A MODEL OF MAXIMAL ISOMETRIC FLEXOR TORQUE IN THE HUMAN ELBOW

by

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A Model of Maximal Isometric Flexor Torque in the Human Elbow

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ABSTRACT

The external application of muscular force is governed by a combination of muscular, anatomic and volitional factors. In this thesis a model of the human elbow flexors is developed using data obtained from a cadaver, and the validity of the model is tested with empirical data gathered from six elite athletes.

During dissection the measurements made on Biceps Brachii, Brachialis and Brachioradialis during full excursion of the elbow joint were effective cross-section, change of length, angle of insertion, and length of lever arm. This data allowed changes in theoretical isometric flexor torque to be produced. Six elite athletes generated curves of maximal isometric flexor torque about the elbow through a joint excursion of 105°, in both abducted and flexed positions of the shoulder.

Comparison of theoretical and actual data suggests that the model is grossly valid in terms of locus of maximal isometric flexor torque, the order of magnitude of torque predicted and the inflexion of the torque/angle of flexion curve at small angles of elbow flexion. Furthermore, the experimental findings suggest that reducing gross biomechanics of elbow flexion to the action of a single flexor acting throughout the excursion of the joint is not justifiable. The empirical curves demonstrate a wide inter-individual variation which is partly accounted for by the nature of the sport undertaken by the individuals.

Consistent differences are found between the theoretical and the empirical curves. The theoretical curves predict a lower maximum torque, a larger angle of elbow flexion at which maximum torque is generated, and predict too-rapid a diminution in isometric flexor torque with increasing elbow extension. These differences may be accounted for, in part, by the necessary simplifications of the model resulting from constraints in the experimental design. These are approximations of effective cross-section due to the shape of Brachialis; the exclusion of series elastic tissue compliance from the model; and the inclusion of only three flexors in the prediction of torque.

It would appear from the consistent differences found between theoretical and empirical curves that, despite the gross validation given the model in this experiment, length change of the three major flexors is not sufficient to account for *in vivo* performance. Consequently there are other factors undefined by the model which contribute in important measure to the production of maximal isometric flexor torque at the human elbow.

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Especially I thank Dr. A. E. Chapman, my Senior Supervisor, for his advice and his patience throughout the long execution of this thesis.

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INTRODUCTION

The force produced by a contracting muscle has been shown to be dependent on a number of factors. In particular, Gordon, Huxley and Julian (1966) have shown that the maximal isometric contractile force of muscle is a function of sarcomere length. Bahler, Fales and Zierler (1968) have shown that the force of contraction is a function of both length and velocity of either shortening or lengthening. Fidelus (1968) has published data on the relative contributions to force by the contractile tissue and the parallel elastic tissue at various lengths of muscle around an optimal length, L_o. Ikai and Fukunaga (1968) demonstrated that maximal isometric tension of the contractile component is related to the muscle cross-sectional area.

Force is also a function of volitional factors such as the number of active motor units and the frequency of impulses sent to them. Chapman (1975) observed that human beings could develop curiously high flexor torques at small angles of elbow flexion (near to full extension) and that this occurred despite an apparent mechanical disadvantage of narrow angles of flexor insertion on the forearm. Furthermore, unpublished data gathered by Noble and Chapman (1974) show that there are features of anatomy in the elbow of the Rhesus monkey which impose particular constraints on the length of the elbow flexors at various angles of elbow flexion. Consequently, the external application of force is governed by a combination of muscular, anatomic, and volitional factors.

This thesis attempts to predict maximal isometric flexor torque at various angles of elbow flexion in the human. It uses the parameters previously investigated by experiments on the linear performance of muscle and, formulating a model of maximal isometric flexor torque about the human elbow, uses direct measurements from a cadaver to generate a theoretical torque/angle of flexion curve. The theoretical curve is compared with empirical data gathered from normal living subjects. It is hoped that the experiment will contribute to an understanding of the torque/angle of flexion curve _in vivo_ - some knowledge of the biomechanical constraints on torque in humans is fundamental to _in vivo_ tests of muscle models. These, in turn, "form the basis for critical examination of hypotheses concerning muscle contractions" (Bahler et al, 1968). The experiment may also give an indication of the relative importance of roles played by endowment and training in performance. Third, the experiment will afford an assessment of some of the techniques used here in gathering and manipulating data for the prediction of torque _in vivo_.

DEVELOPMENT of the MODEL

A simple model of torque exerted at a skeletal joint is illustrated in Figure 1. If AC is fixed and segment AB is free to rotate about point A in the plane of the paper, then the torque, τ, generated by a muscle inserting at D will be

$$F \cdot \sin\beta \cdot r$$

Given the model in Figure 1, an expression for instantaneous flexing torque can be developed as follows:

$$\tau = F \cdot \sin\beta \cdot r \text{ ——————— (1)}$$

$$\text{and } F = F_{CT} + F_{PET} \text{ ———— (2)}$$

where CT is the contractile tissue
PET is the parallel elastic tissue

Now the maximal force exerted by the contractile tissue has been shown to be a function of the length of the contractile tissue (Gordon, Huxley and Julian, 1966) and of its cross section (Wells, 1965). The force exerted by the parallel elastic component has also been shown to be a function of length (Fidelus, 1968). Therefore, we may re-write equation number 1 as follows:

$$\tau = F\left(L/L_0, X\text{-section}\right) \cdot \sin\beta \cdot r \text{ ——— (3)}$$

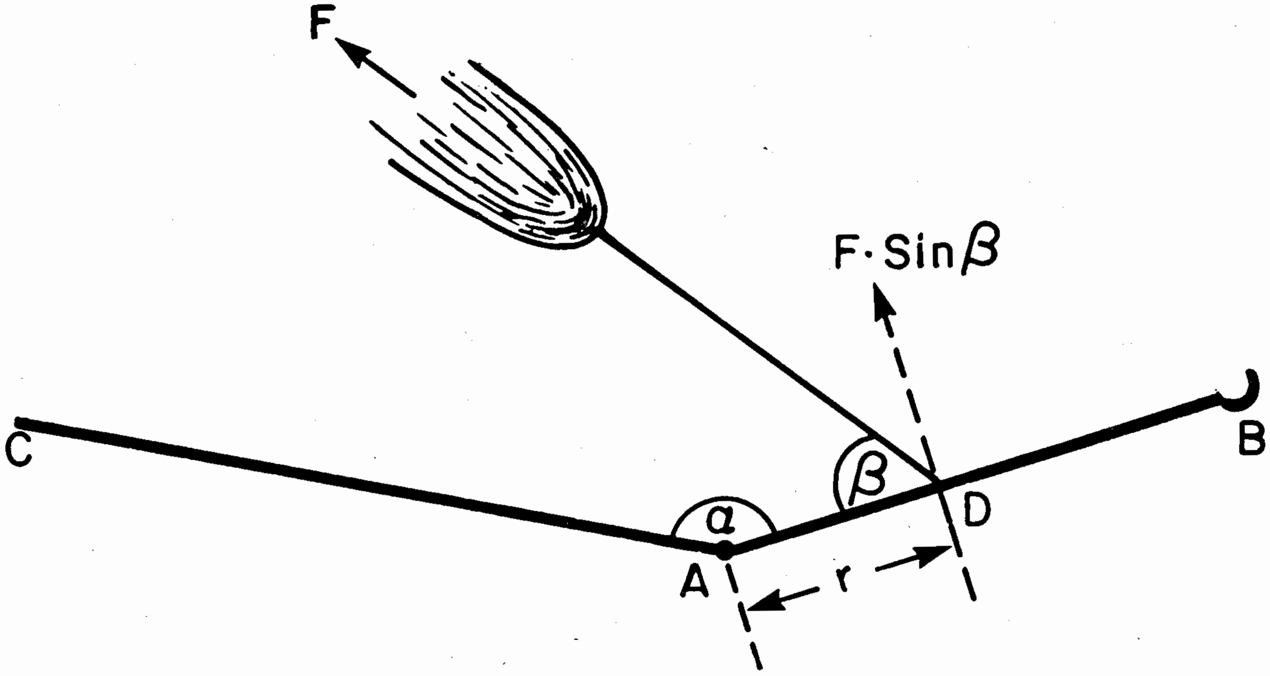


Figure 1. A model of torque exerted at a skeletal joint.

Where there is more than one muscle contributing to flexion, the total flexing torque generated at any angle of the joint, α, may be written as:

$$\text{Total } \hat{\tau}_{\alpha} = \sum F_{i_{\alpha}}\left(\left(L/L_0\right)_{i_{\alpha}}, X\text{-section}_{i_{L_0}}\right)\cdot \sin\beta_{i_{\alpha}} \cdot r_{i_{\alpha}} \qquad \text{—— (4)}$$

where i = Biceps Brachii, long and short heads

Brachialis

Brachioradialis

To simplify the model as much as possible, maximum isometric contraction was used - this eliminated the effects of fiber type, friction and inertia; it was expected that the volitional aspects of performance would be most reliable in maximum efforts by trained athletes.

There were three reasons for selecting the elbow for this study. First, when the forearm is supinated, the centre of rotation of the elbow in flexion-extension is fixed between 0 and 125 degrees of flexion (Fischer, 1911; Hill, 1950; Wilkie, 1950). Second, the shoulder and the wrist are easily fixed and the elbow flexors obliged to shorten across the elbow joint only. Third, it has been shown that the major elbow flexors contract approximately in the line of the humerus in the supinated forearm (Braune and Fischer, 1890).

There is agreement that the major elbow flexors are the

Biceps Brachii, Capita Longus and Brevis, and the Brachialis (MacGregor, 1950). Brachioradialis has been shown to be active during production of flexor torque, even with the forearm supinated, and so it was included in this study (Basmajian and Latif, 1957). In maximal loading, Pronator Teres is thought to contribute to flexor torque but some studies have shown this contribution to be very small (Fidelus,1968), and so a contribution from Pronator Teres was not included in the mathematical prediction of torque. The role of other flexors (namely, Flexor Carpi Ulnaris, Palmaris Longus, Flexor Carpi Radialis, and Extensor Carpi Radialis Longus) is controversial: Some authors ignore them completely (Fidelus, 1968) while others cautiously ascribe 85 percent of the flexing torque to the three main flexors, Biceps Brachii, Brachialis, and Brachioradialis (Bouisset, 1975). Because it was thought that their contribution would be small at most, this paper did not include the contribution of these other flexors to flexor torque at the elbow.

The measurement of the parameters in Equation 3, the collection of data from living subjects and the methods used to compare the theoretical and the empirical torque/angle of flexion curve is discussed in the chapter "Materials and Methods".

REVIEW OF LITERATURE

Hill's paper, "The Maximum Work and Mechanical Efficiency of Human Muscles, and Their Most Economical Speed" (1922), introduced the concept of muscle as a system with both elastic and viscous properties. The expression of these properties in muscle performance became a focus of research. Summarizing the findings of that work, Hill observed in 1938 that:

1. Active muscle contains an undamped elastic element;
2. Active muscle contains an apparently damped element, in series with the undamped elastic one;
3. Resting muscle contains the elastic element (1) but only to a minor degree the apparently damped element (2).

Element (1) has come to be called the series elastic component; element (2) the contractile component. Now, when muscle performance is analyzed over a wide range of lengths, it is apparent that the muscular system, both resting and active, contains a second elastic element acting as if in parallel with elements (1) and (2). This third element is conventionally called the parallel elastic component and its influence on the muscular system is really only apparent in performance at long lengths of the muscular structure. Pringle notes, for example, that the "so-called parallel elastic component...in Frog Sartorius muscle is negligible at or below the muscle length in the body" (1960).

In the last decades each of these functional components of muscle and their integration in muscular work have been widely studied.

Time and again researchers have confirmed that the contractile component behaves according to a "characteristic equation" relating force of contraction, F , and velocity of shortening, V:

$$(F + a)(V + b) = (F_{max} + a)b \qquad \text{(Hill, 1938).}$$

Of particular relevance to the present paper is Gordon, Huxley and Julian's publication, "Variation in Isometric Tension with Sarcomere Length in Vertebrate Muscle Fibres" (1966). Bahler, Fales, and Zierler (1968) published more complex findings about the interrelationship of sarcomere length, maximum contractile force and velocity of shortening in their paper, "The Dynamic Properties of Mammalian Skeletal Muscle". The results of both of these papaers as they relate to the present work are presented below.

If the contractile component of muscle has been widely investigated, so have the properties of the series elastic component. Jewell and Wilkie (1958) investigated the compliance of the series elastic component; over this there has been little concensus. Normal physiologic strain of the series elastic component under maximum isometric contraction of the contractile tissues has been variably placed between 3% of the muscle's resting length (Benedict et al., 1968; Cavagna, 1970) and 10-15% (Wilkie, 1950). Compliance of the series elastic tissues is also of importance to the present paper. Series elastic tissue is not only tendonous but is also distributed in the body of the contractile tissue, some indeed residing in the sarcomeres. This precludes anatomical identification of the series elasticity and, therefore, subsequent correction for the extent to which it modifies contractile component length (Huxley and Simmons, 1971).

Stolov and Weilepp (1966) suggest that the third functional component of the muscular structure, the parallel elastic component, is in fact a composite of the contributions from six anatomic elements including adhesions to neighbouring structures. The role of the parallel elastic component in muscular performance was examined initially by Ramsey et al. (1940) and more recently by Fidelus (1968) and by Hayes et al. (1977). The relevance of their work to the present paper is discussed in the final chapter, "Discussion".

The foregoing research on the components and the integrated function of muscular structure has been conducted _in vitro_. The end point of muscle research must be the understanding of muscular performance _in vivo_. The translation of _in vitro_ findings to an explanation of performance in the live human saddles research with still more complexities and a clear understanding of muscular work _in vivo_ has been difficult to obtain. Apart from the ethics of human experimentation, the control of physiological conditions _in vivo_ is not possible to the degree attained _in vitro_. Researchers have therefore tried other methods to quantify internal work and integrated electromyography has been widely used in attempting to assess the biomechanics of muscular work in vivo (Inman et al., 1952; Messier et al., 1969). Grieve and Pheasant (1976) sound a note of caution here in observing that, "maximum exertable tension...despite its importance, does not account for all the effects of posture upon the EMG-isometric torque relationships". Further on in the same paper they suggest that "muscles may be acting synergistically or antagonistically without contributing to the recorded electrical activity....Caution is hence required when judging stretched muscles to be 'silent'".

While electromyography has struggled with the incongruence of measured myographic potentials and the force output of the muscles under study, some researchers have attempted to simplify the conditions affecting torque by invoking "equivalent flexors". Cnockaert et al. (1975) observe,

> In the case of elbow flexion, the torque is the result of the contraction of the five muscles which form the flexor group. Two muscles are located in the upper arm: brachialis and biceps brachii; the others are located in the forearm: brachioradialis, pronator teres and extensor carpi radialis longus....
>
> We put forward that: (i) biceps brachii is representative of the flexor muscles located in the arm; (ii) brachioradialis is representative of the flexor muscles located in the forearm.

Bouisset is more emphatic (1973):

> ...it is postulated that the torque exerted by the muscle which is being considered as the muscle equivalent remains in a constant relationship with the external torque, measured peripherally, throughout the activity under consideration: that is to say, that its relationship with the resultant of the torques exerted at the point of measurement by all the muscles in this group is constant.
>
> It is difficult to over-emphasize the importance of this assumption, which is regarded as correct and confirmed in virtually all studies on the matter.

Chapman (1973) suggested that the notion of an equivalent elbow flexor acting throughout the excursion of the joint was inconsistent with the observation that individuals could generate curiously high flexor torques at small angles of elbow flexion. This thesis will help to clarify the suitability of choosing equivalent flexors in modelling maximal isometric flexor torque throughout the excursion of the joint.

Acknowledging the difficulties incurred by electromyography in attempting to quantify muscular performance *in vivo*, the present paper attempts to model maximal isometric elbow flexor torque *in vivo* by taking direct measurements on the body (albeit a cadaver) and generating a torque/angle of flexion curve which could be compared with a real execution of the same movement. If the modelled curve resembles the empirical curve generated by live subjects, one could infer that the model and the measurements taken for it were reasonably accurate. This would corroborate the role of the various parameters influencing muscular performance in vivo. A second intention of the present experiment was to account for the curiously high flexor torques generated at small angles of flexion in vivo, noticed by Chapman and described above.

Maximal isometric flexion of the elbow was selected for modelling in order to simplify the experimental conditions as much as possible. In maximal isometric contraction, Wells (1965) demonstrated that the force output per unit cross-sectional area of contractile tissue is independent of fibre type. Furthermore, Ikai and Fukunaga (1968) have estimated that the maximal contractile force of muscle acting at L_o is approximately 4.3 kg-f/cm^2. By selecting interested, trained athletes to generate the curves of maximal isometric torque/angle of elbow flexion, it was hoped that the curves would reflect true maxima at each angle of elbow flexion. The elbow itself was selected for the fixity of its centre of rotation throughout the excursion of the joint. This has been demonstrated by numerous workers including Braune and Fischer (1890), Ficke (1911), and most recently by Morrey et al. (1976).

A question arose about the flexors which ought to be included in the model of torque about the elbow. There is general agreement that the major flexors of the elbow are Biceps Brachii, Brachialis, and Brachioradialis (Davies, 1967). However, Pronator Teres and Extensor Carpi Radialis Longus also pass anterior to the centre of rotation of the elbow. Indeed, Wilkie (1950) considered all five of these muscles significant flexors of the elbow. Basmajian (1961) and Olson et al. (1967) demonstrated that Pronator Teres contributed to supinated elbow flexion only when the movement was strongly resisted; Fidelus (1968) presented data on the moments generated by several flexors at the elbow and concluded that the role of Pronator Teres was constant and relatively insignificant throughout the excursion of the joint. There is little published work about the quantitative role of Extensor Carpi Radialis Longus in supinated elbow flexion. That the three main flexors Biceps Brachii, Brachialis, and Brachioradialis do not account for all flexing torque at the elbow is implied by Morecki et al. (1968) who found "a certain active factor" at 30° of flexion making a difference between their predicted and their measured curves of flexor torque. Basmajian, in his later paper, (Basmajian et al., 1957) imputes a role in elbow flexion to minor flexors in observing "a complete absence of activity in the three flexors during maintenance of flexed postures without a load....The likely explanation of this is that some other of the muscles at the elbow acts sufficiently in some persons to relieve the regular flexors. Proof of this requires further study". In the face of this discussion, the present paper assumes that Biceps Brachii, Brachialis, and Brachioradialis are the major flexors of the elbow (Bouisset et al., 1975, suggest that they constitute 85% of the total flexing torque at the elbow) and the model of maximal isometric flexor torque was generated in this paper from these three flexors alone.

There is intimate relationship between the length of the muscular structures and the contractile capacity of those structures. Of general importance is the influence of length on the capacity of the contractile component and on the role of the parallel elastic component of the muscular structures. Ramsey and Street (1940), Gordon, Huxley and Julian (1966), and Edman (1966) presented data on the length/tension curve of single muscle fibres. It is apparent that, given muscle which is not fatigued and which is then maximally activated, the force generated by the contractile component is determined, in part, by the length of the component relative to some optimal length, L_o. In vitro experimentation (igure 2 following) shows that the optimal sarcomere length in Frog Semitendinosus is 2.05-2.25 microns/ sarcomere. Indeed, Figure 2 demonstrates the length dependence of contractile force over a wide range of absolute sarcomere lengths; these sarcomere lengths, in turn, can be expressed as a percentage of 2.05 microns, the optimal length. Bahler, Fales, and Zierler did, in fact, use relative sarcomere length in presenting similar data on Rat Gracilis Anticus muscle in vitro (1968) and Figure 3 following is a statement of some of their results. The variation in isometric tension with sarcomere length described by Gordon et al. (1966) is a fundamental relationship in the present paper.

