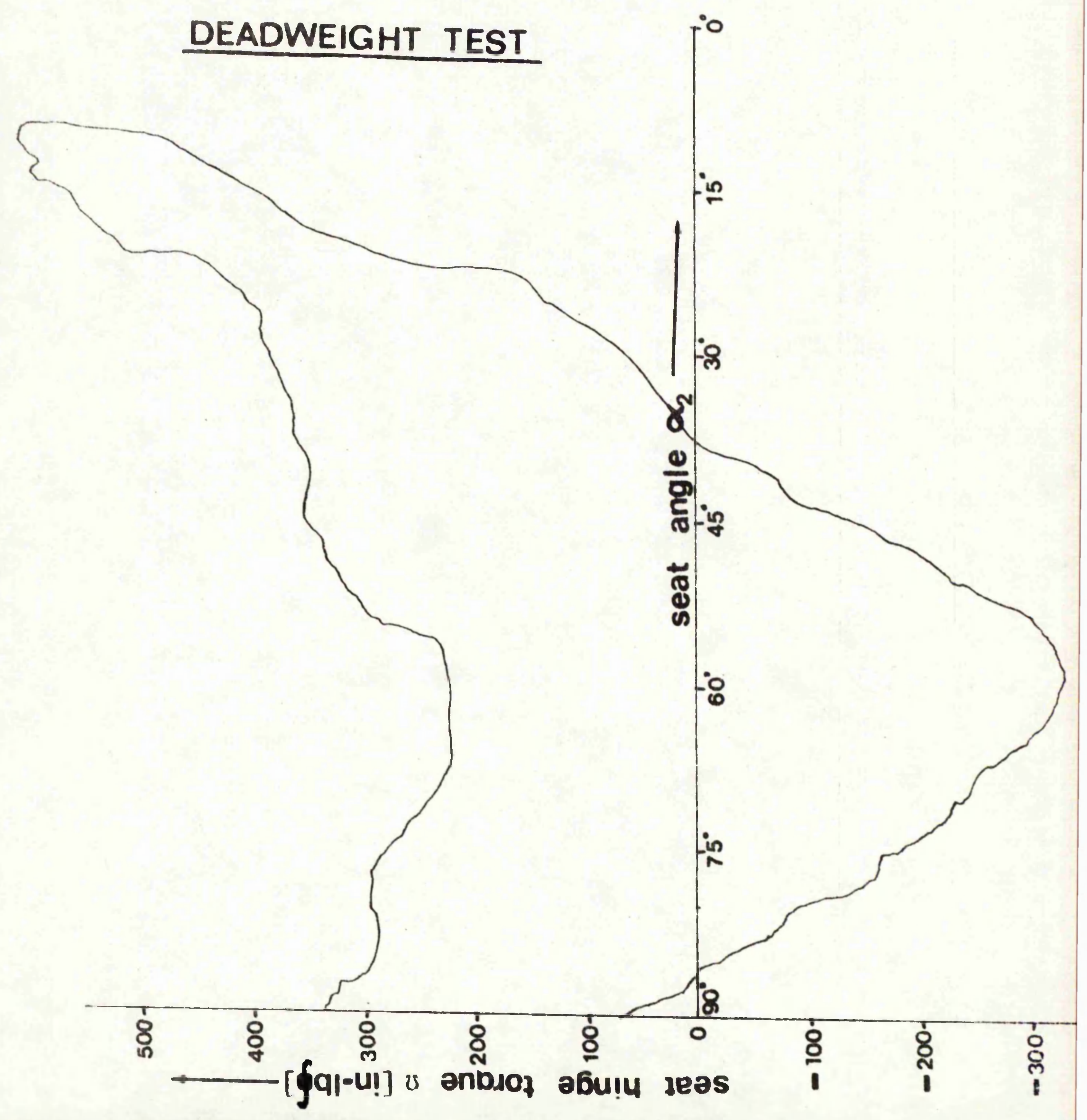
disability ----- Multiple Sclerosis.
Legs only affected. Extremely stiff knees.

physique ----- Medium Height, heavy Build.

sex --- Male **age** --- 55-60

effectiveness of chair aid - - - - Unsuitable for this subject. Able to rise, but knees remain fully extended on sitting down.

DEADWEIGHT TEST



CHAIR AID TEST RESULTS

Subject IV

(figures in in-lbf.)

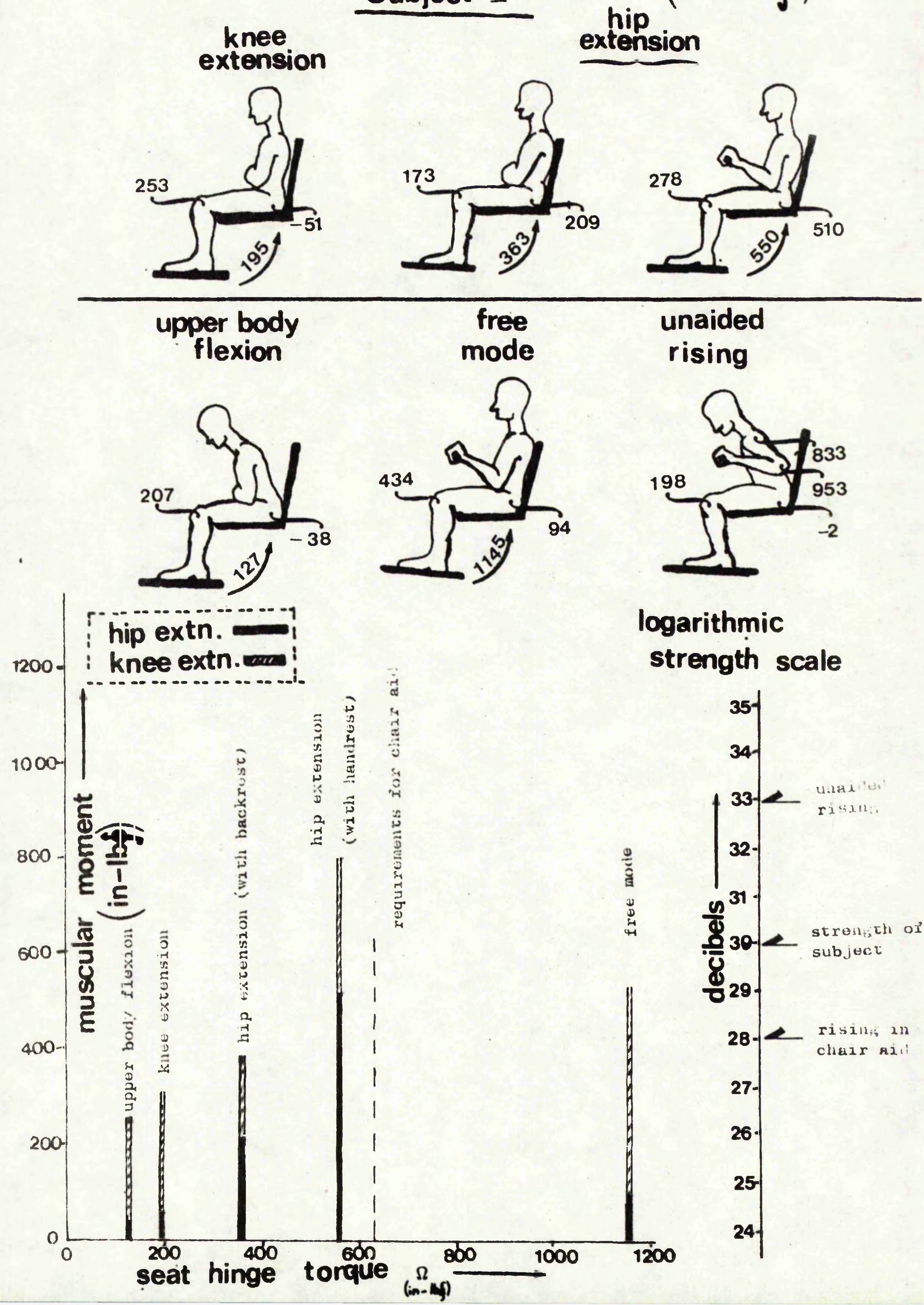


Figure 44

Subject I

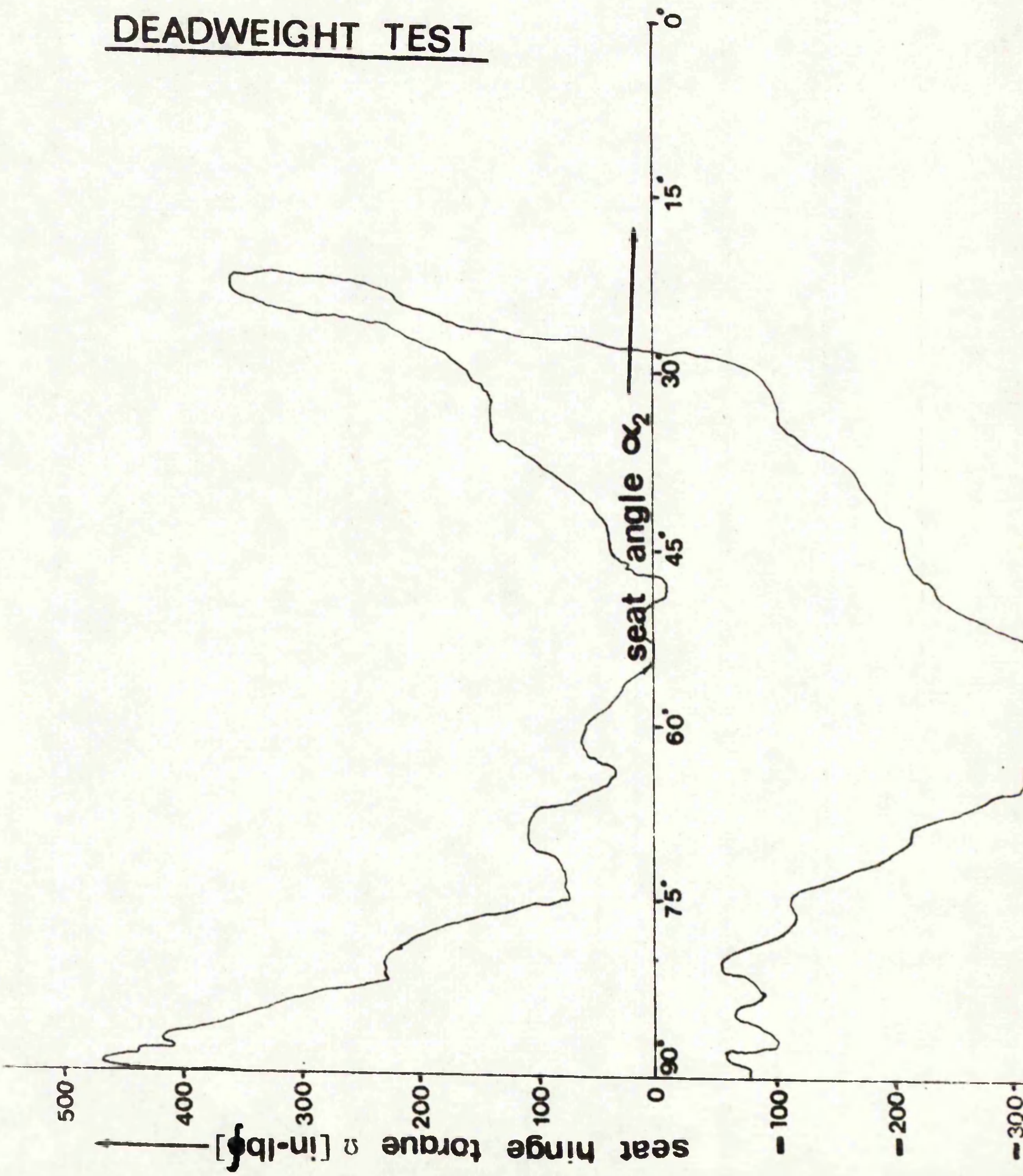
disability ----- Multiple Sclerosis (Weak heart)

physique ----- Small and plump.

sex --- Female **age** --- 45-50

effectiveness of chair aid ---- Able to rise and sit down easily. Found chair "very relaxing"

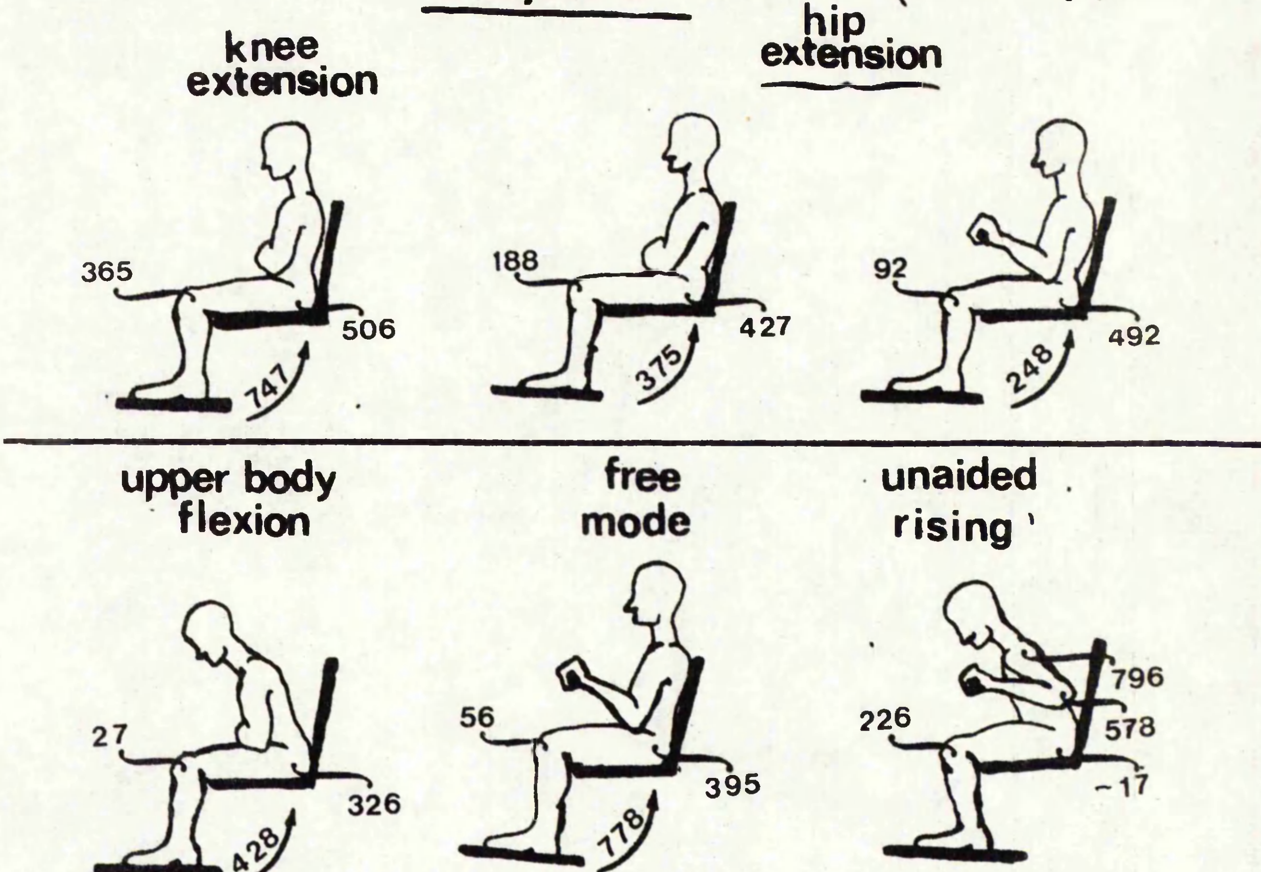
DEADWEIGHT TEST



CHAIR AID TEST RESULTS

Subject V

(figures in in-lbf)



