1975

Dynamic analysis of the motion of linkages with relation to the upper extremity human limb

Neville Thomas Hodkinson

University of Wollongong

UNIVERSITY OF WOLLONGONG

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Recommended Citation

DYNAMIC ANALYSIS OF THE
MOTION OF LINKAGES WITH RELATION TO
THE UPPER EXTREMITY HUMAN LIMB

NEVILLE T. HODKINSON

A Thesis submitted in partial fulfillment of the requirements for the Degree of Master of Engineering Science at the University of New South Wales

Department of Mechanical Engineering
University of Wollongong
(formerly Wollongong University College)

February 1975.
A planar motion dynamic study of the upper extremity human limb is presented in this thesis. The dynamic analysis computer model simulates a planar, rigid linked three member system pivoted at points corresponding to the shoulder, elbow and wrist joints.

High speed camera photographs of hammering a two inch nail into a wooden block provides the essential data for the computer analysis. The magnitude and directions of torques and force reactions at the wrist, elbow, and shoulder joints are computed. Graphed results for the entire motion compare favourable with other methods of analysis.

The dynamic analysis may be used on any real process equipment such as linked mechanism, saws, looms, presses and fast moving machinery, where the evaluation and measurement of inertia and member forces are involved.

The main advantage of this procedure over other methods is the simplicity of the high speed filming-computer evaluation technique as compared to, the elaborate and expensive use of measuring and recording equipment.
ACKNOWLEDGEMENTS

The author wishes to acknowledge the valuable assistance of Professor A. Roberts during the course of this project. Grateful acknowledgement is also made to Mr. P. Costigan who provided assistance during experiments.

Special thanks must go to the staff of the Wollongong Hospital and in particular Sister McKenzie, who made the teaching skeleton available for experiments.

Acknowledgement is made for the generous assistance and encouragement given by Australian Iron and Steel Port Kembla.

The patient and dedicated typing of this thesis by my devoted wife Rosemary and sister Eileen is gratefully acknowledged and appreciated.

This thesis made use of the computer facilities at the University of Wollongong and the Australian Iron and Steel 140" Mill Furnace.
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CHAPTER 1

INTRODUCTION
CHAPTER 1  INTRODUCTION

The most complex of all earthly "machines" the human body has been the subject of continuing research, experimentation and probing investigations from time immemorial.

This Thesis proposes to apply the techniques of high speed photography and digital computation to the study of the dynamic analysis of the upper extremity limb during motion.

Since the turn of the twentieth century, the industrial revolution has heralded great advances in science and technology, allowing deeper study of the physiological, psychological, medical and bioengineering aspects of the human body. Particularly in the past two decades, a wide range of human endeavours relating to human survival and performance in the field of space exploration and medical research has been responsible for moulding together the disciplines of medicine, science and engineering. However for security reasons the great majority of this information is not freely available.

A number of recorded engineering studies of the performance, strength and kinetics of the human body are acknowledged.

As far back as 1680, Borelli (1a) * was measuring the strength of jaw muscles by hanging weights on the lower jaw. Two hundred years earlier, Leonardo da Vinci, was, by dissection, describing in considerable detail the anatomical structure of the body.

Perhaps one of the earliest recorded studies of the mechanical behaviour of tissue was that carried out by Langer (1861)(1b). His experiments involved making a large number of stab wounds with a round-bladed dagger in corpses. The resulting wounds took up an elliptical shape and the

* Numbers in parenthesis refer to REFERENCES page 207
major axes of these ellipses formed a pattern of lines which were regarded as lines of tension. Recent work by Black (1973) (66) provides an introductory background for the mechanical behaviour of human tissue from both the micro and macroscopic point of view.

In 1880 Messerer (1) documented work on the strength of human legs including details of the tensile strength of human bones.

Numerous investigations have been made on the muscular strength of human subjects by means of dynamometers. Early recorded work by Bethe and Franke (1919) (2) and Reijs (1921) (3) relates to the force of the movements in pronation (outward forearm rotation) and supination (inward forearm rotation) of the hand.

The effect of joint position on the force developed by human subjects has been developed from the work of Braune and Fischer (1890) (4), Franke (1920) (5), Hansen and Lindhard (1923) (6), Hvorslev (1928) (7), Garry (1930) (8), Muller (1935) (9), Haxton (1945) (10), Dern, Levene and Blair (1947) (11), Hugh-Jones (1947) (12), Clarke, Elkins, Martin and Wakim (1950) (13), Wakim, Gersten, Elkins and Martin (1950) (14), and Wilkie (1950) (15). Much of this work is confined to flexion of the elbow and extension of the knee.

The effect of different hand, elbow and shoulder positions on the maximum static (isometric torque) force exerted during pronation and supination of the hand was made by Darcus (1951) (16) using an electrical strain gauge dynamometer recorder.

Further work by Caldwell (1962) (17), to determine the maximum force of arm extension (push) for various joint angles and body supports emphasised their importance in the production
of usable muscle forces. Using a dynamometer handle fitted with strain gauges, he showed that the maximum force which can be applied to the control handle is limited by the force which can be supported by the shoulder. Later studies by Caldwell (1963) (18), recorded the endurance effect of prolonged application of relative muscle loadings.

The amplification of man’s capabilities to perform load handling tasks through the use of manipulators, walking machines and powered exoskeletons has also resulted in the engineering requirement for a description and understanding of the biomechanics of the human body. The measurement, recording and analysis techniques of such body motion studies are carried out by Murphy, Garcia and Bird (1966) (19). Specific mention is given to a Hand Accelerometer applied to a manipulative materials-handling task requiring accuracy of positioning. Resultant distances, velocities and accelerations were obtained from vector summation in three axes.

An engineering study of torsional leg fractures as experienced by skiers has been carried out by Outwater and Woodward (1967) (20).

The study entailed the mechanical structural analysis of spiral fractures resulting from torsional overload of the tibia bone of the lower leg. A study by Byers, Kroon and Sweeney (1965) (21), showed that the tensile strength of human bone substance is $8135 \text{ lb/in}^2$, as previously recorded by Messerer in 1880 (1).

By using strain gauges and portable recording equipment Outwater and Woodward recorded a maximum torque of $264 \text{ Kg-cm}$ under extreme snow and speed conditions. The result is approximately one quarter of the maximum computed torsional strength of the tibia. It was concluded from this engineering
FIGURE 1.1

THE MAIN BONES IN THE RIGHT LOWER LEG

They are designed primarily to resist compression and bending. The proximal tibia width (an important dimension for the skier) can be measured with a caliper across the knee.
A readily measurable body dimension that can be related to the torsional resistance of the bone is the proximal tibia width. A reasonable straight-line correlation can be seen above. The torques were calculated using x rays for bone sizes and 8135 psi as bone strength.
LIVING STATURE (Inches)

MALE

2.894 x humerus + 27.811
3.271 x radius  + 33.823
1.880 x femur   + 30.970
2.376 x tibia   + 30.970

FEMALE

2.754 x humerus + 28.140
3.343 x radius  + 31.978
1.945 x femur   + 28.679
2.352 x tibia   + 29.439

FIGURE 1.3
HUMAN PROPORTIONS
HEIGHT VERSUS BONES
**FIGURE 1.4**
EXOSKELETAL GONIOMETER
Refer Lawrence & Lin (23)

**FIGURE 1.5**
ARM AXIS MOTION RECORDINGS FOR EATING WITH SPOON
study that bone fractures occur under non-skiing conditions. Additionally the problem of prediction of an upper limit of torsional failure related to a calculated valve of some readily measurable body dimension introduced the concept of observed or measured data relationships. Considering the engineering structural characteristics of the tibia bone (refer Figure 1.1) and by comparing calculated and corrected torsional strength against known bone dimensions, a straight line relationship between proximal width and torsional strength was plotted as shown in Figure 1.2.

The proximal width is measured across the knee, touching the lateral and medial condyle where the skin is quite thin. The recognised concept of human body proportions supports the concept that the proximal width is related to the area and thickness of the weak neck of the tibia bone. Refer Figure 1.3 for details of the concept of human body proportions, in this case, height versus bone lengths.

A device for externally measuring muscle forces and behaviour called a myrotron was developed by Cornwell Aeronautical Laboratory (1970) (22). The two axis servo-controlled exoskeleton with electronic measuring capability follows elbow flexion-extension and shoulder rotation while measuring related muscle forces and limb position.

A real time algorithm for the computer control of a modified Rancho electric arm was developed by Lawrence and Lin (1972) (23). A set of Exoskeletal Goniometers as shown in Figure 1.4 was used to electronically record movement data of typical daily tasks of the upper extremity limb. Pattern recognition of recorded data was used to decide the task parameters currently being performed and a regression function calculated the associated elbow angle.
This study consisted of a seven degree of freedom recording of nine typical daily tasks, such as, drinking with a cup, eating with a spoon, operating a push button phone, and many others.

A typical recording for the task of eating with a spoon is shown in Figure 1.5.

Prior to 1963 the mechanics of human skeletal joints had been studied chiefly in terms of the kinematics and static forces. The forces actually developed at the joints of a moving limb differ greatly from these static forces due to restraints and demands placed upon the muscular and neurological control systems resulting from the limb weight and inertia force distributions.

Early work in the study of joint forces and torque reactions to weight and inertia loads was recorded by Fischer (1906) (24) and Taylor and Blaschke (1949-55) (25).

The first computer evaluation work in this field was carried out by Pearson, McGinley and Butzel (1963) (26) relating to the Planar Motion study of the joints on the upper extremity limb. In their analysis, a number of simplifying assumptions were made, namely:

(a) Limb segments were treated as solid bodies.

(b) Deformation of soft tissue and blood displacement have some effect on mass distribution and inertia. However with relatively large forces involved, this effect is negligible.

(c) The transverse axes of the joints were considered pinned. Actually joints are fastened by collagenous tissue which acts in tension and permits displacement of bones. This displacement as compared to total arm movements would have little effect on the dynamic forces.
(d) Joints are considered frictionless. The low viscosity of synovial fluid and experiments of Wright and Johns (1960) (27) show this assumption to be reasonable.

The actual computer analysis was restricted to the study of the upperarm and forearm-hand as a combined member. Kinematic data was collected photographically by multiple exposure pictures of the arm with reflective tape attached to the posterior surface.

Physical data was measured by X Ray photography. Weights were determined by the water displacement method and moments of inertia were measured experimentally using laminated cork and linoleum discs models with approximated actual density and mass distributions.

Computer analysis data and results were tabulated and plotted. Pearson, McGinley and Butzel (1963) (26) estimated an error of up to 8 per cent for the joint forces and 16 per cent for the joint reaction torques calculated by this method.

This thesis, proposes to determine force and torque reactions for a three member pin jointed rigid bone structure using high speed filming to obtain kinematic data.

Whereas Pearson, McGinley and Butzel (1963) (26) relied on reflective tape to indicate member position, this study was filmed naturally allowing frame by frame corrections for bone movements, centre of gravity and instantaneous pin jointed frame centre distances to be made.

Weights were determined by the water displacement method and checked by direct weighing of the arm. Limb moments of inertia were calculated using approximate physical sizes and assuming a weight distribution based on the limb volume and the distribution of bone and tissue. Details of limb bones and joints were obtained by superimposing human
skeleton photographs onto the high speed film information.

The computer programme provides for a separate set of physical data and kinematic data for each reference step throughout the motion, and computes, step by step formatted values for displacements, velocities, accelerations, forces and torques. Reference should be made to Chapter 3 for the Simulation Model Theory.

The graphed results compare favourably with actual cineplasty and dynamometer measurements of forces and torques, additionally a close agreement exists with the work recorded by Pearson, McGinley and Butzel (1963) (26). Reference should be made to Chapter 6 for Experimental Model results.
CHAPTER 2

DETAILS OF THE ARM
FIGURE 2.1  HAND MEASUREMENTS OF MEN, WOMEN AND CHILDREN
FIGURE 2.2

Upper Extremity System in Standard Position

Body reference planes: ss, sagittal; hh, horizontal; ff, frontal (coronal). The radioulnar wrist axis is vertical through W. Refer Klofinteg & Wilson (33).
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</tr>
<tr>
<td>Forearm</td>
<td>Radius, ulna</td>
<td>EW</td>
<td>Flexion-extension, pronation-supination (wrist rotation)</td>
</tr>
<tr>
<td>Hand</td>
<td>Carpals, metacarpals, phalanges</td>
<td>WK</td>
<td>Radial and ulnar flexions, volar and dorsal flexions</td>
</tr>
</tbody>
</table>

FIGURE 2.4 MAJOR MOTIONS OF THE UPPER EXTREMITY SYSTEM

The hand, forearm, arm, and shoulder axes are designated WK, EW, HE, and OH, respectively. The corresponding centres are W, E, H and O. Arcs define the major motions in orthogonal planes through each centre as origin. The notation is: F, flexion; Ex, extension; El, elevation; D, depression; S, supination; P, pronation; LR, lateral rotation; MR, medial rotation; RF, radial flexion; UF, ulnar flexion; DF, dorsal flexion; VF, volar flexion. REFER KLOPSTEG & WILSON (33)
CHAPTER 2  DETAILS OF ARM

The upper extremity limb consists of the hand, forearm and the upperarm, attached to the trunk of the body at the shoulder. The limb has a number of rigid central bone structures surrounded by muscles, blood vessels, nerves and an overall skin covering. As a typical illustration the general appearance, relative sizes and average dimensions of the hands of men, women and children is shown in Figure 2.1.

A high level nervous control system of sensory and motor receptors is responsible for the precise movements of the limbs.

It is important at the outset to define carefully the principal component motions of the upper extremity system. The motion of each part upon its proximal joint may be described with respect to the principle planes which intersect at the joint. Figure 2.2 shows the standard position and body reference planes: frontal, sagittal and horizontal. In this standard position, the trunk is erect, the arms hang with their axes vertical, the elbow angle is approximately 90 degrees and the wrist planes are vertical to assume the "shake hands" position. Table 2.3 lists the bones, angles, and axes of the system with chief reference to this standard position, although it is evident that, as the parts move, their reference axes should be shifted so that the angular motions made by each segment are properly referred to the new position of the joint with which the segment articulates.

It should be noted that the standard position as defined in Figure 2.2 provides a convenient engineering reference coordinate system. This differs from the anatomical reference position as shown in Figure 2.4. The anatomical movements are defined in terms of the major muscle group activities and their
Geometrical coordinates of \( E \) about the origin \( H \) (i.e., of the humerus, \( HE \), about the shoulder) are given by colatitude angle \( \theta \) and longitude angle \( \varphi \). Anatomical motion terminology is indicated by elevation—depression of segment axis \( HE \) in plane \( YZ \), and flexion—extension in plane \( YX \). Additional modes of excursions of the arm in the plane \( YX \) are termed parasagittal flexion \( PF \) and parasagittal hyperextension \( PH \). Refer Klopfteg & Wilson (33).
associated limb movements. Therefore the latter would be convenient to analyse the dynamics of muscles producing limb movement but the former, standard reference, would be best for overall limb dynamic analysis. The difficulty of calculating bone and muscle forces operating in various planes is a reality and therefore it appears simpler to standardise on orthogonal planes of reference (Figure 2.2) for filming purposes and to suffer the inevitable problem of group muscle activity during limb movements.

The shoulder-on-chest, arm-on-shoulder, and hand-on-wrist actions as displayed in Figure 2.4 take place through two planes at right angles, as if these articulations were moving about a three-dimensional joint. The insert gives the motion nomenclature of the hand-on-wrist. Thumb and digital motions are left for the later discussions of prehension. Geometrically, the arm motions are better described by a spherical coordinate system where the segment position is given by longitude and colatitude angles. For descriptive purposes, however, the anatomical nomenclature is commonly used, that is parasagittal flexion and hyperextension are simply called flexion and extension. It should be recognised that, for such multiaxial joints, flexion-extension and elevation-depression angles describe motions in the major orthogonal planes only, and intermediate angular excursions must be thought of as combinations of these motions. The geometric and anatomical systems of nomenclature are shown comparatively in Figure 2.5 for the case of the humerus acting on the shoulder.
FIGURE 2.11 BONE TERMINOLOGY

Head
- Anatomical Neck
- Position of Upper Epiphysis
- Surgical Neck
- Shaft
- Position of Lower Epiphysis
- Lower Extremity

Outer End
- Acromial End
  or
- Lateral End

Front (Ant.)

Inner (Sternal) End
  (Medial)

Back (Post.)

Clavicle

FIGURE 2.13 CLAVICLE BONE
FIGURE 2.12  BONE STRUCTURE OF THE UPPER EXTREMITY LIMB
**FIGURE 2.14**

**SCAPULA BONE**

- A, angle;
- AP, acromion process;
- CP, coracoid process;
- GF, glenoid fossa; (lateral angle);
- IB, inner (vertebral) border;
- N, neck;
- S, spine;
- SA, superior angle:
2.1 **Bone Structure of Arm**

The following general terms are used in describing bones, and are summarised in Figure 2.11.

(a) A fossa is a depression or hollow in a bone.
(b) A facet is a small, polished area intended for articulation with another bone.
(c) A process is a bony projection.
(d) A shaft is a cylinder of compact bone.
(e) A head is the shaped upper end of a long bone (normally rounded).

There are 32 bones in each upper limb. The limb is joined on to the trunk by one small bone only, the clavicle, or collar bone at the shoulder joint.

A general arrangement view of the upper extremity limb is shown in Figure 2.12.

The clavicle is a medium sized bone, something like a long drawn out capital S (refer Figure 2.13). Its inner extremity is thick and articulated with the sternum to form the sternoclavicular articulation. The outer extremity is more flattened and broader, and it curves forward to articulate with part of the shoulder blade.

The scapula, or shoulder blade, is a triangular, flat bone which lies over the outer and back part of the thorax, covering the second to the seventh ribs. The anterior surface is rather concave, so as to allow the bone to ride easily over the ribs. The posterior surface has a ridge of bone springing from it, called the spine of the scapula (refer Figure 2.14). This spine gets broader and sticks out more as it passes outwards, ending in a bony process known as the acromion process, which forms the top of the shoulder. To the inner border of this process the clavicle is attached, forming the
The outer or lateral border of the scapula has a shallow, pear-shaped depression at its upper part, the glenoid fossa, which articulates with the head of the arm-bone, forming the shoulder-joint. There is another smaller bony process projecting forwards and outwards from a point of little internal to the glenoid fossa. This is called the coracoid process, and from it arises one of the heads of the biceps muscle.

The shoulder blade is not attached at all to the ribs, and only to the trunk by means of the clavicle.

The humerus is the long bone of the upper limb. Like all long bones it has a head, a shaft, and a lower extremity. Its head is smooth and almost hemispherical and it articulates with the glenoid cavity of the scapula. Just below the head the humerus narrows a little, this part being called the neck. The lower extremity is somewhat expanded and bears a polished, pulley-like surface, which looks as if it had been turned by a lathe, for articulation with the bones of the forearm, to form the elbow-joint. Just above this articular surface there is a hollow, most marked behind. In the latter place the depression is known as the olecranon fossa, and when the forearm is extended (put out straight) the olecranon process of the ulna lodges in this fossa, so that the elbow can never be "bent the wrong way".