That the parallel elastic component modifies the contractile force of the muscular structures when they are working at long lengths is agreed; the quantitative role of parallel elastic component here is

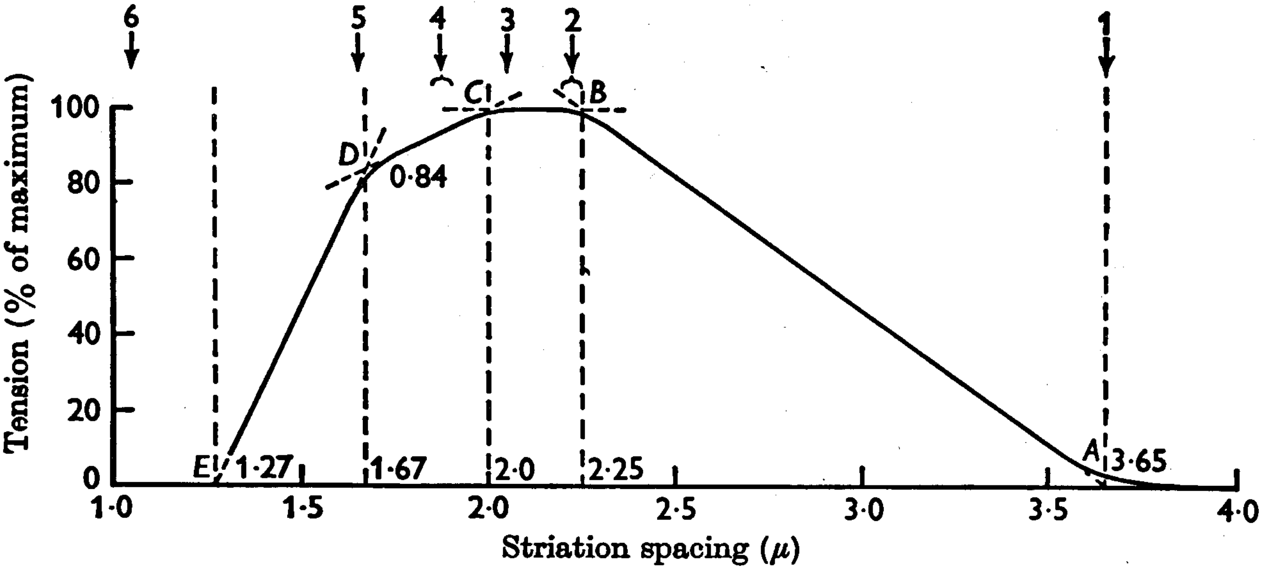


Figure 2. Length/Tension Curve for Frog Semitendinosus at 4° *in vitro*. Adapted from Gordon, Huxley, and Julian, 1966.

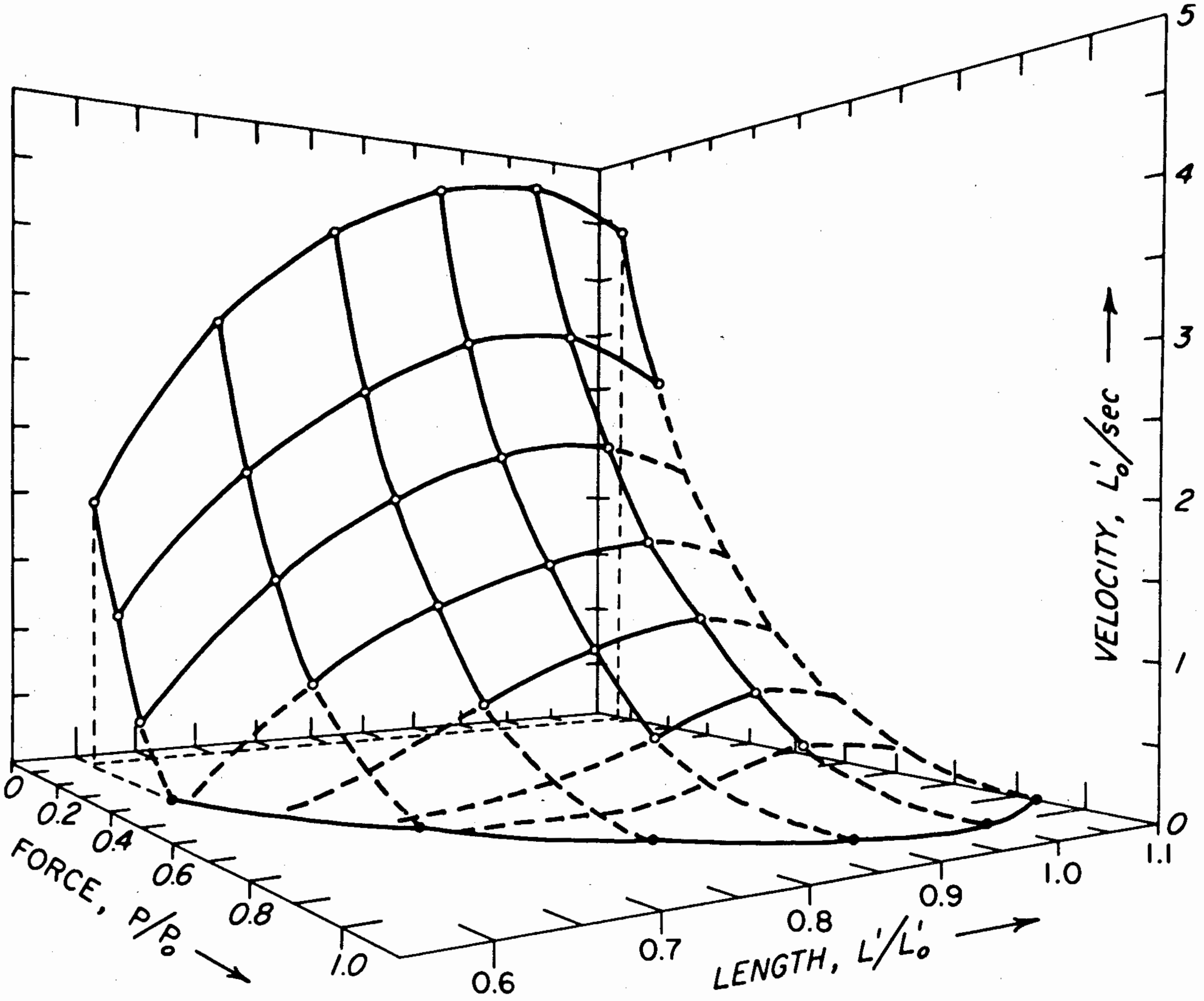


Figure 3. Three-dimensional representation of the dynamic length-force-velocity phase space of the contractile component. Rat Gracilis Anticus *in vitro*. Adapted from Bahler, Fales, and Zierler, 1968.

in debate. Ramsey et al. (1940), Hill (1952), and Thomson (1955) all present data on the passive length/tension curve of intact muscle. Ramsey's data finds that "the resting tension is only 2% at 150% extension and even at 200% extension is only 47% of the maximum tension. In some experiments this fraction was even lower". Hayes et al. in a much more recent paper (1977) find that the absolute measure of passive elastic torque of the structures spanning the human elbow joint is less than 2.5 nt-m's. These two papers stand in contrast to the published work of Fidelus (1968) whose passive length/tension curves for fusiform and penniform muscle are presented below (Figure 4). An additional exercise in the present paper was the incorporation of Fidelus' data into a model of maximal isometric torque allowing for contributions from both the contractile tissues and the parallel elastic tissues. This is reviewed at length in the final chapter, "Discussion".

In addition to the length imposed change on the output of the contractile tissues, which is a general relationship holding for skeletal muscle in a variety of vertebrates, the flexor torque generated by humans at small angles of elbow flexion has been found to be unexpectedly high (Chapman, 1973) and the present paper attempts to reconcile the finding of relatively high flexor torques with the extraordinary elongations of the muscular structures in flexion of the elbow near to full extension.

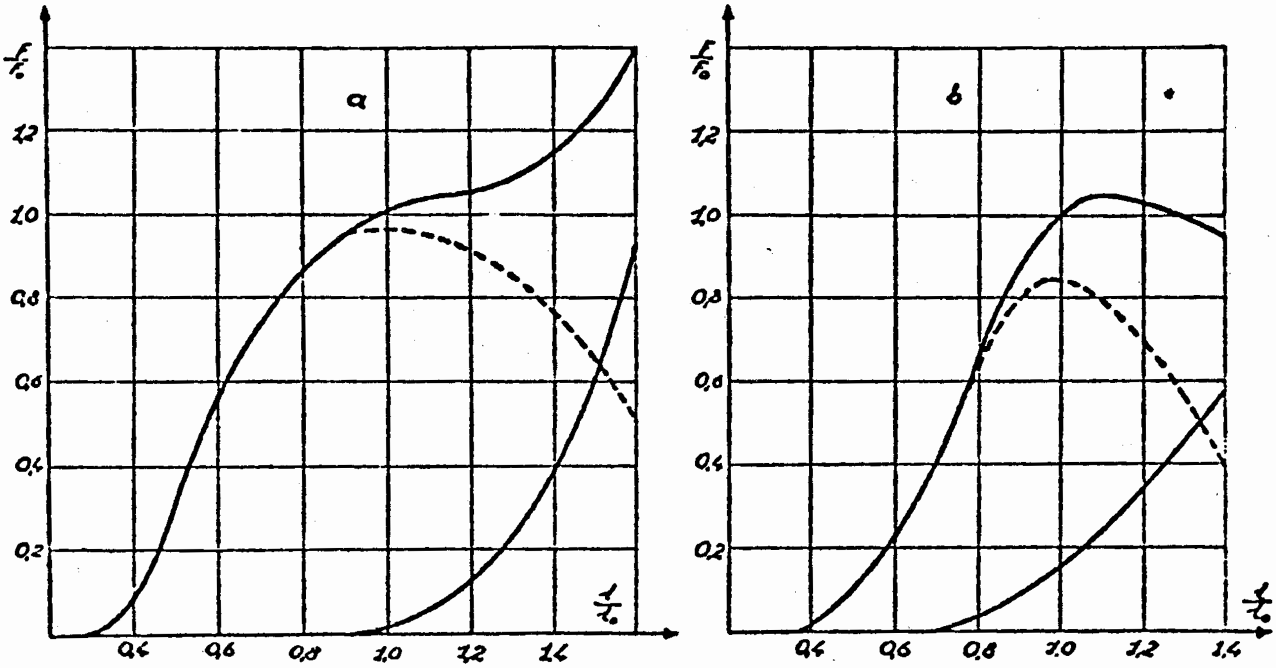


Figure 4. Contractile Force, F/Fo, as a Function of Length, L/Lo, in Rabiit Skeletal Muscle. (Fo is the force developed by the muscle at length Lo, that length where the value of the active force (broken line) was maximal. The lower continuous line indicates contribution from the p.e.c.; the upper line indicates the measured resultant force during isometric stimulation. "a": fusiform muscle; "b": penniform muscle). Adapted from Fidelus, 1968.

Definitive work on assessing the angle of insertion of the elbow flexors and the lengths of their lever arms was published by Braune and Fischer in 1890. They found that the trigonometric calculation of angles of insertion and lever arms using landmark data about the elbow agreed well with the direct measurement of those angles and lever arms. They concluded that the results of the two methods were comparable and, for this reason, both methods were used according to convenience in the present paper in assessing angles of insertion and lever arms of the elbow flexors as these varied with the angle of elbow flexion.

To make an adequate comparison between the modelled curve and the empirical of maximal isometric flexor torque/angle of elbow flexion, the modelled curve needs expression in absolute units of torque. The maximal contractile force per unit cross-sectional area of muscle has been variably reported: Franke (1920) reported the highest value at 11.1 $kg\text{-}f/cm^2$; others including Ficke (1910) and Morris (1948) have reported values slightly less than Franke's. Elftman (1966) has proposed the lowest value, 3.3 $kg\text{-}f/cm^2$. Ikai et al. (1968) in an ultrasonographic study of 245 healthy individuals found that the mean maximal contractile force of skeletal muscle acting at approximately 90^o of elbow flexion was 4.7 $kg\text{-}f/cm^2$ and this value was used in converting the general "torque units" of the torque/angle of flexion curve of the model to absolute units of torque, nt-m's.

MATERIALS AND METHODS

A. Development of the Theoretical Curve

1) Materials

Measurements were performed on a fixed cadaver provided by the Department of Anatomy, the University of British Columbia. It was the body of a burly, middle-aged male of average height and no obvious musculoskeletal abnormalities. However, the cadaver was provided before the experimental protocol was entirely established and, since there was a limited amount of time in which to conduct measurements, final analysis revealed some inadequacies in the breadth of the data collected (for example, quantitative data on the kinanthropometry of the specimen). The skin and subcutaneous fat of the right arm were removed from the hand to the shoulder to expose the underlying anatomy. With the cadaver supine and its shoulder supported by a stable wedge, the right humerus was pinned in two places to angle brackets fixed to the experimental slab. This arrangement kept the upper arm and the centre of rotation of the elbow fixed in space. A plaster cast capable of holding a stylus in the coronal plane was applied to the metacarpals, carpals, and posteriorly over the distal 6 cm of the forearm in order to fix the wrist. With this arrangement, the forearm could be held in full supination and the elbow flexed over 125 degrees without moving the centre of rotation of the elbow. A vertical board was designed to be placed medial or lateral to the upper limb and receive the inscribed arc of flexion. Then, by using Dempster's method (Dempster, 1959), the centre of rotation of this arc was determined and this was necessarily the projection on the vertical board of the centre of rotation of the elbow. In Dempster's method the perpendicular bisectors of several cords drawn through an arc of flexion intersect at the centre of rotation of that arc and in this experiment the perpendicular bisectors of 5 cords drawn through the inscribed arc intersected one another within circumference of a 1 mm circle. The vertical board was then easily converted to a protractor capable of establishing any angle of elbow flexion, α . Figure 5 is an illustration of the preparation and apparatus.

2) <u>Method</u>

Given the model (Figure 1) and the expression for torque (Equation 1) and applying these to conditions in the elbow (Figure 5), it is necessary to measure the lever arms of the several flexors, r , at any one angle of flexion, α ; the length of the contractile tissue of each of the flexors; and, throughout the excursion of the joint, the angles of insertion with respect to the shaft of the radius, β , of each of the flexors as those angles of insertion vary with the angle of flexion, α .

Now in this experiment α is 180 degrees minus the angle subtended by lines passing from the mid-point of the glenoid fossa to the centre of rotation and from the centre of rotation out through the shaft of the radius (Figure 5). The lever arm of each flexor, r ,

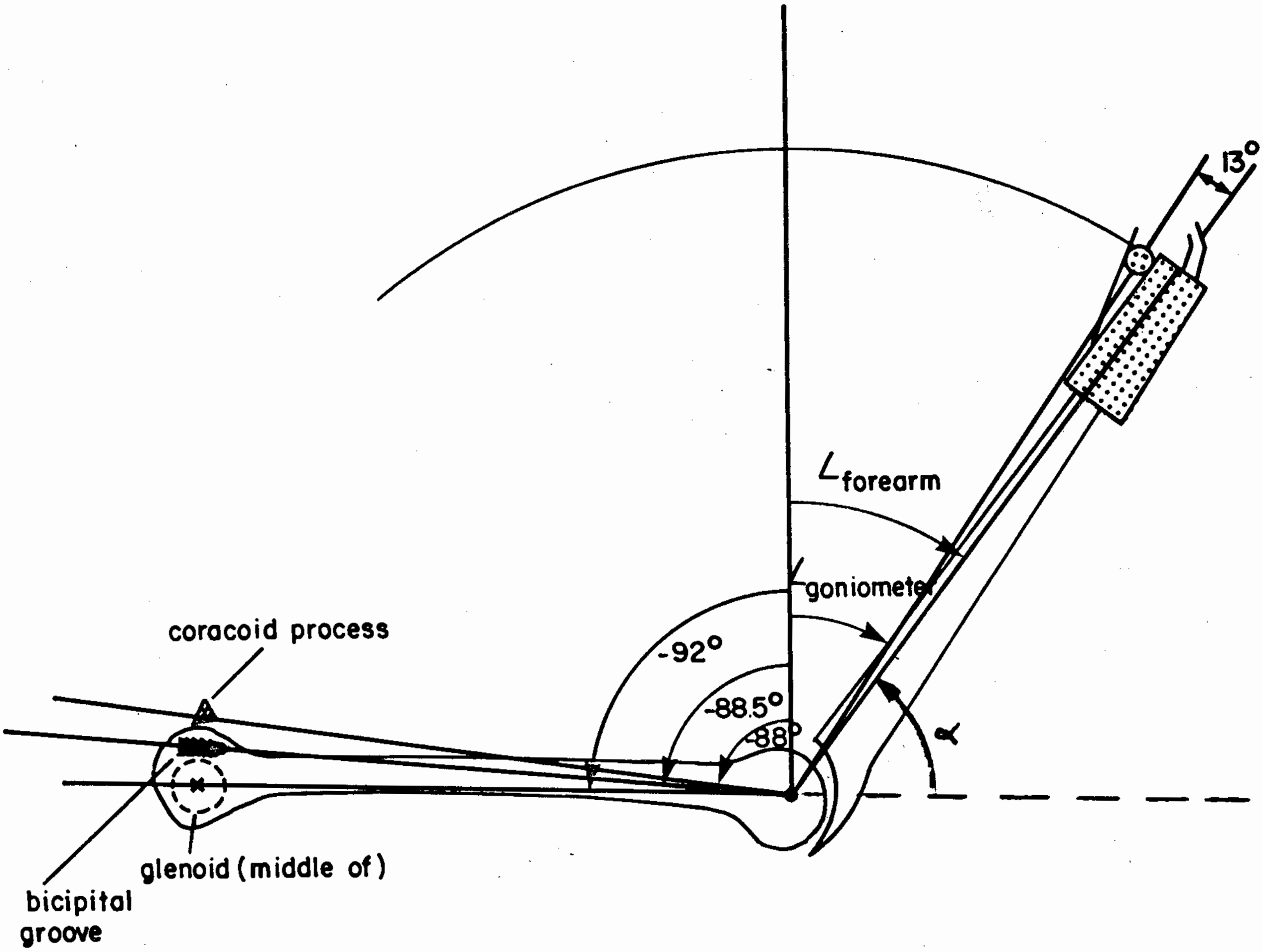


Figure 5. Preparation and fixation of the upper limb to permit determination of the centre-of-rotation of the elbow and selection of angles of flexion, α.

was measured from the centre of rotation out to the point of insertion, which "point" was taken as the geometric centre of the insertion area. This distance was easy enough to measure since both the centre of rotation of the elbow and the origin and insertion of each flexor could be projected onto the vertical board and the various distances separating them read off by caliper.

Following the work of Braune et al. (1890), β was either calculated or measured at various α according to convenience: where the flexors passed straight away from origin to insertion, β was calculated trigonometrically from landmark data on the cadaver; however, both Biceps Brachii and Brachialis incur a pulley effect at smaller angles of flexion with the Biceps Brachii tendon contacting the capitulum of the humerus at α less than or equal to 20 degrees and the Brachialis tendon contacting the trochlea at α less than or equal to 57 degrees - where the flexor tendons incurred this pulley effect, β was measured goniometrically. Figures 6, 7, 8, 9, and 10 present some of the data on lever arms and the lengths of contractile tissue as measured at the embalmed position, α equal to 25 degrees.