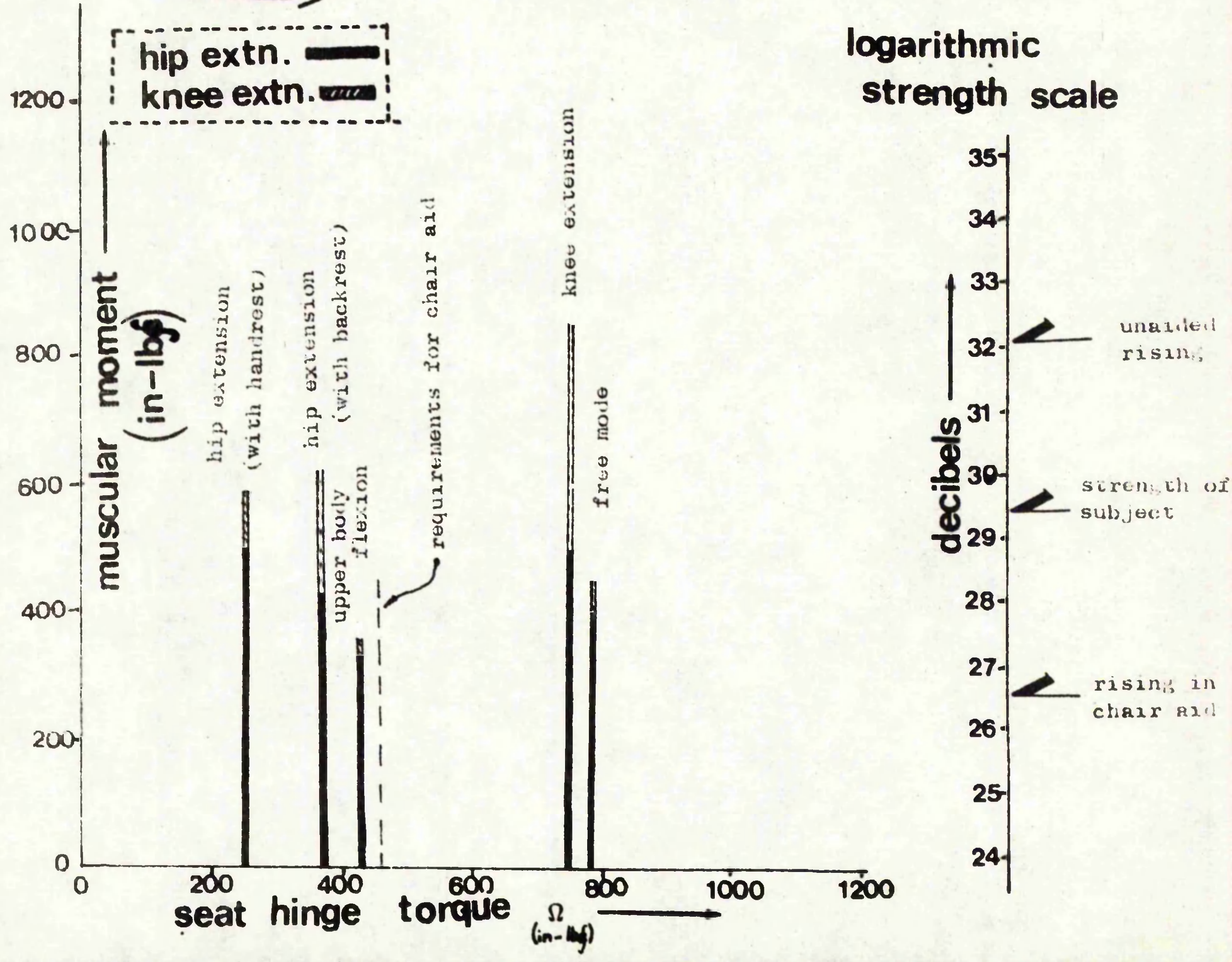


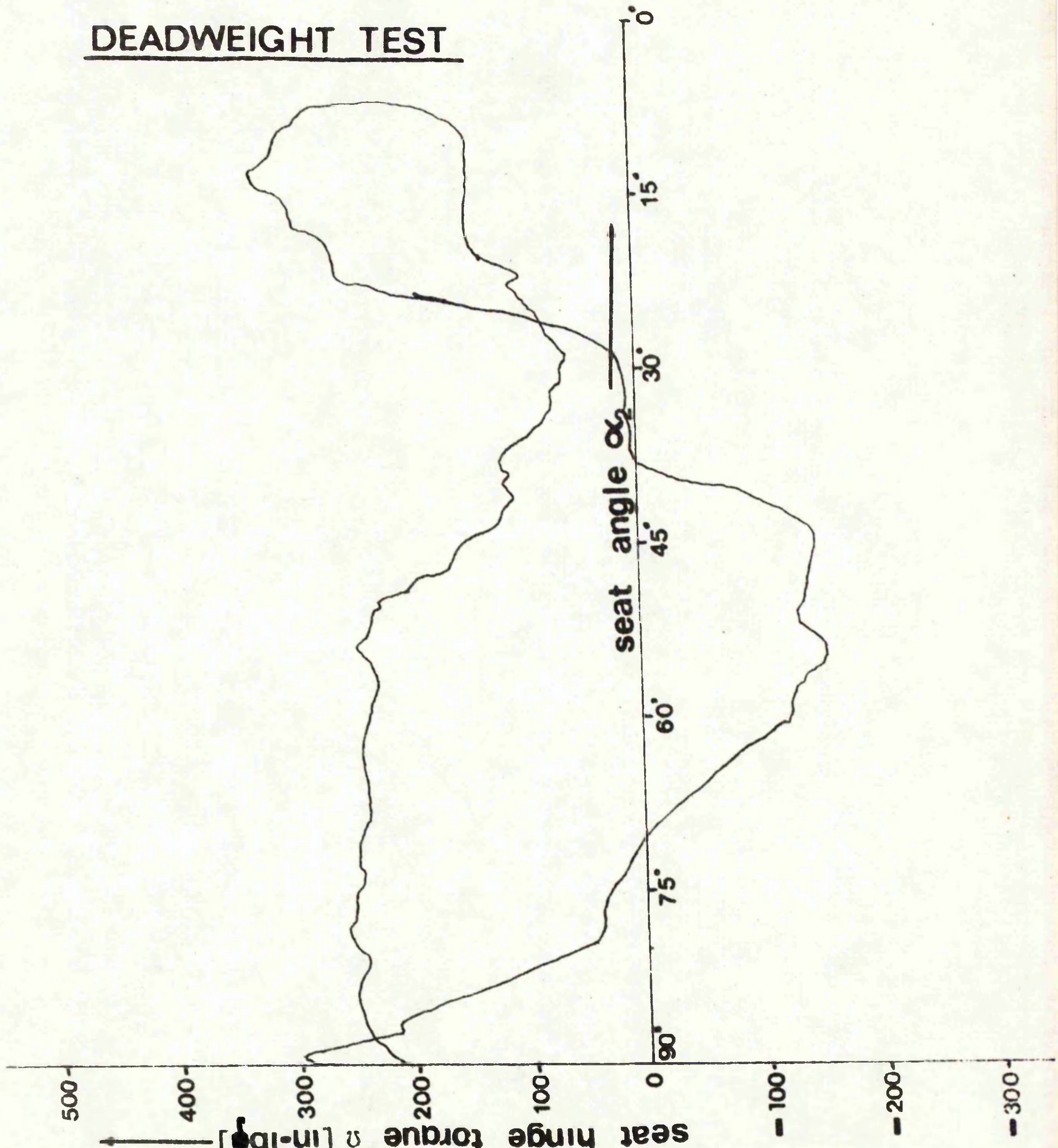
Figure 44

Subject VI

disability ----- Congenital Spastic.
Paralysed on left side of body.
Wears a caliper on left leg.
physique ----- Tall, slim build.

sex --- Female **age** --- 18-20
effectiveness of chair aid ---- Able to rise and sit down in chair aid. Chair aid gives subject control over her movements.

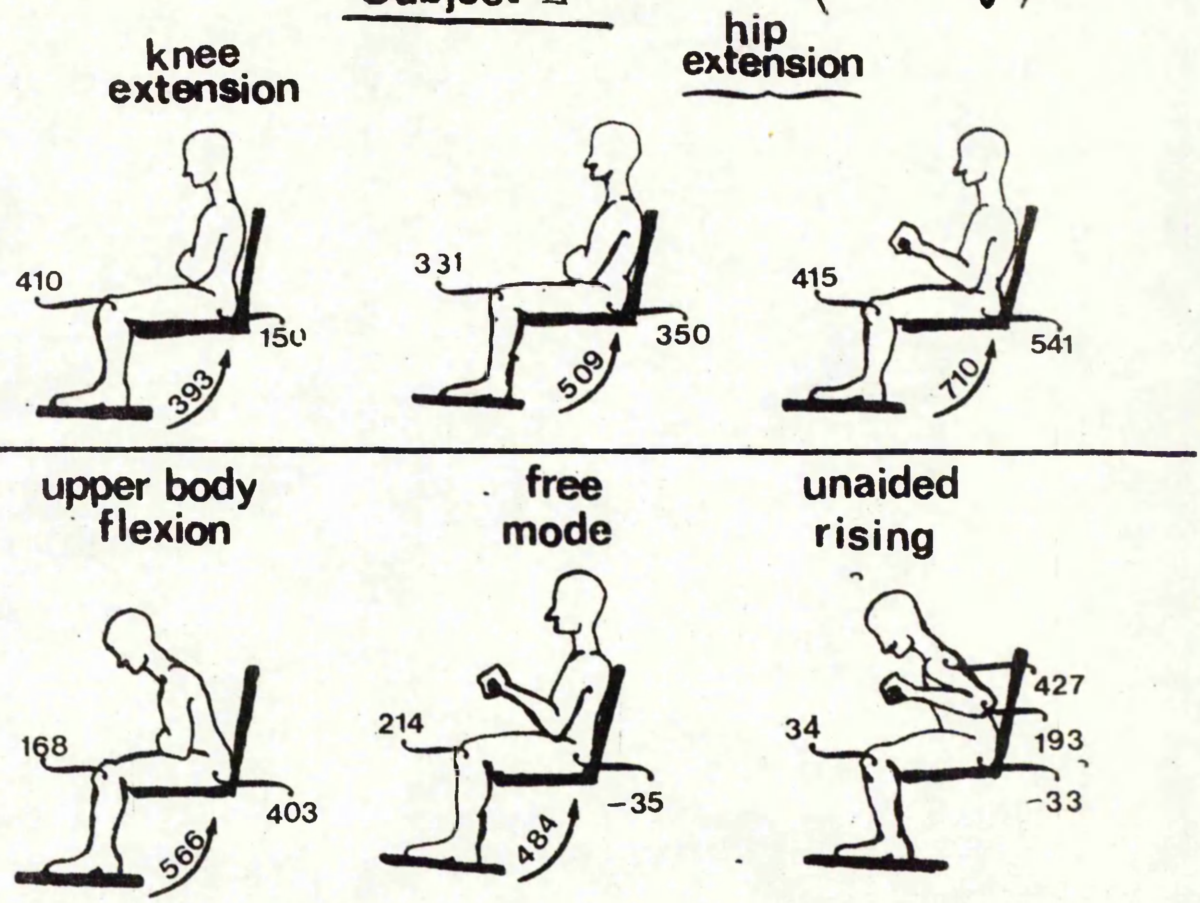
DEADWEIGHT TEST



CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject VI



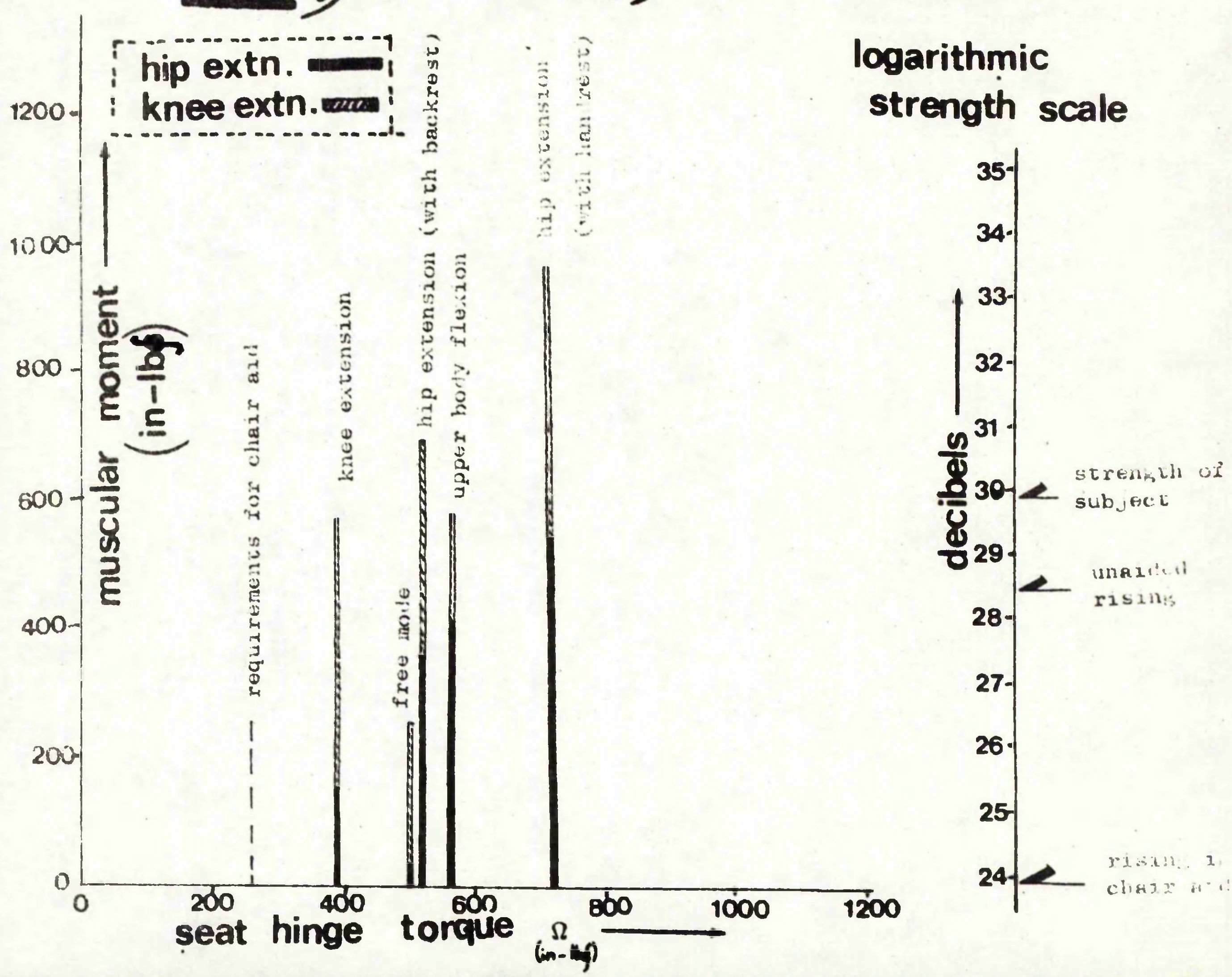


figure 44

Subject VII

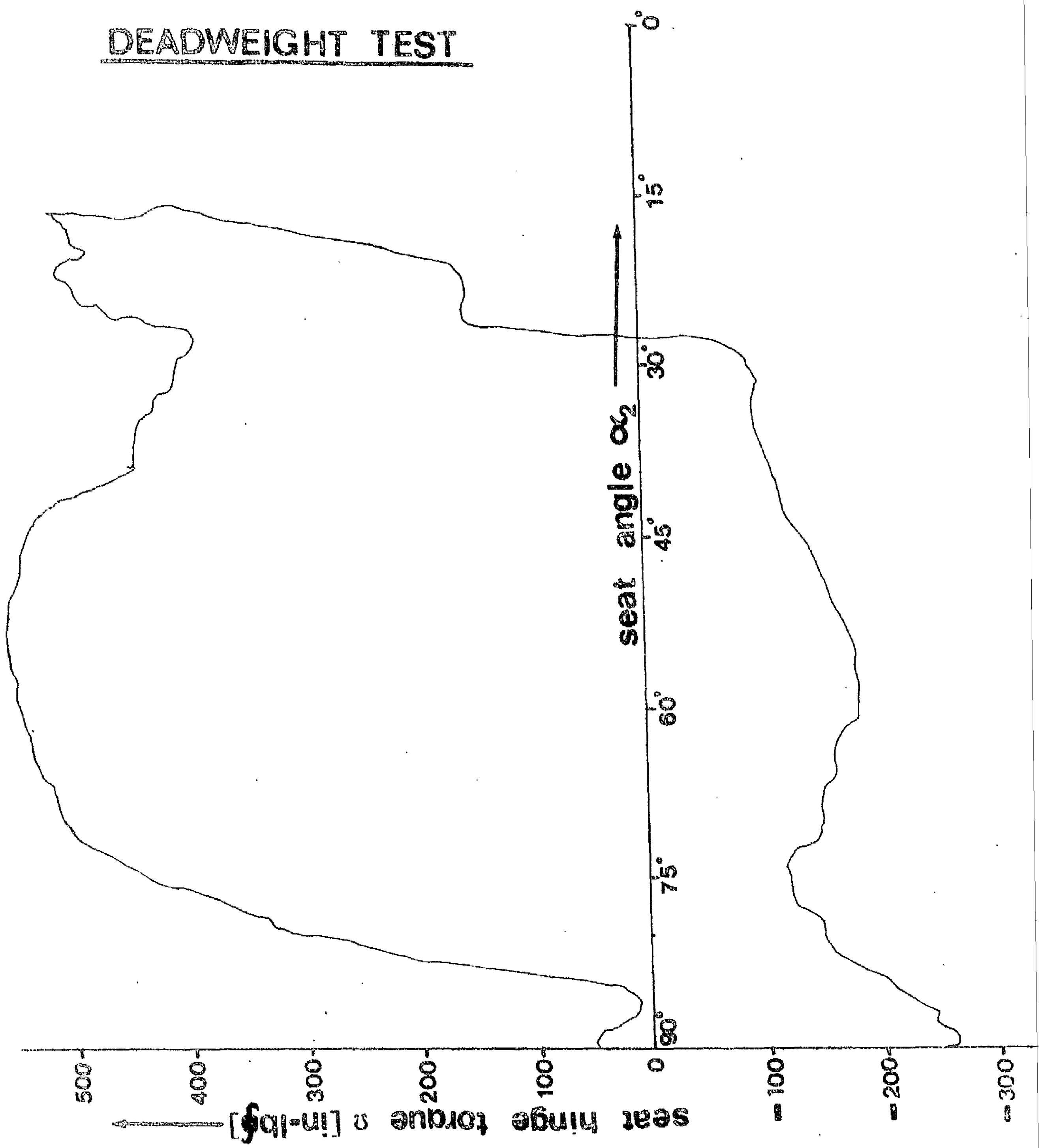
disability ------ Paraplegia.

physique ------ Tall, heavy build, powerful arms and trunk.

sex --- Male   age --- 35-40

effectiveness of chair aid ---- Rose and sat down without difficulty.