The forearm is that part of the upper limb between the elbow and wrist. It is made up of two bones, lying side by side. The outer one, i.e. the one on the thumb side, is the radius, and the inner one the ulna (refer Figure 2.12).

The radius is a little shorter than the ulna. Its head is round, and flat on the top like a saucer, and it articulates with the capitulum (outer part of the lower end) of the
humerus. The shaft is rather thin, but the bone widens out below, just above the wrist, and its lower extremity is smooth and polished for articulation with the wrist-bones.

The ulna is larger than the radius. Its upper end is hollowed out, being known as the trochlear notch or greater sigmoid fossa, into which the inner part of the lower end, that is, the trochlear of the humerus fits, forming the elbow joint. Just below this, on the lateral or outer side of the bone, is another smaller fossa, that is, radial notch which receives the rounded head of the radius. The shaft of the ulna runs down the forearm on its inner side (the little finger side). The top of the bone is narrowed into a tip, called the olecranon process. This process forms the tip of the elbow and it is the part upon which we rest the arm when we place our elbow upon the table with the hand up to the head. The lower part of the bone is quite small.

It will be seen that, of the lower ends of the two bones, that of the radius is the more important as far as the wrist-joint is concerned, whereas, of the upper extremities, that of the ulna is the more important, because it plays the chief part in the formation of the elbow-joint.

The wrist-bones are eight in number, and they are arranged in two rows, an upper and a lower. It is the upper row which articulates with the lower end of the radius, forming the wrist-joint. The lower end of the ulna is really separated from the wrist-bones by a small cartilage. The wrist-bones together are known as the carpus, and each individual bone is spoken of as a carpal bone.

The metacarpus is the name given to the five bones which form the skeleton of the back of the hand. Each metacarpal bone is a miniature long bone, having a base which articulates with the carpal bones, a shaft, and a rounded
head which articulates with the first or upper bone of the
finger to which it corresponds.

The metacarpal bone of the thumb — the first of the
series — is set a little obliquely to the others. For this
reason we are able to bring up the thumb to the tips of the
fingers in a movement which is called "opposition", which is
said to be peculiar to man.

The finger bones are called phalanges, each finger
possessing three, with the exception of the thumb, which has
only two. That phalanx which articulates with the head of the
metacarpal bone is the proximal, the middle phalanx being the
second, while the terminal or distal phalanx which bears the
nail is the third.

2.11 DETAILS OF JOINTS

A joint or articulation is a place of meeting or union
between two or more bones and the degree of flexibility of
this joint is determined by how closely the bones are joined
together by ligaments, and the amount of freedom permitted
them by nearby structures.

Reference is made to details of joint anatomy as
recorded by Fick (1911) (29), Steindler (1935) (32) and

The following terms are used to describe the movement
of joints, and or muscles, and their action (refer also
Section 2.6).

Flexion is bending or decreasing the angle between the
parts as when the arm bends back towards the shoulder.

Extension is the opposite action, stretching out, as
when the arm is straightened.

Rotation is turning on an axis, much as the earth turns
on its axis, except that in the body complete rotation is
FIGURE 2.111  
SKELETAL COMPONENTS OF THE SHOULDER JOINT
H, humerus; C, clavicle; S, scapula.
AC, acromion process of scapula;
CS, capsule of shoulder joint.
Refer Klopste & Wilson (33)
impossible because blood vessels, nerves, and other tissues would be torn.

Abduction is drawing away from the middle line of the body, as the lifting of the arm away from the body.

Adduction, the reverse, is turning toward the midline of the body, as when the arm is brought toward the trunk. The last two examples describe the action of the shoulder joint.

Joints are of two kinds, moveable and immovable but in this analysis only the moveable joints relating to movements of the upper extremity limbs are considered.

Moveable joints (Diathroses) are found in the limbs and spinal column. The bones concerned in a moveable joint do not actually join with each other; they simply touch or glide over one another, being held in place by a covering of fibrous tissue called a capsula. The ends of the bones are always tipped with cartilage and the whole joint-cavity is lined by a membrane, the synovial membrane, which secretes a little sticky fluid, called synovial fluid or 'joint-oil'. This is for purposes of lubrication, so that the joint shall work smoothly, without grating or creaking. Another type of lubricating and cushioning device, a bursa, is found between such bones as those of the elbow and shoulder of the upper extremity limb. Bursae are found in those joints where pressure may be exerted or where the attachment of a tendon rides over the bone.

The shoulder is articulated with the thorax anteriorly by the clavicle, which forms the sternoclavicular and costoclavicular joints with the breastbone and with the first rib (Figure 2.111). These are the only direct skeletal attachments of the shoulder, and therefore of the arm, to the torso. The clavicle then extends laterally and, passing superior and posterior to the coracoid process, articulates
with the acromion of the scapula to form the acromioclavicular joint. The body of the scapula hangs in a muscular suspension on the postero-lateral aspect of the torso and, in addition, receives lateral support from the clavicle at the acromioclavicular joint. The body of the scapula is flat and triangular with two lateral projections, the acromion and the coracoid processes, which are important attachment points for muscles and ligaments. The humerus, with its rounded proximal head articulating in the glenoid cavity of the scapula, completes the shoulder-arm complex.

The skeletal elements of shoulder and arm have been replaced by an equivalent system consisting of axes OH and HE, as defined in figure 2.4. Of these, OH is a fictional axis, but HE has physical reality in defining the axis of the humerus. As shown later on in figure 2.112, the centre line of the humeral articulation lies in the plane of the scapula, which is inclined 30 degrees to the frontal plane of the body. But this does not invalidate the concept of the humerus as a segment possessing two angular degrees of freedom and having a motion field about E. Thus, as justified by the measurements of Taylor and Blaschke (28), both the shoulder and the arm may, to a first approximation, be considered as linked segments, each having a two-angle motion field about its proximal articulation.

The shoulder-joint is basically a ball-and-socket joint of almost perfect form since it allows the greatest possible three degrees of freedom of movement.

The shoulder-joint formed by the articulation of the rounded head of the humerus with the rather shallow glenoid cavity of the scapula relies on tissue, muscles and tendons for its strength. The shallow glenoid cavity is deepened by the fibrous structured glenoid lip. From the nature of the
FIGURE 2.112

SKELETAL GEOMETRY OF ARM AND FOREARM

Front view: H, humeral centre; C, capitulum; U, ulnar distal centre; R₈, radius in supination; R₉, radius in pronation; 170 degree cubital angle; od, elbow axis; CU, forearm rotation axis.

Top view: xx, parasagittal plane; yy, frontal plane; ab, scapular plane; od, elbow axis. Refer KLOPSTEG & WILSON (33)
bony surfaces involved, it is easy to see that the shoulder is able to be "put out" or dislocated. The glenoid fossa is not a deep cut at all, and the humerus has not a neck like the femur of the hip joint. The capsule is large and loose and there are several powerful muscles which help to protect it from injury. Of these the deltoid is the largest.

The shoulder-joint permits six movements to take place, namely the four chief movements, flexion and extension, abduction and adduction and in addition axial rotation, which is a rotation of the humerus about its long axis, and circumduction, in which the whole lower extremity can be moved so as to describe a circle or the base of a cone, the head of the humerus, which does not move much in this movement, serving as the apex of the cone. The elbow-joint is basically a hinge-joint in which only two movements are allowed in one plane, that is, flexion and extension.

The skeletal members of the forearm system are the forearm bones, the radius and the ulna, and the distal end of the humerus, with which they articulate. The details are shown in figure 2.112. The radius, which is lateral to the ulna, makes a joint with the capitulum, a hemispherical prominence on the anterior aspect of the distal end of the humerus. It can rotate in two angles, as well as twist on this articulation. The ulna, however, fitting into a bearinglike groove in the humerus, is constrained to the flexion-extension axis of rotation. Distally, the radius articulates upon the ulna to permit twist of the former on the latter.

The motions of the forearm, flexion at the elbow and rotation throughout the length of the forearm, for which the nomenclature is given in figure 2.4, may readily be understood from the bone and joint mechanics shown in
In the front view, the forearm bones are shown to be capable of rotation upon the elbow axis in the manner of a simple hinge. However, the anatomical centre lines of arm and forearm are not coaxial, and the cubital angle thus formed is about 170 degrees. In consequence, forearm flexion upon the arm describes a slightly conical surface of revolution. The axis of forearm rotation takes a diagonal course passing through the radial capitulum at the elbow and the distal end of the ulna. Pronation and supination features a "wheeling" of the distal radius about this axis, in the manner shown in the front view of figure 2.112.

The components of the upper extremity may be seen in geometric relationship to each other and to the frontal plane of the body proper by reference to the overhead view of figure 2.112, taken from Fick (29). In the naturally hanging arm, the head of the humerus faces into its glenoid articulation with the scapula in an axis which is inclined 60 degrees to the sagittal plane of the body. Proceeding distally, the elbow axis drawn through the epicondyles is somewhat laterally rotated to an 80 degree inclination; and finally the radioulnar wrist axis is shown as perpendicular to the elbow axis. This relationship is obviously variable and is here defined as a "standard" position.

The elbow-joint is formed by the articulation of the lower, pulley-like end of the humerus with the upper ends of the radius and ulna. The ulna takes the largest share in the formation of the joint, its great sigmoid fossa receiving the trochlear of the humerus. The head of the Radius is placed on the lateral or outer side of the joint and, forming another joint with the ulna as well, it moves with that bone in movements of the elbow-joint.
**FIGURE 2.113**  
**ELBOW JOINT**  
Refer Klopsteg & Wilson (33)  
E, capsule of elbow joint (outer side);  
H, humerus;  
O, olecranon process of ulna;  
R, radius;  
U, ulna.

**FIGURE 2.114**  
**SUPERIOR RADIO-ULNAR JOINT AT ELBOW**  
OL, orbicular ligament;  
R, head of radius;  
SU, great sigmoid cavity of ulna;  
U, ulna.
No side to side movement can take place at the elbow, only flexion, produced by the biceps chiefly, and extension, produced by the triceps. The joint is covered in by a strong capsule, strengthened by four ligaments, (refer figure 2.113) but it is not a very strong joint on the whole.

The joints between the upper ends of the radius and ulna, is a Pivot-Joint (or lateral hinge) in which movements of pronation or rotation of the radius and supination occur.

The superior-ulnar joint is formed by the articulation of the rounded, wheel-like head of the radius with the lesser sigmoid cavity on the lateral side of the upper end of the ulna. The head of the radius is kept in place against the ulna by a ligament which tightly embraces the radius, and yet allows sufficient play for the proper movement of the bone. This ligament is called the orbicular ligament (refer figure 2.114).

Although it is closely connected with the elbow-joint, this articulation is quite independent of it as regards movement. The radius rotates pivot-like, in its long axis within the ring formed by the concave lesser sigmoid cavity of the ulnar and the orbicular ligament. While the upper end of the radius is rotating in this fashion at the superior radio-ulnar joint, the rest of the bone is carried half-way round the lower part of the ulnar, where there is another joint between the two bones, the inferior radio-ulnar articulation. As the hand is attached to the radius by the wrist-joint, it moves with every movement of that bone.

The movement of pronation consists of a rotation of the radius so that its lower end comes to lie over and in front of the ulna, the palm of the hand going under at the same time. If the elbow be flexed with the hand in this position, that is, with the palm downwards, the movement
**Figure 2.115** Carpo-Metacarpal Joint of Thumb

M, first metacarpal; SC, synovial sac in joint; T, trapezium bone.

**Figure 2.116** Bone and Joint System About the Wrist

Carpal bones: L, lunate; N, navicular; T, triquetrum; P, pisiform; GM, greater multangular; LM, lesser multangular; C, capitate; H, hamate.

Metacarpal bones: MII

Major articulations: RC, radiocarpal; IC, intercarpal; CM, carpometacarpal.

Refer: Klopfstei, J Wilson (33)
which brings the palm upwards again is called supination. In this last movement the radius has rotated back again to the outside of the lower end of the ulna.

Because supination is a more powerful movement than pronation, all screwing and boring tools are made to be used with this movement. The chief muscle producing supination is the forearm supinator muscle, which is aided by the biceps muscle when heavy work is performed.

A third type of moveable joint is the Saddle-joint as in the carpo-metacarpal joint of the thumb where movements allowed are only those of slight rotation or gliding.

The carpo-metacarpal joint of the thumb is formed by the articulation of the base of the first metacarpal bone with the outermost bone in the lower row of carpal bones, known as the Trapezium (refer figure 2.115). The lower surface of this latter bone is saddle-shaped. The joint has a complete capsule. Owing to the way in which the first metacarpal bone is "set" upon the Trapezium, we are able to bring the thumb up against any of the fingers in the movement called opposition.

The wrist joint is basically an angular joint which is formed by the articulation of an oval-shaped surface with a concave cavity. This type of joint permits movements in two directions.

The bone and joint anatomy of the wrist, as can be seen in figure 2.116, provides many articulations for the accomplishment of major wrist movements. Thus, with reference to figure 2.4, the wrist can be simply flexed toward the volar, dorsal, ulnar, and radial sides, or toward combination fields, such as volar-ulnar, volar-radial, dorsal-ulnar, or dorsal-radial. There is little torsion in wrist and hand, so that the joint complex may be said to have
Ball & Socket Shoulder Joint  
Humerus  
Clavicle  
Soapula  
Radius Pivot Joint  
Radius  
Ulna  
Humerus - Ulna  
Elbow Hinge Joint  
Wrist Angular Joints  
Thumb Saddle Joint  
Finger Hinge Joints

FIGURE 2.117  
UPPER EXTREMITY LIMB BONE AND JOINT
SUMMARY
two principal degrees of freedom, of the type shown by construction in figure 2.5. The major articulations, radiocarpal and intracarpal, share these movements in various proportions, depending upon the direction of movement. Braune and Fischer (30) have measured the motion fields about the major articulations of the wrist. The compound joint actions of the natural wrist afford large angular motion fields with what approaches a curvature rather than the sharp angularity which would be given by a single joint. While volar flexion takes place chiefly in the radiocarpal joint, dorsal flexion is strongly shared by the intracarpal joint. Similarly, in the radioulnar plane, radial flexion occurs largely in the radiocarpal joint, while ulnar flexion depends strongly upon the contribution from the intracarpal joint.

The various types of joints present in the upper extremity limb are shown in figure 2.117. This was prepared as a reference for the general distribution of bones for the upper extremity limb, showing in detail the composite parts, and types of supporting joints of the arm. This knowledge is necessary when preparing a detailed force analysis of the arm where simplifying assumptions are based on the knowledge of joint operation and limb structure.

Reference should also be made to skeleton photos of joints contained in figure 4.311.

2.12 MECHANISM OF JOINTS

The physiological details of joints are well documented, but the actual mechanics of load transmission from one bone member to another at the joints is still a subject for continuing research. The following discussion is based on a report entitled "Mechanism of Human Joints" by Swanson and Freeman (1969) (31).
FIGURE 2.121  SCHEMATIC DIAGRAM OF A HUMAN JOINT

Refer Swanson & Freeman (31)
The junctions - or, in anatomical terms, joints - between the bones of the skeleton fall broadly into two categories: those at which movement is allowed and those at which movement is entirely, or almost entirely, prevented. The former are the synovial joints and - when they function efficiently - these joints transmit compressive loads across their surfaces painlessly whilst presenting little frictional resistance to motion. The geometry of the synovial joints is variable, so that both the number of axes about which movement is allowed, and the range of movement permitted, may differ from joint to joint. It is therefore necessary to understand the normal mode of load transmission and lubrication in synovial joints. Anatomically these functional properties depend in all synovial joints upon five elements in the joint: bone, articular cartilage, the synovial membrane, synovial fluid and ligaments as shown in figure 2.121. In addition, the muscles acting across a joint are not only responsible for movement but also have a large part to play in the maintenance of joint stability.

In a synovial joint the ultimate bearing surfaces, between which relative movement occurs, are composed of cartilage. This material contains relatively few cells, most of its bulk being made up of a fibrous and gelatinous material known as the matrix. The matrix has three components, all of which are produced by the cells: the fibrous protein collagen, several varieties of long chain mucopolysaccharides and water. The collagen is believed - almost certainly correctly - to be arranged in arcade form, each arcade arising in the basal layers of the cartilage and curving to run parallel to the surface. At any one point in a sheet of cartilage the arcades have a predominant direction and the
spatial orientation of the arcades is characteristic for a particular joint surface. The mucopolysaccharide molecules form, with the water present in cartilage, a hydrated gel from which some of the water can be expressed under a compressive load. Cartilage can thus be pictured as a non-random fibrous network the interstices of which are filled by a hydrated gel.

The bearing (or articular) surfaces of cartilage in life are moistened with very small quantities of a fluid known as the synovial fluid. This is produced by the synovial membrane and consists of a water solution of the salts found in the blood, of glucose, of small quantities of protein and of a long chain mucopolysaccharide, hyaluronic acid. The latter is responsible for the viscosity of synovial fluid and may thus be important in synovial joint lubrication. The lubricating properties of hyaluronic acid on cartilage may also be dependent upon the fact that the matrix of cartilage is impermeable to it. In the adult, the cartilage cells depend for their nutrition upon the delivery of chemicals dissolved in the synovial fluid; all the small molecules are able to diffuse through the matrix. Synovial fluid thus certainly has a nutritional function.

The synovial membrane is a soft flexible structure which lines that part of the joint cavity not covered by cartilage. It does not carry load and its function lies not in its mechanical properties but in the production of synovial fluid.

The cartilage/synovial fluid/synovial membrane complex shown in the figure 2.121, may be regarded as a functional unit and, since it is within this unit that motion occurs, the normal function of this complex is essential for the normal function of the joint.
Articular cartilage is firmly attached to the bones participating in the joint. The exact nature of the bone/cartilage bond is unknown but possibly depends upon the inter-connection of the two with collagen and upon the fact that the matrix of the basal layer of cartilage, like the matrix of bone, is calcified.

The bone itself is arranged as a honeycomb structure so that the loads transmitted through the joint are diffused over a wide area. The bones are connected by ligaments composed of collagen fibres arranged in parallel - a structure analogous to that seen in ropes. Ligaments are strong in tension and are flexible.