Applying Equation 4 (see Introduction) presupposes a reliable method for measuring changes of length of the contractile tissue of the major elbow flexors throughout the excursion of the joint. Assuming inextensible tendon (this is contested by Wilkie, 1950, and more recently by others, and will be dealt with subsequently under "Discussion"), the difficulties are twofold: there is interdigitation of contractile tissue and tendonous tissue at one or both ends of the muscles under study; and, the problem of measuring changing length in contractile tissue of live human muscle raises surgical problems beyond the scope of the experiment. In contrast to Morecki's "muscle preparations" (1968) and Pohtilla's soft tissue X-rays of living subjects (1969), this study attempted to measure r, β, and the change in length of contractile tissue in a human cadaver. A pilot project carried out on a Rhesus monkey (unpublished results) measured the change in length of contractile tissue as a function of angle of flexion by measuring the separation of the proximal and distal portion of a severed flexor at various angles of elbow flexion. Although crude, the procedure was accurate enough to detect a pulley effect on the change in length of contractile tissue at small angles of flexion. Several refinements on the pilot procedure were made in order to measure, with greater accuracy, the change in length of contractile tissue in a human subject: Experimentation was conducted on the cadaver supine; Biceps Brachii were severed at the tendons of origin, Brachioradialis at the insertion tendon, and nylon (because of its low constant of friction over the bones) sutured to the tendonous attachments remaining on the muscle bellies; as the humerus was fixed by double pinning to the experimental platform, the scapulae supported, and the trunk immobile, shortening of

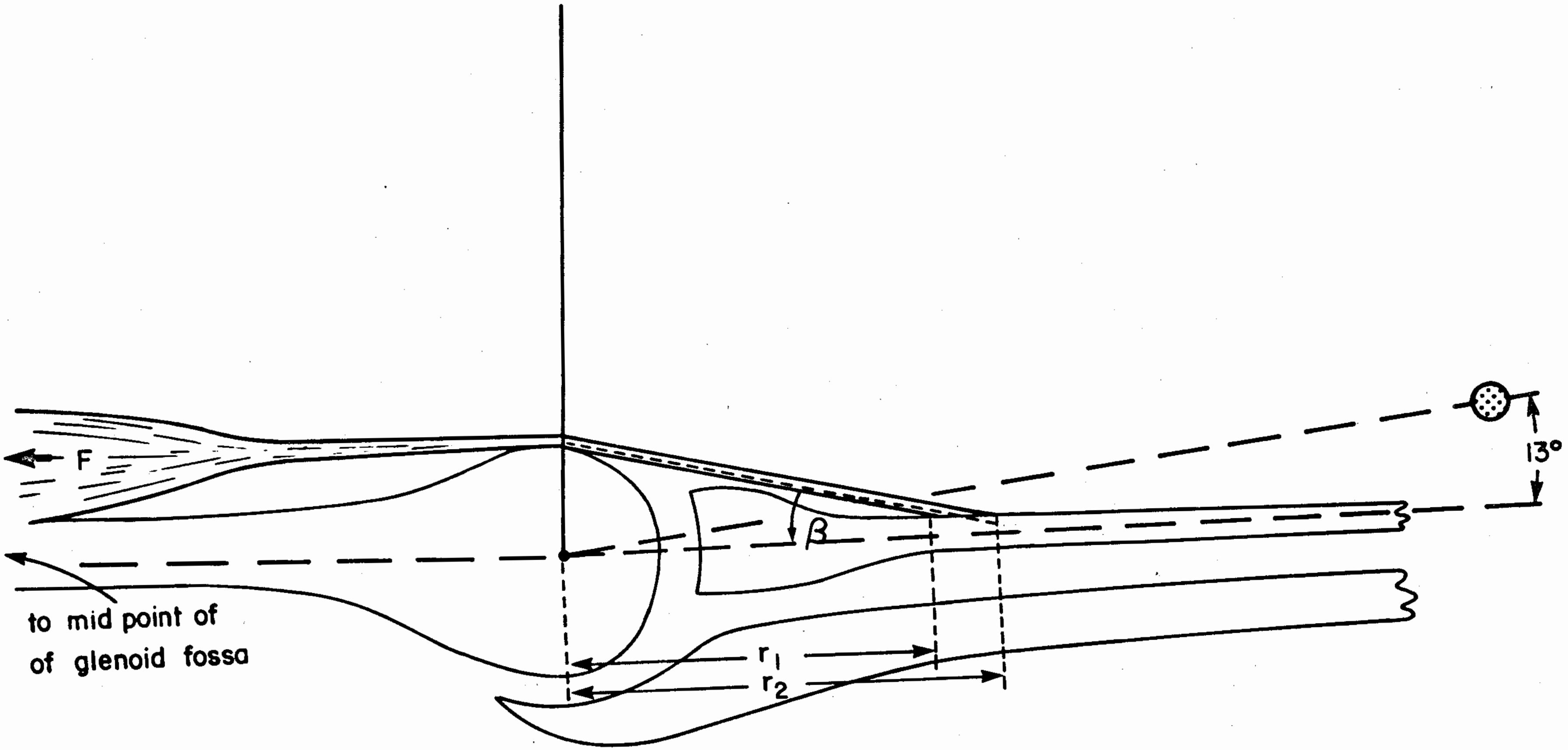


Figure 6. Exposure of Biceps Brachii insertion at the radius.

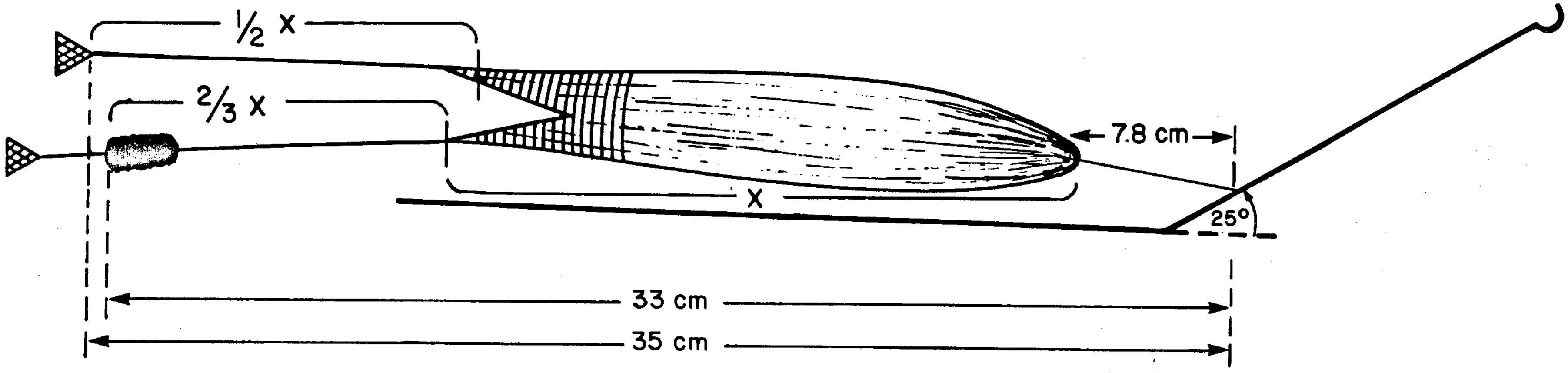
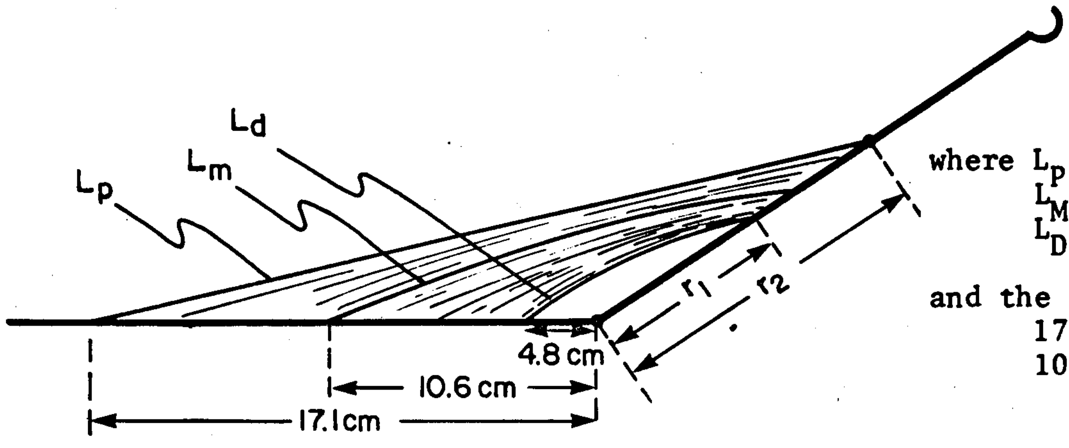


Figure 7. Exposure and dimensions of the contractile tissue and series elastic tissue of Biceps Brachii from origin to insertion.



where L_P = length of proximally originating fibre
L_M = length of middle fibre
L_D = length of distally originating fibre

and the distances along the humerus
17.1 cm. measured
10.6, 4.4 cm. estimations of distance from centre of rotation of the origins of the middle fibre and of the most distally originating fibre (from Grant's #29, 1962)
r_1 and r_2 as measured at the embalmed position, $\alpha = 25^o$
r_1 = 3.4 cm.
r_2 = 5.6 cm.

Figure 8. Exposure and dimensions of Brachialis origin to insertion.

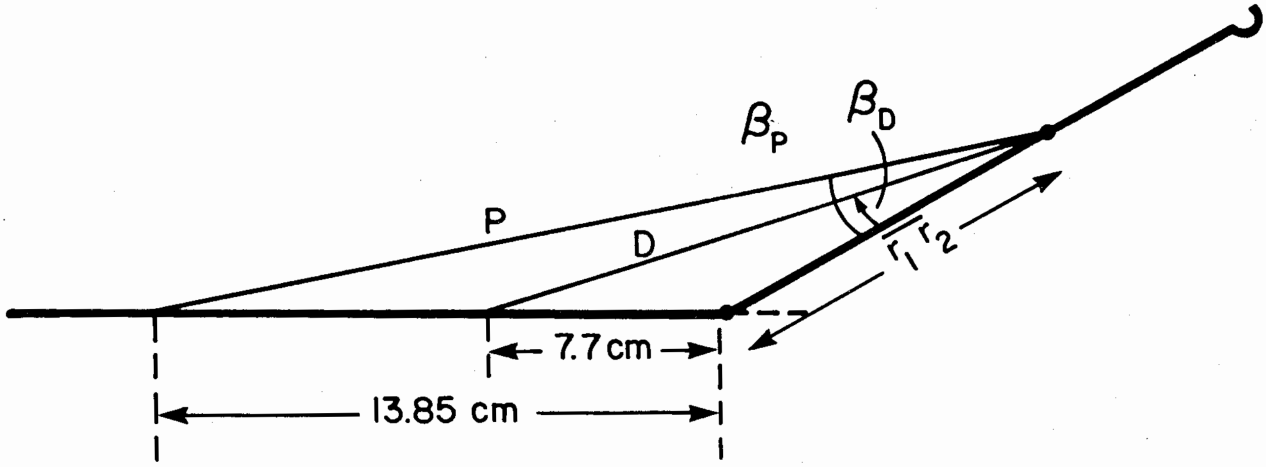


Figure 9. Extrapolations on Brachialis derived from Figure 8. "P" is a fibre originating midway between the origins of L_P and L_M in Figure 8. "D" is a fibre originating midway between the origins of L_M and L_D in Figure 8. Both insert on the common point whose lever arm is $\overline{r_1 r_2}$ = 4.5 cm.

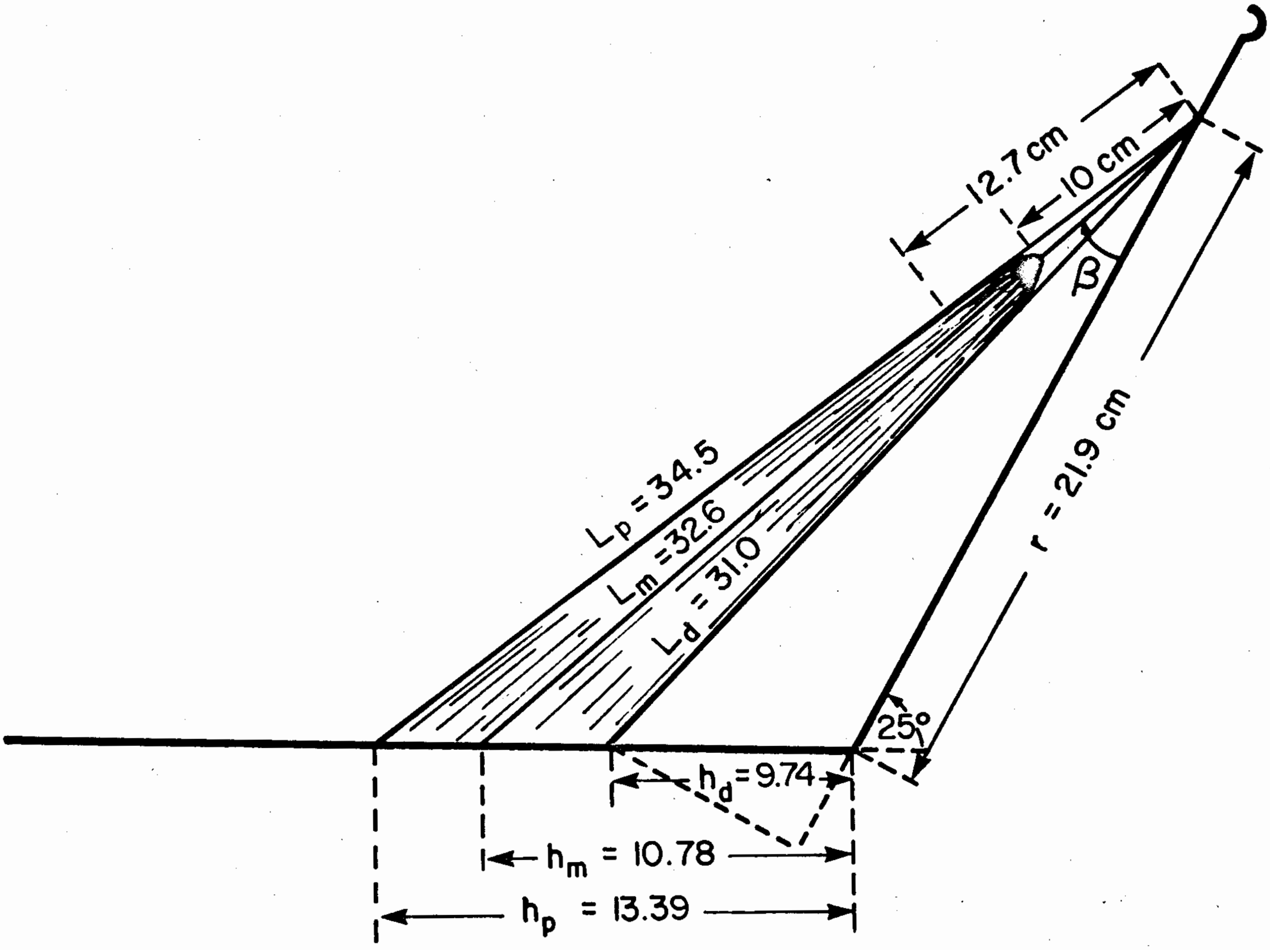


Figure 10. Exposure and dimensions of Brachioradialis.

the elbow flexors was manifest only in flexion of the elbow. The sutured cords were passed over the normal tendonous attachment points and a fixed load (1 kg) was suspended from the cord in order to ensure constant strain and axial alignment in the preparation. The free end of the cord sutured to the free end of the long head of Biceps Brachii was itself sutured to the short head of Biceps Brachii and the load suspended from the centre of this cord loop - this, in turn, insured that normal co-operation could be expected between the long and short heads of Biceps Brachii in flexion. Now the supinated forearm could be flexed from α = 0 degrees to α = 125 degrees and the change in length of the contractile tissue measured as a function of α by measuring the change in marked positions on the nylon cord. The change in length of contractile tissue for Brachialis was not determined in this way because the muscle originates from a large area of the anterior, distal portion of the humerus and inserts deep on the ulna. For this muscle the length of the contractile tissue at each α was computed trigonometrically from direct measurement of distance, centre of rotation to proximal origin, and centre of rotation to proximal origin, and centre of rotation to insertion, and from calculation of the concomitant distance, centre of rotation to distal origin, using Grant's Atlas, plate number 29 (Boileau-Grant, 1962). Because of the particular anatomy of Brachialis, determination of the dependence of contractile force here on angle of flexion will need to involve the summation of the contributions from the range of fibres most distal to most proximal as each is affected by the angle of flexion.

When all these measurements had been completed, each of the muscle groups was removed from the cadaver and weighed so that an estimation could be made of the absolute contribution that each group might make to the total flexing torque acting at the elbow (see Results, Table 1).

3) Manipulation of Data

With respect to Equation 1, the product $\sin\beta \cdot r$ is straightforward; the theorectical computation of force, F, generated by each flexor was more complicated since it depends on the length of the muscular unit with respect to some optimal length, L_o, and also on the weight of the muscle tissue.

It was simple enough to compute the absolute length from origin to insertion at any given α by incorporating measured length changes with the absolute measure of length at the embalmed position, $\alpha = 25$ degrees. The length changes imposed by excursion of the joint were enormous. Indeed, if the muscular structures lengthened 20% in order to accommodate full joint extension, they had to shorten by 40% to bring the elbow in full normal flexion at $\alpha = 125$ degrees. Two approaches were tried in converting the absolute length at a particular α to a percentage of the optimal length, L_o: L_o was initially chosen to be the length at which the length/angle of flexion curve was of greatest slope. This, however, commited the contractile tissue to shorten into the D zone demarcated by Gordon et al., Figure 2 (1966). Teleologically this does not make sense and, indeed, Ramsey et al. (1940)

and Moss et al. (1968) suggest that the cell population of a given muscle adjusts itself to accommodate full, normal shortening without entering into the D zone of Figure 2. Given the large length changes observed in this experiment, it is apparent that the elbow flexors will have to shorten and extend to lengths where the tension developed by the contractile tissue is substantially less than maximal even in the normal range of motion of the joint. The approach finally used in this study was to assume that the muscle in full normal flexion would not shorten into the D zone, but approach it closely. Therefore, the absolute length of the muscular structure at full normal flexion (125 degrees) would be close to 79.5% of L_o (Gordeon et al., 1966). This argument has assumed that the elastic components of muscular structures are inextensible. In fact, under full isometric tension the strain of the elastic structures has been variously put at 3% to 7% (Hill, 1950; Wilkie, 1956; Sonnenblick, 1964; Bahler, 1967); in this study the tendon of Biceps Brachii, for example, was 2/3 as long as the contractile tissue and, under full tension would certainly provide the compliance to reduce the α at which the Biceps Brachii were at L_o. This introduces error into the model which will be discussed in the final chapter.