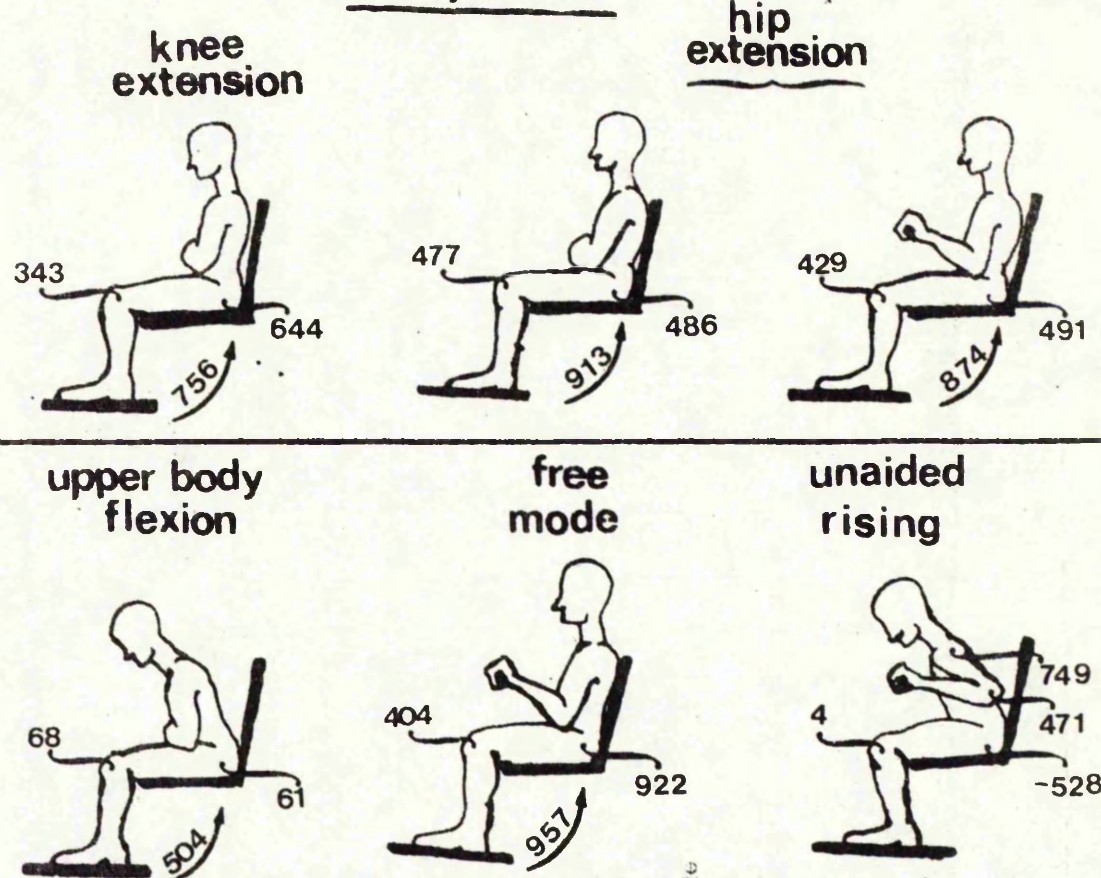
DEADWEIGHT TEST



CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject VII



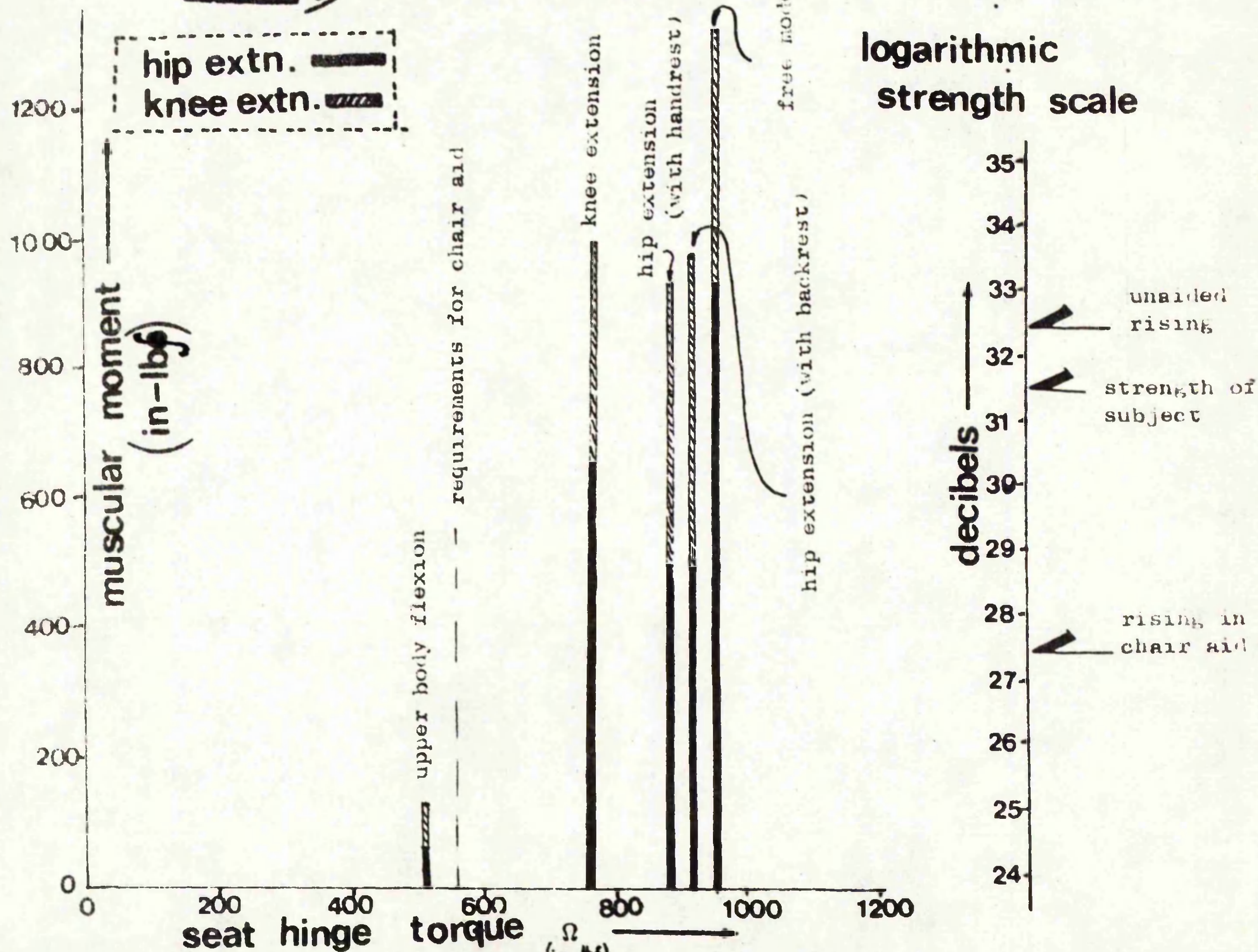


figure 44

Subject VIII

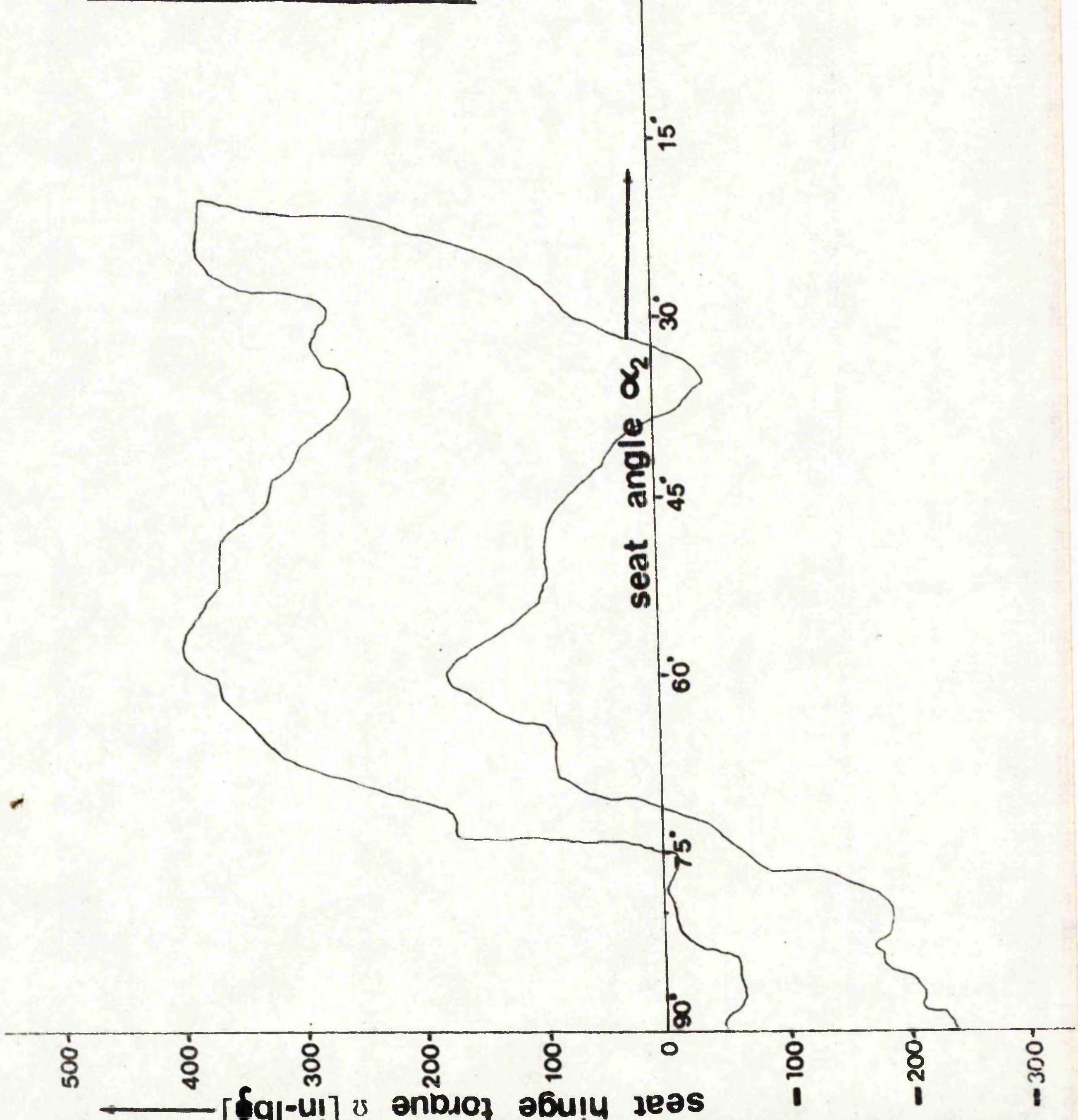
disability ----- Paraplegia.
Calipers on both legs

physique ----- Medium height, Medium build, powerful arms and trunk.

sex --- Male **age** --- 40-45

effectiveness of chair aid ---- Rose with ease, had difficulty sitting down over the last 20° of the cycle.

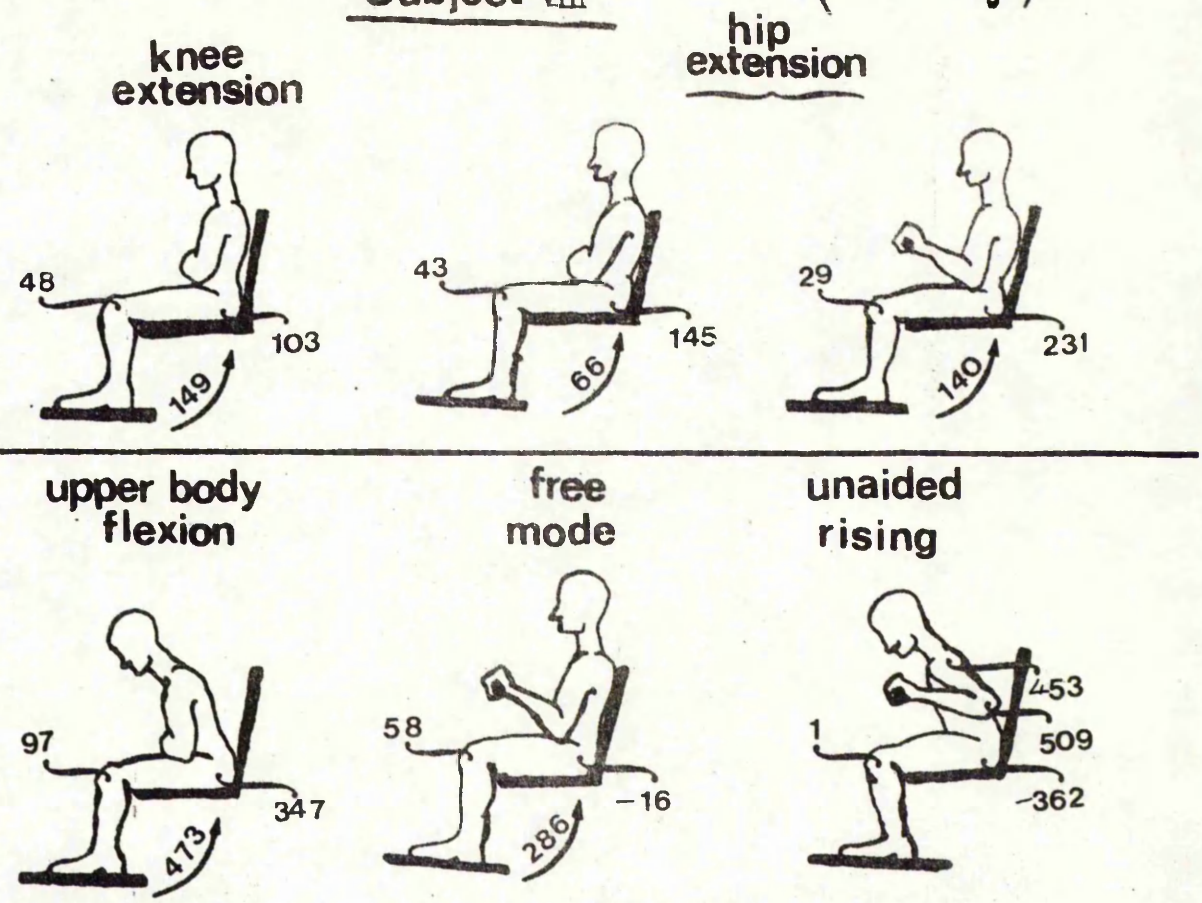
DEADWEIGHT TEST

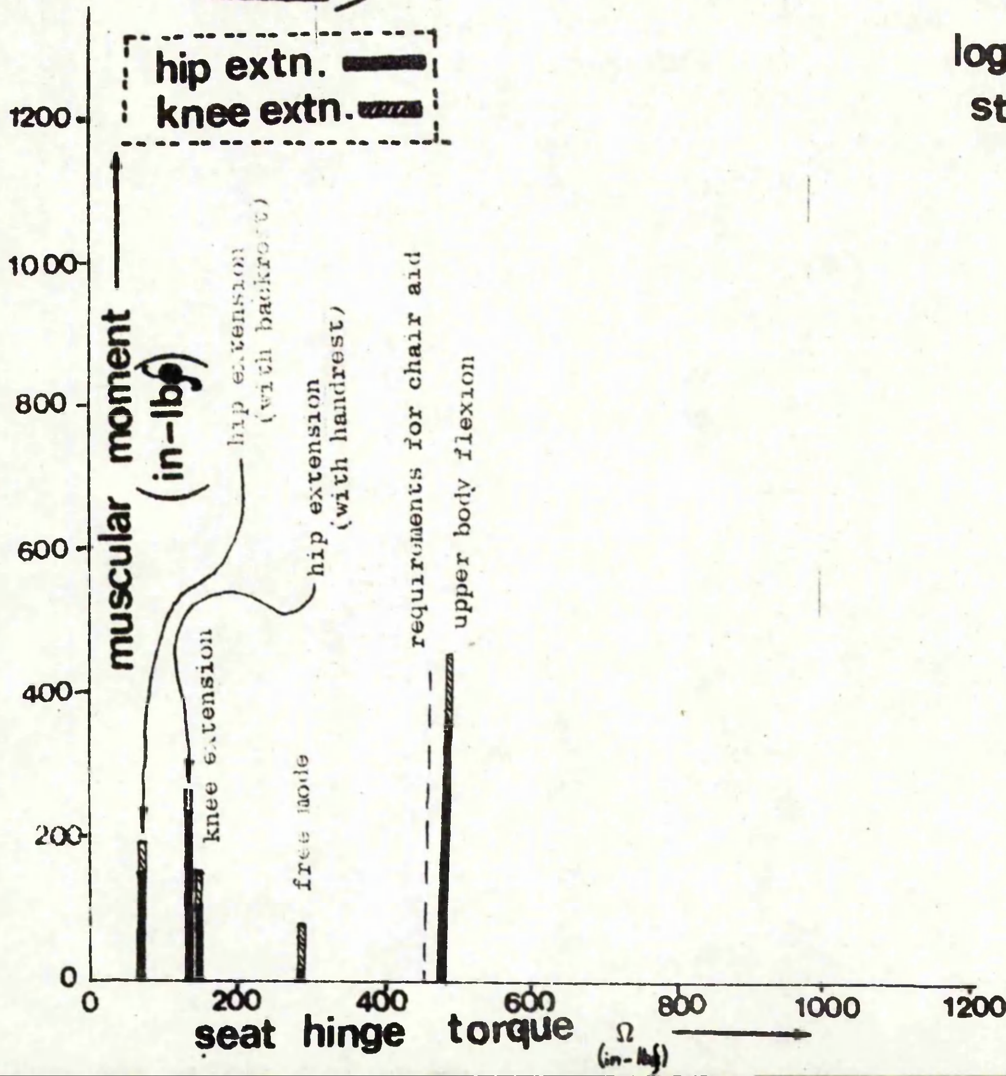


CHAIR AID TEST RESULTS

(figures in in-lbf.)

Subject VIII





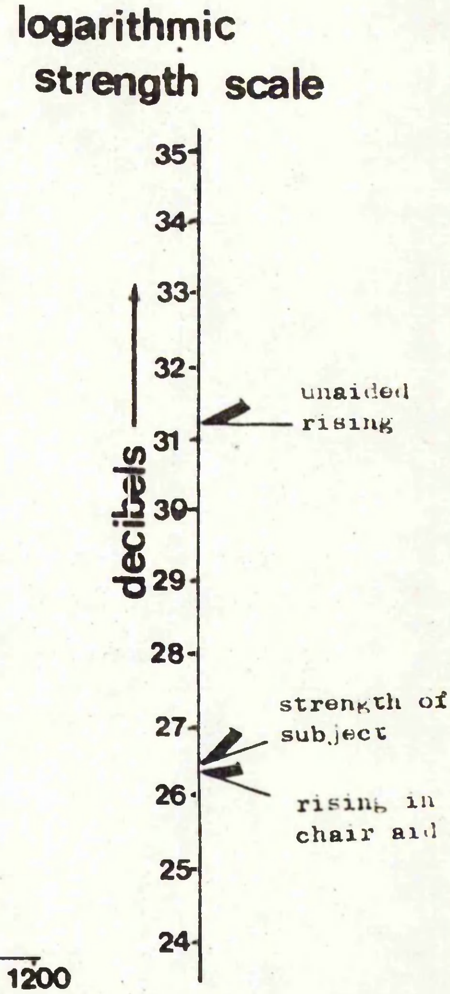


Figure 45

THE EFFECTIVENESS OF MODES OF RISING IN THE CHAIR AID

THE FREE MODE

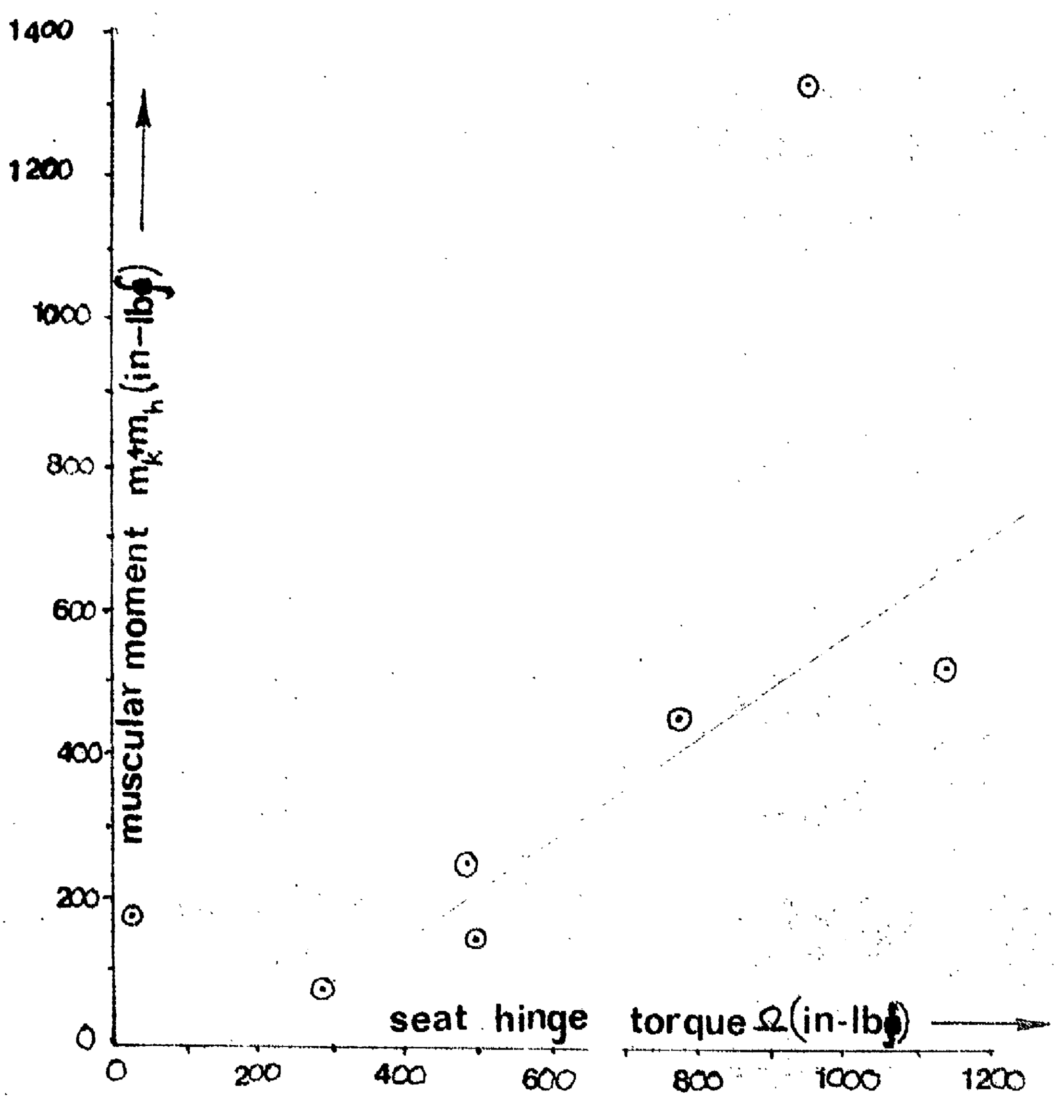


Figure 45(cont.)