Together the bones and ligaments form an outer functional unit in synovial joints: a unit which, together with the cartilage, is responsible for the regulation of the axes and ranges of joint movement, and for load transmission. Within the unit composed of cartilage, synovial fluid and synovial membrane there are two bearings: one, highly loaded, between the cartilage surfaces possibly lubricated by synovial fluid; and second, lightly loaded, between cartilage and synovial membrane almost certainly lubricated by synovial fluid. The first of these bearings is, from the point of view of the function of the joint, the more obvious since it is through this bearing that load is transmitted from one bone to the next. The existence of a low coefficient of friction at the second bearing is, however, equally vital to the function of the joint, since the synovial membrane - which it must be remembered is closely applied to the whole of the unloaded surface of the cartilage - must slide smoothly and easily over the cartilage if it is not to be drawn into the potential gap between the loaded cartilage surfaces. Were
MECHANISM BY WHICH FRICTION IS REDUCED AT SYNOVIAL JOINTS

One hypothesis suggests that a gel of hyaluronic acid forms the lubricant in joints. The application of load (b) causes the cartilages to flatten locally and the fluid to escape sideways. Continued loading (c) flattens the cartilage further so that the fluid cannot escape freely and water and salts take the longer route through the cartilage. The hyaluronic acid remains in the loaded region (d) stabilized by fluid and osmotic forces.

Refer Swanson & Freeman (31)
it ever to be drawn in in this way, it would be crushed and would bleed, so that the joint would be intermittently painful, swollen and stiff.

The present views on the mechanism by which friction is reduced at synovial joints can by summarized as follows. First, the cartilage surfaces are very smooth so that even in the absence of a lubricant they move easily over each other. Secondly, because the cartilage surfaces are deformable and, less importantly, because of the viscosity of synovial fluid, it is likely that the cartilage surfaces are lubricated by a fluid film if they are in loaded contact for times of the order of a fraction of a second. Thirdly, because the cartilage surfaces are permeable to the water but not to the hyaluronic acid in synovial fluid, a concentrated hyaluronic acid water gel can be expected to form between cartilage surfaces loaded for a few seconds (refer figure 2.122). Such a gel might be expected to act as a boundary lubricant. And, finally, current experimental observations suggest that the fat in the cartilage matrix acts as a more efficient boundary lubricant than does the hypothetical hyaluronic acid gel. Thus the lubrication regime between loaded cartilage surfaces depends upon the duration of loaded contact and is probably a varying mixture of elastohydrodynamic and boundary. Elastohydrodynamic lubrication exists where the film of lubricant is so thin, and the fluid pressure acting on the bearing surfaces are so high, that the elastic deformation of these surfaces cannot be ignored. At the 'low pressure bearings' in synovial joints between the synovial membrane, itself, and cartilage, or between unloaded cartilage and cartilage, it seems likely that the lubrication regime is boundary in
nature and that the hyaluronic acid in the synovial fluid is the lubricant.

A number of experiments using a pendulum fitted with a human joint as the pivot have shown that frictional forces exhibit characteristics of fluid lubrication (hydrostatic or hydrodynamic) and not of boundary lubrication. Coefficient of friction between 0.01 and 0.02 exist for such joints as compared to 0.05 - 0.1 for fatty acids in metallic bearings.
Pectoral brings arm to side and across chest i.e. adduction medial rotation Biceps bends elbow, flexion Flexors bends wrist and fingers

FIGURE 2.21 FRONT VIEW OF SKELETAL MUSCLES OF UPPER EXTREMITY LIMBS
Extensors (Straightens wrist and fingers)

Triceps straightens elbow i.e. extension of forearm

Deltoid raises arm i.e. abduction of arm

Trapezius raises shoulder and pulls head back also retraction of shoulder

Latissimus Dorsi draws arm backwards and turns it inwards; it also draws downwards and upstretched arm i.e. adduction, extension & medial rotation

FIGURE 2.22 REAR VIEW OF SKELETAL MUSCLES OF UPPER EXTREMITY LIMBS
FIGURE 2.23  THE BICEPS AND TRICEPS MUSCLES OF THE ARM, SHOWING THEIR ORIGINS AND INSERTIONS
2.2 **SKELETAL MUSCLES OF THE ARM**

Every movement that takes place in the body is carried out by the action of a special form of tissue known as muscular tissue. A muscle is a tissue, no matter how small, that moves other parts of the human body.

There are two classes of muscles in the body (1) voluntary, and (2) involuntary. For this subject we are only interested in voluntary muscles which are under the will, that is, all the muscles attached to the body skeleton. A detailed discussion of the composition and function of muscles follows in Section 2.3 and at this juncture only the general distribution of muscles associated with the arms is shown.

Figures 2.21 and 2.22 show the front and rear views of muscles and their importance related to movement of the arm. Figure 2.23 is included here to show the typical type of "endings" to muscles, which in effect, constitute the mechanical attachments of the muscles to the bony structure.

Muscles are only able to move because of the location of the muscle attachments. These points of attachment are called the origin and the insertion. The attachment to the bone that serves as a relatively fixed basis of movement is the origin. The insertion is the point of attachment to the bone which is moved. Most muscles are attached to the periosteum of a bone by means of a tendon; however, some make direct contact with the periosteum, while others are attached by a sheet of heavy connective tissue. (The covering of the bone and periosteum is thick double-layered membrane containing blood vessels, nerves, and the bone-forming cells).

The movements made possible by the action of a specified muscle can be understood and synthesised in design by knowing its origin and insertion. Consider, for example,
the action of the biceps, a large muscle on the front of the upper arm that can be felt when the arm is bent to the elbow. Because of the placement of its origins, it not only brings about a movement of the forearm in a vertical plane, but also gives it a rotary motion, that is, supination such as used with a screwdriver. In figure 2.23 both the biceps and triceps (on the back of the arm) have more than one origin each; the triceps has three points of origin and derives its name from that fact, and the biceps has two. It will be noted that these muscles have several points of insertion, and because of this, complex action is possible.

The action of the biceps and the triceps when the arm is flexed or extended at the elbow illustrates an important characteristic of muscles, in that, they like most skeletal muscles work in pairs. For each muscular contraction that brings about motion in one direction there is possibly a motion in the opposite direction which depends upon the contraction of the other member of the muscle pair. For example, when you flex your arm, the action is largely the result of the contraction of the biceps muscle. But when you extend your arm, the triceps muscle contracts. These opposing actions are brought about by the so-called antagonistic muscles.

Finally the action of the nervous system in sending contraction impulses to one of a pair of antagonistic muscles, while allowing the other to relax, is termed the reciprocal innervation of the muscle. Thus it can be seen that the combination of muscles and control is responsible for the delicate movements of the limb.
In man and most animals the ability to move depends upon a group of specialized contractile cells, the muscle fibres. Man, and indeed most vertebrates, are quite muscular animals; almost half of the mass of the human body consists of muscle tissue. In the vertebrates three types of muscle fibres have evolved to perform various kinds of movements; skeletal muscle, which is attached to and moves the bones of the skeleton; cardiac muscle, which enables the heart to beat and move the blood through the circulatory system; and smooth muscle, which makes up the walls of the digestive tract and certain other internal organs, and moves material through the internal hollow organs. All three types of muscle have the ability to shorten when stimulated, and ordinarily this stimulation reaches the muscle fibres by a nerve. Both cardiac and smooth muscle can contract in the absence of nervous stimulation, and both the heart and the digestive tract function almost normally even when all the nerves leading to them have been cut.

In contrast, when the nerves to skeletal muscle are severed or blocked, the muscle is completely paralyzed. For a few weeks it will respond to artificial stimulation, such as an electric shock applied to the overlying skin, but even this ability is gradually lost.

A typical skeletal muscle is an elongated mass of tissue composed of millions of individual muscle fibres bound together by connective tissue fibres. The entire structure is surrounded by a tough, smooth sheet of connective tissue so that it can move over adjacent muscles and other structures with a minimum of friction. The two ends of the muscle are
TYPICAL SKELETAL MUSCLE ARRANGEMENT
FIGURE 2.312  PROPRIOCEPTORS OF MUSCLES, TENDONS AND JOINTS

Gerlini Organ  

Muscle spindles  

Muscle fibrils  

Sensory nerve endings in muscle spindle  

Fine motor nerves from cerebellum of brain  

Tendon nerve endings  

Muscle body  

Bone
usually attached to two different bones, although a few muscles pass from a bone to the skin, or even, as in the case of the muscles of the face used in speech and expression, from one part of the skin to another. The end of the muscle which remains relatively fixed when the muscle contracts is known as the origin, the end that moves is called the insertion; and the thick part between the two is called the belly. (Refer figure 2.311). The origins of the biceps muscle are at the shoulder, and its insertion is on the radius bone of the forearm; when the biceps contracts, the shoulder remains fixed and the elbow is bent.

The entire muscle structure is controlled by the central nervous system, the brain, which sends and receives motor and sensory impulses to and from the muscle nerve endings. A typical arrangement of these nerve endings is shown in figure 2.312. The muscle nerve endings are called proprioceptors and are the sense organs stimulated by movement of the body itself. They make us aware of the movement or position of the body in space and of the various parts of the body to each other. They are important as ingoing afferent pathways in reflexes for adjusting "Posture and Tone".

General proprioceptors are found in skeletal muscles, Tendons and Joints and are of two types (a) Tendon sensory nerve endings called Gorli Organ and (b) Muscle motor sensory nerve called muscle spindle. Sensory nerve endings of the tendons of muscles (Gorli Organ) are stimulated by tension which occurs when muscles are stretched and or when they are contracted. Motor nerve endings of muscles or muscle spindle on the other hand are contained in the skeletal muscle and are stimulated when muscle is stretched or shortened.

A similar action in the sensory nerve endings of the
FIGURE 2.313  STRIATED MUSCLE DISSECTION
Refer Huxley (34)
body muscles (muscle spindles) occurs when the muscle fibrils (intrafusal fibres) are stimulated by the muscle either stretching or shortening. It is obvious here that a desired rate of response of muscle fibril movement to tendon tension exists, and this gives rise to a highly developed positioning and displacement control to the limb in its movements. In addition to the above, the close proximity of the fine motor nerve endings (which are stimulated either by contracting or stretching of the muscle) to the sensory nerve endings in the muscle spindle gives a positive signal to the brain the instant that movement occurs and hence further movement can be minutely controlled.

Muscles are usually classified as "striated" or "smooth", depending on how they look under the ordinary light microscope. The classification has a good deal of functional significance. The muscles which vertebrates such as men use to move their bodies or limbs — muscles which act quickly and under voluntary control — are crossed by microscopic striation. The muscles to the gut or uterus or capillaries — muscles which act slowly and involuntarily — have no striations; they are "smooth".

Striated muscle is dissected in the schematic drawings of figure 2.313. (Refer Huxley (1958) (34)). A muscle (A) is made up of muscle fibres (B) which appear striated in the light microscope. The small branching structures at the surface of the fibres are the "end-plates" of motor nerves, which signal the fibres to contract. A single muscle fibre (C) is made up of myofibrils, beside which lie cell nuclei and mitochondria. In a single myofibril (D) the striations are resolved into a repeating pattern of light and dark bands. A single unit of this pattern (E) consists of a "Z-line", then
an "I-band", then an "A-band" which is interrupted by an "H-zone", then the next I-band and finally the next Z-line. Electron micrographs have shown that the repeating band pattern is due to the overlapping of thick and thin filaments (F).

Striated muscles are made up of muscle fibres, each of which has a diameter of between ten and one hundred microns (a micron is a thousandth of a millimeter). The fibres may run the whole length of the muscle and join with the tendons at its ends. About twenty per cent of the weight of a muscle fibre is represented by protein; the rest is water, plus a small amount of salt and of substances utilized in metabolism. Around each fibre is an electrically polarised membrane, the inside of which is about a tenth of a volt negative with respect to the outside. If the membrane is temporarily depolarized, the muscle fibre contracts; it is by this means that the activity of muscles is controlled by the nervous system.

THE STRUCTURE OF FIBRES

The most straightforward way to try to find out how the muscle machine works is to study its structure in as much detail as possible. The contractile structure of a muscle fibre is made up of long, thin elements which are called myofibrils. A myofibril is about a micron in diameter, and is cross-striated like the fibre of which it is a part. Indeed, the striations of the fibre are due to the striations of the myofibril, which are in register in adjacent myofibrils. The striations arise from a repeating variation in the density, that is, the concentration of protein along the myofibrils.

The pattern of the striations can be seen clearly in isolated myofibrils, which are obtained by whipping muscle in a Waring blender. Under a powerful light microscope there is
a regular alternation of dense bands (called A-bands) and lighter bands (called I-bands). The central region of the A-band is often less dense than the rest of the band and is known as the H-zone. When a striated muscle from a vertebrate is near its full relaxed length, the length of one of its A-bands is commonly about one point five microns, and the length of one of its I-bands about point eight micron. The I-band is bisected by a dense narrow line, the Z-membrane or Z-line. From one Z-line to the next the repeating unit of the myofibril structure is thus: Z-line, I-band, A-band (interrupted by the H-zone), I-band and Z-line.

When myofibrils are examined in the electron microscope, a whole new world of structure comes into view. It can be seen that the myofibril is made up of still smaller filaments, each of which is fifty or one hundred angstrom units in diameter (an angstrom unit is a ten thousandth of a micron).

To examine the arrangement of the filaments in considerable detail a piece of muscle is first "fixed", that is, treated with a chemical which preserves its detailed structure during subsequent manipulations. Then the muscle is "stained" with a compound of a heavy metal, which increases its ability to deflect electrons and thus enhances its contrast in the electron microscope. Next it is placed in a solution of plastic which penetrates its entire structure. After the plastic is made to solidify, the block of embedded tissue can be sliced into sections one or two hundred angstrom units thick by means of a microtome which employs a piece of broken glass as a knife. When looked at (in the electron microscope) these very thin sections appear extraordinarily regular and specific in construction.

A myofibril is made up of two kinds of filament, one
of which is twice as thick as the other. In the psoas muscle from the back of a rabbit the thicker filaments are about one hundred angstroms in diameter and one point five microns long; the thinner filaments are about fifty angstroms in diameter and two microns long. Each filament is arrayed in register with other filaments of the same kind, and the two arrays overlap for part of their length. It is this overlapping which gives rise to the cross-bands of the myofibril: the dense A-band consists of overlapping thick and thin filaments; the lighter I-band, of thin filaments alone; the H-zone, of thick filaments alone. Halfway along their length the thin filaments pass through a narrow zone of dense material; this comprises the Z-line. Where the two kinds of filament overlap, they lie together in a remarkably regular hexagonal array. In many vertebrate muscles the filaments are arranged so that each thin filament lies symmetrically among three thick ones; in some insect flight-muscles each thin filament lies midway between two thick ones.

The two kinds of filament are linked together by an intricate system of crossbridges which probably play an important role in muscle contraction. The bridges seem to project outward from a thick filament at a fairly regular interval of sixty or seventy angstroms, and each bridge is sixty degrees around the axis of the filament with respect to the adjacent bridge. Thus the bridges form a helical pattern which repeats every six bridges, or about every four hundred angstroms along the filament. This pattern joins the thick filament to each one of its six adjacent thin filaments once every four hundred angstroms.

The arrangement of the filaments and their cross-bridges, as seen in the electron microscope, is so
extraordinarily well ordered that one may wonder whether the fixing and staining procedures have somehow improved on nature. Fortunately this regularity is also apparent when examining muscles by another method: X-ray diffraction. Muscle which has not been stained and fixed deflects X-rays in a regular pattern, indicating that the internal structure of muscle is also regular. The details of the diffraction pattern are in accord with the structural features observed in the electron microscope. Indeed, many of these features were originally predicted on the basis of X-ray diffraction patterns alone.
Muscles never contract singly, but always in groups. No matter how hard a person tries, he cannot contract the biceps muscle alone - but can only bend the elbow, which involves the contraction of a number of other muscles besides the biceps. Furthermore, muscles can exert only a pull, not a push. Hence they usually are paired as antagonists, one pulling a bone one way, another pulling it the opposite way. The names flexor and extensor are applied to muscles to indicate the type of movement they effect. Thus the biceps, which bends or flexes the arm, is called a flexor, and its opposing muscle, the triceps, with its origin on the shoulder and upper arm, and its insertion on the ulna, straightens or extends the forearm and is called an extensor. Similar pairs of opposing flexors and extensors are found at the wrist, knee, ankle and other joints. Whenever a flexor contracts, the opposing extensor must relax to permit the bone to move, and this requires the proper coordination of the nerve impulses going to both sets of muscles.

When muscles are not contracting to effect a movement, they are not completely relaxed. As long as a person is conscious, all the muscles are contracted slightly, a phenomenon called tonus. Posture is maintained by the partial contraction of the muscles of the back and neck, and of the flexors and extensors of the legs. When a person is standing, both the flexors and extensors of the thigh must contract simultaneously so that the body sways neither forward nor backward on the legs, and the simultaneous contraction of the flexors and extensors of the shank locks the knee in place and holds the leg rigid to support the body. When movement is added to posture, as in walking, a complex
coordination of the contraction and relaxation of the leg muscles is required. It is not surprising that the process of learning to walk is long and tedious.

Some of the larger muscles of the body are extremely strong. Consider the muscle of the calf of the leg, called the gastrocnemius, which is used in rising on one's toes. Its origin is at the knee, and its insertion, by way of the tendon of Achilles, is on the heel bone; because the distance from the toes to the ankle joint is at least six times that from the ankle joint to the heel, the gastrocnemius is working against an adverse lever ratio of six to one. This means that when a person weighing one hundred and fifty pounds stands on one leg and rises on his toes, the one gastrocnemius muscle is exerting a force of nine hundred pounds, and if a man were to hold another person in his arms and perform this action, the muscle would be exerting a force of nearly a ton.

Because of the nervous coordination, no normal person can cause his muscle to contract maximally, but in certain diseases in which the nervous control is impaired, muscles do contract forcefully enough to rip tendons and break bones. In the muscle controlled by the nervous system, an impulse travelling down a motor nerve is transmitted to the muscle membrane at the motor "end-plate"; then a wave of depolarization (the "action potential") sweeps down the muscle fibre and in some unknown way causes a single twitch. When nerve impulses arrive on the motor nerve in rapid succession, the twitches run together and the muscle maintains its contraction as long as the stimulation continues (or the muscle becomes exhausted). When the nerve stimulation stops, the muscle automatically relaxes.
FIGURE 2.321 SINGLE MUSCLE TWITCH
TIP3S OF CONTRACTION

To understand how muscles contract, it is necessary to discriminate carefully between the contraction of a whole muscle and the contraction of its individual fibres. For although a muscle cannot contract maximally a single fibre of it can respond only maximally, or not at all. This phenomenon, described as the "all or none" law, can be demonstrated experimentally by dissecting out a muscle fibre and giving it repeated stimuli of increasing intensity, beginning with ones too weak to cause contraction as the strength of the stimulus is increased, there will be no response until a certain level is reached, at which time the fibre will contract completely. No stimuli of greater intensity can cause any greater contraction. Since a whole muscle is made up of thousands of individual muscle fibres, the nature and strength of its contraction depends upon the number of its constituent fibres which are contracting and upon whether the fibres are contracting simultaneously or alternately.