Now F is a function of L_i/L_{o_i} and we need to quantify the contribution to F from both the contractile tissue and the parallel elastic tissue. Given our measures of absolute length with respect to α and our assumptions about L_o as a consequence of measured length at = 125^0, this study used the data in Figure 2 to express P_i as a percentage of P_{o_i}. Used in Equation number 4, this P_i will give a torque/angle curve in terms of "torque units". But, P_{o_i} reflects the cross-section of the i'th muscle (Ralston, 1953; Wells, 1965) and and estimate of the absolute force in newton-metres, P_{o_i}, generated by the i'th muscle can be made from the data published by Ikai and Fukunaga (1968). They calculated that muscle is capable of exerting 4.3 kg-force per cm^2 of cross-sectional area in maximal isometric contraction. In this study a geometric approximation of the cross-section of each of the major flexors was made by using measures of muscle masses and contractile tissue length in the experimental cadaver. By approximating the contractile tissue as a cylinder, we may use the length, density, and mass of the various flexors to extimate their several cross-sectional areas and the maximal isometric torque proportional to each as follows:

$$\tau = F \cdot \sin\beta \cdot r$$

$$= (\text{area of cross-section})(4.3\ \text{kg-f/cm}^2) \cdot \sin\beta \cdot r \ \text{nt-m}$$

$$= \left(\frac{\text{mass}}{\text{density} \cdot \text{length}}\right)(4.3\ \text{kg-f/cm}^2) \cdot \sin\beta \cdot r \ \text{nt-m}$$

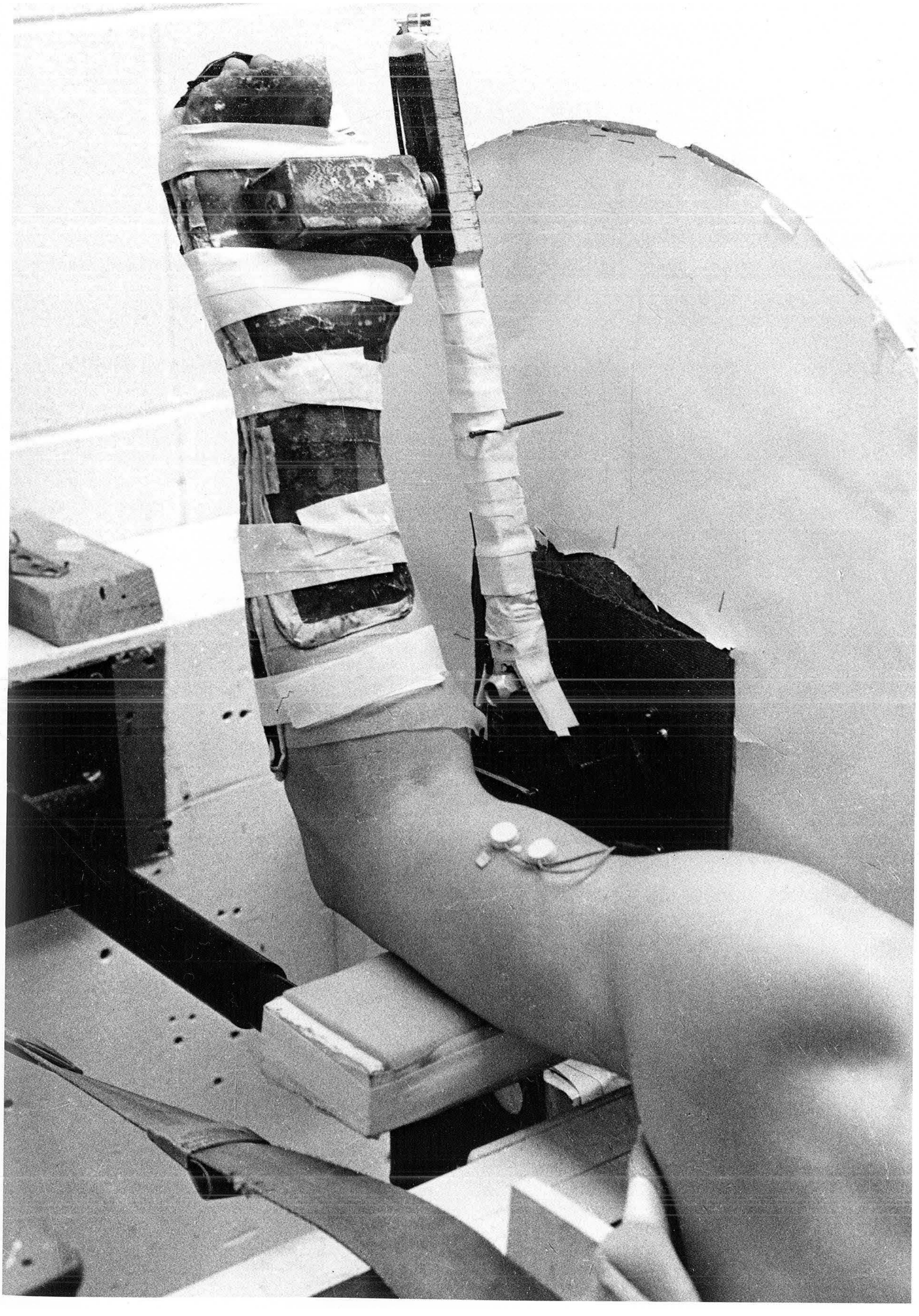
With torque expressed in nt-m's, the model curve can be compared with empirical data according to locus of maximum isometric torque, the ratio of maximum isometric torque to the torque exerted at $\alpha = 5^{o}$, and with respect to the α at which the torque/angle of flexion curve inflects.

Finally, at small degrees of flexion where this experiment found lengths of the muscular structures approaching 140% of L_o, the stretched parallel elastic tissue might be expected to contribute in some measure to flexor torque. There is disagreement about the degree of flexor torque contributed by maximally stretched parallel elastic tissue at the elbow. Bahler et al. (1968) and Morecki et al. (1968) have alluded to the role of parallel elastic tissue in modifying torque; Hayes et al. (1977) find this torque to be less than 2.5 nt-m's in human experiments; Fidelus, however, has published data (1968) on the in vivo isometric force/length curve for rabbit muscle (Figure 4). In sum, he found that in fusiform muscle the role of the parallel elastic component was measurable for any length of the muscular structure where the contractile tissue is at L_o or greater and that, at 140% of L_o, the role of the parallel elastic tissue was very substantial. Hill (1970) cautioned that, "it is precarious to argue from frogs to toads." It is certainly so to argue from rabbits to men. Nevertheless, some computations were made in this study on the basis of Fidelus' data. These are presented in the chapter, "Results," and the contribution from parallel elastic tissue discussed in the last chapter.

B. Collection of the Empirical Curves

1) Apparatus, Including Positioning of Subjects and Recording of Data

Figure 11 following is an illustration of the apparatus:



The pertinent points about this apparatus are these: The angle of flexion can be exactly set by a displacement transducer; the intrinsic moment of the radial shaft of the machine is small; since the shaft is constructed of magnesium alloy, its tensile strength is high and its compliance also small; the equipment accurately measures applied torque by a strain gauge fixed to the radial shaft of the machine; the design of the seat and frame allows the subject to be positioned so that the lie of the humerus is horizontal and so that the centre of rotation of the subject's elbow is co-axial with the axis of rotation of the machine; moreover, the lever arm of the machine's radial shaft can be adjusted by means of a sliding bolt to correspond to the length of the subject's forearm; finally, the output of the elbow flexors can be directed solely across the elbow because the subject's thorax can be immobilized against the machine, the centre of rotation of the subject's elbow can be aligned with the axis of rotation of the machine, the lever arm of the radial shaft can be adjusted to the length of the subject's forearm, and the subject's wrist prevented from flexing, extending, supinating or pronating by the fibreglass cast which covers the hand and distal one half of the subject's forearm and is attached to the radial shaft.

The machine was calibrated to impose specified angles of flexion at the outset of experimentation (5^0, 15^0, 20^0, 50^0, 70^0, 80^0, 90^0, and 95^0) and this remained unchanged throughout the time of data collection. Similarly, the strain gauge output was calibrated for torque at the outset of experimentation and was re-checked prior to each testing session. The performance of each subject was recorded on ultra-violet light directed by outputs from the displacement and strain gauge transducers.

3) Subjects and Trials

Six subjects were tested, three weight lifters, one hammer thrower, and two rowers. All were interested national class athletes and could be expected to perform reliably at each session. This maximized the probability that the efforts of the subjects were consistently maximal. The machine having been calibrated against known torques prior to each testing session, the subject at rest in the apparatus generated a resting torque, a torque arising from the mass of the forearm in the cast and from intrinsic forces generated by elastic components of the muscles and fascia. The dislacement of the strain gauge output away from that resting torque in maximum isometric flexion at any given angle of flexion then was directly a measure, in nt-m's, of the torque generated at that angle by the contractile tissue of all the elbow flexors.

Each subject underwent two testing sessions. Each session consisted of eight maximum isometric flexions for both abducted and flexed positions of the shoulder. At each position of the shoulder the sequence of angles at which flexions were performed was randomly ordered. The highest torque recorded at each angle of flexion was taken as the individual's maximum isometric capability at that particular angle and shoulder position and both records were reproduced as curves of maximum isometric flexing torque vs. angle of flexion at the elbow.

It was thought at the outset that each subject could be used as his own control by having him perform maximum isometric flexions at various angles of elbow flexion in these two positions of the shoulder for, in moving from shoulder flexion to shoulder abduction, the biomechanics of maximum isometric elbow flexion are changed to a large extent only by changing the length of the short head of Biceps Brachii. While the difference in the two curves for each subject has not been helpful in evaluating the model in Equation 4 (This is considered at greater length in the chapter, "Discussion".), data on the change in length of Biceps Brachii Short Head (BBSH) in both shoulder positions *in vivo* was used to generate a second theoretical curve on measurements taken from the cadaver.

The fixed tissue of the cadaver could not be safely stretched to have the shoulder in abducted position. The subjects were easily tested in both positions, however, and the origin of BBSH on the tip of the coracoid process and its insertion on the radius through the common bicipital tendon were quite easily identified on all subjects. The length increase in BBSH in accommodating shoulder movement from flexion to abduction in each subject was measured by calipers and these results and the mean increase in length are tabulated in Table 4. This mean increase was worked back into the model to generate a theoretical curve of maximal isometric flexor torque in shoulder abduction vs. angle of elbow flexion torque in shoulder abduction vs. angle of elbow flexion (Tables 3 and 9). The two theoretical curves then could be compared with the pairs of curves collected from each of the live subjects.

RESULTS

The weights of the muscle groups in the cadaver were (Table 1):

Biceps Brachii	185 gm-f
Brachialis	140 gm-f
Brachioradialis	50 gm-f
Extensors Carpi Radialis (Longus + Brevis)	75 gm-f

Table 1 ----weights of 4 of the muscle groups capable of exerting flexor torque at the elbow.

Now, given Equation #4,

$$\text{Total } \tau = \sum \tau_{i_\alpha}\left(\left(\frac{L}{L_o}\right)_{i_\alpha}, \text{X-section}_{i_{L_o}}\right) \cdot \text{Sin}\,\beta_{i_\alpha} \cdot r_{i_\alpha}$$

and using the following derivation for Brachioradialis as the general paradigm for the conversion of "torque units" to torque in nt-m's (after Ikai et al, 1968) for each of the major flexors,

Wt. Brachioradialis = 50 gm-f

length contractil tissue = 22 cm

let density of contractile tissue be 1 gm/cm^3

if $\tau = F \cdot \text{Sin}\,\beta \cdot r$

then $$\tau_{L_o} = (\text{X-section}_{L_o})(4.3\ \text{kg-f/cm}^2) \cdot \text{Sin}\,\beta_{L_o} \cdot r$$

$$= \left(\frac{\text{mass}}{\text{density} \cdot \text{length}}\right)(4.3\ \text{kg-f/cm}^2) \cdot \text{Sin}\,\beta \cdot r$$

$$= \left(50\ \text{gms} / \text{gms/cm}^3 \cdot 22\ \text{cm}\right)\left(4.3\ \text{kg-f/cm}^2\right) \cdot \sin\beta \cdot r$$

$$= (2.27)(4.3)(\sin\beta \cdot r)(9.8)\ \text{nt-m's}$$

Figure 12: Conversion of "torque units" to torque in nt-m's. The product $\left(\frac{\text{mass}}{\text{density} \cdot \text{length}}\right)$ is proportional to the maximal isometric contractile force which the particular muscle can generate. This is termed "torque units" in the following results and, multiplied by the constants in Ikai et al. (1968), can be converted to nt-m's as above.

Tables 2, 3, 6, 7, and 8 following list the parameters of Equation 4 as they vary with angle of flexion, α, for each of the flexors examined in the cadaver. The torque predicted for each of the flexors from $\alpha = 5^\circ$ to $\alpha = 125^\circ$ is listed in the last column of each table. Table 9 lists the total flexor torque predicted from all contractile tissue of the major flexors as a function of angle of flexion in both experimental positions of the shoulder, flexed and abducted; Table 9 also lists total torque at each α for both positions of the shoulder normalized about the total torque predicted at $\alpha = 90^\circ$, shoulder flexed.

Table 2. Parameters of flexor torque at various angles of elbow flexion for Biceps Brachii, Long Head; prediction of maximal isometric flexor torque from the contractile tissue of this muscle.

Key to the table by column follows:

A ...	angle of flexion
B ...	"r" in cms
C ...	$\sin\beta$
D ...	absolute length of contractile tissue in cms (average of 2 trials)
E ...	$\%L_o$
F ...	$\%P_o$
G ...	"force units"
H ...	τ of the contractile tissue in "force units"
I ...	τ of the contractile tissue in nt-m's

Table 2

A	B	C	D	E	F	G	H	I
125	5.08	.87	11.3	79.5	82	6.6	23.8	10.0
120	5.1	.91	11.38	79.5	82	6.6	25.2	10.6
115	5.12	.95	11.51	81	84	6.6	26.9	11.3
110	5.14	.97	11.78	83	87	6.6	28.7	12.1
105	5.17	.99	11.99	84	89	6.6	30.1	12.7
100	5.19	1.00	12.33	87	92	6.6	31.5	13.3
95	5.2	1.00	12.59	89	94	6.6	32.3	13.6
90	5.22	.99	13.03	92	97	6.6	33.2	14.0
85	5.23	.98	13.43	95	100	6.6	33.8	14.2
80	5.24	.96	13.79	97	100	6.6	33.1	13.9
75	5.25	.93	14.25	100	100	6.6	32.3	13.6
70	5.26	.90	14.70	104	100	6.6	31.2	13.1
65	5.26	.86	15.15	107	100	6.6	29.9	12.6
60	5.27	.82	15.52	109	97	6.6	27.6	11.6
55	5.27	.77	15.92	112	92	6.6	24.6	10.4
50	5.28	.72	16.27	115	88	6.6	22.1	9.3
45	5.29	.66	16.63	117	85	6.6	19.6	8.2
40	5.29	.60	16.94	119	82	6.6	17.3	7.3
35	5.29	.54	17.41	123	75	6.6	14.2	6.0
30	5.29	.48	17.74	125	73	6.6	12.1	5.1
25	5.3	.41	18.03	127	70	6.6	10.1	4.3
20	5.3	.31	18.21	128	68	6.6	7.3	3.1
15	5.3	.29	18.44	130	65	6.6	6.6	2.8
10	5.3	.26	18.64	131	63	6.6	5.8	2.4
5	5.3	.28	18.83	133	61	6.6	6	2.5
0	5.31		19.03	134	60	6.6		

Table 3. Parameters of flexor torque at various angles of elbow flexion for Biceps Brachii, Short Head; prediction of maximal isometric flexor torque from the contractile tissue of this muscle. Summation of maximal isometric flexor torque from contractile tissue of both heads of Biceps Brachii, shoulder flexed. Key to the table by colum follows:

A ... angle of flexion

B ... "r" in cms

C ... $\sin\beta$

D ... contractile tissue absolute length in cms (average of 2 trials)

E ... $\%L_o$

F ... $\%P_o$

G ... "force units"

H ... τ of the contractile tissue in "force units"

I ... τ of the contractile tissue in nt-m's

J ... total torque of the contractile tissues of Long and Short Heads of Biceps Brachii in nt-m's

Table 3

A	B	C	D	E	F	G	H	I	J
125	5.08	.86	11.30	80	82	6.6	23.6	9.9	19.9
120	5.10	.91	11.33	80	82	6.6	24.4	10.3	20.9
115	5.12	.94	11.38	80	82	6.6	26	11.0	22.3
110	5.14	.97	11.65	82	86	6.6	28.2	11.9	24.0
105	5.17	.99	11.84	83	87	6.6	29.3	12.3	25.0
100	5.19	1.00	12.21	86	91	6.6	31.1	13.1	26.4
95	5.2	1.00	12.50	88	93	6.6	31.9	13.4	27.0
90	5.22	1.00	12.92	91	96	6.6	32.9	13.9	27.9
85	5.23	.98	13.26	93	97	6.6	32.9	13.9	28.1
80	5.24	.96	13.70	96	100	6.6	33.4	14.1	28.0
75	5.25	.94	14.20	100	100	6.6	32.5	13.7	27.3
70	5.26	.91	14.67	103	100	6.6	31.6	13.3	26.4
65	5.26	.87	15.17	107	100	6.6	30.2	12.7	25.3
60	5.27	.83	15.66	110	95	6.6	27.3	11.5	23.1
55	5.27	.78	15.96	113	91	6.6	24.7	10.4	20.8
50	5.28	.73	16.38	115	88	6.6	22.4	9.4	18.7
45	5.29	.67	16.71	118	83	6.6	19.6	8.3	16.6
40	5.29	.62	17.18	121	78	6.6	16.8	7.1	14.8
35	5.29	.55	17.49	123	75	6.6	14.4	6.1	12.1
30	5.29	.49	17.90	126	72	6.6	12.3	5.2	10.3
25	5.30	.42	18.16	128	68	6.6	10.0	4.2	8.5
20	5.30	.31	18.47	130	65	6.6	7.0	2.9	6.0
15	5.30	.29	18.67	131	63	6.6	6.4	2.7	5.5
10	5.30	.26	18.87	133	61	6.6	5.5	2.3	4.7
5	5.30	.28	19.07	134	60	6.6	5.9	2.5	5.0
0	5.31		19.24	135	58				

Table 4. Lengths of Biceps Brachii, Short Head, in 6 subjects (coracoid process to insertion on the radius) in flexed and abducted positions of the shoulder. Key to the table by column follows:

A ... subject
B ... length in each of two trials and the mean length, shoulder flexed
C ... length in each of two trials and the mean length, shoulder abducted
D ... elongation

Table 4

A	B	C	D
A.N.	$\frac{34.0}{33.7}$ — 33.9 cms.	$\frac{36.5}{38.0}$ — 37.3 cms.	3.4 cm.
R.C.	$\frac{34.3}{35.2}$ — 34.8 cms.	$\frac{39.5}{39.2}$ — 39.4 cms.	4.6 cm.
W.B.	$\frac{36.0}{35.4}$ — 35.7 cms.	$\frac{41.0}{37.5}$ — 39.3 cms.	3.6 cm.
H.W.	$\frac{34.0}{35.5}$ — 34.8 cms.	$\frac{36.2}{38.0}$ — 37.1 cms.	2.3 cm.
D.E.	$\frac{34.8}{34.3}$ — 34.6 cms.	$\frac{37.2}{34.5}$ — 35.9 cms.	1.3 cm.
M.C.	$\frac{33.7}{33.3}$ — 33.5 cms.	$\frac{37.3}{37.2}$ — 37.3 cms.	3.8 cm.

Mean elongation = 3.2 cm.