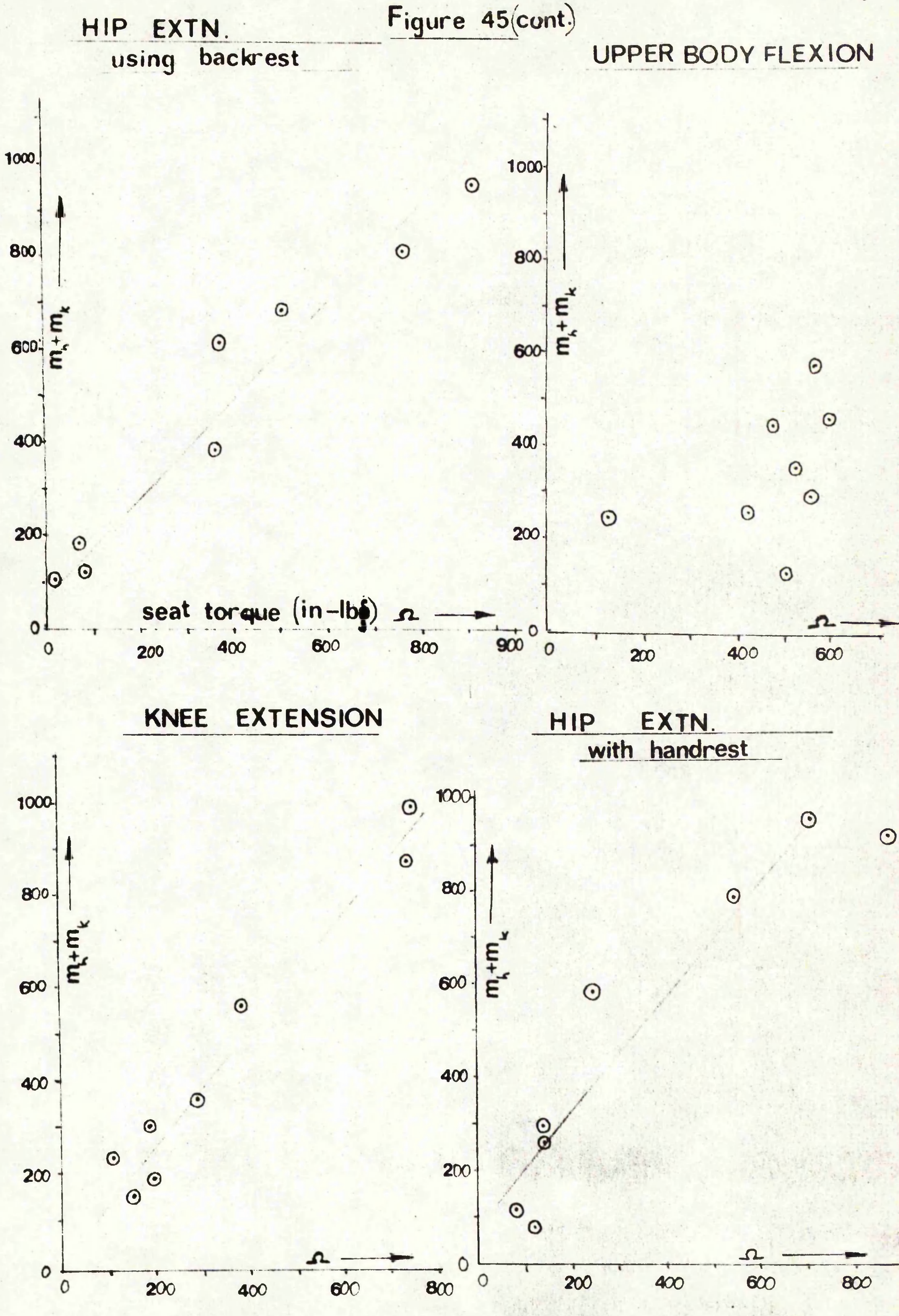


Figure 46

THE LOGARITHMIC STRENGTH SCALE

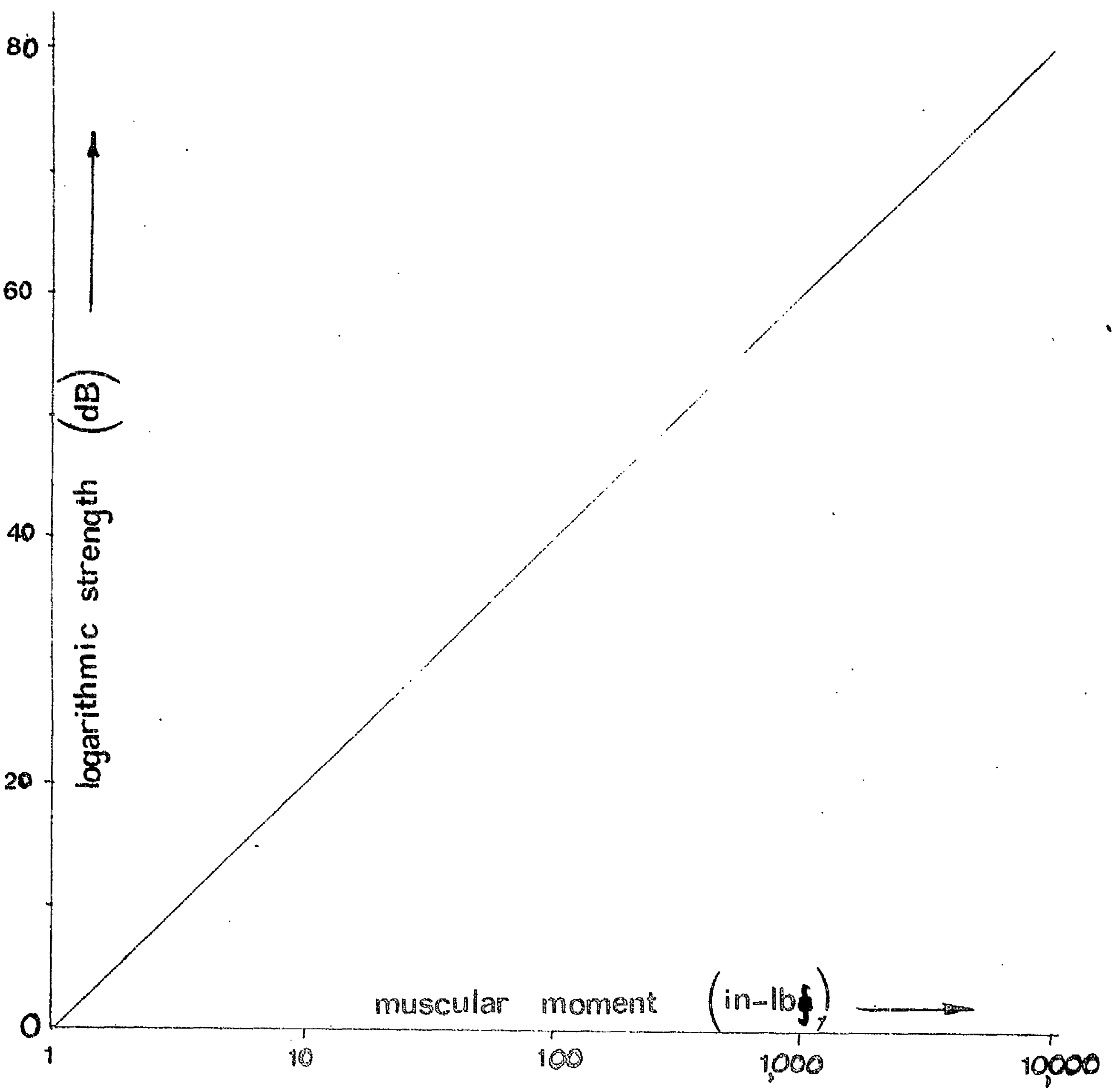


Figure 47

STRENGTH REQUIREMENTS AT THE SHOULDER JOINT IN EXERTING A VERTICAL FORCE AT THE HAND

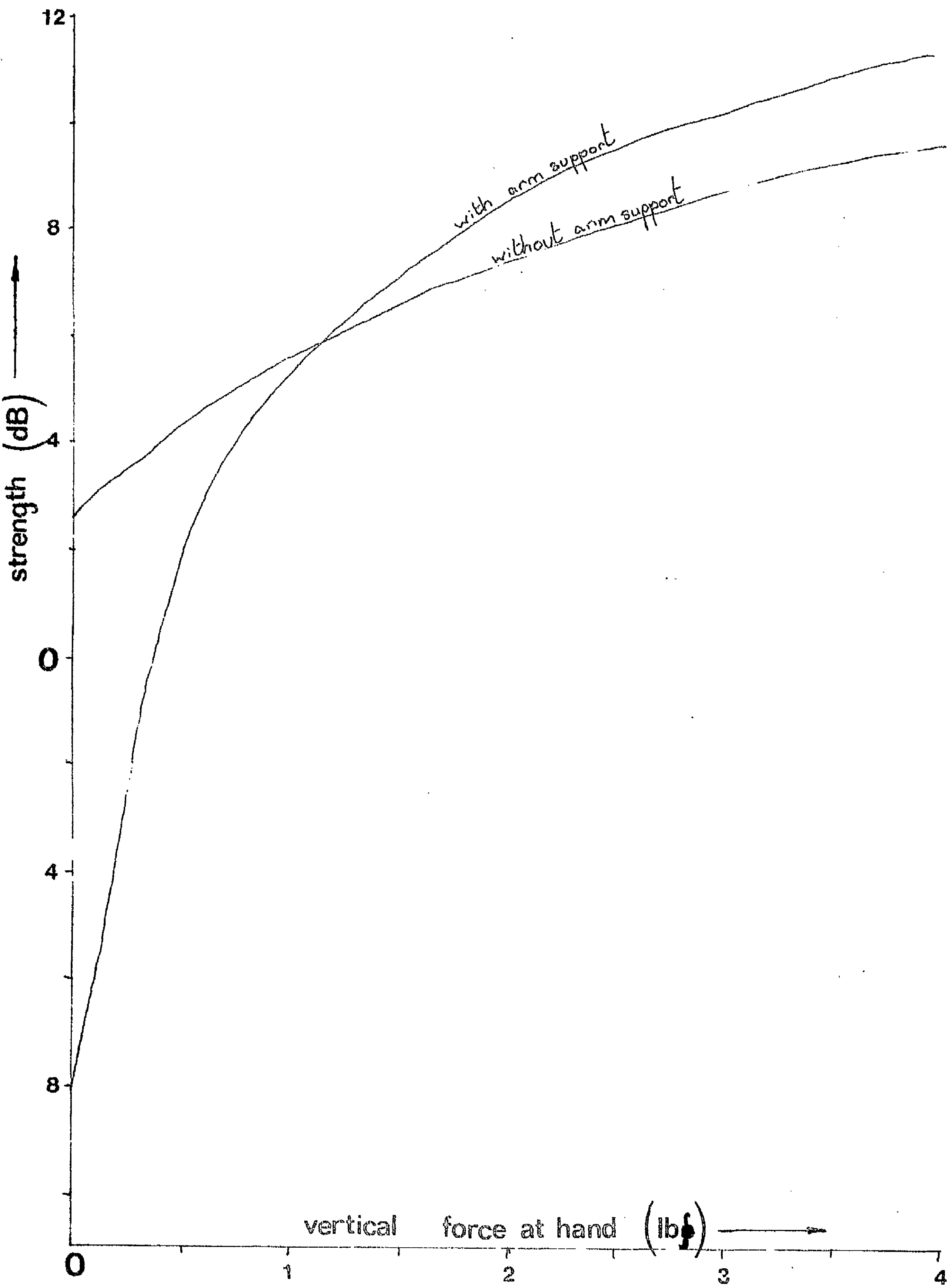


Figure 48

LANDMARKS ON THE LOGARITHMIC STRENGTH SCALE

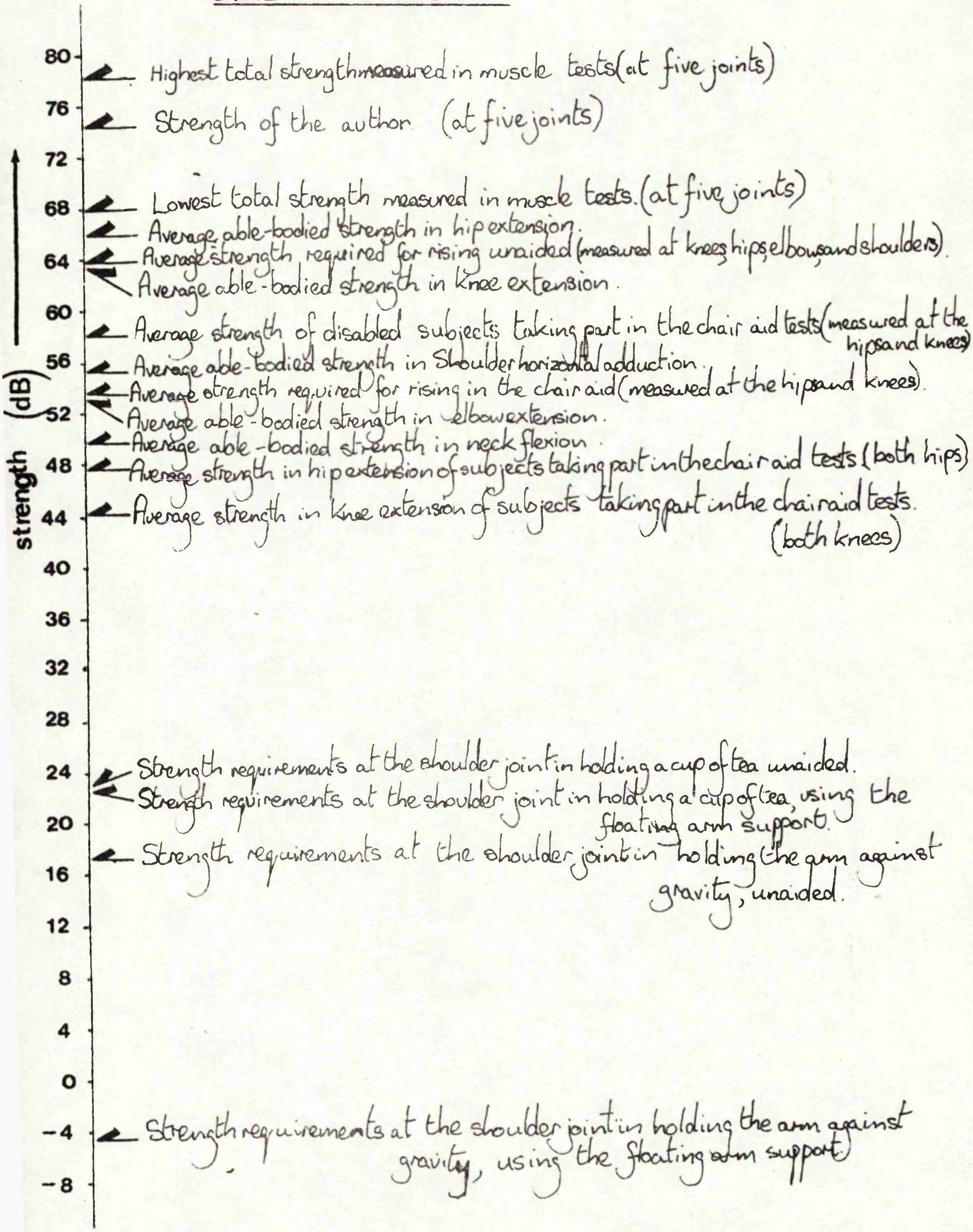


Figure 50

UNAIDED RISING

WITHOUT USE OF ARMS

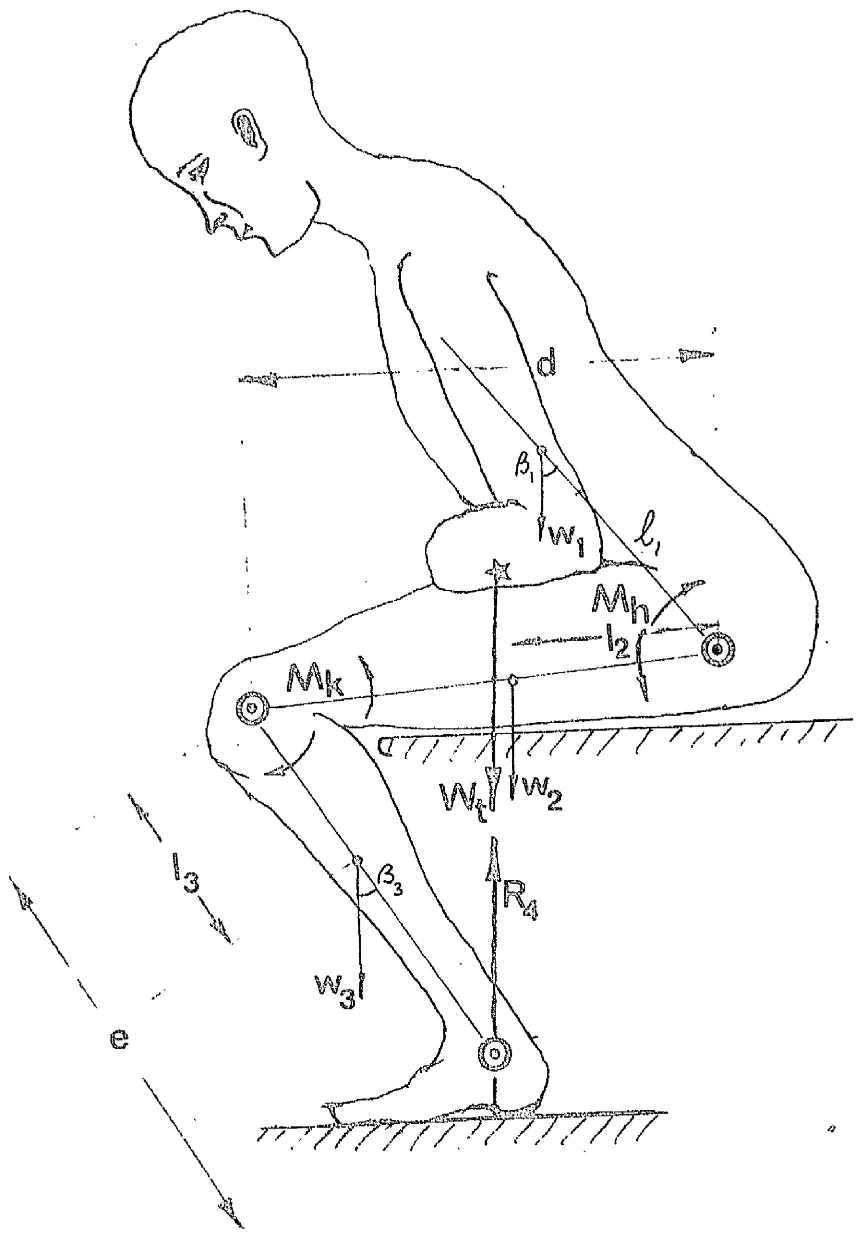


Figure 52

SEAT DYNAMOMETER CALIBRATION

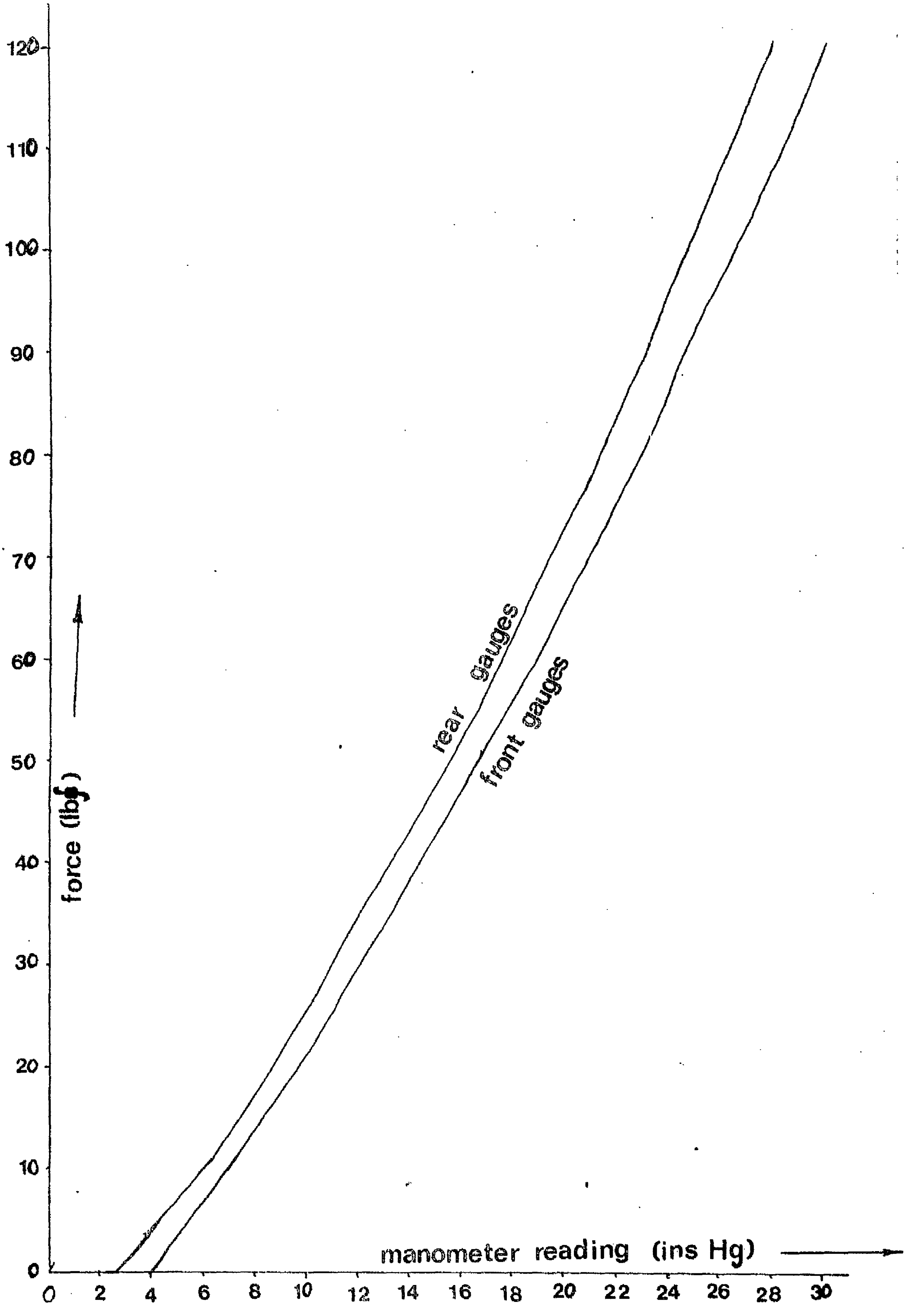


Figure 53

KNEE EXTENSION

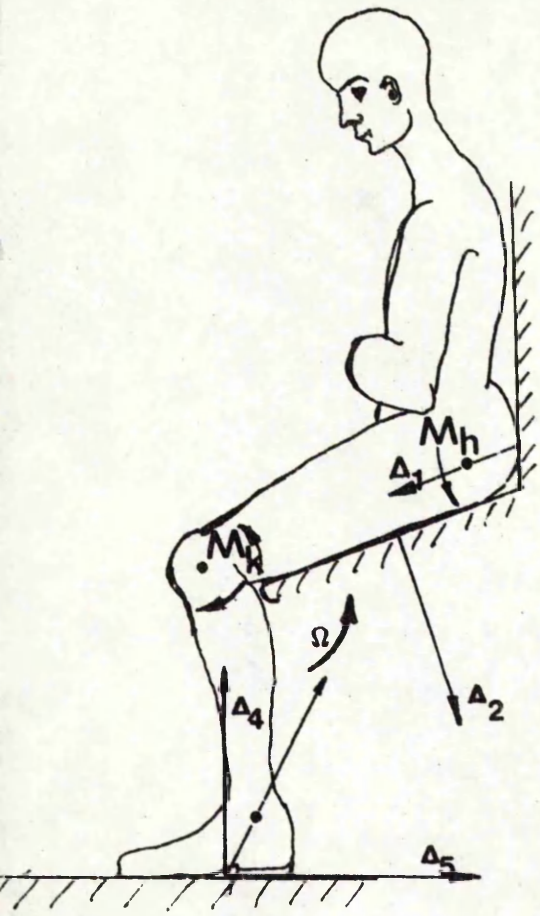


Figure 54

HIP EXTENSION using handrest

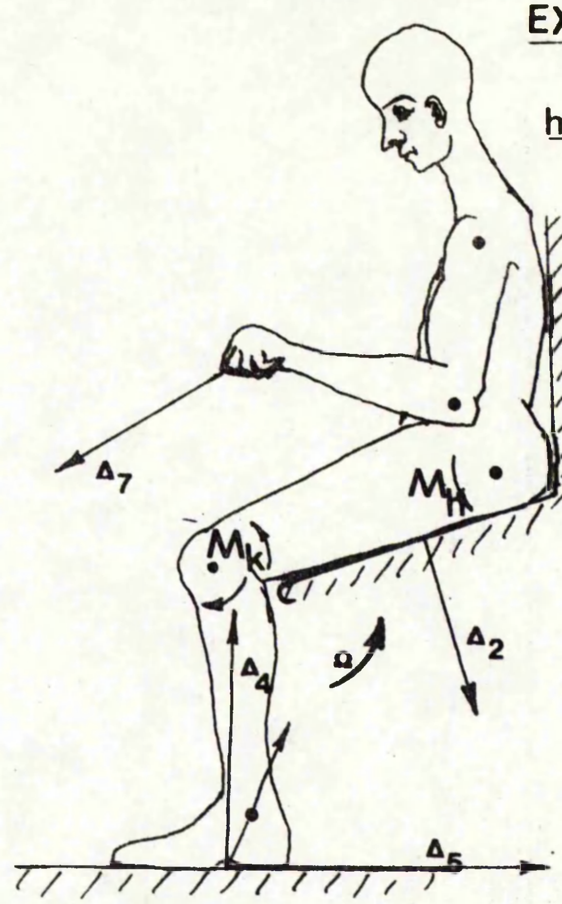


Figure 55

HIP EXTENSION using backrest

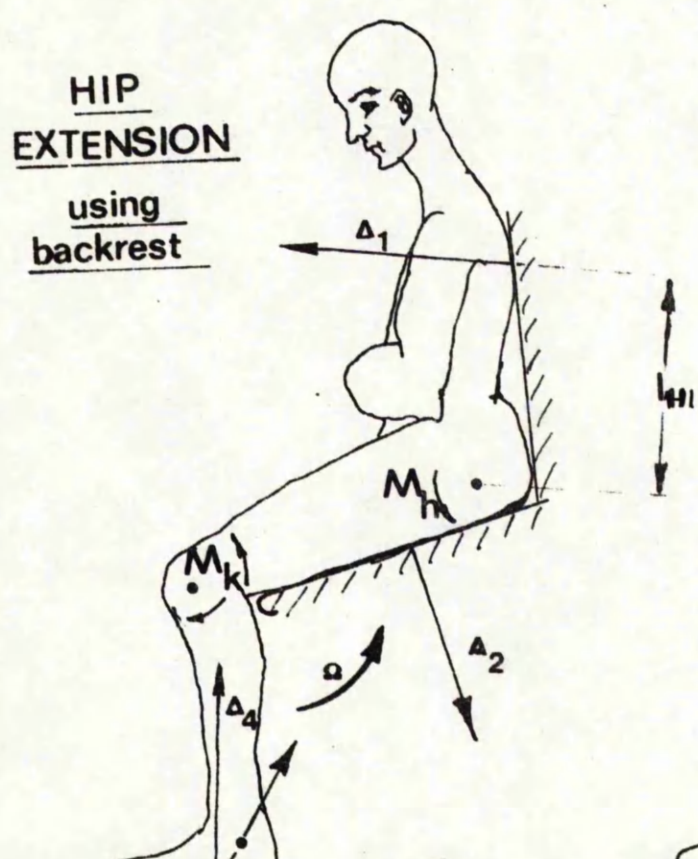


Figure 56

UPPER BODY FLEXION

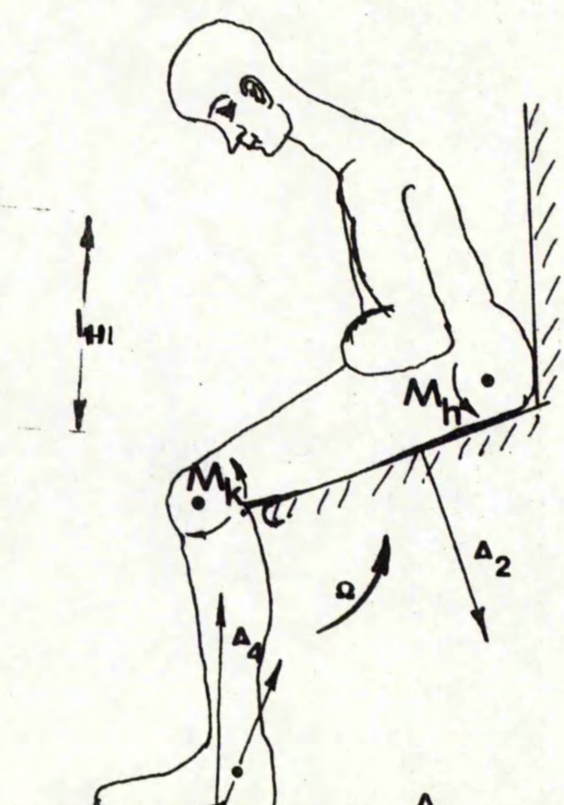
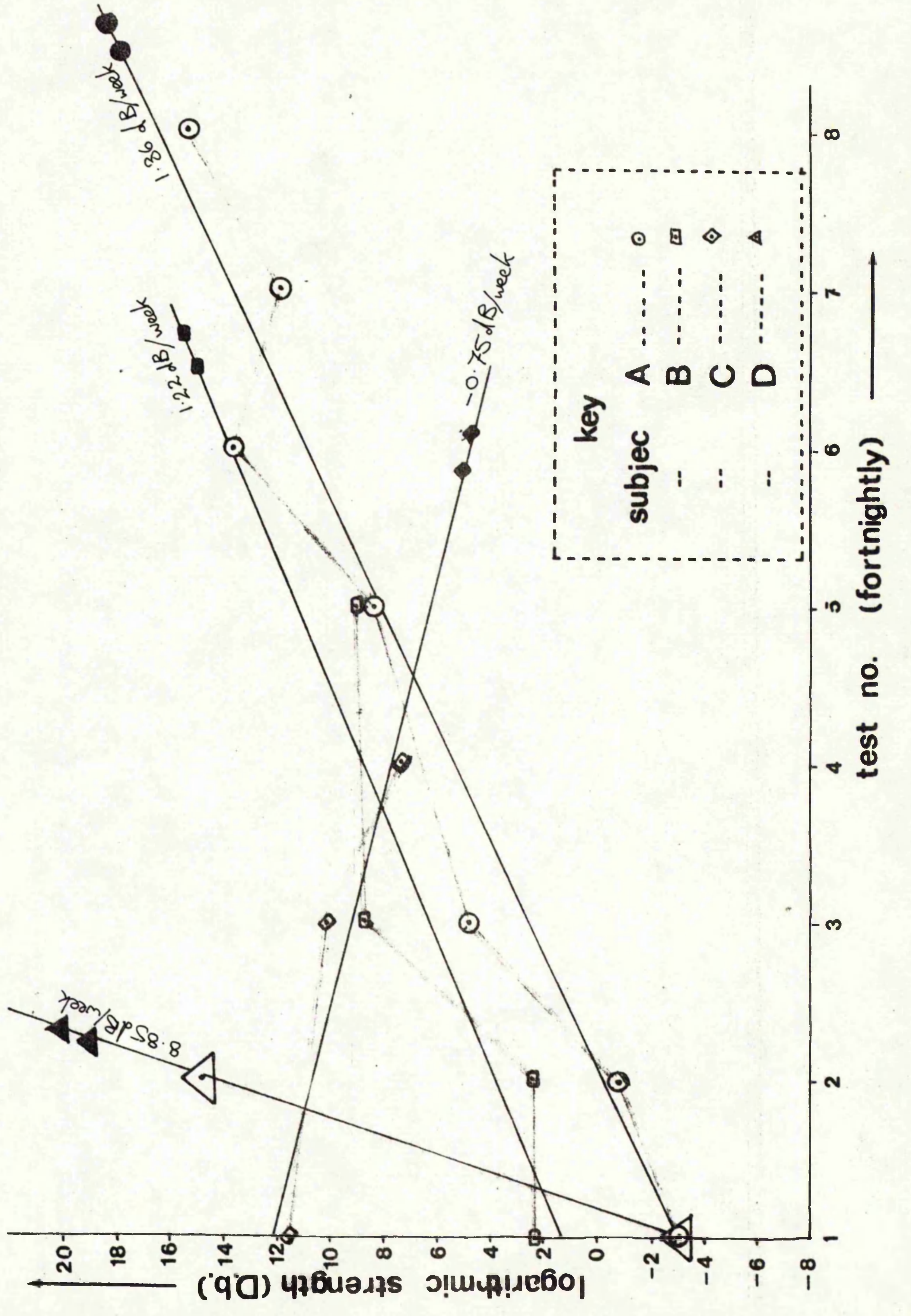


Figure 57

TESTS ON THE THEREPEUTIC EFFECT OF THE FLOATING ARM SUPPORT



APPENDIX SECTION 1

Muscle Strength Testing

For each subject a results sheet was filled in, as shown at the end of this section. Measurements were taken of the force exerted at a point on the body link, the length of the link to the point of application of the force, and where necessary the weight of the limb as measured at that point. The limb weight was subtracted from the value of the measured force to give the force applied by the subject. The product of the link length and the applied force gave the moment about the joint exerted by the muscle group. The sum of the moments about the five joints tested gave the total moment exerted; and the moment at each joint divided by the total moment gave the strength ratio of each muscle group. A complete list of this data for all the subjects tested is given in the computer program.

Figure (9) is a histogram of these results, showing the arithmetic mean and standard deviation of each muscle group. The results are assumed to take the form of a normal distribution.

A.1.2 An Estimation of the Magnitude of the Error in using the Statistical Results to calculate the Residual Strength of a Muscle Group

Let us suppose a subject is being tested whose legs are partially paralysed at the knee. His strength in knee extension is measured as S_K in lbs; while the strengths of his unaffected elbow, shoulder, neck, and hip joints are found to be M_E, M_S, M_N, and M_H respectively.

We can now say that the expected strength of a healthy subject in knee flexion, M_K, will be:

$$M_K \pm \sigma_K = (M_E \pm \sigma_E + M_S \pm \sigma_S + M_N \pm \sigma_N + M_H \pm \sigma_H)\ R \quad \text{......... A.1.1}$$

where σ_E, σ_S, σ_N, σ_H, and σ_K are the standard deviations of M_E, M_S, M_N, M_H, and M_K respectively; and R is the ratio of M_K to $(M_E + M_S + M_N + M_H)$ as taken from the experimental results.

We now have, for the greatest error, and substituting $(\bar{M}_K)$ for $(M_E + M_S + M_N + M_H)R$, the mean value of M_K,

$$\hat{M}_K = (\bar{M}_K) \pm (\sigma_E + \sigma_S + \sigma_N + \sigma_H)\ R + \sigma_K \quad \text{........ A.1.2}$$

Now the data obtained from the statistical results gives:

$$\sigma_K = 0.101\ M_K$$

$$\sigma_E = 0.213\ M_E \quad \text{and} \quad M_E = M_K \cdot \frac{0.093}{0.308}$$

$$\sigma_S = 0.154\ M_S \quad \text{and} \quad M_S = M_K \cdot \frac{0.122}{0.308}$$

$$\sigma_N = 0.229\ M_N \quad \text{and} \quad M_N = M_K \cdot \frac{0.063}{0.308}$$

$$\sigma_H = 0.130\ M_H \quad \text{and} \quad M_H = M_K \cdot \frac{0.415}{0.308}$$

$$\text{also } R = \frac{0.308}{1-0.308} = 0.445$$

Substituting these figures in equation A.1.2 we now have,

$$\hat{M}_K = (\bar{M}_K) \pm (0.093 \times 0.213 + 0.122 \times 0.154 + 0.063 \times 0.229 + 0.415 \times 0.130)$$

$$\times \frac{0.445}{0.308} + 0.101 \quad \therefore \hat{M}_K = (\bar{M}_K) \pm 0.255\ (\bar{M}_K)$$

Now the residual strength is given by

$$\% \text{ R.S.} = \frac{100 \quad S_K}{M_K} \quad \text{.............. A.1.3}$$

where S_K is the measured strength of the subject at the knee joint, and M_K is the expected strength of a healthy subject.

Substituting for M_K in eqn. A.1.3 we have

$$\% \text{ R.S.} = \frac{100 \ S_K}{(\bar{M}_K) \pm 0.255(\bar{M}_K)}$$

$$= \frac{S_K}{(\bar{M}_K)} \ (100 \pm \ 27) \ \%$$

MUSCLE TESTING RESULTS SHEET

Subject No. 14 | Name .... | Date ....

Age 21 | Sex M | Height 73 ins

Weight 152 lbf | R. Handed
R. Footed

Somatotype F_5 T_3

Occupation Student

Physical pursuits Rowing

Comments

Neck flexion length	8.0 ins	Neck flexion moment = 336 in lbf
Neck flexion force	42 lbf	R_{NECK} = 0.064
Thigh length	16.2 ins	Hip Extension moment = 2413 in lbf
Thigh wt.	22 lbf	
Hip Extn.Force	171 lbf	R_{HIP} = 0.458
Lower leg length	16.0 ins	Knee Extension moment = 1456 in lbf.
Knee extn. force	91 lbf	R_{KNEE} = 0.276
Upper arm length	11.9 ins	Shoulder Horiz. Adduction moment = 642 in.lbf
Shoulder Horiz. add. force	54 lbf	$R_{SHOULDER}$ = 0.122
Forearm length	11.0 ins	Elbow extension moment = 423 in lbf.
Forearm wt.	2.5 lbf	R_{ELBOW} = 0.080
Elbow extn.force	41.0 lbf	

Total Moment = 5272 in.lbf.