The following study of the contraction of an isolated muscle serves to illustrate the nature and behaviour of muscles during contraction. The single twitch occurs when a muscle is given a single stimulus — for example, a single electric shock — it responds with a single, quick twitch, which lasts about five hundredths of a second in a human muscle. Laboratory records of a single twitch (refer figure 2.321) indicate that it consists of three separate phases: (1) the latent period, lasting about one hundredth of a second, an interval between the application of the stimulus and the beginning of the visible shortening of the muscle; (2) the contraction period, about four hundredth of a second in duration, during which the muscle shortens and does work;
and (3) the relaxation period, the longest of the three, lasting five hundredths of a second, during which the muscle returns to its original length. The first event after the stimulation of a muscle is the initiation and propagation of an electrical response, the muscle action potential followed by the change in the birefringence of the muscle. After a twitch the muscle consumes oxygen and gives off carbon dioxide and heat at a rate greater than during rest, marking a recovery period in which the muscle is restored to its original condition. This recovery period lasts for several seconds, and if a muscle is stimulated repeatedly so that successive contractions occur before the muscle has recovered from the previous one, the muscle becomes fatigued and the twitches grow feebler and finally stop. If the fatigued muscle is allowed to rest for a time, it regains its ability to contract.

TETANUS

The normal contractions of the muscles do not occur as single twitches, but as sustained contractions evoked by a volley of separate stimuli — the nerve impulses — reaching them in rapid succession.

A sustained contraction is called a tetanus, and while it prevails, the stimuli occur so rapidly (several hundred per second) that relaxation cannot occur between successive contractions. In most tetanic contractions the individual muscle fibres are stimulated in rotation, rather than simultaneously, so that although they contract and relax, the muscle as a whole remains partly contracted.

Muscles of the body can contract in different degrees. This gradation of contraction is controlled through the nervous system: in a weak contraction only a small percentage of the muscle fibres are stimulated at one time; for a stronger
contraction a larger percentage of muscle fibres contract simultaneously.

**TONUS**

The term tonus or "tone" refers to the state of sustained partial contraction present in all normal skeletal muscles as long as the nerves to the muscle are intact.

Each muscle is normally stimulated by a continuous series of nerve impulses which cause a constant, slight contraction or tonus. Tonus is a mild state of tetanus, present at all times and involving only a small fraction of the fibres of a muscle at any moment. It is believed that the individual fibres contract in turn, working in relays, so that each fibre has a chance to recover completely, while other fibres are contracting, before it is called upon to contract again.

**CHEMISTRY of MUSCLE CONTRACTION**

A steam engine can convert only about ten per cent of the heat energy of its fuel into useful work, the rest is wasted as heat. But muscles are able to use, between twenty and forty per cent of the chemical energy of the food molecules, such as glucose, in contraction. The remainder is converted into heat, but is not wholly wasted, since it is used to maintain the body temperature. If one refrains from contracting the muscles, the heat produced elsewhere in the body is insufficient to keep it warm in a cold place. In these circumstances the muscles contract involuntarily (one "shivers"), and heat is produced thereby to restore and maintain normal body temperature.

When the muscle shortens, it exerts less tension; the tension decreases as the speed of shortening increases. One might suspect that the decrease of tension is due to the
internal viscosity or friction in the muscle, but it is not. If it were, a muscle shortening rapidly would liberate more heat than one shortening slowly over the same distance, and this effect is not observed.

Studies of the energy budget of muscle have shown that a shortening muscle does liberate extra heat, but in proportion to the distance of shortening rather than to the speed. Curiously this "shortening heat" is independent of the load on the muscle: a muscle produces no more - and no less - shortening heat when it lifts a large load than when it lifts a small one through the same distance. But a muscle lifting a large load obviously does more work than a muscle lifting a small load, so if the shortening heat remains constant, the total energy (heat plus work) expended by the contracting muscle must increase with the load. The chemical reactions which provide the energy for contraction must therefore be controlled not only by the change in the length of the muscle, but also by the tension placed on the muscle during the change. This is a remarkable property, of great importance to the efficiency of muscle, and new information about the structure of muscle has begun to explain it.

Chemical analysis reveals that muscle is about eighty per cent water, and the rest being mostly protein, with small amounts of fat and glycogen, and two phosphorus-containing substances, phosphocreatine and adenosine triphosphate. From the chemical point of view, about ninety per cent of the contractile structure of muscle is represented by the three proteins myosin, actin and tropomyosin. Myosin is especially abundant; about half the dry weight of the contractile part of the muscle consists of myosin. This is particularly significant because myosin is also the enzyme
which can catalyse the removal of a phosphate group from adenosine triphosphate (ATP) and this energy-liberating reaction is known to be closely associated with the event of contraction, if not actually part of it.

The actual contractile part of a muscle fibre is a protein chain which shortens by folding of the links. Two proteins are involved in this, myosin and actin, neither of which is capable of contracting alone. When they are combined to form a thread of actomyosin, the potassium and adenosine triphosphate are added, the thread undergoes contraction.

Myosin and actin can be separately extracted from muscle and purified. When these proteins are in solution together, they combine to form a complex known as actomyosin. Some years ago Albert Szent-Gyorgyi, the noted Hungarian biochemist made the striking discovery that if actomyosin is precipitated and artificial fibres are prepared from it, the fibres will contract when they are immersed in a solution of ATP. It seems that in the interaction of myosin, actin and ATP we have all the essentials of a contractile system. This view is borne out by experiments on muscles which have been placed in a solution of fifty per cent glycerol and fifty per cent water, and soaked for a time in a deep-freeze. After this procedure, and some further washing, practically everything can be removed from the muscle except myosin, actin and tropomyosin; and this residual structure will still contract when it is supplied with ATP.

Chemical analyses have been made to determine what substances are used up during muscle contraction. Glycogen, oxygen, phosphocreatine and adenosine triphosphate decrease in amount during contraction, and carbon dioxide, lactic
acid, adenosine diphosphate inorganic phosphate increases. The fact that oxygen is used up and carbon dioxide is formed suggests that muscular contraction is an oxidative process. But this oxidation is not essential, for a muscle can twitch a good number of times even when completely deprived of oxygen. Such a muscle, however, becomes fatigued sooner than one contracting in an atmosphere of oxygen. Furthermore, although we breathe more rapidly during muscular exertion, the accelerated breathing continues for some time after the physical work has ceased. This suggests that oxidation is involved, not in muscular contraction but in the process of recovery from contraction.

The appearance of glycogen and the formation of lactic acid are related, for in the absence of oxygen, the amount of lactic acid formed is just equivalent to the glycogen that disappears. Since the breakdown of glycogen to lactic acid requires no oxygen, and since it liberates energy rapidly, it was once thought that this reaction is directly responsible for muscle contraction. When oxygen is present, the muscle oxidizes about one fifth of the lactic acid to carbon dioxide and water, and the energy released by this oxidation is used to reconver the other four fifths of the lactic acid to glycogen. This explains why lactic acid does not accumulate as long as the muscle has sufficient oxygen, and why a muscle becomes fatigued more rapidly when it contracts in the absence of oxygen.

However it was found that a muscle poisoned with idrocetate, which inhibits the chemical reactions by which glycogen breaks down to lactic acid, can still contract, although it is capable of twitching only sixty to seventy times instead of the two hundred or more times achieved by a muscle deprived of oxygen. By the fact that it can
twitch at all when the breakdown of glycogen is prevented shows that this is not the primary source of energy for contraction.

The other change that can be detected chemically during contraction is a splitting off of organic phosphate from phosphocreatine and adenosine triphosphate, accompanied by the release of energy. It is believed that this is the immediate source of energy for contraction. The bond between phosphate and the organic compounds creatine and adenylic acid is different from that of most phosphate compounds and has been called a "high-energy" or "energy-rich" phosphate bond because energy is released when the bond is broken.

After a muscle has contracted, the breakdown of glycogen to lactic acid and the oxidation of lactic acid in the Krebs citric acid cycle provide energy for the resynthesis of adenosine triphosphate and phosphocreatine. In summary, the chemistry of muscle contractions involve the chemical reactions shown below.

(1) Adenosine triphosphate

(2) Phosphocreatine + ATP

(3) Glycogen Intermediate

(4) Part of lactic acid + Oxygen

Inorganic phosphate
+ Adenosine diphosphate
+ Energy (used in actual contraction).
Creatine + ATP
Lactic acid + Energy (used in resynthesis of organic phosphates).
Carbon dioxide + Water + Energy (used in resynthesis of rest of lactic acid to glycogen and in resynthesis of ATP and phosphocreatine).
It is estimated that the energy from organic phosphates alone could give sustained maximal muscular contraction for only a few seconds.

Our muscles are often called upon to do great spurts of work, and although both the rate of breathing and the heart rate increase during exertion, oxygen cannot be supplied in sufficient quantities to permit these exertions. In such circumstances the muscle is said to have incurred an oxygen debt, which is afterwards repaid by rapidly breathing enough extra oxygen to oxidize part of the lactic acid, which furnishes energy for resynthesizing the rest of the glycogen. In other words, during short spurts of extreme muscular activity, muscles use energy from sources that do not require the utilization of oxygen. After the activity has ceased, the muscles and other tissues pay off the "oxygen debt" by utilizing an extra amount of oxygen to restore the energy-rich phosphate compounds and glycogen to their normal condition. During a long race a runner may reach an equilibrium in which he gets a "second wind", and, because of the increase in breathing and heart rate, he takes in enough oxygen to oxidize the lactic acid formed at that moment, so that the oxygen debt is not increasing.

A muscle that has contracted many times, exhausted its store of organic phosphates and glycogen, and accumulated lactic acid, is unable to contract any more and is said to be fatigued. Fatigue is primarily induced by this accumulation of lactic acid, although animals feel fatigue before the muscle reaches the exhausted condition.

The exact spot most susceptible to fatigue can be demonstrated experimentally if the muscle and its attached nerve are dissected out and the nerve stimulated repeatedly by
electric shocks until the muscle no longer contracts. If the muscle is then stimulated directly, by placing the electrodes on the muscle tissue, it will respond vigorously. With the proper apparatus for detecting the passage of nerve currents, it can be shown that the nerve leading to the muscle is not fatigued; it is still capable of conduction. The point of fatigue, then, is the junction between the nerve and the muscle, where nerve impulses instigate muscle contraction.

**THE NATURE OF CONTRACTION**

Electron micrographs show that muscle fibrils are made of longitudinal filaments of actin molecules that extend through both dark and light bands. These are surrounded in the dark (A) bands by aggregations of myosin molecules that are close to, but not combined with the actin; the two are held apart by electrostatic charges. Contraction appears to occur almost entirely in the dark (A) bands, with the light bands serving as "tendons" between adjacent dark bands. According to one theory, the electrical stimulation of the muscle fibre provides changes in the ionic pattern around the actin and myosin which permit them to come together, form actomyosin, and contract. (Refer figure 2.313).

At present there are two hypotheses of the nature of muscle contraction. One states that the energy for contraction is transferred from ATP to the muscle fibre at the moment of contraction, and that after this energy has been used in the physical shortening of the fibre, the fibre automatically relaxes. The other states that contraction resembles the releasing of a stretched spring, the muscle before contraction being analogous to a spring that has been stretched and held by a kind of trigger mechanism. When this trigger is tripped by the nerve impulse, the spring contracts; energy is then required to restretch the spring for the next contraction.
When a muscle is stimulated repeatedly and becomes fatigued, it does not suddenly lose its ability to contract; instead, its relaxations become slower and more labored, and when fully fatigued, it can no longer relax, but must remain contracted. This condition is called "cramps".

When a muscle contracts, it becomes shorter and fatter, but there is no change in its total volume. As soon as the meaning of the band pattern of striated muscle became apparent, it was obvious that changes in the pattern during contraction should give us new insight into the molecular nature of the process. Such changes can be unambiguously observed in modern light microscopes, notably the phase contrast microscopes and the interference microscope. They can be studied in living muscle fibres or in isolated myofibrils contracting in a solution of ATP, and all coming to the same conclusions.

It has been found that over a wide range of muscle lengths, during both contraction and stretching, the length of the A-bands remains constant. The length of the I-bands, on the other hand, changes in accordance with the length of the muscle. Since the length of the A-band is equal to the length of the thick filaments, then the length of these filaments is also constant. But the length of the H-zone - the lighter region in the middle of the A-band - increases and decreases with the length of the I-band, so that the distance from the end of one H-zone through the Z-line to the beginning of the next H-zone remains approximately the same. This distance is equal to the length of the thin filaments which do not appreciably alter their length.

It was concluded from these observations that, when the muscle changes length, the two sets of filaments slide past each other. Of course when the muscle shortens enough,
the ends of filaments will meet; this happens first with the thin filaments, and then with the thick. Under such conditions, in fact, new bands are observed which suggest that the ends of the filaments crumple or overlap. But these effects seem to occur as a result of the shortening process, and not as causes of contraction.

It has often been suggested that the contraction of muscle results from the extensive folding or coiling of the filaments. The observations support a search for processes which would cause the filaments to slide past one another. It is apparent that the sliding concept gives a much more favourable position with respect to what might be called the intermediate levels of explanation; the description of the behaviour of muscle in terms of molecular changes whose detailed nature is not known, but whose consequences can now be computed. If a muscle is treated with an appropriate salt solution, and then examined under the light microscope, it is observed that the A-bands are no longer present. It is also known that such a salt solution will remove myosin from muscle. This demonstrates that the thick filaments of the A-band are composed of myosin, a conclusion which has been quantitatively confirmed by comparing measurements made by chemical methods with those made by the interference microscope. Moreover, when myofibrils which have been treated with salt solution are examined in the electron microscope, they lack the thick filaments. The "ghost" myofibril that remains consists of segments of material which corresponds to the arrays of thin filaments in the I-bands. If the myofibril is treated so as to extract its actin, a large part of the material in these segments is removed. This indicates that the thin filaments of the I-bands are composed of actin and (probably) tropomyosin.
Thus the two main structural proteins of muscle are separated in the two kinds of filaments. As noted earlier, actin and myosin can be made to contract in a solution of ATP, but only when they are combined. It is therefore concluded that the physical expression of the combination of actin and myosin is to be found in the bridges between the two kinds of filaments. It should also be said that the thick and thin filaments are too far apart for any plausible "action at a distance," so it would seem likely that the sliding movement is mediated by the bridges.

The cross-bridges seem to form a permanent part of the myosin filaments; presumably they are those parts of myosin molecules which are directly involved in the combination with actin. In fact, when calculated, the number of myosin molecules in a given volume of muscle, is surprisingly close to the number of bridges in the same volume. This suggests that each bridge is part of a single myosin molecule. How could the bridges cause contraction? It is suggested that they are able to oscillate back and forth, and to hook up with specific sites on the actin filament. Then they could pull the filament a short distance (say one hundred angstroms) and return to their original configuration, ready for another pull. Each time a bridge went through such a cycle, a phosphate group would be split from a single molecule of ATP; this reaction would provide the energy for the cycle.

To account for the rate of shortening and of energy liberation in the psoas muscle of a rabbit, each bridge would have to go through fifty to one hundred cycles or operation a second. This figure is compatible with the rate at which myosin catalyzes the removal of phosphate groups from ATP. When the muscle is relaxed, the removal of phosphate groups from ATP has stopped, and the myosin bridges can no longer
combine with the actin filaments; the muscle can then return to its uncontracted length. Indeed, there is evidence from various experiments that ATP from which phosphate has not been split can break the combination of actin and myosin. The reverse effect - the formation of permanent links between the actin and myosin filaments in the total absence of ATP - would explain the rigidity of muscles in rigor mortis: when the muscles' supply of ATP has been used up, they "seize" like a piston which has been deprived of lubrication.

The system described is sharply distinguished from most other suggested muscle mechanisms by one significant feature; a ratchet device in the linkage between the detailed molecular changes and the contraction of the muscle. This makes it possible for a movement at the molecular level to reverse direction without reversing the contraction. Thus during each contraction the molecular events responsible for the contraction can occur repeatedly at each active site in the muscle. As a result the muscle can do much more work during a single contraction than it could if only one event could occur at each active site.

Earlier it was mentioned that the tension exerted by a muscle falls off as its speed of shortening increases. This phenomenon can now be explained quite simply if we assume that the process by which a cross-bridge is attached to an active site on the actin filament occurs at a definite rate. There is only a certain period of time available for a bridge to become attached to an actin site moving past it, and the time decreases as the speed of shortening increases. Thus during shortening not all the bridges are attached at a given moment; the number of ineffective bridges increases with increasing speed of shortening, and the tension consequently decreases. A
detailed scheme of this general nature has shown that it can account for many features of contraction.

It was also indicated earlier that the total energy (heat plus work) developed by a muscle contracting over a given distance increases with the tension or load placed on the muscle. This can be explained by this mechanism if the chemical reaction which delivers the energy—say the removal of phosphate groups from ATP—proceeds slowly at bridges which are not attached to an actin filament, and rapidly at bridges which attach at any moment.

The load on the muscle, the amount of energy released in a given distance of shortening is automatically varied according to the amount of external work done. This assumption of a difference in the reaction rate at unattached bridges and at attached bridges is plausible; when myosin is placed in a solution approximating the environment of muscle, it splits ATP rather slowly; when the myosin is allowed to combine with actin, the splitting is greatly accelerated.

The sliding-filament model provides a frame of reference in which many different kinds of information can be related to one another; such as, the muscle itself, the artificial contractile system and the muscle proteins.

There remains the most fundamental question of all: Exactly how does a chemical reaction provide the motive force for the molecular movements of contraction? Little progress has been made towards answering this question; indeed, the recent studies have made the problem more difficult by seeming to require that a movement of one hundred angstroms in part of the muscle structure is the consequence of a single chemical event. But it may be that the sliding process is effected by a more subtle mechanism than the one described here; perhaps a
caterpillar-like action, in which one kind of filament crawls past the other by small repetitive changes of length, will be closer to the truth.
Muscles, unless in their relaxed position, exert an overall tension on their attachments which to some degree is related to the position (or displacement), speed of movement and degree of motion resistance of the limb. Some typical relationships and forces related to the upper extremity limb are discussed in order to give an indication of the magnitude and characteristics of the muscle forces involved.

Consider, for instance, the muscle of the calf of the leg, called the gastrocnemius, which is used in rising of the toes. The muscles origin is at the knee, and its insertion, by way of the tendon of Achilles, is on the heel bone; because the distance from the toes to the ankle joint is at least six times that from the ankle joint to heel, the gastrocnemius is working against an adverse lever ratio of six to one. This means that when a person weighing one hundred and fifty pounds stands on one leg and rises on his toes, the one gastrocnemius muscle is exerting a force of nine hundred pounds; and if a man were to hold another person in his arms and perform this action, the muscle would be exerting a force of nearly a ton.

**Hand Forces**

A knowledge of the forces required in the tasks of everyday living in the human hand, provides a useful guide to determine the necessary forces in artificial limbs and human manipulators. For example, the adjustment of strength of grasp to the job requirements (graded prehension) and the maintenance of grasp (isometric prehension) are natural features which should have their counterpart in the real life artificial device.