Table 5. Derived length of contractile tissue of Biceps Brachii, Short Head, in the cadaver at various angles of elbow flexion with the shoulder abducted. Prediction of maximum isometric flexor torque from the contractile tissue of this muscle. Summation of maximum isometric flexor torque from contractile tissue of both heads of Biceps Brachii with shoulder abducted. Key to the table by column follows:

A ... angle of flexion

B ... "r" in cms

C ... Sin β

D ... absolute length of the contractile tissue (average of two trials) with shoulder flexed

E ... absolute length of contractile tissue with addition of mean elongation

F ... $\%L_o$

G ... $\%P_o$

H ... "force units"

I ... torque generated by the contractile tissue in force units

J ... torque generated by the contractile tissue in nt-m's

K ... total torque from contractile tissue, both heads, nt-m

Table 5

A	B	C	D	E	F	G	H	I	J	K
125	5.08	.86	11.30	14.5	103	100	6.6	28.8	12.2	22.2
120	5.10	.91	11.33	14.53	103	100	6.6	30.6	12.9	23.5
115	5.12	.94	11.38	14.58	103	100	6.6	31.8	13.4	24.7
110	5.14	.97	11.65	14.85	105	100	6.6	32.9	13.9	26.0
105	5.17	.99	11.84	15.04	106	100	6.6	33.8	14.2	26.9
100	5.19	1.00	12.21	15.41	109	96	6.6	32.9	13.9	27.2
95	5.2	1.00	12.50	15.7	111	94	6.6	32.3	13.6	27.2
90	5.22	1.00	12.92	16.12	114	89.5	6.6	30.8	13.0	27.0
85	5.23	.98	13.26	16.46	117	84.5	6.6	28.6	12.0	26.2
80	5.24	.96	13.70	16.90	120	79	6.6	26.2	11.1	25.0
75	5.25	.94	14.20	17.40	123	75	6.6	24.4	10.3	23.6
70	5.26	.91	14.67	17.87	127	70	6.6	22.1	9.3	22.4
65	5.26	.87	15.17	18.37	130	65	6.6	19.6	8.3	20.9
60	5.27	.83	15.66	18.86	134	60	6.6	17.3	7.3	18.9
55	5.27	.78	15.96	19.16	136	56	6.6	15.2	6.4	16.8
50	5.28	.73	16.38	19.58	139	51	6.6	13.0	5.5	14.8
45	5.29	.67	16.71	19.91	141	48	6.6	11.2	4.7	12.9
40	5.29	.62	17.18	20.38	144	42	6.6	9.1	3.8	11.1
35	5.29	.55	17.49	20.69	146	38	6.6	7.3	3.1	9.1
30	5.29	.49	17.90	21.10	149	33	6.6	5.6	2.4	7.5
25	5.30	.42	18.16	21.36	151	30	6.6	4.4	1.9	6.2
20	5.30	.31	18.47	21.67	153	28	6.6	3.0	1.3	4.4
15	5.30	.29	18.67	21.87	155	25	6.6	2.5	1.1	3.9
10	5.30	.26	18.87	22.07	151	23	6.6	2.1	0.9	3.3
5	5.30	.28	19.07	22.27	158	21	6.6	2.1	0.9	3.4
0	5.31		19.24	22.47	159	20				

Table 6. Parameters of flexor torque at various angles of elbow flexion for contractile tissue of Brachialis, proximally originating portion; prediction of maximal isometric flexor torque from the contractile tissue of this portion of Brachialis. Key to the table by column follows:

A ... angle of flexion

B ... "r" in cms

C ... Sin β

D ... absolute length of the contractile tissue (average of two trials) in cms

E ... %L_o

F ... %P_o

G ... "force units"

H ... torque generated by the contractile tissue, force units

I ... torque generated by the contractile tissue, nt-m's

Table 6

A	B	C	D	E	F	G	H	I
125	4.15	.95	11.96	79.5	82	3.9	12.5	5.3
120	4.16	.97	12.31	82	87	3.9	13.5	5.7
115	4.2	.99	12.70	85	90	3.9	14.5	6.1
110	4.21	1.00	13.00	87	92	3.9	15	6.3
105	4.24	1.00	13.40	89	94	3.9	15.4	6.5
100	4.26	.99	13.80	92	97	3.9	15.8	6.7
95	4.28	.98	14.1	94	98	3.9	15.9	6.7
90	4.3	.96	14.5	97	100	3.9	16.0	6.7
85	4.33	.93	14.9	99	100	3.9	15.5	6.5
80	4.36	.90	15.2	101	100	3.9	15.2	6.4
75	4.38	.86	15.6	104	100	3.9	14.6	6.2
70	4.4	.82	15.9	106	100	3.9	14.0	5.9
65	4.42	.77	16.2	108	98	3.9	12.9	5.4
60	4.44	.73	16.5	110	95	3.9	11.9	5.0
55	4.46	.63	16.8	112	92	3.9	10.7	4.5
50	4.48	.62	17.1	114	89	3.9	9.5	4.0
45	4.51	.57	17.3	115	87	3.9	8.7	3.7
40	4.54	.53	17.6	117	85	3.9	7.9	3.3
35	4.56	.56	17.8	119	82	3.9	8.1	3.4
30	4.58	.57	18.0	120	80	3.9	8.1	3.4
25	4.6	.59	18.1	121	79	3.9	8.3	3.5
20	4.62	.61	18.3	122	77	3.9	8.4	3.5
15	4.65	.64	18.4	123	76	3.9	8.8	3.7
10	4.67	.67	18.5	123	76	3.9	9.2	3.9
5	4.70	.71	18.5	123	76	3.9	9.8	4.1
0	4.72		18.6	124	75	3.9		

Table 7. Parameters of flexor torque at various angles of elbow flexion for contractile tissue of Brachialis, distally originating portion; prediction of maximal isometric flexor torque from the contractile tissue of this portion of Brachialis; summation of maximal isometric flexor torque from the contractile tissue of both parts of Brachialis. Key to the table by column follows:

A ... angle of flexion

B ... "r" in cms

C ... Sin β

D ... absolute length of the contractile tissue (average of two trials) in cms

E ... $\%L_o$

F ... $\%P_o$

G ... "force units"

H ... torque generated by the contractile tissue, force units

I ... torque generated by the contractile tissue, nt-m's

J ... total torque generated by both the porximally and the distally originating portions of Brachialis, nt-m's

Table 7

A	B	C	D	E	F	G	H	I	J
125	4.15	1.00	6.31	79.5	83	5.83	20.1	8.4	13.7
120	4.16	1.00	6.7	84	89	5.83	21.6	9.1	14.8
115	4.2	.99	7.0	88	93	5.83	22.6	9.5	15.6
110	4.21	.98	7.4	93	98	5.83	23.6	9.9	16.2
105	4.24	.96	7.8	98	100	5.83	23.7	10.0	16.5
100	4.26	.93	8.1	102	100	5.83	23.1	9.7	16.4
95	4.28	.90	8.5	107	100	5.83	22.5	9.5	16.2
90	4.3	.87	8.8	111	94	5.83	20.5	8.6	15.3
85	4.33	.84	9.2	116	86	5.83	18.3	7.7	14.2
80	4.36	.80	9.5	120	80	5.83	16.3	6.9	13.3
75	4.38	.76	9.8	123	75	5.83	14.6	6.2	12.4
70	4.4	.72	10.1	127	70	5.83	13.0	5.5	11.4
65	4.42	.67	10.4	131	64	5.83	11.1	4.7	10.1
60	4.44	.63	10.6	134	58	5.83	9.5	4.0	9.0
55	4.46	.58	10.9	137	54	5.83	8.2	3.5	8
50	4.48	.62	11.1	140	50	5.83	8.3	3.5	7.5
45	4.51	.57	11.4	144	44	5.83	5.8	2.4	6.1
40	4.54	.53	11.6	146	42	5.83	5.9	2.5	5.8
35	4.56	.56	11.7	147	40	5.83	5.9	2.5	5.9
30	4.58	.57	11.9	150	36	5.83	5.4	2.3	5.7
25	4.6	.59	12.0	151	34	5.83	5.4	2.3	5.8
20	4.62	.61	12.1	153	32	5.83	5.3	2.2	5.7
15	4.65	.64	12.3	155	29	5.83	5	2.1	5.8
10	4.68	.67	12.3	155	29	5.83	5.3	2.2	6.1
5	4.70	.71	12.4	156	27	5.83	5.2	2.2	6.3
0	4.72		12.4	156	27				

Table 8. Parameters of flexor torque at various angles of elbow flexion for contractile tissue of Brachioradialis; prediction of maximal isometric flexor torque from the contractile tissue of this muscle.

Key to the table by column follows:

A ... angle of flexion
B ... "r" in cms
C ... $\sin \beta$
D ... absolute length of the contractile tissue (average of two trials) in cms
E ... %L_o
F ... %P_o
G ... "force units"
H ... torque generated by the contractile tissue, force units
I ... torque generated by the contractile tissue, nt-m's

Table 8

A	B	C	D	E	F	G	H	I
125	21.9	.28	20.08	100	100	2.27	13.9	5.87
120	21.9	.29	20.05	100	100	2.27	14.4	6.1
115	21.9	.31	20.02	100	100	2.27	15.4	6.5
110	21.9	.32	19.97	100	100	2.27	15.9	6.7
105	21.9	.33	19.94	100	100	2.27	16.4	6.9
100	21.9	.33	19.92	100	100	2.27	16.4	6.9
95	21.9	.34	19.90	100	100	2.27	16.9	7.1
90	21.9	.34	19.90	100	100	2.27	16.9	7.1
85	21.9	.34	19.90	100	100	2.27	16.9	7.1
80	21.9	.33	19.93	100	100	2.27	16.4	6.9
75	21.9	.33	20.06	100	100	2.27	16.4	6.9
70	21.9	.32	20.29	100	100	2.27	15.9	6.7
65	21.9	.31	20.54	100	100	2.27	15.4	6.5
60	21.9	.29	20.76	100	100	2.27	14.4	6.1
55	21.9	.28	20.80	100	100	2.27	13.9	5.8
50	21.9	.26	21.19	100	100	2.27	12.9	5.4
45	21.9	.24	21.38	100	100	2.27	11.9	5.1
40	21.9	.22	21.56	100	100	2.27	10.9	4.6
35	21.9	.19	21.73	100	100	2.27	9.5	4.0
30	21.9	.17	21.87	100	100	2.27	8.5	3.6
25	21.9	.14	22.00	100	100	2.27	7.0	2.9
20	21.9	.12	22.10	100	100	2.27	6.0	2.5
15	21.9	.09	21.18	100	100	2.27	4.5	1.9
10	21.9	.06	22.24	100	100	2.27	3.0	1.3
5	21.9	.03	22.29	100	100	2.27	1.5	.6
0	21.9		22.33	100	100	2.27		

Table 9. Total maximal isometric flexor torque from the contractile tissue of all 3 major flexors, Biceps Brachii, Brachialis, Brachioradialis, according to angle of elbow flexion and in both positions of the shoulder. Normalization of torques with respect to torque generated at $\alpha = 90^{0}$, shoulder flexed. Key to the table by column follows:

A ... angle of flexion

B ... total torque from the contractile tissues with the shoulder flexed, nt-m's

C ... percentage of normal at $\alpha = 90^{0}$, shoulder flexed

D ... total torque from the contractile tissues with the shoulder abducted, nt-m's

E ... total torque generated in shoulder abduction as a percentage of normal at $\alpha = 90^{0}$, shoulder flexed

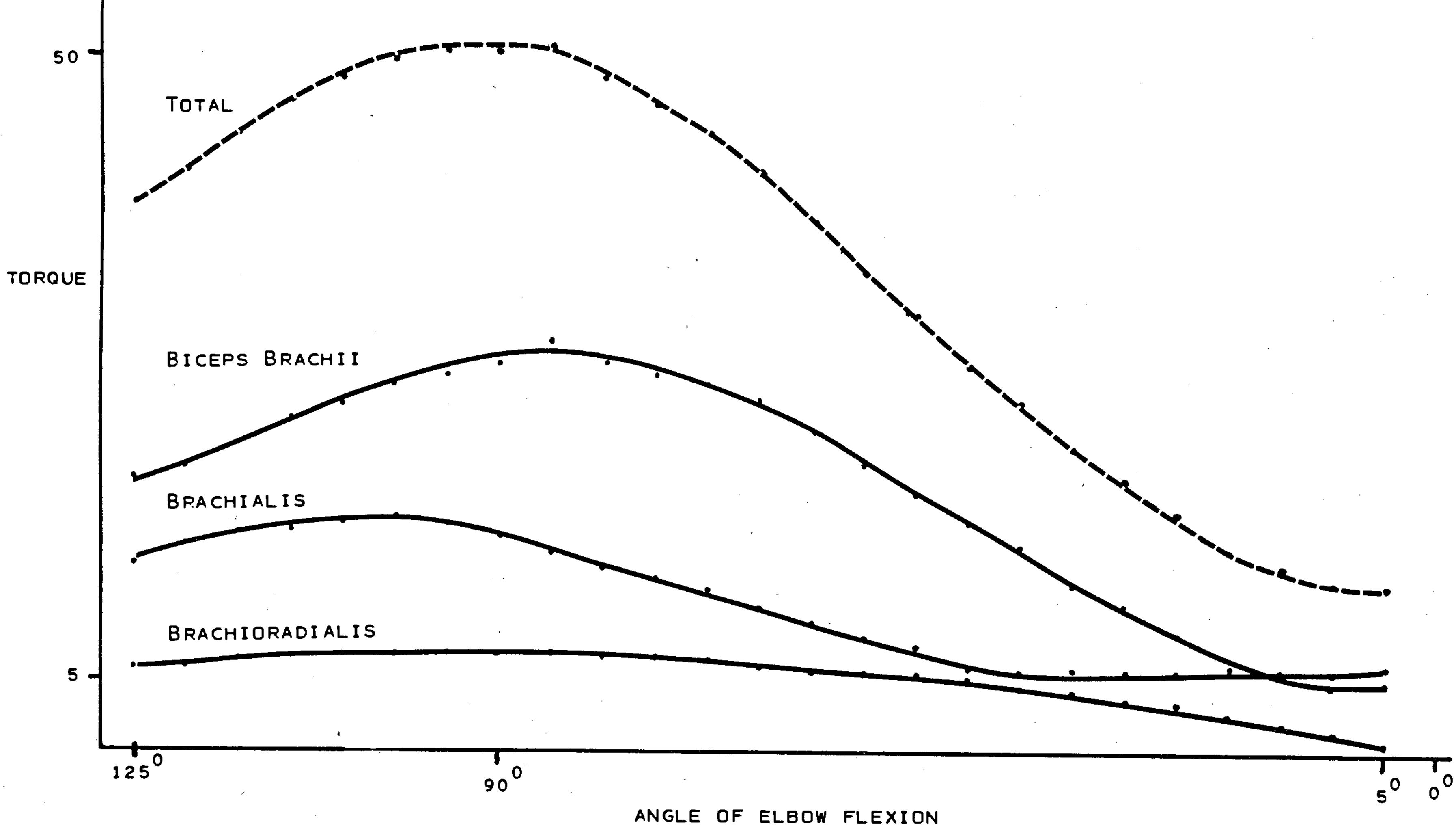
Table 9

A	B	C	D	E
125	39.5	79%	41.8	83
120	41.8	83	44.4	88
115	44.4	88	46.8	93
110	46.9	93	48.9	97
105	48.4	96	50.3	100
100	49.7	99	50.5	100
95	50.3	100	50.5	100
90	50.3	100	49.4	98
85	49.5	98	47.5	94
80	48.2	96	45.2	90
75	46.6	93	42.9	85
70	44.5	89	40.5	81
65	41.9	83	37.5	75
60	38.2	76	34.0	68
55	34.6	69	30.6	61
50	31.6	63	27.7	55
45	27.8	55	24.1	48
40	25.2	50	21.5	43
35	22	44	19.0	38
30	19.6	39	16.8	33
25	17.2	34	14.9	30
20	14.2	28	12.6	25
15	13.2	26	11.6	23
10	12.1	24	10.7	21
5	11.9	24	10.3	20
0	11.9	24	10.3	20

Figure 13, which follows, is the predicted torque/angle of flexion curve for each of the 3 main flexors and the summation of the maximal isometric flexor torques of their contractile tissues at each angle, α.

Figue 13. Theoretical relationship between torque and angle of elbow flexion of muscles individually and combined, shoulder flexed.

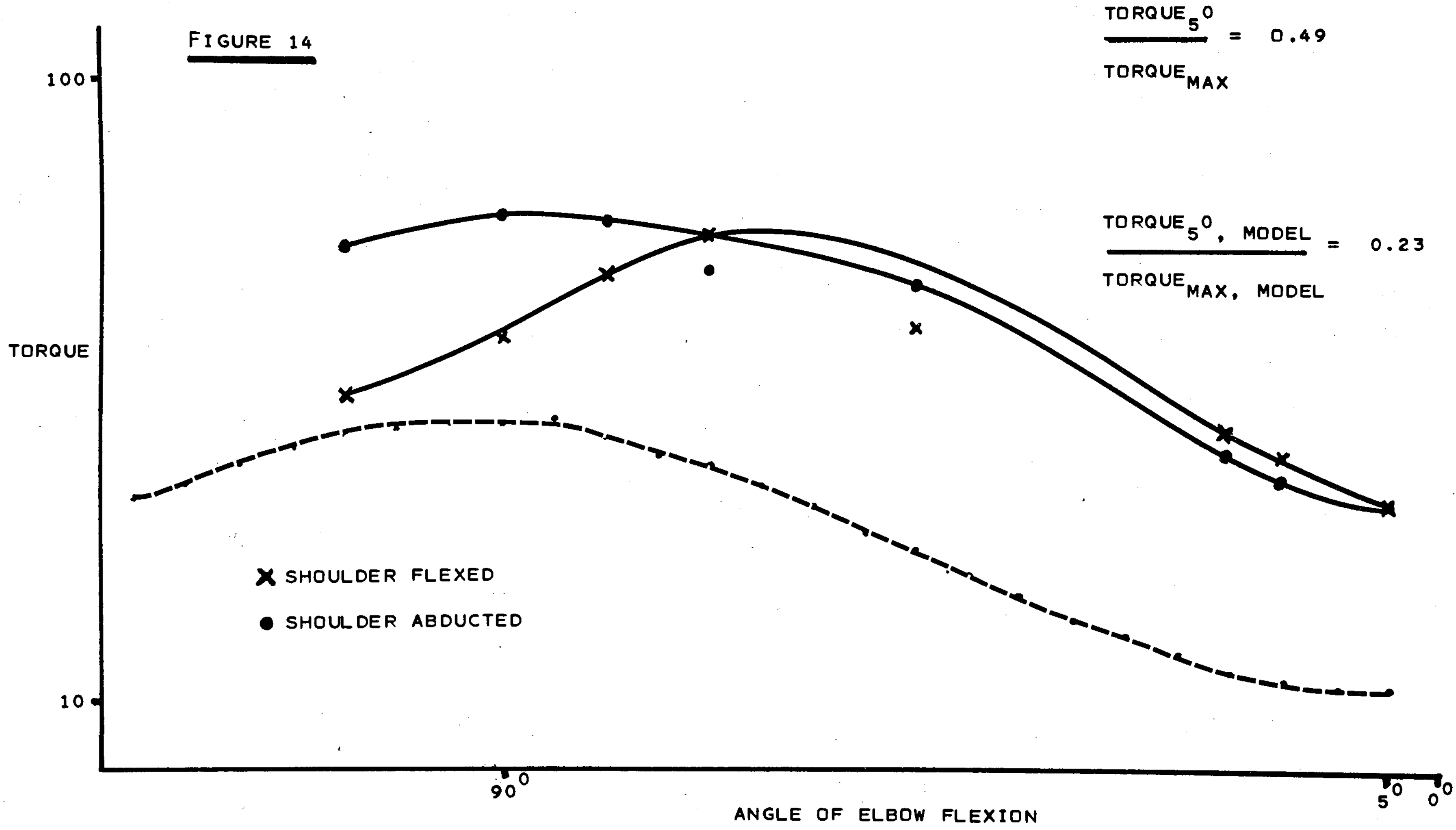
FIGURE 13


50
TORQUE
5
TOTAL
BICEPS BRACHII
BRACHIALIS
BRACHIORADIALIS
125°
90°
5°
0°
ANGLE OF ELBOW FLEXION

Figures 14 to 19 are the empirical torque/angle of flexion curves obtained from maximal voluntary contraction on living subjects. On each graph is superimposed the theoretical curve of Figure 11 drawn to scale. A method of comparing the curves, theoretical and empirical, is to compare the ratio of torque$_5$o to maximum torque in each curve and these numbers are included in the figures.

Figure 14. Maximal isometric flexor torque generated by A. N. at various angles of elbow flexion and in two positions of the shoulder, shoulder flexed in the transverse plane and shoulder abducted in the transverse plane. Super-imposed on the same scale is the model of maximal isometric flexor torque predicted from measurements on the cadaver, ---.