APPENDIX SECTION 2

SYNTHESIS OF THE STRENGTH EQUATION

We wish to solve the equation:

$$\frac{M(5)_i}{\overline{M(5)}} = \left(\frac{W_i}{\bar{W}}\right)^w \left(\frac{H_i}{\bar{H}}\right)^h \left\{\frac{R(5-r_i) + T\,(5-t_i) + M\,(r_i + t_i - 6)}{R(5-\bar{r}) + T\,(5-\bar{t}\,) + M\,(\bar{r} + \bar{t} - 6)}\right\}\left\{\frac{\text{Sin } F\,(Y_i)}{\text{Sin } F\,(\bar{Y})}\right\}^y$$

where $F\,(Y_i) = \dfrac{Y_m - Y_i}{Y_m}$ when $Y_m > Y_i$

and $F\,(Y_i) = \dfrac{100 - Y_i}{100 - Y_m}$ when $Y_m < Y_i$

and $F\,(\bar{Y})$ is the same function with $\bar{Y}$ substituted for Y_i.

The following is a list of symbols used in the text, with the equivalent as used in the program.

Text	Program	
$M(5)_i$	S(I)	The strength of the subject in inch pounds as given in the text in chapter 2.
$\overline{M(5)}$	4814.0	The average strength of the population tested
W_i	W(I)	The body weight of each subject in inch pounds
$\bar{W}$	162.0	The average body weight of the population tested
w	EW	The body weight index
H_i	H(I)	The height of each subject in inches
$\bar{H}$	69.6	The average height of the population tested.
h	EH	The height index
R	ER	The rotundity coefficient
r_i	R(I)	The somatotype coordinate of rotundity (see Figure 7)
T	ET	The thinness coefficient.

Text	Program	
t_i	T(I)	The somatatype coordinate of thinness
M	A	The muscularity coefficient
$\bar{r}$	4.000	The average somatatype coordinate of rotundity of the population tested.
$\bar{t}$	3.655	The average somatatype coordinate of thinness of the population tested.
y	EY	The index of the age function
Y_m	EZ	The age of maximum strength
Y_i	Y(I)	The age, in years, of each subject
Y	X(I)	An abbreviation for the age function (see previous page)
$\bar{\bar{Y}}$	EX (21.53)	The average age of the population tested
	EA	The standard deviation of the calculated strengths from the measured strengths.
	PI	π

All other letters used in the program are there for programming convenience, and are not otherwise of importance.

The program itself is a paragon of simplicity. Each unknown parameter in the equation is given an initial value, the strength is computed from the strength equation, and the standard deviation of the calculated strengths from the measured strengths, for all 29 subjects is then calculated. The computer then returns to the beginning of the program, where it recomputes the equation using a different value of one of the parameters. Whenever the standard deviation falls below a specified value, the computer prints out the value of the parameters it has just used. The computer continues to cycle the parameters, until it has gone through the whole range of each.

In cycling the six parameters, care must be taken to restrict the range and interval of the value of each parameter as it is cycled; or the number of cycles to be carried out soon becomes astronomical. In practice, a large interval was used for each cycle at first, with a large specified value for the standard deviation. It is possible, by running the program several times, narrowing down the range and interval of the parameters each time, to "home in" on their best values.

On the following page is a sample of the results printed out by the computer. It can be seen how the results with the lowest standard deviation are grouped around a narrow set of values of the parameters.

It is also interesting to see how individual values of the calculated strength vary, as the parameters vary.

The results given by the optimum set of parameters are plotted in figure (11); and the strength equation is given in the completed form in chapter 2 of the text.

THE STRENGTH EQUATION SYNTHESIS PROGRAM

```
      DIMENSION W(29),S(29),H(29),R(29),T(29),Y(29),X(29)
      READ (1,101) (W(I),S(I),H(I),R(I),T(I),Y(I),I=1,29)
101   FORMAT (6F8.1)
      L=0
      EO=1000000.0
      E=0.0
      PI=3.14159
      A=100.0
      DO 1 J2=7,21,2
      EY=0.05*FLOAT(J2-11)
      DO 1 J1=1,3,1
      EZ=FLOAT(J1-3)
      DO 1 J3=19,41,2
      EH=0.05*FLOAT(J3-1)
      DO 1 J4=1,13,2
      EW=0.05*FLOAT(J4-11)
      DO 1 J5=19,31,1
      ET=FLOAT(J5+19)
      DO 1 J6=19,31,1
      ER=FLOAT(J6+24)
  2   DO 3 I=1,29
      IF (Y(I)-25.0) 4,4,5
  4   X(I)=Y(I)/(25.0+EZ)
      EX=(21.53+EZ)/(25.0+EZ)
      GO TO 6
  5   X(I)=(100.0-Y(I))/(75,0-EZ)
      EX=49.083/(75.0-EZ)
  6   ES=((W(I)/162.0)**EW*(H(I)/59.6)**EH*((5.0-R(I))*ER+(5.0-T(I))*ET
     1   +(R(I)+T(I)-6.0)*A)/(ER+1.345*ET+1.655*A)*(SIN(X(I)*PI/180.0)
     1   /SIN(EX*PI/180.0))**EY)*4814.0
      E=SQRT(E*E+(ES-S(I))**2)
      IF (L.EQ.1) WRITE (2,201) ES,S(I)
201   FORMAT (1H, 2F8.1)
  3   CONTINUE
      L=0
      EA=E/5.3852)
      IF (EA.GT.759) GO TO 1
      WRITE (2,202) EW,EH,ER,ET,EY,EZ,EA
202   FORMAT (1H, 7F8.1)
      IF (EO.LE.E) GO TO 1
      EO=E
      L=1
      GO TO 2
  1   E=0
      CONTINUE
      STOP
      END
```

DATA

W (lbf)	M(5) (in-lbf)	H (ins)	r	t	Y (years)
147	4920	70	4	3	22
155	4856	69	3	4	30
170	4922	67	2	5	47
161	5237	72	5	3	22
186	6288	72	5	5	24
203	7845	76	5	4	24
196	4279	71	3	4	62
150	5392	65	4	4	24
161	3691	63	4	3	58
124	3748	62	4	3	48
163	4528	68	3	5	50
168	4528	68	4	3	52
152	5272	73	5	3	21
156	5373	71	4	4	51
147	3769	67	5	3	25
188	3759	71	3	5	51
168	3307	70	3	4	58
164	8049	72	4	5	19
168	5938	74	5	3	19
147	4646	69	5	2	20
96	2473	60	5	1	19
140	3539	74	5	2	20
140	5108	71	5	2	23
133	5650	70	4	4	23
176	4184	68	3	5	34
154	4564	72	5	3	18
224	3671	74	2	5	20
158	4730	71	4	4	23
207	5337	69	3	5	48

RESULTS OF THE STRENGTH EQUATION SYNTHESIS PROGRAM.

w	h	r	t	y	(Y + 27)	S.D.
-.2	1.4	46.0	43.0	0.0	-2.0	758.9
-.2	1.4	45.0	44.0	0.0	-2.0	759.0
-.2	1.4	46.0	44.0	0.0	-2.0	758.7

M(5) Calculated	M(5) Measured
4277.5	4920.0
4183.4	4856.0
3974.6	4922.0
5377.7	5237.0
7256.4	6288.0
6614.5	7845.0
4154.5	4279.0
4759.5	5392.0
3624.3	3691.0
3734.0	3748.0
5020.6	4528.0
3999.1	4520.0
5545.9	5272.0
5343.6	5373.0
4951.5	3769.0
5183.4	3759.0
4200.4	3307.0
6436.8	8049.0
5540.6	5938.0
4156.4	4646.0
2823.4	2473.0
4629.0	3539.0
4368.4	5108.0
5408.4	5650.0
4944.2	4184.0
5425.7	4564.0
4322.7	3671.0
5330.0	4730.0
4885.1	5337.0

-.2	1.4	46.0	44.0	0.0	-2.0	758.7
-.2	1.4	47.0	44.0	0.0	-2.0	758.8
-.2	1.4	47.0	45.0	0.0	-2.0	758.9
-.2	1.5	46.0	43.0	0.0	-2.0	758.9
-.2	1.5	45.0	44.0	0.0	-2.0	759.0
-.2	1.5	46.0	44.0	0.0	-2.0	758.5

4279.9	4920.0
4179.8	4856.0
3959.5	4922.0
5395.9	5237.0
7281.1	6288.0
6673.0	7845.0
4162.8	4279.0
4727.0	5392.0

RESULTS OF THE STRENGTH EQUATION SYNTHESIS PROGRAM (Contd)

w	h	r	t	y	$(\hat{Y} + 27)$	S.D.
M(5) Calculated	M(5) Measured					
3588.4	3691.0					
3691.1	3748.0					
5009.0	4528.0					
3989.8	4520.0					
5572.5	5272.0					
5354.3	5373.0					
4932.6	3769.0					
5193.7	3759.0					
4202.8	3307.0					
6458.7	8049.0					
5574.6	5938.0					
4152.8	4646.0					
2781.8	2473.0					
4657.5	3539.0					
4377.1	5108.0					
5411.5	5650.0					
4932.7	4184.0					
5444.1	4564.0					
4349.2	3671.0					
5340.6	4730.0					
4880.9	5337.0					
-.2	1.5	46.0	44.0	0.0	-2.0	758.5
-.2	1.5	47.0	44.0	0.0	-2.0	758.4
4282.4	4920.0					
4199.7	4855.0					
3994.6	4922.0					
5376.1	5237.0					
7254.3	6288.0					
6648.4	7845.0					
4182.6	4279.0					
4725.9	5392.0					
3590.5	3691.0					
3693.2	3748.0					
5024.7	4528.0					
3992.2	4520.0					
5552.0	5272.0					
5352.9	5373.0					
4914.5	3769.0					
5210.0	3759.0					
4222.8	3307.0					
6453.5	8049.0					
5554.1	5938.0					
4137.5	4646.0					
2771.6	2473.0					
4640.3	3539.0					

RESULTS OF THE STRENGTH EQUATION SYNTHESIS PROGRAM (Contd)

w	h	r	t	y	$(\hat{Y} + 27)$	S.D.
M(5) Calculated	M(5) Measured					
4361.0	5108.0					
5410.1	5650.0					
4948.2	4184.0					
5424.1	4564.0					
4387.9	3671.0					
5339.3	4730.0					
4896.3	5337.0					
-.2	1.5	47.0	44.0	0.0	-2.0	758.4
-.2	1.5	48.0	44.0	0.0	-2.0	758.8
-.2	1.5	46.0	45.0	0.0	-2.0	758.7
-.2	1.5	47.0	45.0	0.0	-2.0	758.4
-.2	1.5	48.0	45.0	0.0	-2.0	758.5
-.2	1.5	48.0	46.0	0.0	-2.0	758.9
-.2	1.6	46.0	44.0	0.0	-2.0	759.0
-.2	1.6	47.0	44.0	0.0	-2.0	758.8
-.2	1.6	48.0	44.0	0.0	-2.0	759.0
-.2	1.6	47.0	45.0	0.0	-2.0	758.6
-.2	1.6	48.0	45.0	0.0	-2.0	758.6
-.2	1.6	49.0	45.0	0.0	-2.0	759.0
-.2	1.6	48.0	46.0	0.0	-2.0	758.8
-.2	1.6	48.0	46.0	.1	0.0	759.0

APPENDIX SECTION 3

STATISTICAL ANALYSIS OF PRELIMINARY ARM SUPPORT TESTS

Given a set of data from the preliminary arm support tests, we wish to answer the following questions statistically (Reference 8).

1. Does the floating arm support significantly increase the maximum force applied at the hand ?

2. Is the maximum force applied at the wrist significantly greater than the force applied at the hand ,

   a. With the arm support in use ?

   b. Without the arm support ?

3. Is the force applied at the final position (at the mouth), on the "hand to mouth" path, smaller than at the other positions on the path, with the arm support in use ?

For 1 and 2 the same method is used. The results are of the form of sets of measurements of the applied force (usually of about 4 measurements per set). All the sets of results tested in the same manner, go together to form a group of results (e.g. the results of the force measured at the hand with the arm support in use comprise one group).

Firstly sets of data are discarded which do not have a corresponding set, measured on the same subject to compare directly e.g. if a subject was too weak to carry out the tests without the arm support, the results taken in using the arm support are not included in the comparison of the results without the arm support.

From the results that remain, the arithmetic mean of each group of sets was calculated. We now wish to examine whether the difference between the means can be explained by the scatter in the results. The arithmetic mean of each set in the group is now calculated, and the individual measurements are scaled to make the mean of each set equal to the mean of the whole group.

i.e. For three subjects, we have for example, measurements at the hand with the arm support in use of:

$$(x_{11}, x_{21}, x_{31}, x_{41}), \quad (x_{12}, x_{22}, x_{32}),$$

$(x_{13}, x_{23}, x_{33}, x_{43}, x_{53})$. The arithmetic mean of the group will be

$$A_G = \frac{\sum x_{ij}}{N}$$

where N is the number of results in the group. The arithmetic mean, for example of set 2 will be:

$$A_j = \frac{\sum x_{i2}}{n}$$

where n is the number of results in the set. The measurements are now scaled to the mean.

Thus x_{ij} becomes $x_{ij}\dfrac{A_G}{A_j}$

Using these figures, the standard deviation of the group can now be calculated.

This will be $$\text{S.D.} = \sqrt{\frac{\sum \left(A_G - x_{ij} \cdot \frac{A_G}{A_j}\right)^2}{N}}$$

This standard deviation gives an indication of the scatter of individual results within a group. We wish to compare the scatter of the groups. This will be given by the standard error of the group, which will be considerably lower than the standard deviation.

The standard error, $\text{S.E.} = \dfrac{\text{S.D.}}{\sqrt{N}}$

Also comparing the standard errors of the two groups $(\text{S.E.})_1$ and $(\text{S.E.})_2$, we can say that the combined error will be:

$$\sqrt{(\text{S.E.})_1^{\,2} + (\text{S.E.})_2^{\,2}}$$

We can now compare this directly with the difference between the arithmetic mean of the groups, e.g. If the error is equal to the difference in arithmetic means, there is a 68% probability that there is a systematic difference between the two groups of results. If the error is twice the difference, the probability is 95%; and if the error is three times the difference, the probability is 99.73%, which is highly significant.

Using this technique we can now commence calculating the results. Taking the first subject; his measurements without the arm support of the force in pounds exerted at the hand are:

9.1 9.9 8.6 7.4

The arithmetic mean of this set is 8.75 lbf. Now the arithmetic mean of the whole group of results of the force exerted at the hand, without the arm support is: 6.9 lbf.

We thus scale his results down, multiplying each result by $\frac{6.9}{8.75}$ They now become:

7.2 7.8 6.8 5.8

The deviations from the mean of the group will be,

0.3 0.9 0.1 1.1

and the squares of the deviations will be

0.09 0.81 0.01 1.21

Thus sum of the squares for the whole set is now calculated (for 79 results in 19 sets) this is:

$$D^2 = 198.51$$

thus $$\frac{D^2}{N} = 2.513$$ where N is the number of results.

Thus the standard deviation	=	$\sqrt{2.513}$
	=	1.58 lbf
and the standard error	=	$\frac{1.58}{\sqrt{79}}$
	=	<u>0.178 lbf</u>

Now the standard error of the corresponding group of results, with the arm support in use is 0.158 lbf.

The standard error between the two groups of results is thus:

$$\sqrt{0.178^2 + 0.158^2}$$

$$= \underline{0.238 \text{ lbf}}$$

The difference between the means, however, is 0.97 lbf. Therefore the difference between the means is four times the standard error. The difference can thus be said to be statistically highly significant. i.e. The difference is too great to be accounted for by the scatter of the results.

The following are the statistical results calculated.

1. Comparison of the maximum force at the hand using the arm support with the force when not using the arm support.

   Arithmetic mean of the force with the arm support in use = 5.98 lbf.

   Arithmetic mean of the force without the arm support = 6.91 lbf.

   Standard error between the two groups = 0.238 lbf.

$\therefore$ Difference in means = 4 x S.E.

This is a highly significant difference of 13%.

2a Results using the arm support, applying the force at the hand, compared with the force applied at the wrist are:

Arithmetic mean of force at hand	=	7.88 lbf
Arithmetic mean of force at wrist	=	6.12 lbf
Standard error between two groups	=	0.51 lbf

Difference in means = 3.5 x S.E.

This is a highly significant difference of 22%.

2b. Results in applying the force at the hand, compared with the force applied at the wrist, without the use of the arm support are:

Arithmetic mean of the force at the hand	=	9.96 lbf
Arithmetic mean of force at the wrist	=	6.05 lbf
Standard error between two groups	=	0.49 lbf
Difference in means	=	8 x S.E.

This is a highly significant difference of 40%.

3. Comparison of the force in the final position (at the mouth) on the "hand to mouth" path with the force exerted at the other positions gives:

Arithmetic mean of force exerted at the mouth	=	4.70 lbf
Arithmetic mean of force exerted at the other positions on the "hand to mouth" path	=	6.36

This is a 26% decrease. (It is felt that calculation of the standard error is not required here).

APPENDIX SECTION 4

THE FUNCTIONAL ANATOMY OF THE UPPER LIMB

This section is based on reference (1), which is a report on electromyographical studies of the upper limb over the two previous decades. Commentry on arm movements irrelevant to this thesis have been omitted, and relevant comments have been added.

The Shoulder Girdle

In static loading of the shoulder girdle, the upper fibres of the trapezius "play no active part in the support of the shoulder girdle in the relaxed, upright posture" (Bearn 1961). One can only surmise that the clavicle is then suspended by the first rib and ligaments at the sternoclavicular joint, all these being passive structures.

Muscular activity in the shoulder girdle is thus voluntary. If the subject wishes while applying a force at the hand with the arm slightly abducted (less than 90°), he may relax the shoulder girdle. It will now be supported as a structure, with no muscular involvement.

Sometimes shoulder girdle rotation (i.e. upward rotation of the glenoid cavity) does take place. In general this is a small, natural movement, associated with lifting objects with the hands. However in pathological cases of subjects with very weak arms, in applying a force at the hand, many will use "trick movements" of the shoulder girdle. This is done by "stiffening" the arm, and rotating the shoulder girdle. A small force will then be applied at the hand.

Shoulder Girdle Elevation with Upward Rotation of the Glenoid Cavity

In this movement the acromion rises, the superior angle of the scapula descends, and the inferior angle swings laterally. It is almost always part of a larger movement involving either abduction or flexion of the shoulder joint, as when the hand reaches for some object above the head.

The upper part of the trapezius, the levator scapulae and the upper digits of serratus anterior constitute a unit which acts as the upper component of a force couple that rotates the scapula. The lower part of the trapezius, and the lower half or more of the serratus anterior constitute the lower component of the scapular rotatory force couple. They act with increasing vigour through elevation of the arm. The lower part of the trapezius is the more active during abduction, but in flexion is less active than the serratus anterior, apparently because the scapula must be pulled forward during flexion. In general, the middle part of the trapezius serves to fix the scapula, but must relax to allow the scapula to slide forward during the early part of flexion. (Luman, 1944).