Prehension forces of the hand required in the manipulation of common objects and other activities of everyday
### TABLE 2.331
**PREHENSION FORCES IN MANIPULATION OF COMMON OBJECTS** *Refer Klopsteg & Wilson (33)*

<table>
<thead>
<tr>
<th>Object</th>
<th>Description</th>
<th>Force (lb)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Drinking glass</td>
<td>3 in diameter, filled with water; total weight 0.7 lb.</td>
<td>1.75</td>
</tr>
<tr>
<td>Doorknob (survey of 13)</td>
<td>2.2 in average diameter, average force to operate, 3.1 in lb.</td>
<td>0.6</td>
</tr>
<tr>
<td>Pulling on sack</td>
<td>Prehension of folds of sack</td>
<td>2.7</td>
</tr>
<tr>
<td>Manipulation of screw cap</td>
<td>Conventional toothpaste tube, 1/4 in screw cap</td>
<td>2.5</td>
</tr>
<tr>
<td>Holding tablespoon</td>
<td>Spoon weight 0.12 oz held as for drinking glass</td>
<td>1.6</td>
</tr>
</tbody>
</table>

*Not the maximum forces which could be utilized but rather the minimum necessary to perform the activities.*

### TABLE 2.332
**MEASURED FORCES OF PREHENSION**

*Refer Klopsteg & Wilson (33)*

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Palmer</th>
<th>Tip</th>
<th>Lateral</th>
<th>Cross</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>21.5</td>
<td>21.9</td>
<td>21.2</td>
<td>21.3</td>
</tr>
<tr>
<td>Standard deviation</td>
<td>5.8</td>
<td>4.8</td>
<td>4.8</td>
<td>5.8</td>
</tr>
<tr>
<td>Range</td>
<td>15.2–31.5</td>
<td>19.5–26.7</td>
<td>18.5–32.5</td>
<td>18.2–32.4</td>
</tr>
</tbody>
</table>
FIGURE 2.333  

EFFECT OF WRIST ANGLE ON PREHENSION FORCE  

Palmar prehension of a half-inch block.  
Standard position, SP, is at 35 degrees  
dorsal flexion relative to the forearm–hand coaxial reference, 180 degrees.  
Solid line, average; broken lines,  
standard deviations. (Data are from  
Eberhart et al. (33) for eight  
onamputee male subjects.)
living have been measured by Keller (1947) (36) by placing a strain gauge over the palmar pad of the thumb. Typical examples are illustrated in Table 2.331. A further study of actual prehension forces of the human hand have been made by Pick (1911) (29) using the cross section and superficial flexors of the digits of 21.5 square centimeters as a basis of calculation. Applying the Haxton (1944) (37) force factor for human muscles of four (4) Kilograms per square centimeter, a force of 85.8 Kilograms or 190 pounds is obtained, and this represents the force available for finger prehension, but does not directly give the components of force which can act to oppose an object in prehension. Table 2.332 gives the prehension forces for the human hand and it shows that only about 10 per cent of the intrinsic force can be applied in a palmar prehension, while nearly 50 per cent is applied in full grasp. The difference is due partly to shifts of mechanical advantage and to the mobilization of additional muscles, such as the lumbricals which can act more strongly in grasp.

The effect of wrist position upon prehension forces is shown in Figure 2.333. As the wrist is dorsiflexed, the palmar prehension force rises to a maximum at about 35 degrees of dorsal flexion and then declines. The explanation is that the finger flexors are stretched by dorsiflexion, so that they are placed in a more favourable position on their tension length diagrams at moderate excursions - refer Section 2.34. The falling off at more extreme dorsiflexion is due to other causes such as a reduced mechanical advantage.

The human hand is richly supplied with receptors which mediate touch, pressure and muscle and joint senses. These receptors make it possible to recognize objects by shape, to sense magnitudes and directions of external
### Table 2.334
**Prehension Force and Thickness Discriminations**

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Force discrimination (lb)</th>
<th>Thickness discrimination (in)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normal</td>
<td>0.11</td>
<td>0.08</td>
</tr>
<tr>
<td>Amputee, harness</td>
<td>0.15</td>
<td>0.4 (or more)</td>
</tr>
<tr>
<td>Amputee, cineplasty</td>
<td>0.12</td>
<td>0.06</td>
</tr>
</tbody>
</table>

### Table 2.335
**Principal Muscle Groups Action on the Wrist**

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volar flexion</td>
<td>Flexors carpi ulnaris and radialis, palmaris longus, the long thumb flexors</td>
</tr>
<tr>
<td>Dorsal flexion</td>
<td>Extensors carpi radialis longus and brevis and carpi ulnaris, the digital extensor pollicis longus</td>
</tr>
<tr>
<td>Radial flexion</td>
<td>Extensor pollicis brevis, abductor pollicis longus</td>
</tr>
<tr>
<td>Ulnar flexion</td>
<td>Flexor and extensor carpi ulnaris, extensor digitorum ulnaris</td>
</tr>
</tbody>
</table>

### Table 2.336
**Wrist-Flexion Forces (lb)**

Refer Klopstein & Wilson (33)

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscle force</th>
<th>Measured force*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volar flexion</td>
<td>292</td>
<td>50 ± 11</td>
</tr>
<tr>
<td>Dorsal flexion</td>
<td>147</td>
<td>24 ± 7</td>
</tr>
<tr>
<td>Radial flexion</td>
<td>98</td>
<td>38 ± 9</td>
</tr>
<tr>
<td>Ulnar flexion</td>
<td>91</td>
<td>30 ± 9</td>
</tr>
</tbody>
</table>

*Mean ± standard deviation.
forces, to differentiate textures, and to perceive movements of objects. The finger for instance can sense a minimum pressure of 3 gm/sq mm in contrast to 26 for the abdomen, while the volar surface of the finger can distinguish two points only 2.3 mm apart, as compared with a minimum of 67 for the back.

Additionally the large areas of the cerebral cortex given over to the coordination of motion and sensation in the hand (refer figure 2.411) and the multiplicity of peripheral sense receptors is responsible for these abilities of the hand.

Perception of forces and positions for normal and amputee subjects is given in Table 2.334. This illustrates the point that the attachment of the prosthesis to the body at the stump and the suspensory and control harnesses is responsible for replacement of a large amount of the lost sensory feedback information to the brain for dedicated tasks. In the case of cineplasty where muscle sensors are still fully utilised, the discriminations are the same as for the normal hand.

**WRIST FLEXION FORCES**

The principal muscle groups acting on the wrist have been described in Table 2.335. Fick (1911) (22) gives the total cross-sectional area of these muscle groups; and applying the factor of 4 kg/sq cm of Hauston (1944) (37), the total muscle forces as in Table 2.336 were calculated. Experimental results by the University of California found the measured forces were much less than the calculated values - indicating a mechanical advantage of approximately 0.2 to 0.33 relative to the carpometacarpal joint.

The effect of wrist position upon wrist-flexion forces is given in figure 2.337. The volar-flexion forces are
FIGURE 2.337

WRIST-FLEXION FORCES Refer KLOPSTE1 & WILSON (33)
Maximum forces of flexion measured at the carpometacarpal joint at various wrist positions. Solid lines, average; dotted lines, standard deviations. (Unpublished data, University of California at Los Angeles, for 15 nonamputee male subjects.).

FIGURE 2.338

JOINT TORQUE CURVE OF FOREARM FLEXORS Curve A indicates measurements of Miller (14). Curve B shows the sine of the angle of pull as a function of elbow angle.
TORSION FORCES IN THE FOREARM
Standard position, SP. Both curves are averages for 20 nonamputees.
(Unpublished data, University of California at Los Angeles.).
approximately constant at all but the most flexed positions where force diminishes in accordance with the rule of the force-length diagram (refer figure 2.346). Likewise, at full dorsal flexion, forces should be weaker on account of the shortened muscle, but the reduction at the volar position results from the less favourable skeletal leverage.

ELBOW JOINT TORQUE

Forearm flexion depends upon an interaction of skeletal levers and muscle forces which, together, determine the flexor torque about the elbow. Figure 2.338 shows the curve of maximum isometric torque for a normal male subject as reported by Miller (1942) (38). The torque rises to a maximum of about 50 ft lbs at 110 degrees of flexion, after which it again declines. The sine of the angle of pull, which directly represents the effectiveness of the muscle upon the forearm lever, describes a curve of approximately the same shape. But in accordance with the force-length diagram (refer figure 2.346), muscle length and therefore muscle force decreases with flexion angle, and the actual torque falls faster than does the curve of lever effectiveness. This data gives little support for the theory that the combination of lever and muscle effects sufficiently interact to preserve constancy in flexor torque. Actually, declining torque, caused by reduction in muscle force at flexion angles above 100 degrees, displays an opposite trend which, if anything, introduces a fractional disadvantage.

FOREARM TORQUES

Pronation and supination forces at the wrist are graphed in figure 2.339. These isometric measurements were taken from a special wrist cuff and hence they are free of augmentation from the digital flexors and extensors. The familiar effect of muscle length upon force is readily apparent. Greatest
supination force is found at extreme pronation, when the supinators are strongly prestretched; then it declines at each position until, at 90 degrees of supination, a minimum force is registered. Similarly but in the opposite sense, pronation force is also seen to be a function of wrist position.
2.34 CINEPLASTY and MUSCLE MECHANICS

Cineplasty is the surgical formation of a skin lined tube (muscle tunnel) constructed in the muscle as a durable and efficient linkage between the human body muscle and the mobile mechanism of an artificial aid - prosthesis. The insertion of a pin and linkage enables energy of the contraction of muscles to be transmitted to the prosthesis. Cineplasty relies on the known ability of a muscle to learn a new and unnatural function. Actual muscle forces have been measured directly as a result of this procedure.

Early workers suggested the use of muscles and tendons, but Vanhetti (1928) was the first to use muscular control directly onto prosthesis. Some of his ideas were tried out on human subjects by Casi (1935) involving loops of tendon or moveable " clubs " at the end of the human limb stump. The work of Sauerbruch (1945) resulted in the skin lined muscle tunnel, that is considered standard today. The forces and displacements for a considerable series of tunnels was first measured byeyer (1928) (42). Muscle strengths, and excursions of amputees, and muscle physiology studies were carried out by U.S. Army (1946) (43). Typically a limbs muscle force of 5 lbs was measured for a ninety degrees extension flexion elbow angle.

Floydray and Wilson (1954) (33) have recorded details of hand prehension forces, wrist flexion and extension flexion forces and torsions.

A basic understanding of the principal features of muscle mechanics is essential. For instance, individual muscles have a limited useful excursion proportional to its total length. Moreover the tensions developed are not constant throughout the range of excursion but vary as
different lengths of muscles, becoming less as the muscle shortens and greater as it lengthens. These variations in tension are not observed in one's ordinary movements because of the existence in the body of the compensatory mechanism which maintains relatively constant movements around joints.

The arrangement of very short effective lever arms through which the muscles act in proportion to the much larger resistance lever arms, results in exceedingly high muscle forces to produce utilisable forces at the ends of the resistance arms. This is also necessary to give the speed of motion manifest in everyday activities.

The fundamental principles of muscle mechanics have been recorded using two types of experimental procedures. In one, a muscle is caused to contract without being allowed to shorten. The muscle generates force (tension) but does no external work. Such a contraction is called "isometric" and is involved in trying to crack a nut between the teeth. In the second type, a muscle, caused to contract, is allowed to shorten, and external work is done. A contraction of this type, called an "isotonic" or "free" contraction (load excursion), takes place when, for example, a weight is raised in the hand. With cineplastic amputees, both types of contraction are required to activate a prosthetic device.

**MUSCLE LENGTH - TENSION DIAGRAM**

The intact muscle, as found in the body, consists essentially of two elements, contractile muscle, made up from the aggregation of fleshy muscle fibres; and elastic tissue, later called "passive", contributed predominantly by the connective tissues and partly by the sarcolemma of the fleshy fibres. Because of the elastic nature of the connective tissues, a muscle isolated from the body will
FIGURE 2.341  TYPICAL MUSCLE LENGTH–TENSION CURVE
assume a certain length to which it will return if passively stretched and released. This length, defined as the "rest length", is used as a reference point for comparison of data from different muscles.

The series of curves obtained by plotting tension against length is called the "length-tension diagram" of a muscle (refer figure 2.341). It is evident that, if an isolated muscle is to be stretched by graded increments beyond its rest length, external force or tension will have to be applied to overcome the resistance of the passive elastic elements. The resulting logarithmic curve is called the "passive tension curve" since it is obtained without stimulation of the contractible elements and is due wholly to the elasticity of the connective tissue.

Another fundamental relationship exists between the tension which the muscle is capable of developing during contraction and the existing length of that muscle. If the muscle is held at lengths, successively from the rest length to shorter and longer lengths in increments and then stimulated to contract without shortening (isometric contraction), plotting the measured tension against length produces the total tension curve.

Above the rest length, the passive tension required to stretch the muscle if subtracted from the Total Tension (Passive + Contractile) at a length will isolate the tension resulting from the contractile elements of the muscle tension. The resulting contractile length-tension curve is parabolic with the maximum tension in the region of the rest length.

Two basic types of variations exist for the length tension curves of muscles - namely, (1) Variations in passive tension and (2) Variations in developed tension resulting from
FIGURE 2.342
IDEALISED LOAD–EXCURSION DIAGRAM
the number of muscle fibres undergoing contraction.

The passive tension of a muscle is due predominantly to the connective tissue. Generally muscles which lie more distally in the extremities appear to possess a proportionately longer amount of connective tissue and therefore exhibit a steep passive-tension curve. Additionally, "muscle bound" individuals yield steep passive-tension curves, indicating a high content of connective tissue.

In 1933 Hill (44) established that velocity of muscle shortening is reduced as the tensile load is increased. Hill's equation indicates that velocity and force are related by a rectangular hyperbola.

Fenn and Harsh (45) found force to be an exponential function of velocity while Ramsey (46) showed only approximate agreement to both of these findings. However, these formulations must be strongly qualified in terms of the condition of observation - such as, type of muscle fibre arrangement, and the amount and elasticity of sarcolemma and other connective tissue investing the muscle.

THE LOAD - EXCURSION DIAGRAM

In obtaining the above Parabolic length-tension diagram the muscle is permitted to do no work (isometric contraction). If, however, a plot of load versus excursion (length) is made for a muscle lifting increasing loads (isotonic contraction) a linear curve results (refer figure 2.342). That is, the greater the load, the less the muscle can shorten.

EXPERIMENTAL RELATIONSHIPS

Actual experimental measurements are recorded by Buchthal and Kaiser (1949) (47), Blaschke and Taylor (1953) (48) and Klapsteg and Wilson (1954) (33).
FIGURE 2.343  MUSCLE DYNAMOMETER FOR ISOMETRIC LENGTH–TENSION MEASUREMENTS

Refer Klopfsteg & Wilson (33)
Total-tension 1, passive P, and developed-tension Δ curves for flexor muscles of the human forearm. Refer Klopsteg & Wilson (33)
BICEPS FORCE-LENGTH CURVES
Fine lines are the curves for biceps tunnels of five amputees; ••• mean of the five cases; o-o "most probable curve" (11).
Isometric Length Tensions Relationships are measured for flexors and extensors of the forearm, some through biceps brachii, some through the triceps, and some through the pectoralis major muscles using cineplastic muscle tunnels. Since these muscles had been freed from their long insertions and deprived of their compensating skeletal lever arms, they are ideal for the study of muscle action. Length and Tension were measured by strain gauge dynamometers (refer Figure 2.343). A compression ring was used to guarantee that the limb was not moved relative to the dynamometer.

Typically Figure 2.344 shows the relationship between length and tension in the passively stretched (P) and isometrically contracted (I) flexors of the forearm. Subtraction of these curves yielded the contractile tension \( \Delta \) Curve. Experimental results particularly with the gastrocnemius muscle of the frog and rat show that maximal tension occurs at a length somewhat greater than the rest length of the muscle. The pectoralis major muscle of the human subject also exhibits this property.

The ability of a muscle to develop tension is related directly to the cross sectional area of the muscle. That is, halving the muscle bundle section-halves the tension, while halving the muscle length does not affect the ability of the muscle to develop tension, but obviously affects the rest length.

A typical cineplastic biceps force-length curve is shown in Figure 2.345. The variations in results emphasize that the force-length curve for a given muscle cannot be predicted accurately, but that each muscle tunnel must be individually tested. All measurements were made with strain gauge dynamometers whose muscle forces can be registered with
FIGURE 2.346

EFFECT OF ELBOW ANGLE UPON BICEPS FORCE-LENGTH CURVE Refer Klopsteo & Wilson (33)
Both curves are averages for three amputees. Shaded area indicates loss in force-length area caused by elbow position.

FIGURE 2.347

PECTORAL FORCE-LENGTH CURVES
Fine lines are the curves for pectoral tunnels of five amputees; ••• mean of the five cases; o-o "most probable curve" (11).
FIGURE 2.348

ADAPTATION OF MUSCLE DYNAMOMETER FOR ISOTONIC LOAD-EXCURSION MEASUREMENTS

REFER KLOPSTEG & WILSON (33)
LOAD EXCURSION CURVE FOR PECTORALIS MAJOR MUSCLE. Refer Kloosteg & Wilson (33).
Linear relation between load and excursion for freely weighted right pectoralis major of human subject.

PARABOLIC RELATION BETWEEN LOAD AND WORK FOR MUSCLE

Parabolic relation between load and work for muscle of Figure 2.349. Values used to calculate curve were determined from straight line of Figure 2.349 extrapolated out to zero shortening.
negligible shortening, thus giving an isometric force measurement. The average maximum isometric curve is virtually linear throughout the entire range from extreme shortening to the tolerable extent of stretching. Additionally the force length diagram varies with the degree of forearm flexion. In figure 2.346 the forearm extended reduces the force about three pounds at the rest length and progressively more with shortening, but at a forearm flexion of ninety degrees the curve is almost linear. Moreover, surgical freeing of the biceps from lateral attachments with fascia or surrounding muscle elements does not explain this effect.

Experimental results for the shoulder rectoral muscle tunnel is given in the force-length curve (figure 2.347). In this case the maximal isometric force curve is convex upwards in the shortening phase and is in contrast to the linearity of the biceps curve.

Isotonic Load Excursion Relationships were measured using the dynamometer fitted as in figure 2.348. Muscle contractions were rather slow at two to five seconds, with the emphasis on maintaining uniform speed of contraction.

Figure 2.349 shows the linear relationship between load and shortening in the pectoralis major of the shoulder. Extrapolating this line out to zero shortening yields a value for the load which agrees with the maximal isometric tension.