FIGURE 14



$$\frac{\text{TORQUE}_{5^0}}{\text{TORQUE}_{\text{MAX}}} = 0.49$$

$$\frac{\text{TORQUE}_{5^0}\text{, MODEL}}{\text{TORQUE}_{\text{MAX, MODEL}}} = 0.23$$

Figure 15. Maximal isometric flexor torque/angle of flexion curve generated by R. C. The modelled curve is super-imposed as ---.

FIGURE 15

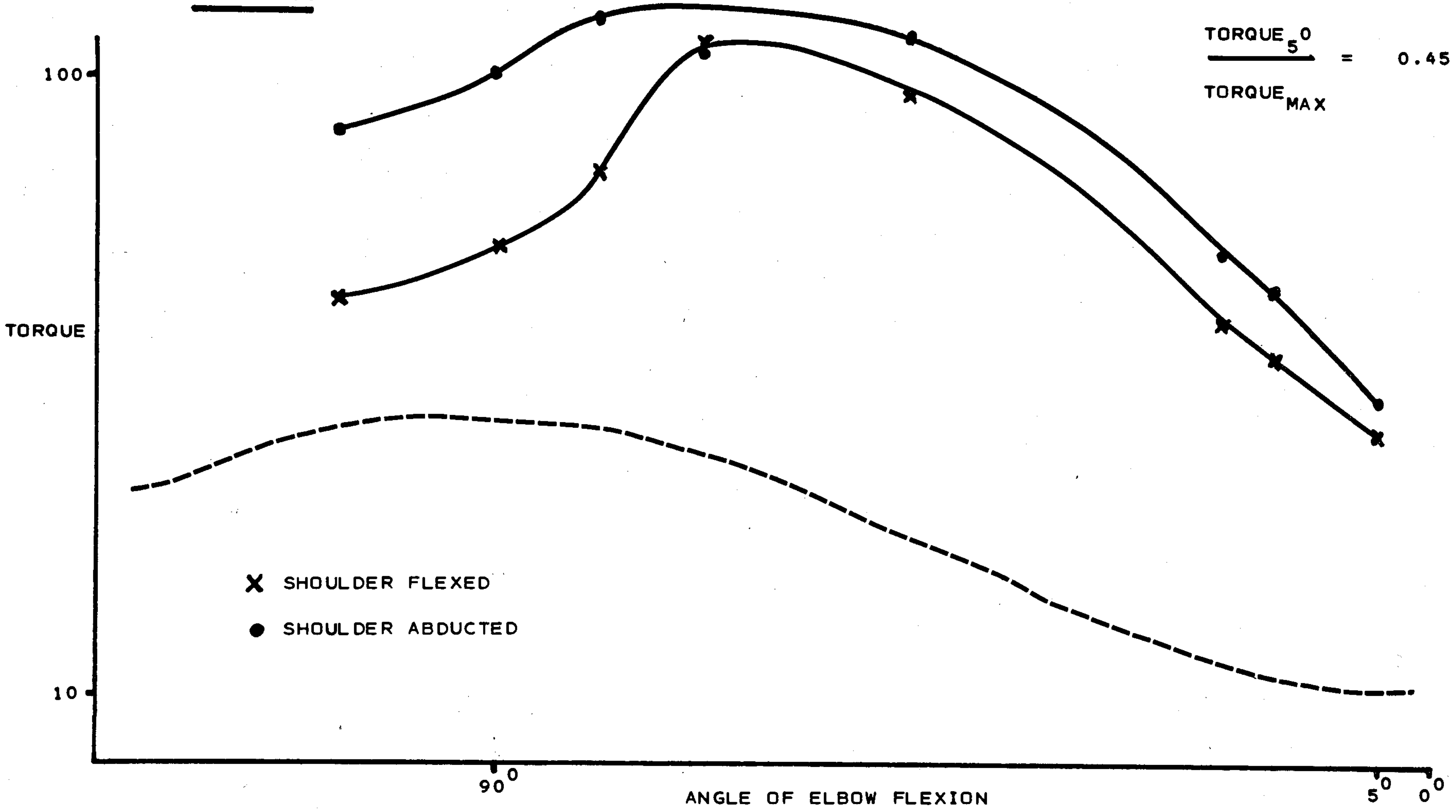

TORQUE
100
10
ANGLE OF ELBOW FLEXION
90°
5°
0°
TORQUE$_{5°}$ / TORQUE$_{MAX}$ = 0.45
X SHOULDER FLEXED
● SHOULDER ABDUCTED

Figure 16. Maximal isometric flexor torque/angle of flexion curves generated by W. B. The modelled curve is super-imposed as ---.

FIGURE 16

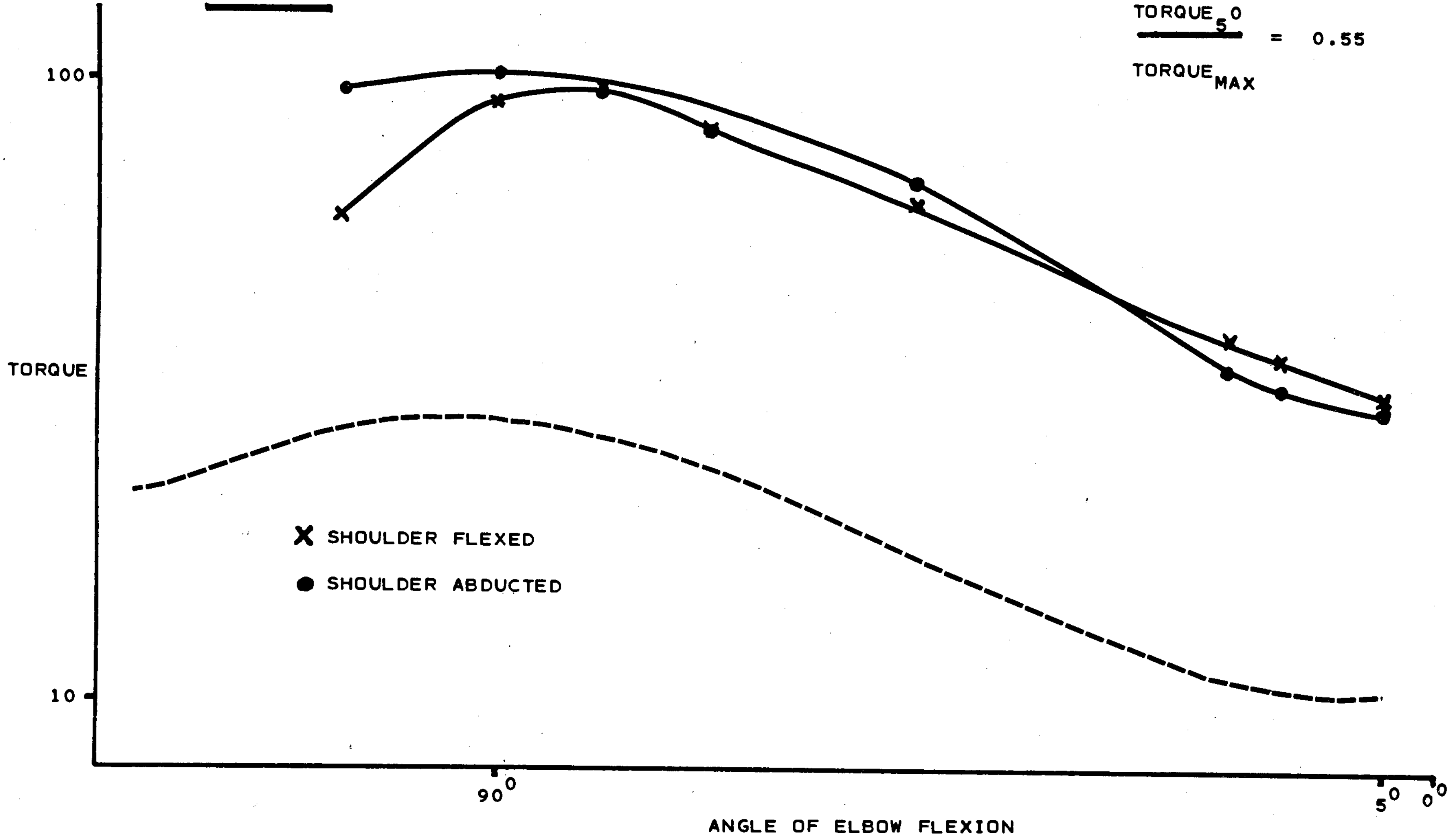

$\frac{TORQUE_{5^0}}{TORQUE_{MAX}} = 0.55$
100
TORQUE
10
X SHOULDER FLEXED
● SHOULDER ABDUCTED
90°
5° 0°
ANGLE OF ELBOW FLEXION

Figure 17. Maximal isometric flexor torque/angle of flexion curves generated by H. W. The modelled curve is super-imposed as ---.

FIGURE 17

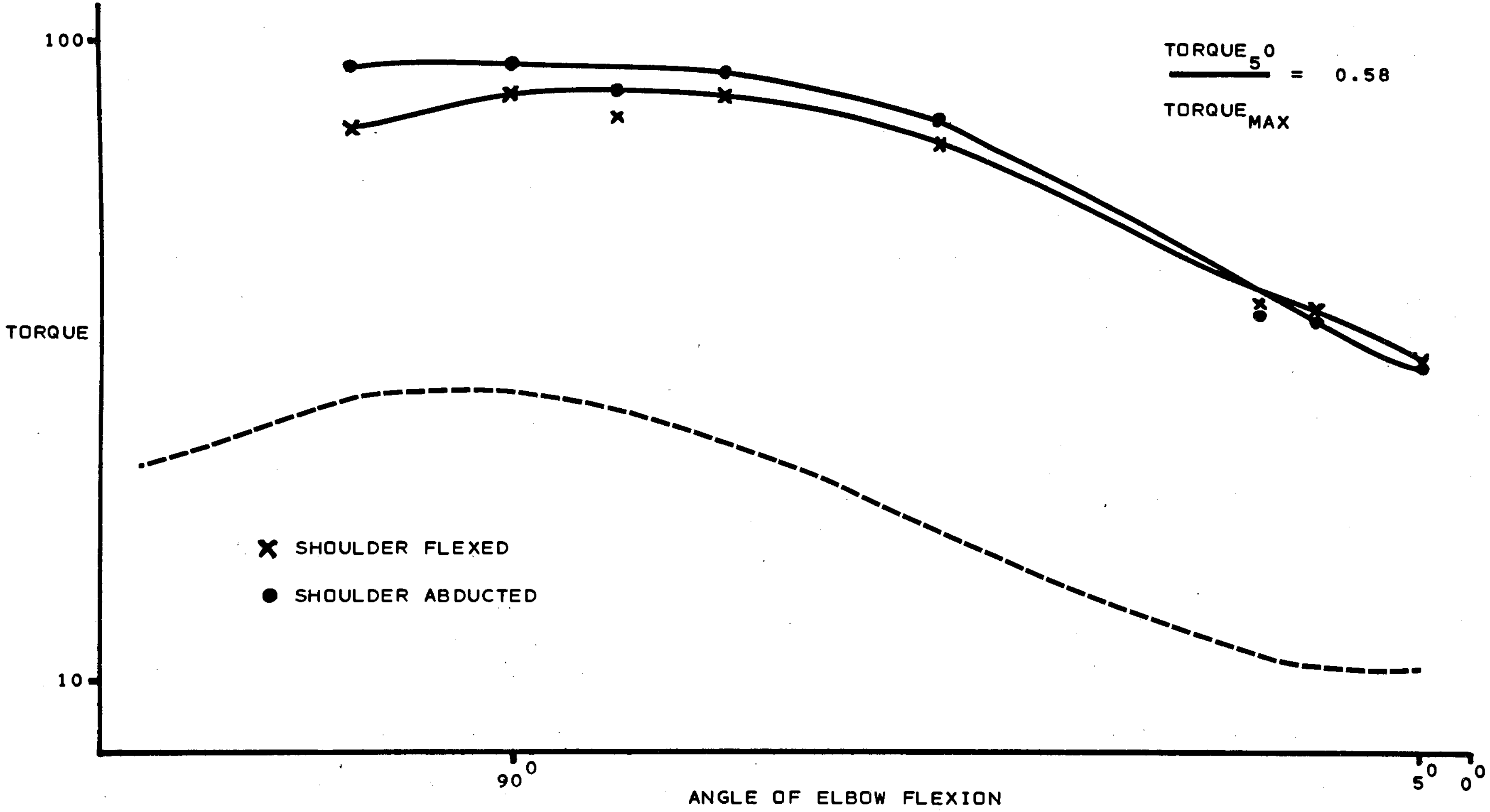

$\frac{TORQUE_{5^0}}{TORQUE_{MAX}} = 0.58$
100
TORQUE
10
90°
ANGLE OF ELBOW FLEXION
5°
0°
✕ SHOULDER FLEXED
● SHOULDER ABDUCTED

Figure 18. Maximal isometric flxor torque/angle of flexion curves generated by D.E.

The modelled curve is super-imposed as ---.

FIGURE 18

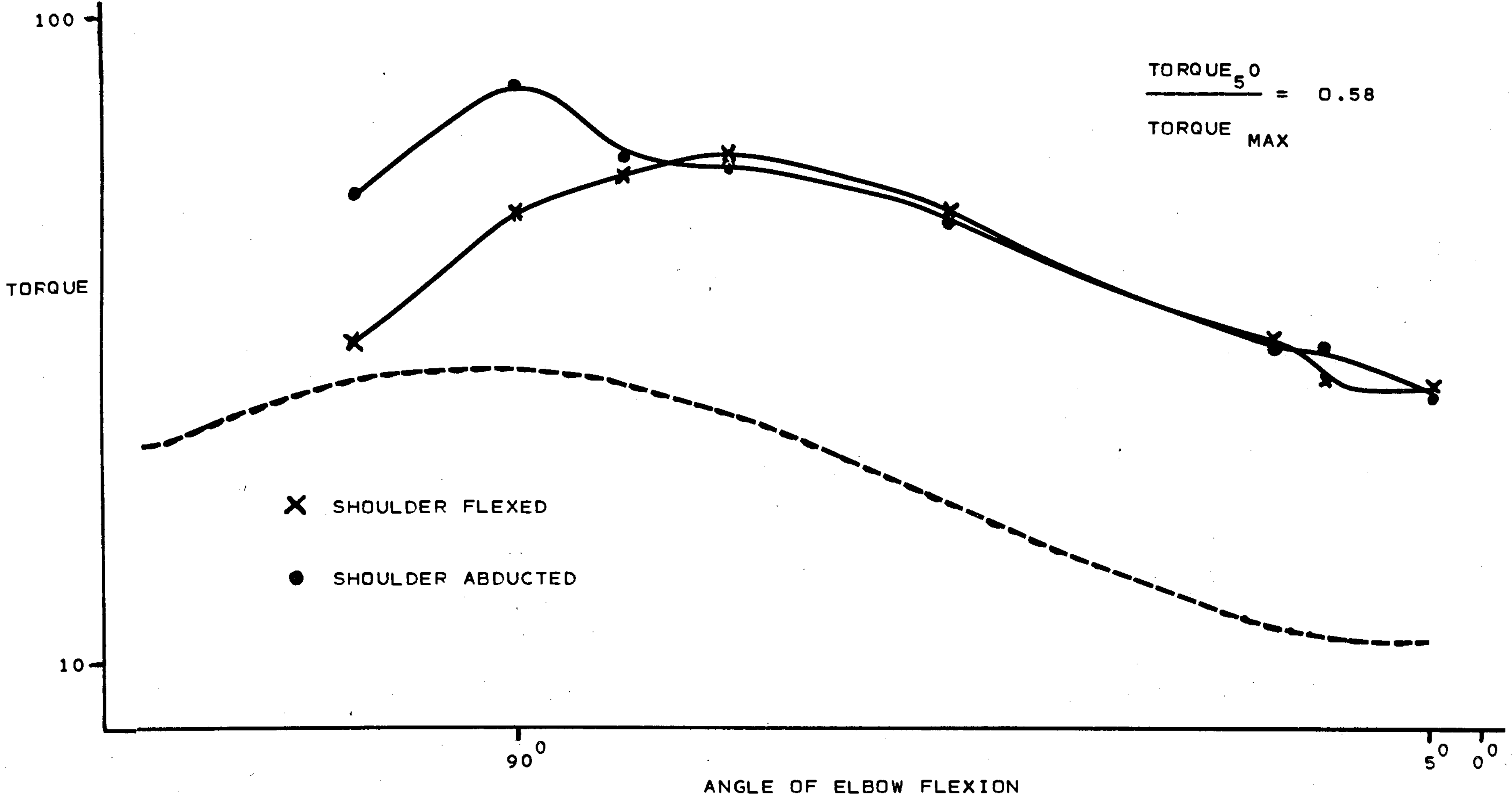
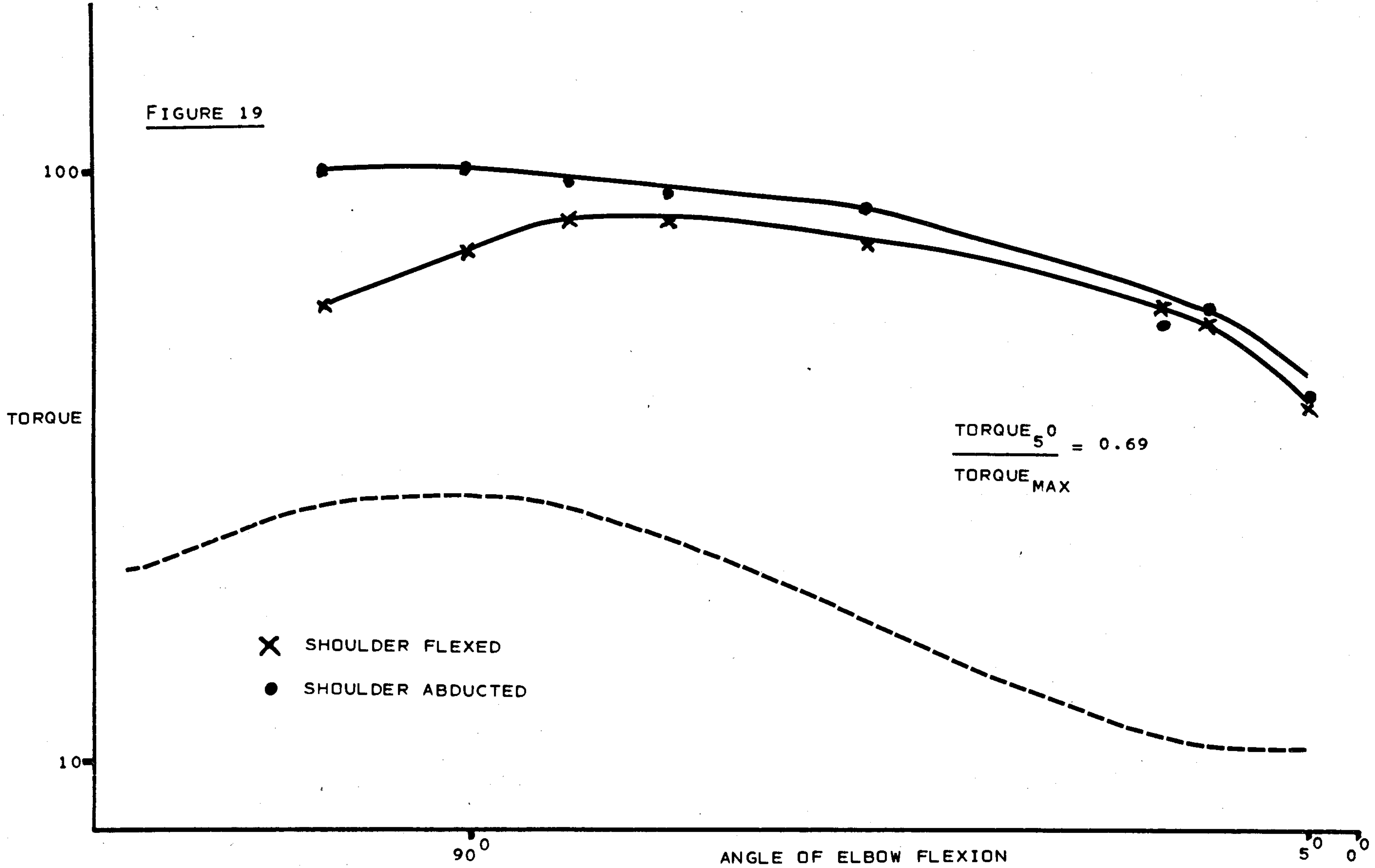

TORQUE
100
10
TORQUE$_5$0 / TORQUE MAX = 0.58
SHOULDER FLEXED
SHOULDER ABDUCTED
90^0
5^0
0^0
ANGLE OF ELBOW FLEXION

Figure 19. Maximal isometric flexor torque/angle of flexion curves generated by M. C. The modelled curve is super-imposed as ---.