Shoulder Joint Abduction

The obvious activity in the deltoid increases progressively, and becomes greatest between 90° and 180° of elevation. The activity of the supraspinatus also increases progressively. Thus it is not simply an initiator of abduction. No part of the pectoralis major is active during abduction. The biceps also do not contribute to abduction when the arm is medially rotated and the forearm prone. (This probably would also apply to the position of the arm on the "hand to mouth" path used in the arm support tests).

Shoulder Joint Flexion

The clavicular head of the pectoralis major, along with the anterior fibres of the deltoid, are the chief flexors, the former reaching its maximum activity at 115° of flexion. Both heads of the biceps brachii are active in flexion of the shoulder joint, the long head being the more active.

Elbow Flexion

Although they can be felt quite easily, the biceps brachii, the brachialis, and the brachioradialis have not been fully understood as far as their integrated functions are concerned. In the movements produced by the flexors, there is a fine interplay between them, and a wide range of response from person to person. There is therefore no unanimity of action. For example, the brachialis is generally active during quick flexion of the supine forearm, but occasionally it is completely inactive. There is also no set pattern in the sequence of appearance and disappearance of activity in these muscles.

There is little difference in the activity of the long head and the short head of the biceps during isometric contraction. The more prone is the forearm, the less the activity of the biceps. The brachialis is also a flexor of the semiprone forearm, and supplies a more constant force than the biceps in all positions of the forearm. The brachioradialis was found to be inactive during isometric contraction, and is only active during quick flexions of the elbow.

Pronator teres contributes to elbow flexion when resistance is offered to the movement.

Wrist Abduction

In abduction, the flexor carpi radialis, and the extensor carpi radialis (longus and brevis) act reciprocally, the antagonistic muscles relaxing. Extensor digitorum contracts during abduction, but this contraction is not limited to the radial part of the muscle, and the flexor digitorum superficialis may be active too. Apparently this type of activity has a synergistic function.

APPENDIX SECTION 5

UNAIDED RISING WITHOUT THE USE OF THE ARMS

Figure (50) shows a subject about to rise in the critical position, with the thigh link horizontal, and having just lost contact with the seat. The body is simply divided into three links: the lower leg link, the thigh link, and the upper body link. The latter includes the arms. The feet are treated as an extension of the floor, with the floor reaction R_4 acting at the ankle joint. A muscular moment, M_K, acts at the knee joint, and a moment, M_H, acts at the hip joint.

The values of M_K, M_H, and β_1 , the angle of the upper body link, are now expressed in terms of β_3, the lower leg angle.

We have, resolving vertically for the whole body,

$$W_T - R_4 = 0 \qquad \text{A.5.1}$$

where W_T is the body weight minus the weight of the feet.

Taking moments about the knee joint for the lower leg, we have,

$$M_K - R_4 e \sin\beta_3 + W_3 \ell_3 \sin\beta_3 = 0 \qquad \text{A.5.2}$$

and taking moments about the hip joint for the lower leg and thigh gives

$$R_4 (d - e\sin\beta_3) - W_3 (d - \ell_3 \sin\beta_3) - W_2 \ell_2 - M_H = 0 \qquad \text{A.5.3}$$

Finally, taking moments about the hip joint for the upper body, we have

$$M_H - W_1 \ell_1 \sin\beta_1 = 0 \qquad \text{A.5.4}$$

Eliminating M_H between A.5.3 and A.5.4, and re-arranging, we have

$$\sin\beta_1 = \frac{(W_T - W_3)\, d - W_2 \ell_2 + (W_3 \ell_3 - W_T e)\sin\beta_3}{W_1 \ell_1} \qquad \text{A.5.5}$$

Eliminating R_4 between A.5.1 and A.5.2, we have,

$$M_K = (W_T e - W_3 \ell_3) \sin \beta_3 \quad \text{..........} \quad \text{A.5.6}$$

and eliminating R_4 between A.5.1 and A.5.3 gives

$$M_H = W_T d - W_3 d - W_2 \ell_2 + (W_3 \ell_3 - W_T e) \sin \beta_3 \quad \text{...} \quad \text{A.5.7}$$

Figures for a fifty percentile subject (as given in the appendix section 10) are now substituted into equations A.5.5, A.5.6 and A.5.7, where β_1, M_K, and M_H are found as a function of β_3. Figure (22) shows these functions plotted over a range of values of β_3.

APPENDIX SECTION 6

ANALYSIS OF UNAIDED RISING WITH USE OF THE ARMS

Figure (23) shows a subject in the critical position to be analysed, about to rise. For purposes of the analysis, the body is divided into seven links.

1. The head, shoulders, and trunk link
2. The thigh link
3. The lower leg link
4. The foot link
5. The upper arm link
6. The forearm link
7. The hand link

Firstly the joint centres of the links are located on the photograph of each subject, and the centreline of each link is marked (with the exception of the foot link which is ignored altogether, and is regarded as an extension of the floor; and the hand link, the centre of gravity of which is assumed to be at the centre of the handgrip).

Using figures given by Dempster (Reference 3), which are reproduced in the appendix (Section 10), the centre of gravity of each link is located. A vertical line XX is now marked, passing for convenience through the ankle joint. The horizontal distance is now measured of the centre of gravity of each body segment (given by Dempster as a proportion of the link length) from the line XX, and these distances are called X_1 to X_7.

The weight of each body segment is now found. It is assumed that the weights will be proportional to those given by Dempster for a fifty percentile subject, corrected for each subject here by multiplying the ratio of the link lengths of the subject to the link lengths of a fifty percentile subject. The calculated segmental weights are finally scaled, to add up to the total body weight of the subject, and are referred to as W_1 to W_7. We wish to find the position of the centre of gravity of the body. This is simply located by taking moments about the line XX for all the body segments. This gives:

$$W_T\bar{\bar{X}} = W_1X_1 + W_2X_2 + W_3X_3 + W_5X_5 + W_6X_6 + W_7X_7 \qquad \ldots\ldots \qquad A.6.1$$

where W_T is the weight of the body minus that of the feet, and $\bar{\bar{X}}$ is the distance of the centre of gravity of the whole body from XX.

The three basic equations for the analysis may now be written. Resolving horizontally for the forces in Figure (23) we have:

$$R_{HH} - R_{4H} = 0 \qquad \ldots\ldots\ldots\ldots\ldots \qquad A.6.2$$

and resolving vertically,

$$R_{HV} + R_{4V} - W_T = 0 \qquad \ldots\ldots\ldots\ldots\ldots \qquad A.6.3$$

Taking moments about the ankle joint A, we have:

$$W_T\bar{\bar{X}} - R_{HV}\,y\sin\phi - R_{HH}\,y\cos\phi = 0 \quad \ldots\ldots \qquad A.6.4$$

In addition to these equations, it is necessary to give a value to one of the variables, in this case the angle of the reaction at the hand, θ_H. This gives as a fourth equation:

$$R_{HH} - R_{HV}\tan\theta_H = 0 \qquad \ldots\ldots\ldots\ldots\ldots \qquad A.6.5$$

Substituting for R_{HH} from equation A.6.5 into equation A.6.3, we now have:

$$W_T\bar{X} - R_{HV}\ (y\sin\phi\ + y\cos\phi\ \tan\theta_H\ ) = 0$$

or $$R_{HV} = W_T\bar{X}\ /\ (y\sin\phi\ + y\cos\phi\ \tan\theta_H\ ) \qquad \ldots\ldots\ldots\ \text{A.6.6}$$

The moment of the muscle groups about each joint is assumed to be equal and opposite to the moments of the forces about that joint. Thus to calculate the moment of the muscle groups, we simply take moments about the joints for all external forces acting. About the knee joint for the lower leg we have,

$$M_K = R_{4V}\,e\,\sin\beta_3\ +\ R_{4H}\,e\cos\beta_3 - W_3\,\ell_3\sin\beta_3$$

$$\ldots\ldots\ \text{A.6.7}$$

Taking moments about the hip joint for the leg and thigh, we have:

$$M_H = R_{4V}\ (d - e\sin\beta_3) - R_{4H}\,e\cos\beta_3 - W_3(d - \ell_3\sin\beta_3\ )$$

$$- W_2\,\ell_2 \qquad \ldots\ldots\ldots\ \text{A.6.8}$$

The links of the upper limb will not necessarily lie in the two dimensional plane of the rest of the body links, in certain cases the plane formed by the arm and forearm links is perpendicular to the plane of the rest of the body links. This is taken into account in the measurement of the angles and lengths of the links of the upper limb.

We can now write the moment at the shoulder joint as:

$$M_S = R_{HV}\,r_s\ \text{Sin}\ \phi_S\ + R_{HH}\,r_s\ \cos\phi_S\ + W_5\,\ell_5\cdot\sin\beta_5 + W_6$$

$$.\ (f\sin\beta_5 - \ell_6\sin\beta_6\ ) \qquad \ldots\ldots\ldots\ldots\ \text{A.6.9}$$

Finally the moment at the elbow joint will be :

$$M_E = R_{HV}\,r_E\ \sin\phi_E\ +\ R_{HH}\,r_E\cos\beta_E - W_6\,\ell_6\sin\beta_6\ \ \ldots\ \text{A.6.10}$$

These equations are now solved on a computer for the eight subjects tested, over a range of values of θ_H, the angle of the handrest force.

A.6.2 The Unaided Rising Computer Program

This is a very simple program in which a large number of lengthy, but straightforward calculations are carried out. At the start of program the data for the first subject is read into the computer. It then calculates the position of the centre of gravity of the subject from equation A.6.1. In the heart of the program is a group of linear equations to be solved. These are within a "DO LOOP" which cycles the value of the handrest force angle (labelled R in the program) over its entire range. The program then goes on to calculate and print out all the data of interest, and is then returned to the beginning where the data is read in for the next subject. The program is written in Fortran and is given at the end of this section. For various reasons, when compiling the program it was necessary to change some of the symbols used in the analysis. These are as follows: (see Figure (23)).

Program	Analysis
X	$\bar{x}$
R	θ_H
TH	ϕ
PI	π
Z_1	θ_H
Z_2	θ_4
A_3	β_3
RS	r_s
PS	ϕ_s
A5	β_5
LE	r_e
A6	β_6
PE	ϕ_E

A.6.3 Results of the Unaided Rising Program

At the end of this section the results are given, as printed out by the computer. The results are selected using the last two columns of figures. The first of these gives the sum of the moments in the four body joints (on both sides of the body, this will in fact be eight joints). The second gives the value of the moment at each joint, divided by the ratio of its inherent strength to the total strength of the subject (as found in chapter 2). The resulting figures are summed for the whole body. This gives a good indication of the best mode for rising; or the mode in which the lowest muscular moments are required, taking into account the inherent strength of each muscle group. Having located this model, the simple sum of the muscular moments only, is used for subsequent comparisons with rising in the chair aid.

THE UNAIDED RISING PROGRAM

```
      READ L,L2,L3,LE,MH,MK,MS,ME
  2   READ (1,101) N
101   FORMAT (I2)
      READ (1,102) WT, D, E,L, VS, VE,A1,A3,X1,X2,X3,X5,X6,X7,TH,Y,RS,LE,
     1   W1,W2,W3,W4,W5,W6,W7,G,F,A5,A6,HS,HE
102   FORMAT (10F7.1)
      READ (1,101) M
      WRITE (2,201)
201   FORMAT (1H ,///)
      PI=3.14159
      L2=D*0.433
      L3=E*0.433
      X=(W1*X1+W2*X2+W3*X3+W5*X5+W6*X6+W7*X7)/WT
      DO 1 I=1,121,2
      R=FLOAT(I-31)
      RHV=WT*X/(Y*SIN(TH*PI/180.0)+Y*TAN(R*PI/180.0)*
     1    COS(TH*PI/180.0))
      RHH=RHV*TAN(R*PI/180.0)
      RFH=RHH
      RFV=WT-RHV
      RH=SQRT(RHV+RHH*RHH)
      R4=SQRT(RFH+RFV*RFV)
      Z1=180.0*ATAN2(RHH,RHV)/PI
      Z2=180.0*ATAN2(RFH,RFV)/PI
      MH=RFV*(D-E*SIN(A3*PI/180.0))-RFH*E*COS(A3*PI/180.0)-
     1   W3*(D-LS*SIN(A3*PI/180.0))-W2*L2
      MK=RFV*E*SIN(A3*PI/180.0)+RFH*E*COS(A3*PI/180.0)-
     1   W3*L3*SIN(A3*PI/180.0)
      MS=RS*(RHV-W7)*SIN(VS*PI/180.0)+RHH*RS*SIN(HS*PI/180.0)-
     1   W5*G*0.436*SIN(A5*PI/180.0)-W6*(G*SIN(A5*PI/180.0)-
     1   F*0.430*SIN(A6*PI/180.0))
      ME=LE*(RHV-W7)*SIN(VE*PI/180.0)+LE*RHH*SIN(HE*PI/180.0)-
     1   W6*F*0.430*SIN(A6*PI/180.0)
      Z3=ABS(MH)+ABS(MK)+ABS(MS)+ABS(ME)
      Z4=ABS(0.241*MH)+ABS(0.324*MK)+ABS(0.820*MS)+ABS(1.075*ME)
      WRITE (2,202) RH,Z1,R4,Z2,MH,MK,MS,ME,Z3,Z4
202   FORMAT (1H0,F6,1,F7,1,F8,1,F7,1,5F7,0,F8,2)
    1 CONTINUE
      IF (M.EQ.0) GO TO 2
      STOP
      END
```

RESULTS OF UNAIDED RISING PROGRAM

R_H (lbf)	θ_H	R_4 (lbf)	θ_4	M_H (in.lbf)	M_K (in.lbf)	M_S (in.lbf)	M_E (in.lbf)	ΣM (in.lbf)	Σ(M x Strength Ratio)
SUBJECT I									
225.7	-18.0°	79.7	-119.0°	-337	-695	-154	716	1900	1201.39
211.3	-16.0°	64.3	-115.0°	-256	-602	-62	704	1625	1064.84
198.9	-14.0°	51.1	-109.5°	-186	-521	19	694	1420	975.39
188.1	-12.0°	39.9	-101.6°	-123	-449	91	685	1348	986.28
178.6	-10.0°	31.0	-89.9°	-66	-384	155	677	1283	996.05
SUBJECT II									
65.1	82.0°	118.1	33.0°	763	677	462	249	~~2152~~	1049.99
66.1	84.0°	120.6	33.0°	790	690	462	235	2177	1045.69
67.3	86.0°	123.2	33.0°	817	703	462	221	2203	1041.25
68.5	88.0°	125.9	33.0°	845	717	461	207	2230	1036.65
69.9	90.0°	128.7	32.9°	874	732	461	192	2258	1031.87

RESULTS OF UNAIDED RISING PROGRAM (Contd)

R_H (lbf)	θ_H	R_4 (lbf)	θ_4	M_H (in.lbf)	M_K (in.lbf)	M_S (in.lbf)	M_E (in.lbf)	$\sum M$ (in.lbf)	$\sum$(M x Strength Ratio)
SUBJECT III									
109.6	14.0°	49.4	32.5°	-117	399	917	211	1645	1136.59
104.9	16.0°	55.3	31.5°	-51	438	907	204	1601	1117.66
100.8	18.0°	60.8	30.8°	9	474	899	197	1580	1104.83
97.0	20.0°	65.8	30.3°	66	507	891	191	1655	1115.84
93.7	22.0°	70.5	29.9°	118	538	884	185	1724	1126.07
SUBJECT IV									
125.9	18.0°	49.3	52.1°	-122	191	802	970	2085	1791.49
123.6	20.0°	54.2	51.3°	-61	194	818	961	2034	1781.83
121.5	22.0°	58.9	50.6°	-2	198	833	953	1986	1772.51
119.6	24.0°	63.4	50.1°	56	201	848	946	2050	1790.38
118.0	26.0°	67.9	49.6°	111	204	862	938	2116	1808.45

RESULTS OF UNAIDED RISING PROGRAM (Contd)

R_H (lbf)	θ_H	R_4	θ_4	M_H (in.lbf)	M_K (in.lbf)	M_S (in.lbf)	M_E (in.lbf)	ΣM (in.lbf)	Σ(Mxx Strength Ratio)
SUBJECT IV									
152.8	4.0°	28.7	21.8°	-185	127	804	607	1723	1397.41
144.2	6.0°	38.7	22.9°	-96	179	800	592	1667	1372.80
136.7	8.0°	47.6	23.5°	-17	226	796	578	1616	1350.78
130.0	10.0°	55.7	23.9°	54	268	792	565	1679	1357.10
124.2	12.0°	63.1	24.2°	119	306	789	554	1768	1370.35
SUBJECT VI									
118.9	-4.0°	8.6	-73.8°	-180	-100	436	169	885	614.97
109.2	-2.0°	12.5	-17.8°	-100	-27	431	182	741	582.00
101.1	0.0°	19.9	0.0°	-33	34	427	193	687	576.16
94.2	2.0°	27.1	7.0°	25	87	423	202	738	598.56
88.2	4.0°	33.5	10.9	76	134	420	210	840	631.97

RESULTS OF UNAIDED RISING PROGRAM (Contd)

R_H (lbf)	θ_H	R_4	θ_4	M_H (in.lbf)	M_K (in.lbf)	M_S (in.lbf)	M_E (in.lbf)	ΣM (in.lbf)	Σ(M x Strength Ratio)
SUBJECT VII									
164.0	-4.0°	19.3	-143.7°	-532	-142	389	547	1610	1080.67
153.5	-2.0°	7.6	-135.4°	-442	-66	423	527	1457	1040.99
144.5	0.0°	3.5	0.0°	-362	1	453	509	1326	1006.73
136.6	2.0°	12.4	22.5°	-291	61	480	494	1325	1014.25
129.7	4.0°	20.7	25.9°	-227	114	504	480	1325	1021.00
SUBJECT VIII									
236.4	-4.0°	29.8	-146.4°	-895	-299	650	546	2390	1432.56
217.0	-2.0°	9.6	-127.9°	-697	-135	704	505	2041	1332.05
200.8	0.0°	10.2	0.0°	-528	4	749	471	1752	1249.04
187.0	2.0°	24.9	15.2°	-383	123	789	441	1736	1253.02
175.2	4.0°	38.2	18.7°	-256	228	823	415	1722	1256.49

APPENDIX SECTION 7

CALIBRATION OF THE CHAIR SEAT DYNAMOMETER

Figure (27) shows a plan view of the position on the seat of the pressure gauges. Gauges F_1 and F_2 are interconnected, and record the force acting along the line F_1 F_2; while B_1 and B_2, also being interconnected, record the force along the line B_1 B_2.

The calibration of the gauges is divided into three sections:

(a) To measure the response of the gauges as a point load is moved along the lines F_1 F_2 and B_1 B_2

(b) To measure the response of the gauges as a point load is moved along the line FB

(c) To measure the response of the gauges for a range of values of a point load acting at B, and at F.

(a) Response of the Gauges across the Seat

This is to examine whether the gauge readings are consistent across the width of the seat, as a precaution against false readings caused by the subject sitting eccentrically on the seat. As a preliminary test, a subject was seated on the chair, and was asked to sit as far to one side as possible, while still remaining comfortable. It was found to be fairly noticeable when the subject was 1 inch off the centre line, and the maximum shift possible was found to be 4 inches off the centre line.

A point load of 75 lbf. was applied to the seat at positions along F_1 F_2 and B_1 B_2. Figure (51) is a graph of the results. It can be seen that there is no significant variation in the gauge readings as the load is applied up to 5 inches eccentrically.