LOAD-WORK DIAGRAM

Figure 2.3410 shows the load-work diagram for the muscle of figure 2.349. It follows that if the shortening \( s = x - 1 \) (load) + b and the work done is the product of "s" and "I" that the work \( l.s = ml^2 + bl \), a parabolic equation. Experiments show that the maximal work obtained by measuring the area under the \( \Delta \) curve (refer figure 2.344)
FIGURE 2.3411  RELATION BETWEEN LOAD AND MAXIMAL VELOCITY FOR HUMAN PECTORALIS MAJOR MUSCLE
Broken line is theoretical curve for same muscle after longitudinal section.

FIGURE 2.3412  RECORD OF HUMAN PECTORALIS MAJOR SHORTENING
Record of human pectoralis shortening under a load of 0.32 kg. Lower line is a reference line for measurement purposes.
Refer Klopsteg & Wilson (33)
### FREE CONTRACTIONS IN HUMAN MUSCLE
Starting at near rest length.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Maximal velocity (cm./sec.)</th>
<th>Maximal excursion (cm.)</th>
<th>Minimal time for total shortening (sec.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pectoralis</td>
<td>145</td>
<td>8.7</td>
<td>0.10</td>
</tr>
<tr>
<td>Biceps</td>
<td>92.5</td>
<td>0.4</td>
<td>0.17</td>
</tr>
<tr>
<td>Triceps</td>
<td>55</td>
<td>4.0</td>
<td>&lt;0.10</td>
</tr>
</tbody>
</table>
is twenty per cent greater than experimental results as in figure 2.3410.

**LOAD VELOCITY DIAGRAM**

Figure 2.3411 shows the load-velocity diagram for the human pectoralis major muscle. The record of the pectoralis shortening under the minimal load of 0.32 Kg is shown in figure 2.3412. The entire shortening of 2 cm requires only about 0.12 seconds or 75 cm/second average velocity. However in this case the maximal velocity is approximately twice the average velocity at 143 cm/sec.

The load-maximal velocity curve can be fitted to an empirical equation as shown by Hill (1938) (44) by:

\[ (p + a) \cdot (v + b) = (Po + a) \cdot b = \text{constant} \]

where \( P \) = load, \( v \) = maximal velocity, \( Po \) = maximal isometric tension at initial length, and \( a \) and \( b \) are constants to be determined.

Table 2.3413 provides data on free contractions of the sternal portion of the pectoralis major, the biceps, brachii, and the triceps.

For zero load, the maximal velocities are proportional to the fibre length and maximal excursion. For Table 2.3413 the ratio of fibre lengths are 0.32, 0.65 and 0.59 respectively. Haines (1934) (49) found that the maximal excursion of straight fibred muscles in the intact human body averaged 57 per cent of the maximal extended length.

**POWER**

Since the curve relating load and maximal velocity can be fitted by an equation of the form \((P + a) \cdot (v + b) = \text{a constant}\), it follows, as Hill (1938) (44) has shown, that the load at which maximal power is developed may be determined by differentiating the product \(P \cdot v\) with respect to \(P\) and setting this slope or differential equal to zero for maximum conditions. Subsequently the maximal
FIGURE 2.3414  FORCE-VELOCITY CURVE FOR MUSCLE GROUP
CAUSING ELBOW FLEXION Refer Harrison (50)
Curve (1), values apply at the hand.
Curve (2) constant power curve, value 112 ft lb/sec.
power occurs when \( \frac{P}{P_0} = \frac{a}{P_0} \left( \sqrt{1 + \frac{P_0'}{a}} - 1 \right) \), Where (\( P/P_0 \)) is a fraction of the isometric tension and is insensitive to large changes of (\( a/P_0 \)). Thus it follows that an isolated muscle will develop maximal power when lifting a load equal to about one quarter to two fifths of the maximal isometric tension the muscle can develop at initial length. Using the value of \( a/P_0 = 0.81 \) for the Muscle of figure 2.3411, the maximal power of 37 Watts or 0.049 Horsepower occurs when the load is 8.1 Kilograms, corresponding to a velocity of 45 cm/sec.

The variation of muscle speed during the contraction period of the motion cycles has been recorded by Harrison (1963) (50), particularly in respect to externally forcing the motion of muscles to increase power output. Typically figure 2.3414 shows the Force-Velocity diagram for the muscle group causing elbow flexion. In this case the maximum power and the average power output is 112 ft lb/sec. Results showed that the maximum average power values were 114 ft lb/sec. for a constant velocity input, 104 ft lb/sec. for the constant acceleration, constant velocity and constant retardation input, and 99 ft lb/sec. for the simple harmonic input. Comparing these results shows the greater the initial acceleration and final retardation of the input motion, the greater may be the average power output.

**ABSOLUTE MUSCLE FORCE**

The variability of the value of the absolute muscle force renders it useless for comparison purposes. However the ratio of isometric force to the muscle physiological cross section yields comparable results. Haxton (1944) (37) found an average tension value of 3.9 Kg/sq cm for ankle flexors using mid portion muscle.
MAXIMAL DEVELOPED FORCE IN ISOLATED HUMAN MUSCLE

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Cross section (sq. cm.)</th>
<th>Maximal developed force (kg.)</th>
<th>Force per unit cross section (kg./sq. cm.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pectoralis</td>
<td>12.8</td>
<td>20.9</td>
<td>1.63</td>
</tr>
<tr>
<td>Biceps</td>
<td>9.15</td>
<td>21.8</td>
<td>2.38</td>
</tr>
<tr>
<td>Triceps</td>
<td>15.98</td>
<td>20.9</td>
<td>1.31</td>
</tr>
</tbody>
</table>
lengths and muscle fibre corrections. Table 2.3 provides data on the maximal developed force in isolated human muscles for the sternal portion of the pectoralis major, the biceps brachii, and the triceps.
FIGURE 2.351

DIAGRAMMATIC REPRESENTATION OF A MUSCLE CONTRACTING AGAINST AN EXTERNAL LOAD

---

FIGURE 2.351(a)

ELBOW MUSCLE TIME CONSTANT VERSUS FORCE CURVE

Typical curves showing relation between the product $G(dP/dv)$ and force in muscles causing elbow flexion; the values apply at the hand. Refer Harrison (50)
The human muscle consists of a contractile element and non-linear spring element of negligible mass connected in series. At one end A the contractile element is fixed and the muscle power output is taken from the point B at one extremity of the spring element. (Refer figure 2.351 by Harrison (1963) (50) for details).

The relationship between the velocity of contraction of the muscle $V$ and the load force $P$ is given by the equation $V = v + G \frac{dP}{dt}$ (1)

where $V$ is velocity of contraction of muscle, ft/sec.
$v$ is velocity of power transfer point B, ft/sec.
$P$ is muscle or load force, lbs.
$G$ is compliance (reciprocal of stiffness) of series elastic component, ft/lb.

Since the compliance $G$ varies with the force $P$ by $G = f(P)$, equation (1) becomes $V = v + (G \frac{dP}{dV}) \frac{dV}{dt}$ (2)

The product $G \frac{dP}{dV}$ is called the muscle time constant. Typically figure 2.351a shows the relationship between the muscle time constant and muscle forces during elbow flexion. Since the muscle time constants are relatively small compared to the muscle contraction time under load, the power output is large. For the elbow flexion muscle group, the muscle time constant is 0.03 seconds and the contraction time is of the order of 0.3 seconds.

The relationship between the force exerted by a muscle and the speed of contraction was shown to be hyperbolic in Section 2.34. From the mechanical point of view this force speed relationship makes muscles very different from other devices from which energy may be released.
The muscle force for a given muscle is independent of displacement but dependent on the instantaneous speed. Consequently there exists a pair of corresponding force-speed values whose product is a maximum. Thus knowing the mean muscle speed over a large number of cycles is not sufficient to determine maximum power output, since muscles generate maximum power at a particular speed of contraction.

The third property of interest is that exhibited by the connection between the muscular contractile machinery and the skeletal members. The characteristic of this connection is called the "series elastic component" and its behaviour is similar to a non-linear spring. The connection spring stiffness increases with increase in its load. Measurements of the magnitude of the elastic effect of the connections was made by Wilkie (1950) (15). During elbow flexion, the upperarm was fixed, and the elastic effect of the connections between the forearm and the muscles causing flexion was measured at the hand. The actual measured value was smaller than the actual stiffness of the connections, the stiffness ratio being equal to the square of the lever ratio existing for the muscle group.

The spring stiffness at the connection is given by

$$\frac{n \cdot \Delta P}{(1/n) \Delta S} = n^2 \frac{\Delta P}{\Delta S} = n^2 \times \text{Stiffness}$$

This follows from the fact that an increment of force $\Delta P$ causing an increment of displacement $\Delta S$ at the hand corresponds to an increment of force $n \cdot \Delta P$ and of displacement $(1/n) \Delta S$ at the muscle connection, where $n$ is the lever ratio.

Typically Wilkie (1950) (15) showed values of stiffness for muscles causing elbow flexion from 1.4 lb/in. measured at the hand with no tension in the muscle, rising to 6 lb/in. with a tension corresponding to the maximum
FIGURE 2.352  
BICEPS FORCE VERSUS STEADY STATE VELOCITY OF SHORTENING.
Both measured at the hand. (From Hill (44), Wilkie (15) and Bigland and Lippold (52).
FIGURE 2.353  SCHEMATIC MODEL OF A SINGLE MUSCLE

FIGURE 2.354  EFFECTIVE SPRING CONSTANT OF BICEPS VERSUS MUSCLE TENSION
Both measured at the hand. (From Wilkie (15)).
isometric muscle force.

A similar analysis of the human muscle contractile element including the dynamics of two identical muscles acting on a simple limb member is recorded by Vickers (1968) (51). This differs from the previous study in that the neuromuscular (motor) system dynamics are also included. The neuromuscular system includes sensory and motor neurons at the spinal cord level and their associated muscles, joints, and receptors in the periphery.

The inclusion of a percentage neuromuscular stimulation variable \( Z \) into the hyperbolic force velocity curve is graphically shown in figure 2.352 and is given by the equation

\[
F = (Z) F_0 \left( 1 - \left(1 + \frac{1}{n}\right) \frac{V}{V_0} \right)
\]

where \( n \) is a constant.

This neuromuscular stimulation \( Z \) (the ratio of the actual stimulation to the maximum) was shown to be proportional to the electromyographic (EMG) muscle signals by Bigland and Lippold (1954) (52). Although results were for steady state conditions the same linear relationship holds satisfactorily up to the maximum allowable force, which may be as much as twice the maximum isometric force. The interesting point about this relationship is that the stimulation \( Z \) in effect controls the damping as well as the isometric (no work) force.

The addition of a typical muscle contracting spring model in series with the steady state model is shown in figure 2.353. Muscle spring constants \( K \) plotted against muscle tension as measured by Wilkie (1949) (15), and as shown in figure 2.354 were in agreement with the muscle model equations. This assumes that the effect of increasing the stimulation \( Z \) is to increase the active cross section of
FIGURE 2.355  SCHEMATIC OF THE LIMB AND MUSCLE SYSTEM
REFER HARRISON (50)
the muscle allowing the spring constant $K$ to be related to the average stimulation $\bar{Z}$. The effect of forearm elbow flexion angle variations with muscle length (refer figure 2.346) is partly cancelled by the changing lever arms and was neglected in this model.

The basic neuromuscular model was then extended by placing two such identical muscles together onto a limb system as shown in figure 2.355.

The equations of motion then became as follows:

Velocity $V_1 = \ell \cdot \dot{\theta} + \frac{1}{Ko \cdot \ell \cdot \bar{Z}_1} \cdot \ddot{t}_1$

Velocity $V_2 = -\ell \cdot \dot{\theta} + \frac{1}{Ko \cdot \ell \cdot \bar{Z}_2} \cdot \ddot{t}_2$

Torque $T_1 = To \cdot \bar{Z}_1 \left(1 - \beta \cdot \frac{V}{V_0}\right)$

Torque $T_2 = To \cdot \bar{Z}_2 \left(1 - \beta \cdot \frac{V}{V_0}\right)$

Inertia Torque $I \cdot \ddot{\theta} = Md + T_1 - T_2$

Where $\ell$ is the lever arm

$\beta$ is a constant = $1 + 1/n$

$Md$ is disturbance moment

$I$ is moment of inertia of limb

$V_0$ is maximum isometric Velocity

$\bar{Z}$ is the average stimulation

The resulting equations were non-dimensionalised, and linearised by defining $Z (r) = \bar{Z} + Z' (r)$ where $Z'$ is the time varying average stimulation. In these terms it is evident that the dynamic properties of the neuromuscular system can be changed directly by changing the level of stimulation.

The above simultaneous equations are solved using $A$, $B$, $C$ and $D$ to represent the Laplace Transforms of $w(r)$,
FIGURE 2.356

BLOCK DIAGRAM OF MUSCLE SYSTEM WITH AN ASSUMED FEEDBACK LOOP
Refer Wilkie (51)
\[ \phi(r), \delta(r) \text{ and } m_d(r) \text{ respectively, and the transformation becomes:} \]

\[ B(s) = \frac{1}{s} A(s) = \frac{C(s) + (s + 1) D(s)}{S(s^2 + R_s + \delta)} \]

Where \( s \) is the Laplace Transform variable

\[ R = \frac{I \cdot \frac{V_o^2}{T_o} K_o}{\beta^2} \]

\[ \delta = \frac{Z_1}{Z_2} \]

\[ W = \frac{\beta \cdot \frac{d}{V_o} \cdot \frac{d \theta}{dt}}{\theta} \]

\[ \phi = \frac{\frac{\phi^2}{K_o}}{T_o} \cdot \theta \]

\[ \delta = \frac{z_1'}{z_2'} \]

\[ m_d = \frac{M_d}{T_o} \]

\[ v = \frac{V_o \cdot \frac{E \cdot K_o}{\beta}}{T_o} \cdot t = \gamma \cdot t \]

If \( \delta \) is assumed proportional to the error in position with \( \delta \) constant, such as in a tracking task or holding against a disturbance force, then the Block Diagram of such a muscle system can be drawn as in figure 2.356.

Impulse response experiments by Young and Starks (1965) (53) were found to agree with these equations for values of \( \gamma = 22 \cdot s^{-1} \) and \( k/R = 0.054 \).

Similar results were obtained by McRuer, Magdaleno and Moore (1968) (54) in their work entitled "A Neuromuscular Actuation System Model". Additionally, detailed models for the muscle spindle and spinal nervous activity in the neuromuscular system were shown to agree with data available for the dynamic properties of neuronal and muscular elements. Again as in the studies of Harrison and Vickers (1968) (50 & 51) mentioned above, a simple muscle model was
Figure 2.357
Agonist/Antagonist Muscle Pair

Figure 2.358
Isometric Tension-Length Curve
Refer: M. Ruer et al (54)
developed for a limb muscle system. In this case the model was restricted to relatively small movements such as involved in a tracking task by an operator using a control stick manipulator. In such a system as shown in figure 2.357 the agonist/antagonist muscle pair activity is in operation and each muscle has an average tension Po. For motion to occur, one muscle must generate a force greater than the other. The relationship between tension and length in response to motor nerve commands is expressed in terms of the tension length and force velocity curves for muscles. (Refer Section 2.34 for details).

Typically figure 2.358 shows a family of isometric tension length curves for varying levels of stimulation by the motor nerve firing frequency f. This result is consistent with electromyographic movements of muscles made by Bigland and Lippold (1954) (52) and Close, Nickel and Todd (1960) (55).

The equation for the tension - length curve expressed as first order terms in a Taylor series expansion is:

\[ P = P_o + C_f \Delta f - K_m \Delta L \]

Where \( P_o \) is tension at the operating point
\( \Delta f \) is the change in average firing rate or average electrical activity.
\( \Delta L \) is the change in muscle length.
\( K_m \) is the slope of the tension-length curves for constant \( f \).
\( C_f \) is the slope of the tension-length curves for constant length. This tends to increase as \( f_o \) increases.

Note that \( C_f \) and \( K_m \) are evaluated at the operating point defined as \( f_o \) and \( L_o \).

Similarly the equation for the force-velocity data
FIGURE 2.359 SCHEMATIC OF LIMB/MANIPULATION DYNAMICS FOR NEUROMUSCULAR ACTUATION SYSTEM

REFER McRuer et al (54)
FIGURE 2.3510  **Bode Diagram for Limb/Manipulator Dynamics**

FIGURE 2.3511  **Simplified Diagrammatic View of a Muscle Spindle and Its Orientation with Muscle**

Refer: McRuer et al (54)
curves as shown in figure 2.354 is the same as used by Vickers (1968) (51). Moreover, the use of a Taylor's series for the steady rate tracking of small perturbation motions yields the following force-velocity equation:

\[ F = P_o + C f \cdot \Delta f - B_m \cdot \Delta V - K_m \cdot \Delta L \]

Where \( B_m = P_o(1/b + 1/V_m) \), a direct function of \( P_o \)

\( K_m \) is function of \( P_o \)

Data recorded by Granit (1958) (56) indicates that \( K_m \) is a linear function of \( P_o \) for isometric tension.

Typically figure 2.359 shows the limb/manipulator system schematic for a simple spring mass damper manipulation and the small signal level muscle model. The linearised equation for a muscle given above corresponds to a force source \( P_o + C f \cdot \Delta f \), coupled to a parallel spring/viscous damper combination. The damper element has a damping coefficient which is linearly related to the operating point tension. Since the limb and manipulator are in parallel, the two are lumped together in the single effective inertia \( M \).

The system Transfer Function between limb rotation and differential firing rate is given by the equation:

\[ \frac{C_f}{\Delta f} = \frac{(K_m + K_c)}{(T_{M_1} \cdot s + 1)(T_{M_2} \cdot s + 1)} \]

The dynamics of this equivalent system are illustrated for two cases of tension by the \( jw \)-Bode diagram - Refer figure 2.3510. From this it is apparent that the effect of changing the tension of the muscle group is to decrease the low-frequency pole and increase the high-frequency pole, but in terms of the neuromuscular system dynamics the increase in steady state tension is most significant.

The control of neuromuscular behaviour is dependent on a complex organ called the muscle spindle location in
FIGURE 2.3512  REFLEX ARCS OF MUSCLE SPINDLE

REFER MC ROER ET AL (54)
**Figure 2.3513**: Schematic of Muscle Spindle Model

Refer: McRuer et al. (54)
the muscles.

A simplified diagrammatic view of a muscle spindle is shown in figure 2.3511 and is further described in Section 2.3. The muscle spindle is in itself a complex neuromuscular integrative system receiving a continuous barrage of motor control or command signals from the central nervous system and sending a constant stream of sensory signals via its several paths back to the central nervous system.

The overall position of the muscle spindle to the surrounding muscle and spinal cord command system is shown in figure 2.3512. The sensory fibre is seen to form a feedback path to alpha command cells in the cord, which supply the muscle on which the spindle lies. Thus an increase in muscle length or contraction of spindle, by increasing spindle firing (stimulation), reflexly induces increased motor cell firing, thus tending to offset the increase in length. For the same reason, command signals to cause muscle contraction can be initiated in two ways: either directly via the alpha or indirectly via the gamma control to the spindle and, by the feedback loop described, back to the alpha motor neuron.