FIGURE 19


100
TORQUE
10
$\frac{\text{TORQUE}_{5^0}}{\text{TORQUE}_{\text{MAX}}} = 0.69$
✕ SHOULDER FLEXED
● SHOULDER ABDUCTED
90⁰
ANGLE OF ELBOW FLEXION
5⁰
0⁰

In the curve predicted by the model for the flexed position of the shoulder, the maximal isometric flexor torque is exerted at $\alpha = 90^{\circ}$. To facilitate comparison between the two predicted curves and the pairs of curves generated by each of the subjects, the curves were all normalized about the torque at $\alpha = 90^{\circ}$ in the flexed position of the cadaver (Tables 9, 10). The data from Table 10 was examined for means and standard deviations (Table 11). Finally, using the mean lengthening of Biceps Brachii, Short Head, in accommodating movement of the shoulder from flexed to abducted positions in all the subjects, the theoretical torque/angle of flexion curve in the abducted position of the shoulder was generated for the cadaver (Table 5) and normalized (Table 9). A final juxtaposition of all the curves is presented in Figure 20.

Table 10. Torques (nt-m's) generated in two shoulder positions for each subject. For each subject the upper figure is the torque generated with shoulder abducted, the lower figure with the shoulder flexed. The torques are normalized for each subject about the torque he generates at $\alpha = 90^\circ$, shoulder flexed.

Table 10

Subject	105°	90°	80°	70°	50°	20°	15°	5°
A.N.	76 - 123% 54 - 87%	80 -129% 62 -100%	79 -127% 72 -116%	73 -118% 77 -124%	70 -113% 64 -103%	46 -74% 48 -77%	42 -68% 46 -74%	38 -61% 38 -61%
R.C.	92 -123% 68 - 91%	100-133% 75-100%	108-144% 86-115%	103-137% 105-140%	106-141% 98-131%	74 -99% 64 -85%	69 -92% 59 -79%	52 -69% 48 -64%
W.B.	98 -101% 80 -82%	100-103% 97-100%	98-101% 98-101%	93 -96% 93 -96%	86 - 89% 81 - 84%	58 - 60% 62 - 64%	56 -58% 59 -61%	52 -54% 54 -56%
H.W.	97 -105% 87 - 95%	97 -105% 92 -100%	92 -100% 89 - 97%	96 -104% 92 -100%	89 - 97% 86 - 94%	61 - 56% 63 - 69%	60 -65% 62 -67%	53 -58% 53 -58%
D.E.	75 -104% 55 - 76%	90 -125% 72 -100%	81 -113% 78 -108%	79 -110% 81 -113%	72 -100% 71 - 99%	53 - 74% 55 - 76%	53 - 74% 49 - 68%	46 -64% 48 -67%
M.C.	100 -115% 80 - 92%	100-115% 87-100%	98 -113% 92 -106%	97 -112% 92 -106%	95 -109% 89 -102%	77 - 89% 79 - 91%	79 - 91% 77 - 89%	65 -75% 65 -75%

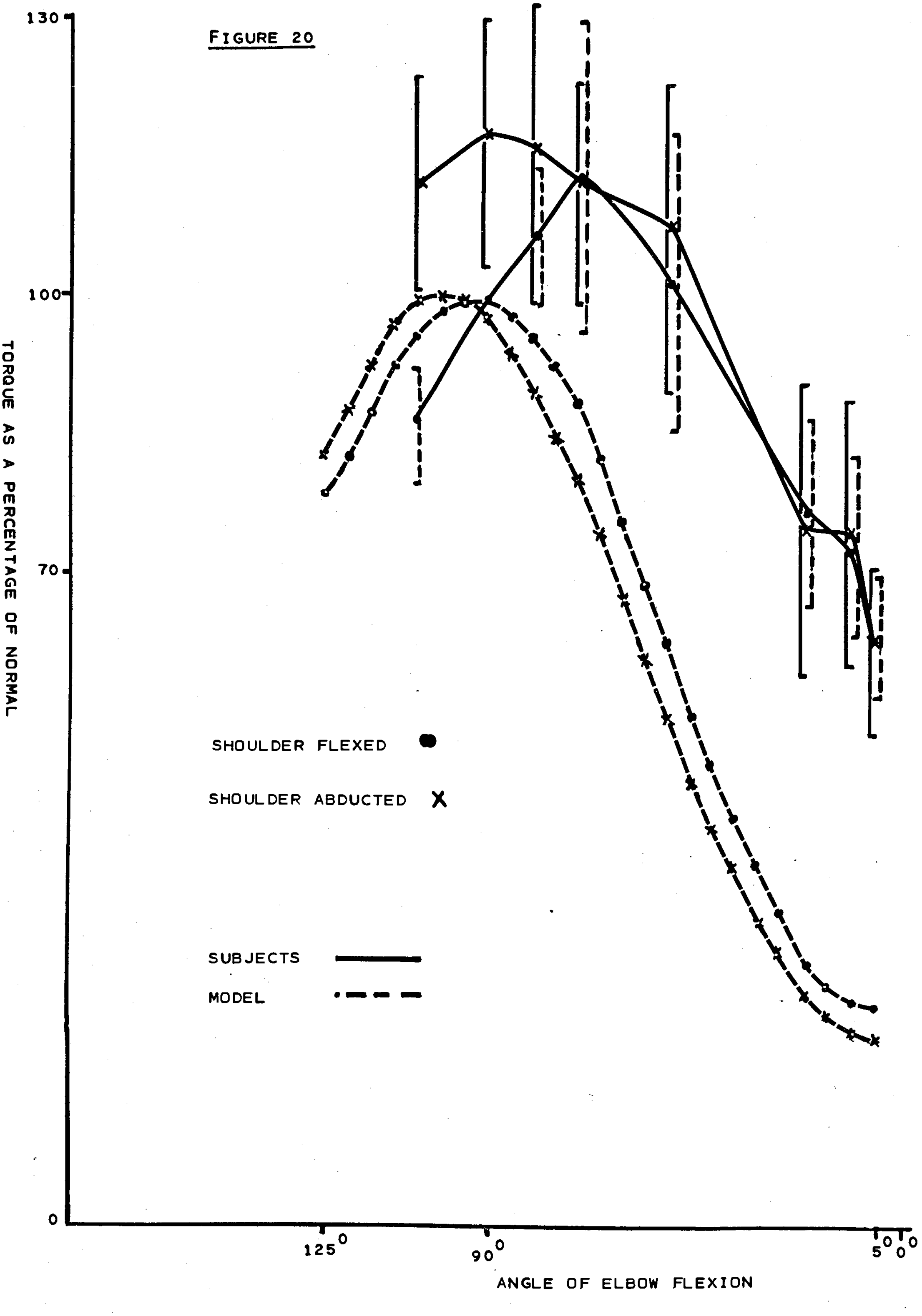
Table 11. Mean and standard deviations for torque at various α in both positions of the shoulder, torque expressed as a percentage of normal at $\alpha = 90^{\circ}$, shoulder flexed.

Table 11

		105°	90°	80°	70°	50°	20°	15°	5°
Shoulder flexed	$\bar{X}$	87	100	107.1	113.2	102	77	73	63.5
	δ^2	50	0	56.6	271	248	99	99.6	47.5
	δ	7	0	7.5	16.5	15.7	9.9	9.98	6.89
Shoulder abducted	$\bar{X}$	111.8	118.3	116.3	112.8	108.2	75.3	74.7	63.5
	δ^2	96.9	159	280	196	332	272	197	57.9
	δ	9.85	12.6	16.8	14	18.2	16.5	14	7.6

Figure 20. Torque/angle of flexion curves predicted for both abducted and flexed positions of the shoulder in the cadaver and the mean torque/angle of flexion curves with standard deviations for the 6 subjects in both abducted and flexed positions of the shoulder--expressed as a percentage of the predicted torque at $\alpha = 90^{\circ}$ with the shoulder flexed.

FIGURE 20


130
100
70
0
TORQUE AS A PERCENTAGE OF NORMAL
125°
90°
50°
ANGLE OF ELBOW FLEXION
SHOULDER FLEXED ●
SHOULDER ABDUCTED X
SUBJECTS
MODEL

DISCUSSION

There are similarities and, yet, important differences between the predicted and the measured curves of maximal isometric flexing torque/angle of elbow flexion. This chapter will compare the model with the empirical curves, attempt to account for the differences between them, and suggest improvements in the design of the experiment.

In terms of similarities, the model of maximal siometric flexing torque generated here using the change in length of the contractile tissue, the lever arm, and the angle of insertion of each of the "three major flexors" (Basmajian, 1957) resembles the empirical curves in the following ways: The model has the same general shape as the empirical curves including an inflexion at small angles of elbow flexion; the maximum torque predicted by the model is of the same order as that generated by each of the six subjects, i.e., 55 nt-m's compared to 90 nt-m's; the angle at which the model predicts maximal isometric flexing torque is close to (but always less than) the angles of flexion at at which the subjects generated mazimal torque.

In addition to finding these similarities between the modelled and the empirical curves, the experiment has demonstrated two conditions *in vivo*: First, that enormous length changes are imposed upon the contractile tissue in flexion of the elbow through its range of motion. Second, it has disclosed at least 3 features of anatomy which determine the inflexion in the curve at small angles of elbow flexion. These are the pulley effects imposed on the flexors by the trochlea and the capitulum; the increase in β for Brachialis as the elbow moves through the last 30^o to full extension and, thereby, the increase in torque generated by Brachialis as the elbow works at smaller and smaller angles of elbow flexion; the bulking effect of Brachialis which, in contraction, displaces Biceps Brachii anteriorly and increases the angle, β , at which the Biceps Brachii insert on the radius (or, at least, disturbs the precise α at which Biceps Brachii comes into contact with the capitulum and, thereby, begins to incur a pulley effect.

There are 3 striking differences between the modelled curve and the experimental curves. The model predicts a lower maximal isometric flexor torque than consistently was observed empirically--the model anticipates a maximum torque of 55 nt-m's whereas the subjects could generate 90 to 100 nt-m's. The magnitude of the maximum torque is dependent largely on the estimate of effective muscle cross section. The model reduced muscle architecture to a uniform cylinder and this simlification could underestimate the real torque generated by the muscles in vivo. In addition, these measurements were made on a sedentary, middle-aged male and the empirical curves were generated by national class athletes trained in strength events. Still, the order of the torques, modelled and empirical, is the same.

A second and important difference between the modelled and the empirical curves is that, while the angles of flexion at which the model predicts and the subjects generate maximal isometric flexor torque are quite close to each other, the angles differ consistently. The model predicts that maximum torque will be generated at 90° of elbow flexion; the empirical curves have their maxima at a smaller angle of flexion, generally between 70° and 80°. It appears that there are at least two reasons for this difference. First, the model assumes that the length change imposed on the muscular structures by the excursion of the joint is expressed only in the contractile tissue. In vivo, however, there is a compliance of the elastic elements and in maximal isometric flexion the contractile tissue will shorten with this compliance. The compliance of the series elastic tissue has been variably estimated bweeen 3% (Hill, 1938) and 15% (Wilkie, 1950). With such a compliance L_o would be found at a smaller angle of elbow flexion than the model infers and this, in turn, would move the locus of maximal isometric flexor torque out to smaller angles of elbow flexion. A second explanation for the difference in the locus of maximal isometric flexor torque between the model and the empirical data is that no quantitative role has been ascribed to the several other elbow flexors in the model (see below) and, extrapolating from the present data on the work of Brachioradialis, torque generated by the minor flexors would maintain the plateau of the torque/angle of flexion curve and indeed displace its maximum towards smaller angles of elbow flexion in vivo.

In examining the curves for both positions of the shoulder, Biceps Brachii,Short Head, was longer in shoulder abduction at any given and this shifted the locus of the curve approximately 10°. This argues that, to shift the theoretical curves to the right to more closely approximate the empirical data, the lengths of all the flexors whould be shorter at any given α than the present model assumes. Now, if the series elastic compliance in vivo did not accommodate this effective shortening, the consequence would be that the contractile tissue entered the D zone at the shortest anatomical lengths in contrast to the published work of Ramsey and Street (1940) and of Moss (1968).

The most important difference between the model and the empirical data, however, is that the ratio of torque generated at $\alpha = 5^\circ$ and the maximum torque generated anywhere was 0.23 in the model and between 0.45 and 0.69 in the 6 subjects. This may reflect, in part, the smaller muscle mass of the cadaver and, thereby, a larger effect from parallel elastic tissue; it may also reflect the assumptions and simplifications in the model referred to below.

As well, the model predicts a more rapid fall in torque than the subjects recorded in efforts at smaller angles of elbow flexion even though inspection of the graphs suggests that the slopes are similar. There are two possibilities here: Either the modelled curve should not drop off as quickly as it does (that it would implies an error in concept and/or in measurement), or else there are forces at work in vivo but not measured in the model which tend to keep torque high at small angles of elbow flexion.

In dealing with the first possibility, i.e., that the model predicts too rapid a fall in torque, consider the treatment of F, β, and r in the model. In this experiment the measure of length change of the contractile tissue was reliable, separate trials yielding the same result ± 3%. Full excursion of the elbow required the contractile tissue to move between a length of -20% and +40% of an optimal length, L_o. This enormous change in length would reduce the contractile force of the contractile tissue at the extremes of lengthening and shortening (Gordon et al., 1966). Is it possible that elastic tissues contribute to flexor torque at small angles of flexion?

Several workers have measured the passive force of contraction developed by intact muscles at lengths greater than L_o (Ramsey et al., 1940; Inman et al., 1953; Fidelus, 1968). In view of the extraordinary length to which the contractile tissues extend at small angles of elbow flexion, a curve of isometric flexor torque was generated using Fidelus' data to incorporate a contribution from the parallel elastic tissue in the total flexor torque. The result is tabulated in Table 12, and the curve of maximal isometric flexor torque/angle of flexion then becomes Figure 21. While this manipulation raises the ratio of torque at $\alpha = 5^{\circ}$/maximum torque from 0.23 to 0.42, the large contributions which it predicts from the parallel elastic tissue at small angles of elbow flexion cannot be justified. For example, at $\alpha = 5^{\circ}$ this modification to the model predicts a torque of 11.9 nt-m's from the contractile tissues of Biceps Brachii, Brachialis, and Brachioradialis, and a total of 10.2 nt-m's of torque from all the parallel elastic tissues. It is unreasonable to expect that half the torque at small angles of elbow flexion should be generated by the parallel elastic tissue and that this does not pertain is apparent in the empirical curves where active flexion generated five times the absolute falue of the resting torque measured at small angles of elbow flexion. Moreover, a rather small contribution from parallel elastic tissue <u>in vivo</u> is confirmed by Hayes (1977) who found it to be less than 2.5 nt-m's at full elbow extension.

A clearer shortcoming in the computation of $\mathcal{F}$ in the model and one which would inappropriately reduce torque at small angles of flexion is the approximation of the contribution from Brachialis. This muscle originates from a large area of humerus and the model halved the muscle into a proximally originating and distally originating part. This simplification commits the distally originating portion of Brachialis to an extraordinary elongation at small angles of elbow flexion so that the capacity of this muscle to generate force at small angles of elbow flexion is unrealistically compromised. Indeed, measurements on the cadaver showed that at small angles of elbow flexion the torque generated by Brachialis begins to rise. This finding was also mentioned by Sanderson (1975). Therefore the model seriously underestimates the contributions to maximal isometric flexor torque at small angles of elbow flexion by this muscle which has been called the "workhorse of elbow flexion" by other researchers (Basmajian et al., 1957).

A second cogent criticism of the computation of $\mathcal{F}$ here is

Table 12. Flexor torque generated by the parallel elastic tissues at various angles of elbow flexion (after Fidelus, 1968); summation of torques from both the contractile tissues in maximal isometric flexion and the parallel elastic tissues of the 3 major flexors at various angles of elbow flexion. Key to the table by column follows:

A ... angle of flexion

B ... torque generated by parallel elastic tissue of Biceps Brachii, Short Head, nt-m's

C ... torque generated by parallel elastic tissue of Biceps Brachii, Long Head, nt-m's

D ... torque generated by parallel elastic tissue of Brachialis, proximal portion, nt-m's

E ... torque generated by parallel elastic tissue of Brachialis, distal portion, nt-m's

F ... torque generated by parallel elastic tissue of Brachioradialis, nt-m's

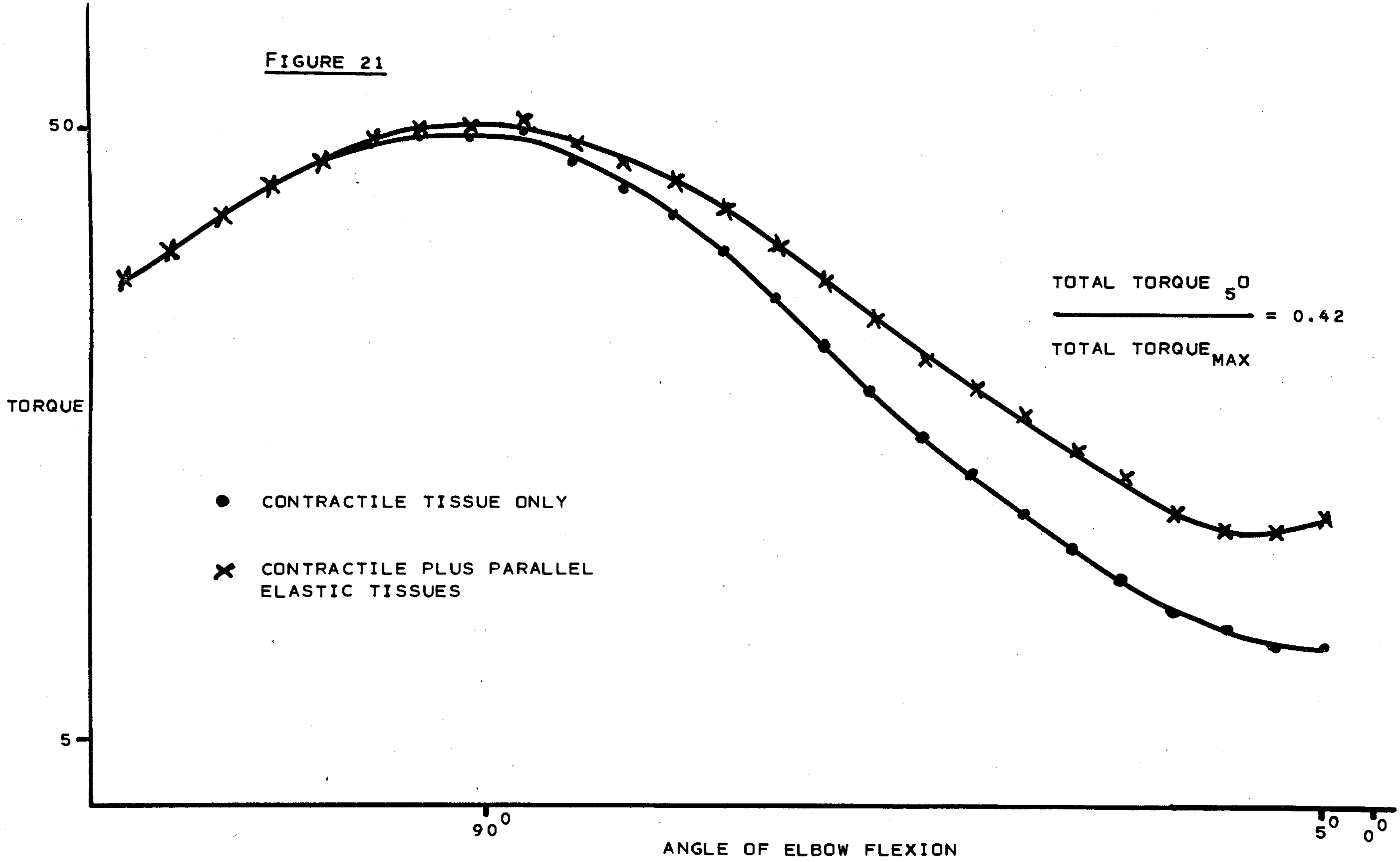
G ... total torque generated by parallel elastic tissue of all 3 major flexors, nt-m's

Table 12

A	B	C	D	E	F	G
125	0	0	0	0	0	39.5
120	0	0	0	0	0	41.8
115	0	0	0	0	0	44.4
110	0	0	0	0	0	46.9
105	0	0	0	0	0	48.5
100	0	0	0	0.3	0	50
95	0	0	0	0.5	0	50.8
90	0	0	0	0.6	0	50.9
85	0	0	0	1.0	0	52.2
80	0	0	0.1	1.3	0	49.6
75	0	0	0.3	1.4	0	48.3
70	0.4	0.5	0.3	1.6	0	47.3
65	0.6	0.6	0.4	1.7	0	45.2
60	0.7	0.6	0.4	2.1	0	42.0
55	1.0	0.9	0.4	2.3	0	39.2
50	1.0	1.0	0.5	2.7	0	36.8
45	1.4	1.1	0.4	3.0	0	33.7
40	1.5	1.2	0.5	3.2	0	31.6
35	1.5	1.4	0.5	3.6	0	29.0
30	1.4	1.3	0.6	3.8	0	26.7
25	1.3	1.2	0.7	4.1	0	24.5
20	1.1	1.0	0.7	5.0	0	22.0
15	1.1	1.0	0.8	5.6	0	21.7
10	1.1	1.0	0.9	5.9	0	21.0
5	1.3	1.3	1.0	6.6	0	22.1
0						

Figure 21. Prediction of total maximal isometric flexor torque/angle of flexion curve for Biceps Brachii, Brachialis, and Brachioradialis incorporating contributions from both the contractile tissues and the parallel elastic tissues (after Fidelus, 1968), shoulder flexed.