(b) The Response of the Dynamometer along the Seat

The chair seat can be regarded in two dimensions as a rigid beam supported at two points along its length; with a resultant point load acting at a third point along its length. As the point load moves from one support to the other (from F to B), the magnitude of the support forces in gauges F and B should theoretically vary according to the laws of simple statics. Measurements were carried out to verify this.

A point load of 75 lbf. was applied to the centre line of the seat at positions along the line FB. Gauge readings were taken, and using the calibration curves for the gauges (Figure 52)), the readings were plotted as values of the support force at each gauge, from which the magnitude and position of the load were calculated and compared to their true values. From these results it can be said that an applied load of 75 lbf. can be measured to an accuracy of ± 0.8 lbf. and its position can be determined to ± 0.2 ins.

(c) The Dynamometer Calibration

A point load was applied to the chair seat, firstly at B, and then at F. The load was increased from 0 to 125 lbf. in increments, and then back to zero. Figure (52) shows the calibration curves. These are used throughout the subsequent experiments in the form shown.

APPENDIX SECTION 8

A.8.1 A Biomechanical Analysis of the Modes of Rising on the Chair Aid

There are basically four distinct modes of rising on the chair aid. The following is an analysis of the biomechanics of the modes. Throughout the analysis, the body is simplified to a two-dimensional chain of rigid links. Muscle forces are assumed to cause a torque about their respective joint centres. In each chair position analysed, the mechanism is correctly adjusted, and when the subject is relaxed, the torque about the seat hinge is zero. In the analysis, the equilibrium forces throughout the system are ignored. Only additional forces and torques, due to the disturbance of the equilibrium by muscle forces applied by the subject, are analysed.

A.8.2 Application of a Knee Extension Moment

Figure (53) shows the force system generated as a muscular moment is applied at the knee. We are interested in finding the applied muscular moment in terms of the measurable forces between the subject and the chair, and in terms of the geometry of the system. We are also interested in the seat hinge torque, generated by these muscular moments. There are two unknowns in the force system to be resolved before any equations can be written. Firstly the position of the footrest force, and secondly the point of action of the resultant of the force parallel to the chair seat. Firstly the resultant force at the footrest is assumed to pass through the ankle joint. This is the same assumption as was made in the unaided rising analysis, and it eliminates muscular moments at the ankle joint. The components of force parallel to the chair seat is assumed for large values of α_2 to be supplied by the lower part of the backrest, and for smaller values of α_2 by the knee-rest. (It can be easily seen in practice, whether the knee-rest is acting during a test).

In carrying out the chair tests, the subject is asked to lean forward slightly from the backrest. This will eliminate any backrest forces, apart from the force at the buttocks). Any hip moment will be applied by moving the upper body slightly forward. Only a small movement is required here for the small hip moments generated in this mode. This will tend to shift the seat reaction R_2 slightly forward.

From equation 5.1 in section 5.4.5 of the text, we have the seat hinge torque given by,

$$\Omega = \Delta_2 (AU) - \Delta_1 \left((AD) \sin(\alpha_2 - \theta_1) + L_1 \sin(\theta_1 - \alpha_1) \frac{d\alpha_1}{d\alpha_2}\right)$$

$$+ \Delta_4 \left(\Lambda_{FII} . (JF) + \Lambda_{FIII} (IF)\right) + \Delta_5 \Lambda_{FI} \left[+ \Delta_3 . \Lambda_K . \sin(ZKL)\right] \quad \ldots\ldots\ldots\ldots\ldots \text{A.8.1}$$

where the dimensions and angles are given in Figure (31); and the values for the mechanism geometry functions, Λ, are given in Figure (30).

The moments of the muscle groups about the body link joints are assumed to be equal and opposite to the moments of the forces about the joints. For the knee joint, summing the forces in the lower leg, we have,

$$M_K = \Delta_4 L_{K4} + \Delta_5 L_{K5} \quad \ldots\ldots\ldots\ldots \text{A.8.2}$$

where L_{K4} and L_{K5} are the perpendicular distances of their respective forces to the knee joint; and M_K is the moment in knee extension.

Similarly for the hip joint,

$$M_H = \Delta_2 L_{H2} + \Delta_5 L_{H5} - \Delta_4 L_{H4} \quad \ldots\ldots \text{A.8.3}$$

where L_{H2}, L_{H5}, and L_{H4} are the perpendicular distances of the forces from the joint, and M_H is the hip flexion moment.

A.8.3 Application of a Hip Extension Moment using the Handrest

Figure (54) shows the force system generated as a muscular moment is applied at the hip. In testing subjects, it is ensured that the main muscular forces will be applied at the hip by removing the tangential component of the footrest force Δ_5, simply by removing the footrest tangential force dynamometer. (This was only done in testing healthy subjects, due to the difficulties and dangers of applying a force like this. Disabled subjects were just asked to press straight down with the feet, and any measurable value of Δ_5 was included in the calculations).

In the system there are two unknown parameters: Δ_7 and θ_H. These are found graphically, by drawing a force triangle Δ_2, Δ_4, and Δ_7 (or a force quadrilateral if Δ_5 is acting).

We can now write an equation for the seat hinge torque generated by this mode,

$$\Omega = \Delta_2 (AU) + \Delta_4 (\Lambda_{FII} + \Lambda_{FIII} \cdot (JF)) - \Delta_7 (\Lambda_{HI} \cdot \sin (XPR) - \Lambda_{HII} \cdot \text{Sin} (XPQ)) \quad \left[+ \Delta_5 \Lambda_{FI}\right] \quad \ldots\ldots \quad A.8.4$$

and can write expressions for the moments about the body link joints: about the knee joint for knee extension:-

$$M_K = \Delta_4 L_{K4} \quad \left[+ \Delta_5 L_{K5}\right] \quad \ldots\ldots\ldots \quad A.8.5$$

and about the hip joint for hip extension:-

$$M_H = \Delta_4 L_{H4} - \Delta_2 L_{H2} \quad \left[- \Delta_5 L_{H5}\right] \quad \ldots \quad A.8.6$$

In chapter 6, section 6.2.1 of the text, an explanation is given for the need to find only the muscular moments at the hip and knee joints, and not in the upper limbs or in the trunk.

A.8.4 Application of a Hip Extension Moment using the Backrest

This mode (figure 55) is similar to the previous mode, except that the moment applied at the hip is balanced for the upper body by forces on the backrest, instead of at the hands. The one difference is that the position of the force is at present unknown. This can be found fairly easily.

As before the magnitude and direction of the force is found graphically. The position of the force at the backrest is found by taking moments about the hip joint for forces acting on the subject.

This gives,

$$\Delta_1 \; L_{H1} + \Delta_4 \; L_{H4} - \Delta_2 \, L_{H2} = 0 \qquad \ldots\ldots \text{ A.8.7}$$

Knowing the magnitude and direction of the force Δ_1, the length L_{H1}, the position of the force is easily located.

We can now write an equation to give the magnitude of the seat hinge torque generated in this mode:

$$\Omega = \Delta_2 \,(AU) - \Delta_1 \left(d \sin (\alpha_2 - \theta_1) - L_1 \sin (\theta_1 - \alpha_1) \; \frac{d\alpha_1}{d\alpha_2} \right)$$

$$+ \Delta_4 \left(\Lambda_{FII} \,(JF) + \Lambda_{FIII} \,(1F) \right) \qquad \left[+ \; \Delta_5 \; \Lambda_{FI} \right]$$

$$\ldots\ldots\ldots \text{ A.8.8}$$

and in the same manner as with the previous mode, the moments at the knee and hip joints will be given by:

$$M_K = \Delta_4 \, L_{K4} \qquad \left[+ \Delta_5 \, L_{K5} \right] \qquad \ldots\ldots\ldots \text{ A.8.9}$$

where M_K is the moment in knee extension, and $M_H = \Delta_4 L_{H4} - \Delta_2 L_{H2}$

$$\left[- \Delta_5 L_{H5}\right] \qquad \text{........ A.8.10}$$

where M_H is the moment in hip extension.

A.8.5 Upper Body Flexion (Figure 56)

Analytically this is the simplest mode, in which the subject flexes the hip joint, and the spine, to tilt the upper body forward. The main effect of this mode on the kinematics of the system will be to move the seat reaction forwards. A secondary effect will be to generate forces Δ_4 and Δ_5 on the footrest.

The seat hinge torque generated by this mode will be;

$$\Omega = \Delta_2 (A \cup) + \Delta_4 (\Lambda_{FII} (JF) + \Lambda_{FIII} (1F)) + \Delta_5 \Lambda_{FI} \qquad \text{......... A.8.11}$$

and the muscular moment required at the knee will be given by:

$$M_K = \Delta_4 L_{K4} + \Delta_5 L_{K5} \qquad \text{............ A.8.12}$$

where M_K is the knee extension moment.

Lastly we have,

$$M_H = \Delta_4 L_{H4} - \Delta_2 L_{H2} - \Delta_5 L_{H5} \qquad \text{.......... A.8.13}$$

where M_H is the hip extension moment.

APPENDIX SECTION 9

A.9.1 Calculations and Results of the Preliminary Chair Aid Tests

Preliminary tests were carried out on healthy subjects to verify the biomechanical theory given in the previous section of the appendix. Four modes of rising are analysed in section 8 of the appendix; and in this section, calculations are performed on one set of results for each mode to give values for the seat hinge torque generated, and the muscular moments applied at the hip and knee joints.

A.9.2 Knee Extension (Figure 53)

From section 8 of the appendix, the seat hinge torque generated is given by

$$\Omega = \Delta_2 (AU) - \Delta_1 (d \sin(\alpha_2 - \theta_1) - L_1 \sin(\theta_1 - \alpha_1) \frac{d\alpha_1}{d\alpha_2}$$

$$+ \Delta_4 (\Lambda_{FII} . (JF) + \Lambda_{FIII} . (IF)) + \Delta_5 \Lambda_{FI} \qquad \ldots\ldots\ \text{A.8.1}$$

We can now use figures obtained in the preliminary tests. When the subject is sitting relaxed in the chair, we have the normal force on the seat,

$$R_2 = 108.8 \text{ lbf}$$

also the force normal to the footrest,

$$R_4 = 32 \text{ lbf}$$

As a knee extension moment is applied, the following measurements are taken from the photographic record of the tests:

$\alpha_2 = 85.5^\circ$	$AU = 10.4$ ins.	$R_4 = 91$ lbf
$JF = 2.6$ ins	$IF = 5.4$ ins.	$R_5 = 25.0$ lbf

Subtracting the forces from those measured in the relaxed position, we have:

$\Delta_2 = 55.6$ lbf $\quad \Delta_4 = 59.0$ lbf and $\Delta_5 = 25.0$ lbf.

A vector quadrilateral is now constructed of sides Δ_2, Δ_4 and Δ_5. The fourth side will give the magnitude and direction of Δ_1 ($\Delta_1 = 38.8$ lbf).

We can now find $\alpha_2 - \theta_1$ (= 6.9°) and $\theta_1 - \alpha_1$ (= 86.1°) and TD (= 3.4 ins). From figure (30), values can be found for $\frac{d\alpha_1}{d\alpha_2}$ (= 0.685), Λ_{FI} (= 0.70), Λ_{FII} (= 0.41), and Λ_{FIII} (= 0.44).

These numerical values are now substituted into equation A.8.1, giving

$$\Omega = 55.6 \times 10.4 - 38.8\,(18.0 \sin (6.9) + 3.4 \times 0.685)$$
$$+ 59\,(0.41 \times 2.6 + 0.44 \times 5.4) + 25 \times 0.70$$
$$= 576 - 174 + 204 + 17$$
$$= \underline{623 \text{ in.lbf.}}$$

This compares with the value of the seat hinge torque, measured directly, of 630 in.lbf.

The perpendicular distances from the knee joint to the lines of action of the forces Δ_4 and Δ_5 are now measured to find the muscular moment exerted at the knee joint.

$$M_K = \Delta_4 \times 1.9 + \Delta_5 \times 18.2$$
$$= \underline{567 \text{ in.lbf.}} \text{ in knee extension}$$

and the perpendicular distances from the hip joint for the forces Δ_4, Δ_5, and Δ_2 are measured to find the muscular moment exerted at the hip.

$$M_H = \Delta_4 \times 16.9 - \Delta_5 \times 22.1 - \Delta_2 \times 5.35$$
$$= \underline{149 \text{ in.lbf}} \text{ in hip extension}$$

A.9.3 Hip Extension using the Handrest (Figure 54)

As in the previous section, measurements are taken from the photographic record of the test. These give:

$\alpha_2 = 85.5^{\circ}$ $\qquad R_2 = 24.9$ lbf $\qquad AU = 11.8$ ins. $\qquad R_4 = 98$ lbf.

$JF = 5.3$ ins. $\qquad IF = 2.7$ ins. $\qquad R_5 = 0$ lbf.

and subtracting the forces from those measured in the relaxed position, we have

$$\Delta_2 = 83.9 \text{ lbf} , \quad \text{and} \quad \Delta_4 = 66 \text{ lbf}$$

A force triangle is now constructed to find the force vector at the handrest. This gives:

$$\Delta_7 = 27.8 \text{ lbf}$$

We also have $\angle XPR = -17.5^{\circ}$, $\angle XPQ = 18.2^{\circ}$ $\quad \Lambda_{FII} = 0.41$,

$\Lambda_{FIII} = 0.44$, $\Lambda_{HI} = 23.4$ and $\Lambda_{HII} = 27.8$.

Substituting these figures into equation A.8.4, we have,

$$\begin{aligned} \Omega &= 83.9 \times 11.8 + 66\ (5.3 \times 0.41 + 2.7 \times 0.44) \\ &\quad - 27.8\ (23.4 \sin (18.2^{\circ}) + 27.8 \sin (17.5^{\circ})\ ) \\ &= 989 + 222 - 437 \\ &= \underline{774 \text{ in.lbf.}} \end{aligned}$$

This compares with a directly measured value of 985 in.lbf.

The muscular moment exerted at the knee is given by:

$$\begin{aligned} M_K &= \Delta_4 \times 5.3 \\ &= \underline{350 \text{ in.lbf.}} \text{ in knee extension;} \end{aligned}$$

and the muscular moment at the hip joint is given by:

$$\begin{aligned} M_H &= \Delta_4 \times 13.6 - \Delta_2 \times 3.4 \\ &= 895 - 285 \\ &= \underline{610 \text{ in.lbf.}} \quad \text{in hip extension.} \end{aligned}$$

A.9.4 Hip Extension using the Backrest (Figure (55))

As in the previous two sections, measurements are taken from the photographic record of the test. These give

$\alpha_2 = 59.5^\circ$, $\Delta_2 = 112.3$ lbf, AU = 10.6 ins,

$\Delta_4 = 100$ lbf. JF = 9.9 ins, IF = 1.9 ins, and $\Delta_5 = 25.0$ lbf.

A force quadrilateral is constructed from the forces Δ_4, Δ_5, and Δ_2; the fourth side of the quadrilateral gives Δ_1 in magnitude and direction.

The point of action of Δ_1 on the backrest is critical, and is found by taking moments about the hip joint for the four forces acting on the subject. This gives L_{H1}, the distance of the force from the hip joint along the centre line of the trunk link.

$$L_{H1} \times \Delta_1 \sin 87.9^\circ - \Delta_4 \times 9.61 + \Delta_2 \times 5.25 = 0$$

$$\text{giving } L_{H1} = 5.52 \text{ ins.}$$

By constructing the force vector at the appropriate position on the photograph, TD is found.

$$TD = 7.28 \text{ ins.}$$

From figure (30), values are found for $\dfrac{d\alpha_1}{d\alpha_2}$ (= 0.27),

Λ_{FII} (= 0.36) and Λ_{FIII} (= 0.41)

These numerical figures are now substituted into equation A.8.8, giving

$$\Omega = 112.3 \times 10.6 + 67.2\ (18 \sin (29^\circ) + 7.28 \sin (88.6) \times 0.27)$$
$$+ 100\ (9.9 \times 0.36 - 1.9 \times 0.41)$$
$$= 1190 - 717 + 278$$
$$= \underline{751 \text{ in.lbf}}.$$

This compares with a directly measured value of 683 in.lbf.

The muscular moment exerted at the knee is given by:

$$M_K = \Delta_4 \times 5.45$$

$$= \underline{545 \text{ in.lbf.}} \text{ in knee extension}$$

and $$M_H = \Delta_4 \times 9.61 - \Delta_2 \times 5.25$$

$$= \underline{371 \text{ in.lbf.}} \text{ in hip extension}$$

A.9.5 Upper Body Flexion (Figure 56)

We have for this mode, from the photographic record of the test;

$\alpha_2 = 85.5^\circ$, $\Delta_2 = 37.7$ lbf., AU = 17.8 ins.

$\Delta_4 = 37$ lbf. JF = 7.2 ins , IF = 0.8 ins.

$\Lambda_{FII} = 0.41$ and $\Lambda_{FIII} = 0.44$

Substituting these values into equation A.8.11, we have:

$$\Omega = 37.7 \times 17.8 + 37 \, (0.41 \times 7.2 + 0.44 \times 0.8)$$

$$= 672 + 122$$

$$= \underline{794 \text{ in.lbf.}}$$

This compares with a directly measured value of 876 in.lbf.

The muscular moment exerted at the knee is given by:

$$M_K = \Delta_4 \times 6.4$$

$$= \underline{237 \text{ in.lbf.}} \text{ in knee extension,}$$

and at the hip,

$$M_H = \Delta_4 \times 10.6 - \Delta_2 \times (-4\tfrac{1}{2}8)$$

$$= \underline{573 \text{ in.lbf.}} \text{ in hip extension}$$

A.9.6 Results of the Preliminary Tests

Figure (42) shows the results of the preliminary chair tests; given as the measured value of the seat hinge torque, plotted against the calculated value for the different modes (as given by the key).

There is a standard deviation of 17.2% between the measured and the calculated results, but errors of this magnitude are to be expected in a complex system involving the human body such as this. It would thus appear that the results verify the biomechanical theory used in analysing the subject/chair aid interaction.

The values of the muscular moments exerted at the knee and hip joint are plotted against the measured value of the seat hinge torque generated by them, in figure (43).

APPENDIX SECTION 10

PHYSICAL DATA ON THE FIFTY PERCENTILE SUBJECT

The data on the fifty percentile subject (a subject of average height, weight, and build) used throughout this thesis is based on the findings of W.T. Dempster (reference3~~7~~4). Those parts of the report used are:

(a) The mass of the body parts expressed as a percentage of the body weight

Trunk, head, and neck	...	61.3%
Upper arms	...	5.0%
Forearms	...	2.9%
Hands	...	1.1%
Thighs	...	18.4%
Lower legs	...	8.6%
Feet	...	2.7%

(b) The body link lengths

Hip to apex of head	...	33.0 ins
Upper arm link	...	11.9 ins
Forearm link	...	10.7 ins
Thigh link	...	17.1 ins
Lower leg link	...	16.1 ins

(c) The position of the centre of gravity of each link, expressed as a percentage of its length

Head, neck, and trunk	...	39.6%
Upper arm	...	43.6%
Forearm	...	43.0%
Thigh	...	43.3%
Lower leg	...	43.3%

(d) <u>The average body weight</u>

131.5 lbf.

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