Examination of the properties of the component parts of the muscle spindle revealed that the macro model as shown in figure 2.3513 can be applied directly to the fibre elements since they are micro muscles. Such a muscle spindle model is shown in figure 2.3513 where the model inputs are the dynamic and static fibre firing rates and the muscle length, with the primary ending firing rate being the output. The primary endings firing rate is assumed to be proportional to the nuclear bag deformation $X_s$ and hence proportional to spring deflection. The intrafusal muscle
FIGURE 2.3514  ELEMENTARY NEUROMUSCULAR SYSTEM MODEL.

REFER McROER et al. (54)
dynamic \( Y_d \) and static \( Y_s \) fibre inputs act upon these muscle fibres. The model comprises a force generator, an equivalent spring and damper. In the case of the static fibres, the viscous elements are not applicable and this enables simplification of the model since the effect of the static fibre force generator on the nuclear bag deformation can be described by an equivalent length input \( \gamma c \) and an equivalent nuclear bag deformation change \( X_n \).

The equations of motion are obtained by summing the forces at each node to zero and solving for the nuclear bag deformation \( X_s = (\gamma c + X) - X_n \) which is proportional to the primary ending firing rates. The equation becomes

\[
X_s = \frac{K_r (S + 1/\tau_k) (\gamma c + X) + Pd/B_n}{(S + 1/a \tau_k)}
\]

where \( \frac{1}{\tau_k} = \frac{K_n}{B_n} \)

\( \frac{1}{a \tau_k} = \frac{K_r + K_n}{B_n} \)

\( \frac{1}{K_r} = \frac{1}{K_h} + \frac{1}{K_s} \)

Thus the primary ending has a lead/lag response to muscle length and/or static fibre input. The response to dynamic fibre input is a simple lag. The lead break frequency is dependent on the operating point dynamic fibre stimulation frequency, whereas the lag break frequency depends on both dynamic and static fibre inputs.

The muscle spindle model is further expanded by applying it to the single loop feedback system as shown in figure 2.3512. The resulting neuromuscular system model as shown in figure 2.3514 provides four functions in one
entity; namely (1) sensory feedback of limb position, (2) some lead/lag series equalisation whose parameters are controlled by the steady-state gamma bias signal $\gamma_b$, (3) the source of one command to the system $\gamma_c$ and (4) a means for adjustment of the steady-state bias signal to the muscle. The spindle output incremental firing rate $\Delta f_{sp}$ is summed with an alpha motor neuron command input $\alpha_c$ with the result, (after conduction and synaptic delays) being an alpha motor neuron incremental firing rate $\Delta f$. This in turn perturbs the muscles and manipulator, giving rise to limb movement, which is then sensed by the spindle system.

The effective damping in the limb-manipulator dynamics transfer function is set by the operating point muscle tension $P_0$ which is due to the alpha motor neuron firing rate $f_0$, and is shown by the total gamma bias $\gamma_o$ entering the Gm block.

Close agreement exists between the dynamics of the overall human neuromuscular and manipulator system when tracking a random appearing command input (refer McRuer, Graham, Krendel and Reisener (1955) (57)) and those of the human neuromuscular system to force disturbance inputs. (Refer Houk (1963) (58) and Sun, Eisentein and Bomze (1966) (59).
FIGURE 2.41  SPINAL CORD AND CEREBRO-SPINAL NERVES
2.4 NERVE STRUCTURE OF THE ARM

The central guiding and controlling force for all systems, muscles, and tissues of the body is the central nervous system, consisting of (i) the brain, (ii) the spinal cord, and (iii) the cerebro-spinal nerves.

The brain is universally recognised to be the organ of "mind", containing all the mental faculties of "will, judgement and reason", thought and memory. (Refer Nomenclature). It is contained within the skull-box or cranium, and from it, twelve pairs of nerves - the cranial nerves - are given off, which pass through various holes in the base of the skull.

The spinal cord is the continuation of the brain down the neural canal of the spinal column. The spinal cord portion of the brain lies within the Vertebral Canal and is a continuation from the Ventricles of the brain. Branching off from the spinal cord are thirty one pairs of cerebro spinal nerves which travel to all parts of the body.

Figure 2.41 shows the general distribution of the spinal nerves, emerging from the spinal cord. The spinal nerves fall into four main categories; the cervical Thoracic, Lumbar and Cauda, Equina Sacral and Cocoygeal Nerves. Of main interest here are the Thoracic and Cervical nerves since they supply all the nerves, both motor and sensory, to the upper extremity limbs and trunk of the body.

In order to simplify the content of figure 2.41 the following explanation is included related to terminology used.

Central Canal contains central spinal fibres and connects with Ventricles of the brain.

Cervical Enlargement of spinal cord where nerves to the arms originate.
FIGURE 2.42  NERVOUS CONNECTION BETWEEN THE CENTRAL NERVOUS SYSTEM AND THE PERIPHERY (Muscle and Sense Organs).
Thoracic Section of spinal cord and upper part of the lumbar region—lateral horns contain nerve cells from which sympathetic nerves arise.

Lumbar Enlargement of spinal cord where nerves to legs originate.

White Matter contains nerve fibres travelling to and from the brain and also linking various parts of the cord itself.

Grey Matter contains nerve cell bodies.

Posterior Horns contain cells which synapse (description of linking of nerves) with ingoing (afferent) nerves whose cell bodies lie in the Posterior Root Ganglia outside the cord.

Anterior Horns contain cell bodies whose fibres carry outgoing (efferent) or motor messages to voluntary muscles.

The Posterior Sensory Fibres travel with the Anterior or motor fibres in the same Spinal nerve.

The cerebro—spinal nerves are those nerves which are given off as branches from the brain and spinal cord. A nerve itself is like a telegraph wire which transmits messages either to or from the brain, and terminates at what is called a nerve—ending. These nerve endings fall into two categories of either (a) Motor or (b) Sensory.

Figure 2.42 shows diagrammatically the structure of some of the receptors or sense organs with which the greatest percentage of "feedback" to the brain is obtained. It should be noted here that some sensory fibres have no end organs at all; and the "naked axis cylinder" (Nerve Fibre central conducting media) branches repeatedly and the branches end among the ordinary cells of the skin or connective tissues.

The nerve trunk itself, that is the nerve connecting receptors to the spinal cord contains several thousand fibres,
<table>
<thead>
<tr>
<th>Forearm (Cont.)</th>
<th>Hand</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>Flex.digitor.sublinus</td>
<td>Flex.poll.brev.</td>
</tr>
<tr>
<td>Flex.digitor.profund.</td>
<td>Flex.poll.brev.</td>
</tr>
<tr>
<td>Pronator quadrat.</td>
<td>Opponens poll.</td>
</tr>
<tr>
<td>Flex.carpi.uln.</td>
<td>Flexor digit.V.</td>
</tr>
<tr>
<td>Palmaris long.</td>
<td>Opponens dig.V.</td>
</tr>
<tr>
<td>Abduct.poll.brev.</td>
<td>Adduct.poll.</td>
</tr>
<tr>
<td>Abductor dig.V.</td>
<td>Palmaris brev.</td>
</tr>
<tr>
<td>Lumbricalis</td>
<td>Interossei</td>
</tr>
</tbody>
</table>
FIGURE 2.43 SEGMENTAL DISTRIBUTION OF MUSCLES OF THE UPPER EXTREMITY AND THEIR ASSOCIATED LOWER MOTOR NEURONE (SPINAL NERVES SUPPLIED TO MUSCLES FROM SEGMENTS OF THE SPINAL CORD).

<table>
<thead>
<tr>
<th>Cervical Segments</th>
<th>Dorsal Segments (Thoracic Segments)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>6 6.7 8 1</td>
</tr>
<tr>
<td>Supraspinat</td>
<td></td>
</tr>
<tr>
<td>Teres min.</td>
<td></td>
</tr>
<tr>
<td>Deltoideus</td>
<td></td>
</tr>
<tr>
<td>Infraspinatus</td>
<td></td>
</tr>
<tr>
<td>Subscapularis</td>
<td></td>
</tr>
<tr>
<td>Teres major</td>
<td></td>
</tr>
<tr>
<td>Biceps</td>
<td></td>
</tr>
<tr>
<td>Brachialis</td>
<td></td>
</tr>
<tr>
<td>Coracobrachialis</td>
<td>Triceps brach Anconaeus</td>
</tr>
<tr>
<td>Extensor carpi radial</td>
<td></td>
</tr>
<tr>
<td>Pronator teres</td>
<td></td>
</tr>
<tr>
<td>Flexor carpi radial</td>
<td></td>
</tr>
<tr>
<td>Flexor polli long</td>
<td></td>
</tr>
<tr>
<td>Abductor polli long</td>
<td></td>
</tr>
<tr>
<td>Extensor polli brevis</td>
<td></td>
</tr>
<tr>
<td>Extensor polli long</td>
<td></td>
</tr>
<tr>
<td>Extensor digit comm</td>
<td></td>
</tr>
<tr>
<td>Extensor indicis prop</td>
<td></td>
</tr>
<tr>
<td>Extensor carpi uln</td>
<td></td>
</tr>
<tr>
<td>Extensor dig. V. prop</td>
<td></td>
</tr>
</tbody>
</table>
FIGURE 2.44 (CONT.)

Lumbar Segments  Sacral Segments  Coc

1  2  3  4  5  1  2  3  4  5

Long Deep Muscles of the Back

Levator and sph. ani. Rectal Muscles
M. cocoyg

Quadratus Lumb
FIGURE 2.44 (CONT.)

Dorsal (Thoracic) Segments

| 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 | 1 |

Long Deep Muscles of the Back

Serratus post
sup.

Serratus post
inf.

Rectus abdominis

Oblique. int. abdom.

Transversus abdom.

Oblique. int. abdom.

Quadratus Lumb

Intercostal muscles
FIGURE 2.44 SEGMENTAL DISTRIBUTION OF TRUNK MUSCLES AND THEIR ASSOCIATED LOWER MOTOR NEURONE (SPINAL NERVES SUPPLIED TO MUSCLES OF "BACK" FROM SEGMENT OF THE SPINAL CORD).

<table>
<thead>
<tr>
<th>Cervical segments</th>
<th>D</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2</td>
</tr>
</tbody>
</table>

Long Deep Muscles of the Back

Short
depth
Cervical
Muscles

Trapeziums
Latissum
Lev. ang. scap.
Rhomb.
Longus Capitis
Longus colli
Scaleri
Pectoral maj.
Subol. Pect. mins
Serrat ant.

Diaphragm
of both motor fibres, to the various muscles of the body, and sensory fibres, from the various receptors in the skin, subcutaneous tissue, muscles and joints.

The nerve fibres contained in the nerve trunk and leaving the spinal cord are distributed to localised areas of the body according to a fixed pattern and this connection of the brain to the motor and sensory nerve endings is rather complicated in view of the large number of endings involved.

**DISTRIBUTION OF NERVES OF THE UPPER EXTREMITY OF THE BODY**

The simplest method of classifying the distribution of nerves of the upper extremity of the body is to consider; (a) segmental distribution of muscle nerves, and (b) segmental distribution of skin sensory nerves.

Figure 2.43 and 2.44 Appendix 6 for example, relate the Cerebro-spinal nerve (C5 and C6 Cervical) segments with muscle location and type as detailed by Brain (60). Consider say the Biceps of the arm. Thus at a glance it can be seen where the motor nerve fibres of that particular muscle of the upper extremity limbs originate in the spinal column.

Figure 2.45 Appendix 6 further classifies the segmental nerves associated with particular muscles and relates these to the action obtained by the motor and sensory nerve fibres. It is therefore possible, because of this subdivision of the nervous system, to predict and conceive that such a system for individual control of synthetic muscles could be devised using these same neuro-brain motor signal as the human system does. This of course is assuming that medically, such devices would be similar in design to the "synapse" of the nerves and could be connected to the
CUTANEOUS (SKIN) SENSORY SEGMENTATION
POSTERIOR VIEW
Diagram to show cutaneous areas of
distribution of spinal segments and the
peripheral nerves - posterior aspect.
Trigeminal (V. Cereb.)

Cervical Plexus
Superficial Branches

Intercostal Nerves D2, 12

Brachial Plexus

**FIGURE 2.47**

CUTANEOUS (SKIN) SENSORY SEGMENTATION
ANTERIOR VIEW

Diagram to show cutaneous areas of distribution of spinal segments and the peripheral nerves - anterior aspect.
"line" nerve fibres.

As an indication of the degree of sensory control used by "real life" limbs the following distribution of the cerebro-spinal nerves relating to sensory nerve fibres on and throughout the upper extremity of the body was prepared. This is shown in figure 2.46 and 2.47 for both Posterior and Anterior views of the body.

It is to be noted that a single branch from the spinal cord serves a large number of nerve endings on and through the region of the body indicated by the diagrams. Within these regions are contained many thousands of sensory receptors, all of which, when and if stimulated give a degree of refinement to the motion of the body, initiated by the motor signal from the brain. It is of particular importance here to discuss the basic action of the nerve ending mechanisms.

**NERVE ENDINGS**

Of interest here are muscle and skin ending nerves which constitute the largest percentage of sensory and motor nerves of the body, particularly in respect to movements of the limbs and trunk. However it should be noted that Pacinian Corpuscles nerve endings similar to those in the skin are found in deep connective tissue and around joints. They are stimulated by pressure of surrounding structures and thus protect the joint from damage as well as indicating to the brain the magnitude of limb forces in its members.

The muscle nerve endings are called Proprioceptors and are the sense organs stimulated by movement of the body itself. They are important as ingoing afferent pathways in reflexes for adjusting "Posture and Tone".

General Proprioceptors are found in skeletal
FIGURE 2.48

PROPRIORECEPTORS (NERVE ENDINGS) IN SKELETAL MUSCLES, TENDONS AND JOINTS

- Gorli Organ
- Tendon nerve endings
- Muscle body
- Muscle spindle
- Muscle fibrils
- Sensory nerve endings in muscle spindle
- Fine motor nerves from cerebellum of brain
Figure 2.49 Diagrammatic Section Through the Skin

A - Epidermis Layer.
B - Dermis Layer.
C - Subdermal Layer.
D - Muscle that erects Hair.
E - Hair Oil Gland.
F - Nerve Plexus of Follicle.
G - Hair Root.
H - Papilla of Hair.

Branch nerve endings in Epidermis and Dermis register and probably other skin sensations. (PAIN)
muscles, Tendons and Joints and are to two types (a) Tendon sensory nerve endings called Gorli Organ and (b) Muscle motor sensory nerve called muscle spindles. Figure 2.48 shows both types of Proprioceptors in simplified form.

Sensory nerve endings of the tendons of muscles (Gorli Organ) are stimulated by tension which occurs when muscles are stretched and/or when they are contracted. Motor nerve endings of muscles or muscle spindle on the other hand are contained in the skeletal muscle and are stimulated when muscles are stretched or shortened.

Similarly, sensory nerve endings of the body muscles (muscle spindles) contained in the muscle fibrils (intrafusal fibres) are stimulated by the muscle either stretching or shortening. It is evident that a desired rate of response of muscle fibril movement to tendon tension exists and this gives rise to a highly developed positioning and displacement control to the limb in its movements.

In addition to the above, the close proximity of the fine motor nerve endings (which are stimulated either by contracting or stretching of the muscle) to the sensory nerve endings in the muscle spindle gives a positive signal to the brain the instant that movement occurs and hence control of further movement can be minutely controlled. This sense is called a kinesthetic sense.

It is commonly accepted that free nerve endings are responsible for pain reception and that particular type of encapsulated nerve endings are responsible for the others. Figure 2.49 is a diagrammatic section through the skin showing the types of sense organs present. It is generally agreed that the following sense organs respond to the following stimuli. Cold-end bulb or Krause; Warmth — Ruffini's end-organs; Touch — Meissner's corpuscles and
FIGURE 2.410  BRAIN TO BODY RELATIONSHIP
Merkel's discs; Deep pressure - Pacinian corpuscles, and pain by free nerve endings. In addition to these the hairy skin of the body is sensitive to touch related to hair movement at the nerve plexus of the follicle.

The skin nerve endings make up what is called Cutaneous sensation comprising five basic skin sensations of Touch, Pressure, Pain, Warmth and Cold. There is much controversy as to how these are registered. In some areas they appear to be served by special nerve endings (sensory receptors or end-organs) in the skin. These end-organs are not uniformly distributed over the whole body surface; for instance, touch nerve endings are very numerous in hands and feet but are much less frequent in the skin of the back.

The "overstimulation" of any sensory receptor can give rise to the sensation of pain. Also the blending of two or more simultaneous stimulated sensory receptors gives rise to sensations in the brain of tickling, itching, softness, hardness, wetness and dryness.

A detailed description of the known action of these sense organs and their distribution throughout the skin is given in Section 2.5. A typical section of skin is shown in figure 2.51 showing the type of skin sense organs.

The complicated nervous processes associated with the arm and hand are demonstrated by figure 2.410 which shows the brain cerebral motor cortex activity for the arms, trunk and legs of the human body.
ZONES OF THE SKIN are displayed in an idealized section. The underlying dermis, the thickest part of the organ, is supported by a fat-rich subcutaneous stratum. Intermingled with the cells of the dermis are fine blood vessels (color), tactile and other nerves, the smooth muscles that raise the hair when contracted, and a variety of specialized glands. Above the dermis are the twin levels of the epidermis: a lower zone of living cells capped by a horny layer of dead cells filled with the fibrous protein keratin. Melanocytes, the pigment organs that produce the granules responsible for varying skin colors, lie at the base of the epidermis.
The distribution of skin over the body serves not merely as the outside wrapper, covering in the bones, fat and muscles, but also as an organ of the body. On account of the large area which it occupies, the whole structure of the skin is collectively called the cutaneous system. The appearance, surface structure and distribution of the sensory receptors throughout the upper extremities of the body is of most interest here. Thus the skin functions of body temperature regulation, sweat excretion, preparation of vitamin D to be stored in the subcutaneous fat are only mentioned here and not discussed in detail. Reference is made to articles by Villee (35) and Montagna (1965) (61) throughout this section.

In general it is true to say that the entire skin, except the palms of the hands and the soles of the feet, is equipped with countless hair follicles—inpocketing of cells from the inner layer of the epidermis. Refer figure 2.51. These cells undergo division and give rise to the hair cells, just as the inner layer of the epidermis gives rise to the outer layers. The hair cells die while still in the follicle, and the hair visible above the surface of the skin consists of tightly packed masses of their remains. The hair's colour depends on the amount and kind of pigment present, on the number of air bubbles, and on the nature of the surface of the hair which may be smooth or rough. The hairs alter in length and colour depending on their locality, for example, underneath the forearm hairs are small and whitish and tend to be rubbed off by rubbing of the skin surface.