FIGURE 21



$$\frac{\text{TOTAL TORQUE}_{5^0}}{\text{TOTAL TORQUE}_{\text{MAX}}} = 0.42$$

the assumption that change in length of the muscular structures with excursion of the joint is registered solely by the contractile tissue. Other works have investigated the dynamics of the contractile and elastic tissue in muscular contraction and suggested that the elastic tissue extends between 3% and 15% when the contractile tissue is contracting maximally. In vivo, this would allow the contractile tissues to generate P_o at a smaller angle of flexion and thereby maintain higher torques than predicted at smaller angles of flexion.

As the computation of F has been somewhat inaccurate, so, in retrospect, are the assumptions made in estimating the product $\sin\beta \cdot r$. While the experiment measured and calculated β by a method validated as long ago as 1890 (Braune and Fischer), the model assumed that the flexors inserted through discreet tendons. Even if this were so, there is in the live human a bulking effect of the shortening muscles by which, at any angle of elbow flexion, β is necessarily modified. But, this would be most prominent at large angles of elbow flexion and of little consequence at small angles where the muscles are elongated. Of more concern in accounting for the discrepancy between the model and the empirical curves at small angles of elbow flexion is the model's assumption that the contractile force of each flexor is directed simply through its tendon of insertion. In vivo this is not so and this is explained in the following discussion of the measurement of the lever arm, r.

r was established as the distance between the centre of rotation of the elbow and the "point" of tendonous insertion of each of the flexors. Once these "points" were established by inspection, the error in measuring r was ± 5%. Point insertion is a simplification for not only does the tendon insert upon an area of bone, but there is, in vivo, a network of antibrachial fascia interdigitating muscles and bone - this, and the bicipital aponeurosis of which we took no account, would cause a functional increase in r and would be of most influence at the smaller angles of elbow flexion where the network of fascia is stretched.

The foregoing section enumerated some of the features of the model which permitted the modelled curve to fall away too quickly at small angles of elbow flexion. A second way in which the discrepancy between the modelled and the empirical curves could be accounted for has to do with forces at work in vivo which were not allowed for by the model. In analyzing the model and the experimental conditions, there are indeed other forces contributing to torque which the model ignored and of which the subjects had advantage. The assumption that the flexors of the elbow are the Biceps Brachii, Brachialis and Brachioradialis is simplified. While these are commonly regarded as the major flexors Braune and Fischer (1890), Basmajian (1957) , Basmajian (1957), Davies (1967), and others have listed minor flexors of the elbow. These are muscles originating on the humerus, passing anteriorly to the axis of rotation of the elbow and inserting either on the forearm or at the hand and they include Pronator Teres, the common flexors including Flexor Carpi Radialis, Palmaris Longus, Flexor Carpi Ulnaris, Flexor Digitorum Superficialis, and, indeed, Extensor Carpi Radialis Longus. None of these minor flexors were included in the model even though, for example, Extensor Carpi Radialis weighed 75 grams and Brachioradialis only 50. The apparatus in which the subjects generated their curves of maximal isometric flexor torque inadvertently permitted them to use to full advantage these minor flexors of the elbow. The cast in which the hand and distal forearm were enclosed effectively fixed the wrist so that all the minor flexors were able to generate flexor torque at the elbow. The living individual, in generating maximal flexor torque at the elbow, will stabilize his wrist in much the way that the cast on the dymamometer stabilized the wrists of the experimental subjects, but this is not the movement modelled in this experiment where only Biceps Brachii, Brachialis and Brachioradialis were considered. The incongruency between the movement modelled and the movement measured makes a difference between the two curves inevitable.

There are two incidental findings of interest. For every subject, maximal isometric flexor torque in shoulder flexion was generated at a smaller angle of elbow flexion than the maximum torque in shoulder abduction. In moving the arm from shoulder flexion to shoulder abduction, in the transverse plane, only the short head of Biceps Brachii is affected very much and it is lengthened in order to accommodate the movement. It is inferred that in shoulder flexion the Short Head of Biceps Brachii achieves L_o and, thereby, P_o at a smaller angle of elbow flexion because at that smaller angle it will have lengthened enough to accommodate the shortened distance from insertion to origin at the corocoid process.

When the theoretical curves in the two shoulder positions are compared to the mean and standard deviations of each group of empirical curves (Figure 20), some important features are apparent. Firstly, the magnitude of torques differ as has been rationalized above. Secondly, there is great interindividual variation in the flexor torques recorded from the subjects. Thirdly, the empirical torques at small are quite scattered and inflections in the curve not always apparent in individuals. Indeed, Figure 20 has, at small α , a step with a final descent. This suggests that either $\hat{\tau}_{15^\circ}$ is too high (for reasons previously discussed) or $\hat{\tau}_{5^\circ}$ too low, perhaps because of pain experienced in sustained maximal isometric contraction at this angle. Fourthly, the role of change in length of the contractile tissue may not be as important in modifying output at two anatomical positions of the shoulder as other parameters undefined by the model. At large α the model predicts the trend in torque but the actual differences found between the two trials for each subject were greater than predicted; at small α the values were too close together for significant comparison.

The second incidental finding in the experiment was that empirical curves fell into three discreet groups, at least insofar as the ratio torque at $\alpha = 5^\circ$/maximum torque was concerned (Table 13):

$$\frac{\tau_{\alpha=5^\circ}}{\tau_{\mathrm{MAXIMUM}}}$$

ROWERS	WEIGHT LIFTERS	HAMMER THROWER
0.45	0.55	0.69
0.49	0.58	
	0.58	

Table 13: Ratio of maximal isometric flexor torque at $\alpha = 5^\circ$ to maximal isometric flexor torque in the six subjects, shoulder flexed.

Whether this ratio is a consequence of specific training (for example, a training which has more or less emphasis on the development of the common flexors) or whether the grouping of the athletes by ratio and sport is an instance of selection at the elite level of those, who, through endowment, are most likely to succeed, is uncertain.

Much of the recent work in integrated electromyography has attempted to quantify internal work. This experiment has attempted to measure in vivo the principle parameters of torque as they change with angle of flexion. Admittedly, the subject was a cadaver, the technique changed the architecture of the intact limb by dissecting out three major flexors from their investing fascia, and it ignored the minor flexors which, at small angles of elbow flexion, must contribute significantly to flexor torque. Nevertheless, this experiment has validated rather grossly the model $T = F \cdot \sin\beta \cdot r$ as it has been used here. The experiment demonstrated that the contractile tissue does pass through an extraordinary range of lengths in the full excursion of the elbow joint. The technique used in measuring change in the length of the contractile tissue was reliable, albeit Draconian, and quantified the role that the trochlea and the capitulum of the humerus play in maintaining flexor torque at small angles. The technique also demonstrated that each of the main flexors has a unique relationship to α in terms of the change in length of its contractile tissue and the variation in β. This clarifies the inaccuracy of invoking "a flexor equivalent" to stand for all flexor activity throughout the excursion of the joint. At the present time, it is not possible to rationalize this length change, the contribution of parallel elastic component to torque, and the published work of Fidelus (1968).

There are several limitations in the usefulness of the present model. Predictions from measurements on a single cadaver have been compared with empirical data which, within itself, shows wide inter-individual variation. There were no direct correlations found. It would appear that data collected from a large number of cadavers might be of value in modelling the performance of groups but that information taken from a single cadaver, as has been done here, would not be very useful in qualifying individual performance in vivo. Furthermore, the present model considers only 3 flexors, incorporates no quantitative contribution from series elastic tissue and makes only a crude approximation of muscle cross-section.

Significant improvements in the experimental design (which would improve the model if not completely rid it of its limitations) would include a more accurate estimation of muscular volume and effective cross-section. The wide disparity in assessments of cross-sectional areas in various experiments is pointed up by comparing the results of the present experiment with findings of typical cross-section in other experiments. Table 14 following tabulates these cross-sections as they were determined in the present experiment and in 3 others:

	Biceps Brachii	Brachialis	Brachioradialis
present data	13.2	9.7	2.3
Braune et al.	3.0	4.2	1.6
Schumacher et al.	3.59	4.62	1.37
Morris	10.66	11.48	4.88

Table 14: Cross-sectional areas of the 3 major flexors as determined in the present experiment, and by Braune and Fischer (1890), Schumacher and Wolff (1966) and Morris (1949). In cm^2.

It would be very difficult to arrive at a more accurate estimate of the functional lever arm of each flexor. A greater congruence between the parameters of the model and the *in vivo* performance to which we are comparing it, would be desirable. Finally, the empirical curves were drawn from torque measured at only eight angles of elbow flexion - several more points at small angles of flexion and around the locus of maximum torque would help to clarify the dependence of maximal isometric flexor torque on the angle of elbow flexion in the living subject.

SUMMARY AND CONCLUSIONS

There is a detailed literature of biomechanics studied in in vitro linear systems. This experiment develops a model of flexor torque in vivo, interpolates data from both the literature and present experimentation into a theoretical curve of maximal, isometric flexor torque/angle of flexion in the human elbow, and compares the predicted curve with curves generated by six athletes. In collecting data for the theoretical curve, some of the peculiarities of performance in vivo were accounted for; the comparison between the theoretical and the empirical curves was a test of the model.

A human cadaver was used in developing the theoretical curve. Because it simplified both the model and the experiment, maximal, isometric flexor torque at the elbow was modelled. After consideration of the literature, three flexors only were incorporated in the model, Biceps Brachii, both Long and Short Heads, Brachialis, and Brachioradialis. For each of these flexors, estimations were made of effective cross-section, and measurements were taken of length changes, lever arms and angles of insertion as each of these varied with angle of elbow flexion. These parameters are related in the following expression for maximal, isometric flexor torque, τ, at any angle of elbow flexion, α :

$$\tau_{\alpha} = \sum F_{i_{\alpha}}\left(\left(L/L_{0}\right)_{i_{\alpha}}, X\text{-section}_{i_{L_{0}}}\right) \cdot \sin\beta_{i_{\alpha}} \cdot r_{i_{\alpha}}$$

F is contractile force produced by the i'th muscle,
L_{0} is optimal length of the i'th muscle,
β is the angle of insertion of that muscle on its terminal link,
r is the lever arm on which the i'th muscle acts,
i is Biceps Brachii, Long Head, Biceps Brachii, Short Head, Brachialis, Brachioradialis.

The empirical curves of maximal, isometric flexor torque/angle of elbow flexion against which the theoretical curve was compared were generated by six national class athletes each working at two positions of the shoulder. Their performance was assessed for means and standard deviations; information gathered from the subjects was subsequently used to generate a second theoretical curve of maximal, isometric flexor torque/angle of elbow flexion for elbow flexion, this with the shoulder abducted. The four curves so produced were compared and their similarities and differences are discussed.

The model is grossly validated in this experiment. In particular, there are similarities in the empirical and theoretical curves with respect to the locus of maximum torque, in the order of torque predicted and measured, and in the presence of an inflexion of the torque/angle of flexion curve at small α . However there are consistent differences between each of the empirical curves and the theoretical curves and these are discussed.

The consistent differences were three. First, a smaller maximum torque was predicted (approximately 55 nt-m's) than was measured (79-110 nt-m's). This may be accounted for, in part, by the model's approximation of each flexor as a cylinder and, thereby, its reduction of the effective cross-sectional area of muscle available for force production.

Second, the angle of elbow flexion, α , at which the model predicts maximal, isometric flexor torque production is in every case larger than the angle of elbow flexion at which the subjects generated maximum torque. It is thought by the writer that this difference reflects, in part, the fact that the model allows for no compliance from series elastic tissue -- which compliance would <u>in vivo</u> effectively reduce the α at which the contractile tissue was at L_o. Furthermore, the model considers torque produced only by three major flexors, the Biceps Brachii, Brachialis, and Brachioradialis. <u>In vivo</u> there are other muscles which would contribute to maximal, isometric flexor torque (Pronator Teres, Flexor Carpi Radialis, Palmaris Longus, Flexor Carpi Ulnaris, Flexor Digitorum Superficialis, and Extensor Carpi Radialis Longus) and, while individually their contribution might be small, together it might be very significant in maximal, isometric flexor torque production. A significant contribution from the minor flexors would plateau the torque/angle of flexion curve in vivo and displace the locus of maximum torque production out to smaller α .

A third consistent difference between the theoretical and empirical curves is that the model predicts too rapid a fall-off in maximal, isometric flexor torque as the angle of flexion goes to zero — the ratio $\tau_{\alpha=5^\circ}/\tau_{maximum}$ is 0.23 in the model and ranges from 0.45 to 0.69 amongst the subjects. Such a rapid diminution in maximal flexor torque production cannot be accounted for by the fact that the model ignores flexor torque generated by elongated parallel elastic tissue at small angles of elbow flexion. While this conflicts with the published work of Fidelus (1968), most researchers have concluded that parallel elastic tissue contributes very little to net contractile force at long lengths of muscle in vivo. There are, however, four other circumstances, at least, which would, in the model, underestimate maximal, isometric flexor torque production at small α . The first of these is the approximation used in the model of reducing Brachialis to two representative fibres, a proximally originating and a distally originating one, both of them inserting at a single point. This approximation underestimates the torque contributed by Brachialis at lengths greater than L_o and would be particularly significant at those small α where the torque produced by the muscle, this "workhorse of flexion" (Basmajian et. al., 1957), begins to rise. A second condition which would serve to underestimate torque production at small α is the assumption of a "point insertion". *In vivo* the lever arm is determined by an area of tendonous insertion, there is interdigitation of antebrachial fascia which, at small α , could enlarge the area of effective insertion, and there is the bicipital aponeurosis which was not accounted for in this model at all. A third condition is compliance from the series elastic tissue which would help to contain elongation of the contractile tissues and, thereby, maintain their force production at smaller angles of flexion. A fourth condition which could result in the too-rapid diminution of predicted torque as the angle of elbow flexion diminishes is the exclusion of any contribution from the minor flexors mentioned above; furthermore, by fixing the subjects' wrists as was done by the casting on the dynamometer, the minor flexors were able to contribute all their force production to flexor torque recorded at the elbow by each subject. Therefore, some of the incongruence between the theoretical curve and the empirical curves must be attributed to the simplifications made by the model.

The experiment brings additional information to hand. In the cadaver it was apparent that enormous length changes are imposed on the contractile tissues of the flexors in accommodating full excursion of the elbow joint. Furthermore, the role of the capitulum and the trochlea as effective pulleys for the action of Biceps Brachii and Brachialis at smaller angles of elbow flexion was clarified. These length changes and anatomical pulleys make the notion of a single equivalent flexor acting through the excursion of the joint untenable; the notion has been used as a simplification by some researchers up to the present time.

Data gathered from the cadaver also rationalizes the inflection of the torque curve as the angle of inflection goes to zero. This inflexion would appear to result from at least three conditions: the bulking effect of the contracting muscles which would increase the angle of insertion that Biceps Brachii makes on the radius; the increasing angle of insertion which Brachialis has on the ulna as α goes from approximately 40° to 5°, and the two anatomical pulleys.

Data from the subjects, on the other hand, demonstrates two conditions: First, a wide inter-individual variation in the torque/angle of flexion curve and, second, an apparent grouping of individuals by sport in the ratio of torque at $\alpha = 5^{\circ}$ to maximum torque (Table 13). Whether the latter is a function of selection at the elite level, a function of training, or a function of both is not examined.

Given the use of a cadaver, important shortcomings in the experimental design are these: The estimation of effective cross-sections for the flexors from an approximation of each of them as cylinders; the particular simplifications made in reducing Brachialis to two fibres and a point insertion; the exclusion of compliance from the series elastic tissues in determining length of the contractile tissues at each angle of flexion; the inclusion of only three flexors in the model, all be they commonly held the major flexors; and the fixation of the wrists of each of the subjects, thereby permitting them full advantage of the minor flexors. In retrospect, the existence of so much inter-individual variation in vivo severely limits the application of results from a single cadaver such as are at hand here.

In sum, the experiment has provided a gross validation of the model,

$$\tau_{\alpha} = \sum F_{i_{\alpha}}\left(\left(\frac{L}{L_0}\right)_{i_{\alpha}}, X\text{-section}_{i_{L_0}}\right) \cdot \sin\beta_{i_{\alpha}} \cdot r_{i_{\alpha}}$$

in maximal, isometric flexion of the human elbow. It suggests, however, that length change alone, when considered in the three major flexors, Biceps Brachii, Brachialis, and Brachioradialis, is not sufficient to account for the dependence of maximal, isometric flexor torque on angle of elbow flexion -- it would appear that there are other parameters, some of them perhaps undefined in the model, which contribute in important measure to the dependence of torque on angle of flexion. The experiment demonstrated some of the conditions which permit the production of curiously high torques at small angles of elbow flexion. The validity of substituting for all the flexors a single equivalent flexor acting throughout the excursion of the joint is not upheld as a justifiable simplification of gross mechanics in the elbow. Finally, wide inter-individual variation in torque/angle of flexion curves generated by each of the subjects is readily apparent and the experiment points up concrete features of anatomy which may help to cause this -- the application of this is not examined.

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