Fingernails and toenails likewise develop from inpocketings of cells from the inner layer of the epidermis, and the growth of nails is similar to that of the hair.
Branch nerve endings in Epidermis and Dermis register and probably other skin sensations. (PAIN)

A - Epidermis Layer.
B - Dermis Layer.
C - Subdermal Layer.

FIGURE 2.52  TYPICAL HAIRLESS SKIN DETAILS
Both hairs and nails are derivatives of the skin. Nails are composed of densely packed dead cells which are translucent, allowing the underlying capillaries to show through and give the nails their normal pink colour.

The colour of the skin depends on three factors; the yellowish tinge of the epidermal cells, their translucent quality which allows the pink of the underlying blood vessels to show through, and the kind and amount of pigment - red, yellow or brown - contained in the inner layer of the epidermal cells.

The subdermal tissues containing fat cells, connective tissue, blood vessels, and nerves, serves primarily as a connection between the skin and the deeper tissues. In some areas of the body, such as the neck, the connecting fibres are loose, so that the skin can be moved quite freely. However in other regions like the palms of the hands, for example, the skin is attached much more firmly, so that very little skin motion is possible.

The dermis like the epidermis varies in thickness and texture in different regions of the body. It is especially thick over the palms of the hands where large concentrations of nerve endings exist and where excessive contact with objects occur. The ridges and irregularities of the skin surface are accounted for by the indentations of the lower skin layer of Stratum Germinativum on the stratum corneum. These indentations are generally irregular in outline (refer figure 2.52) and are known as papilla of the skin. In the hands of the body however, the papillae are so pronounced that they raise the surface into permanent ridges - commonly referred to as skin prints. This occurs not only in the skin in the fingers but in the palms of the hands, and the soles.
of the feet. The ridges are so arranged that they offer maximum resistance to slipping, either in walking or in grasping an object such as in the case of the hand. For this reason they are called "friction ridges".

Below the papilla of the skin in the subdermal layer lie the all important sensory receptors such as the Merkel discs (touch) and the Pacinian corpuscles (pressure), to mention a few. Merkel discs are found in areas of hairless skin in the epidermis of the finger tips, mouth, and lips, where the senses of touch are particularly delicate. Refer figure 2.52 for a typical section of hairless skin. Also Meissner corpuscles, formed in the papillae of hairless skin regions, vary from 40 to 150 micron in length to 20 to 60 micron in width. There may be as many as twenty or thirty of them packed into a square millimeter.

The largest and in some respects the most elaborate of the encapsulated endings are the Pacinian corpuscles. These are 0.5 to 4.5 millimeters long and 1.0 to 2.0 millimeters wide. They occur deep in the corium and subcutaneous tissue and are abundant in the hand and foot regions. They are found also in the tissues of the joints, ligaments of the leg and forearm, in the external genitals, and in the coverings of bones. Pacinian corpuscles have a thick coat composed of ten to fifty successive layers of connective tissue arranged concentrically around a central granular core in which the nerve fibrils terminate. These are clearly related to Gorli-Mazzoni corpuscles of muscle endings.

It is pointed out here that very little work has been done to find the "matrix" of distribution of these nerve endings and only limited details are available.
However it is evident that skin sensitivity varies from one part of the body to another. Pain receptors, for instance, are quite plentiful on the forehead and the upper arm, where there are about two hundred per square centimeter of skin surface, but on the thumb there are only fifty per square centimeter. Touch spots also differ in their distribution, where on the lips they are less than one millimeter apart while in the middle of the back, this distance is increased to over ten millimeters.
FIGURE 2.61 MOVEMENTS AT THE SHOULDER

ABDUCTION, PLANE OF SCAPULA

EXTENSION

ABDUCTION, FRONTAL PLANE

FLEXION
2.6 MOVEMENTS OF ARM WITH ASSOCIATED MUSCLE ACTIVITY

In general the range of movement of the upper extremity limb on the trunk of the body is governed by the action of four joints, the shoulder, sternoclavicular, elbow and wrist joints. Moreover, the motion of limbs is a direct result of the action of muscles which are attached (at points called the origin and insertion) to the body skeleton. The mechanical energy in the form of skeleton motion is a transformation of muscular energy which results directly from chemical processes within the individual muscle cells.

Below is a detailed discussion of the degree of movement possible at each joint of the upper extremity limb and the associated muscle actions that power these movements. Reference is made to Royle (1938) (62) and Klopsteg and Wilson (1954) (33).

MOVEMENTS OF THE SHOULDER

The range of movement of the upper limb on the thorax depends mainly upon two joints, the shoulder joint and the sternoclavicular joint (Refer Section 2.11 for details). It is necessary to distinguish the part which each of these joints plays.

Anatomical movements which are referenced to the plane of the scapula (thirty degrees lateral from the standard position) are summarised in figure 2.61 and are described below.

Abduction is the movement of the upper limb (arm) away from the body in a anterolateral direction in the plane of the scapula. It has a range of about one hundred and eighty degrees (refer figure 2.61) and is limited by the bony ligamentous conformation of the shoulder joint.
Figure 2.61a  Shoulder abduction showing scapula rotation

Drawings from tracings of radiograms of a right shoulder taken during abduction in a frontal plane. A, arm is at the side. B, arm has been abducted to a right angle. C, arm is almost fully elevated. Note the degree to which the scapula rotates. Note also the lateral rotation of the humerus as evidenced by the change in position of the tubercles and the intertubercular groove. For every three degrees of arm elevation, two come from the glenohumeral joint (arm on shoulder) and one by scapular rotation (shoulder on body).
FIGURE 2.62  
FLEXION OF THE SHOULDER
FIGURE 2.63
EXTENSION OF THE SHOULDER
FIGURE 2.64
ADDITION PLUS FLEXION OF THE SHOULDER
Additionally for every three degrees of arm elevation, two come from the glenohumeral or shoulder joint (arm on shoulder) and one degree comes from the scapula rotation or shoulder on body. (Refer figure 2.61a).

Flexion is the movement of the upper limb which carries it anteromedially (forwards and upwards) in a plane perpendicular (Antero-posterior) to the plane of the scapula; it has a similar range of about ninety degrees (refer figure 2.61 and 2.62).

Extension is the movement of the upper limb which is the reverse of flexion and carries it posterolaterally (backwards) in the Antero-posterior plane; it has a range of about forty five degrees (refer figure 2.61 and 2.63).

Adduction is the reverse of abduction and carries the arm towards the body in a posteromedial direction. Adduction as a standard position movement is limited by the body, but in the anatomical movement the arm adducts past the body. When combined with flexion this movement has additional range (refer figure 2.64). It may also be combined with extension to a lesser extent.

Rotation is also possible at the shoulder joint and its range is also about ninety degrees.

**ABDUCTION**

This is the function of the lateral part of the Deltoid and the Supraspinatus. In two recorded instances only the Supraspinatus was able to abduct the arm and these were both in very muscular patients in whom the Deltoid was paralysed. Usually, however, when the patient is unable to abduct or maintain the arm abduction, the medial rotation of deltoid muscle must be considered paralysed or defective in function according to the degree of disability. Some authorities
credit the supraspinatus with drawing the head of the humerus into the glenoid cavity as a preliminary movement to abduction by the deltoid.

**FLEXION**

This is the movement due to the anterior portion of the deltoid. The coracobrachialis also assists in flexion on the shoulder in the first part of the movement, but a complete flexion is not possible in paralysis of the clavicular portion of the deltoid. (Refer figure 2.62).

**EXTENSION**

This movement is of small range and is due to the contraction of the posterior part of the deltoid. The scapular head of the triceps and the teres major can produce a small degree of extension when the deltoid is paralysed (refer figure 2.63).

**ADDUCTION**

If the arm is carried against resistance from the abduction position into adduction, the movement is performed by the pectoralis major and the latissimus dorsi. If adduction is combined with flexion (refer figure 2.64) the clavicular position of the deltoid and the pectoralis major are the principal movers and the latissimus dorsi relaxes. If the arm is carried into adduction and extension, the latissimus dorsi and the posterior part of the deltoid are the muscles concerned. It is well to understand that the pectoral muscle is not a flexor of the shoulder but the clavicular portion of this muscle does flex the arm. It produces both adduction and medial rotation. It is possible for the different parts of the pectoral muscle to act independently. For example, adduction plus flexion with the upper limb horizontal at the shoulder calls into action the clavicular portion of the pectoralis major as well as
the clavicular portion of the deltoid. The sternal part of the pectoralis major, on the other hand, comes into action as a principal mover when the degree of flexion is less and it extends the fully flexed arm powerfully. It is possible for a single muscle to have several functions because of an extended attachment. The deltoid and pectoralis major are two examples.

**Rotation**

The range of rotation is independent on the position of the upper limb. It is much the same whether the limb is adducted at the side of the body or abducted at the shoulder. The muscles concerned in lateral rotation are the teres minor and the infraspinatus. The subscapularis is the prime mover in medial rotation. The pectoralis major and the latissimus dorsi are sometimes described as medial rotators, but true medial rotation is due to the subscapularis above.

**Movements of the Scapula and Sterno-Clavicular Joints**

The movements of the shoulder on arm joints are much more limited than the overall movements of the upper limb with respect to the body. The greater mobility is due to the movements at the sterno-clavicular joint and the movements of the scapula at the acromio-clavicular joint. The scapula may be moved forwards and laterally and adducted medially and drawn backwards. It may be elevated cranially (towards the brain) and depressed caudially. The clavicle takes part in these movements and moves backwards and forwards during medial and lateral movement of the scapula. The scapula may also be moved round an antero-posterior axis. This is rotation of the scapula.

**Antero-Lateral Movement or Abduction**

The principal mover in this movement is the serratus
anterior. Owing to the structure of this muscle a contraction of the whole muscle leads to a relatively greater exclusion of the inferior angle of the scapula. The serratus anterior is not a respiratory muscle.

**ADDUCTION OF THE SCAPULA**

Movement towards the vertebral column or adduction is brought about by the rhomboid muscle as principle movers. Owing to the disposition of these muscles the movement is upwards as well as medially.

**ELEVATION OF THE SCAPULA**

Movement in a cranial direction is due to the action of the levator scapulae and the part of the trapezius attached to the superior lip of the spine of the scapula and to the clavicle.

**DEPRESSION OF THE SCAPULA**

This is due to the action of the pectoralis minor and the subclavius muscles.

**ROTATION**

Rotation of the scapula is a difficult movement to understand. When the upper limb has been abducted to a right angle further movement is possible only because the scapula is able to rotate. In this action the upper part of the trapezius elevates the shoulder girdle, the part of the trapezius attached to the base of the spine fixes this point, while the serratus anterior draws the inferior angle forwards and outwards. The serratus anterior muscle is not the main elevator of the arm, operating over only one third of the range of movement (rest supplied by deltoid and supraspinatus), but paralysis of the serratus anterior causes inability to lift the arm upwards above the head either in flexion or abduction. The vertebral border of the scapula stands out prominently.
Protraction and retraction of the shoulder (shows abduction)

Extension of head

Upper part of trapezius elevates scapula

Figure 2.6: Shoulder—Action of the Trapezius
FIGURE 2.66  
SCHEMATIC OF MUSCLE ACTING ON THE SHOULDER-ARM SYSTEM

H, humerus; C, clavicle; S, scapula.

a. Top view: Solid arrows represent muscles which flex or extend the arm on the shoulder. Dotted arrows show muscles acting on the scapula to flex or extend the shoulder.

b. Front view: Solid arrows represent muscles which elevate or depress the arm on the shoulder. Dotted arrows show muscles acting on the scapula to elevate or depress the shoulder.
FIGURE 2.67 MUSCLES OF THE SHOULDER-ARM SYSTEM Refer Klopsteg & Wilson (33) Muscles acting on the arm: a, posterior view; b, anterior view (thorax removed); c, anterior view. Muscles acting on the scapula: d, posterior view; e, anterior view (thorax removed).
from the chest wall if such a movement is attempted and this condition is called a "winged scapula". If the trapezius be paralysed it is possible to lift the arm in a flexed state above the head, but the range of movement is diminished in the respect that the shoulder girdle remains depressed and the reach above the head is sub-normal (refer figure 2.65).

THE SHOULDER-ARM MUSCULATURE

The muscular anatomy of the shoulder-arm system is highly complex, and a considerable degree of simplification is necessary to gain an elementary understanding of the basic actions of the muscles as well as of their interactions in the various phases of motion. For clarification, the schematic views of figure 2.66 show the skeletal members and principal muscle groups. The muscles can be divided into three groups: those joining scapula to thorax, those joining arm to scapula, and those running from arm to thorax. The neuromuscular organization is capable, within the limits set by the musculoskeletal mechanism of performing shoulder motions alone, arm motions alone, or coordinated motions of both. The latter is the normal mode. For example, Inman et al. (1944) (63) have shown that, for every three degrees of arm elevation, two are contributed by the glenohumeral joint (arm-on-shoulder), and one is contributed by scapular rotation (shoulder-on-body).

The principal muscles are illustrated in figure 2.67, where they are divided into those acting upon the scapula, chiefly controlling shoulder motion, and those acting on the arm, either originating from the thorax or from the scapula. Table 2.68 lists the muscles concerned in the various actions of shoulder and arm.

The shoulder-arm musculature is most realistically
### Table 2.68
**Muscles Acting on Shoulder and Arm**

<table>
<thead>
<tr>
<th>Motion</th>
<th>Thorax-to-shoulder</th>
<th>Shoulder-to-arm</th>
<th>Thorax-to-arm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elevation</td>
<td>Trapezius I</td>
<td>Deltoid I, II, III</td>
<td>Supraspinatus</td>
</tr>
<tr>
<td></td>
<td>Levator scapula</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Depression</td>
<td>Trapezius III</td>
<td>Teres major</td>
<td>Pectoralis major</td>
</tr>
<tr>
<td></td>
<td>Rhomboid</td>
<td></td>
<td>Latissimus</td>
</tr>
<tr>
<td></td>
<td>Pectoralis minor</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>Serratus</td>
<td>Deltoid I</td>
<td>Pectoralis major</td>
</tr>
<tr>
<td>Extension</td>
<td>Trapezius II</td>
<td>Infraspinatus</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Rhomboid</td>
<td>Teres minor</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Deltoid III</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Teres major</td>
<td></td>
</tr>
</tbody>
</table>

* Motions follow the terminology illustrated in Fig 2.4

### Table 2.69
**Muscles Causing Parasagittal Motions and Rotation of Arm** *(Refer: Klopsteg & Wilson (33))*

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parasagittal flexion</td>
<td>Deltoid I, II</td>
</tr>
<tr>
<td></td>
<td>Biceps</td>
</tr>
<tr>
<td>Parasagittal extension</td>
<td>Latissimus</td>
</tr>
<tr>
<td></td>
<td>Deltoid III</td>
</tr>
<tr>
<td></td>
<td>Supraspinatus</td>
</tr>
<tr>
<td>Medial rotation</td>
<td>Pectoralis major</td>
</tr>
<tr>
<td></td>
<td>Teres major</td>
</tr>
<tr>
<td></td>
<td>Subscapularis</td>
</tr>
<tr>
<td></td>
<td>Latissimus</td>
</tr>
<tr>
<td>Lateral rotation</td>
<td>Teres minor</td>
</tr>
<tr>
<td></td>
<td>Infraspinatus</td>
</tr>
</tbody>
</table>
treated in terms of functional groups. Further, the composition of these groups will vary with the motion concerned. While the motions listed in Table 2.68 cover the principal muscular actions of the shoulder-arm system, an additional set of motions of the arm deserves emphasis because of its importance in amputee biomechanics. Included are arm rotation and arm flexion-extension in the parasagittal plane. It will be recalled that the latter motions (figure 2.5) require special mention because of the importance of movements begun from the standard vertical arm position. Accordingly, Table 2.69 lists the muscles concerned in these motions.

**MOVEMENTS OF THE ELBOW JOINT**

**FLEXION**

The muscles concerned in this movement are the brachialis, the biceps and the brachioradialis. Of these, the brachialis is the prime mover. It is possible to flex the elbow without the biceps, but if flexion be performed against resistance the biceps and the brachioradialis assist. If acting alone the brachioradialis can flex the elbow to about a right angle, but if joined to the tendon of a paralysed biceps it may bring about full, though weak flexion.

**EXTENSION OF THE ELBOW JOINT**

This is the function of the triceps and anconeus muscles.

**MOVEMENT OF THE FOREARM**

These movements are pronation and supination and have a range of about one hundred and thirty degrees. The muscles producing pronation are the pronator teres and the pronator quadratus. Of these the more powerful is the pronator teres. Supination is brought about by the supinato
FIGURE 2.610  PRONATION AND SUPINATION RANGE OF MOVEMENT
**PRINCIPAL MUSCLES ACTING ON THE FOREARM**

**a.** Front view, forearm flexors; B, biceps; Br, brachialis; BR, brachioradialis.

**b.** Front view, forearm rotators; BR, brachioradialis; PT, pronator teres; PQ, pronator quadratus.

**c.** Back view, forearm extensors; T, triceps; A, anconeus.

Refer Klöpfel & Wilson (33)

**TABLE 2.612**

**MUSCLES ACTING ON THE FOREARM**

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forearm flexion</td>
<td>Biceps, brachialis, brachioradialis</td>
</tr>
<tr>
<td>Forearm extension</td>
<td>Triceps, anconeus</td>
</tr>
<tr>
<td>Pronation</td>
<td>Pronator teres, pronator quadratus</td>
</tr>
<tr>
<td>Supination</td>
<td>Biceps, supinator brevis, brachioradialis</td>
</tr>
</tbody>
</table>
muscle and the biceps. (Refer figure 2.610).

**FOREARM MUSCULATURE**

The principal muscles acting upon the forearm are illustrated in figure 2.611 and listed in Table 2.612. Flexion is caused by the powerful biceps–brachialis group having origins in the shoulder and humerus and inserting in radius and ulna. Thus, they make up the power units of a third-class lever system, the mechanical characteristics of which will be described presently. Flexion force is also contributed by the brachioradialis, particularly in the more flexed positions of the forearm. The extensor system is chiefly composed of the triceps, with a contribution by the anconeus. The olecranon of the ulna, serving as the insertion of the triceps, makes up a first-class lever system with the triceps as the force element.

The forearm rotators are divided into two groups, pronators and supinators. Pronation, caused principally by two specialized muscles, the pronators teres and quadratus, is also aided by the digital flexors, which originate in the medial epicondyle and which, running obliquely across the volar forearm, contribute force to this motion. Supination also receives strong support from muscles for which supination is a secondary action. The biceps, owing to its insertion on the radius, is a strong supinator when the forearm is in the intermediate range of flexion. Likewise, the digital and wrist extensors, because of their dorsal oblique course from the lateral epicondyle to the radial aspect of the wrist, exert additional supinatory action. (Refer Steindler, Arthur (1935) (32).

**MOVEMENT OF THE WRIST AND CARPUS**

**FLEXION**

Flexion or volar flexion has a range of about ninety
degrees and is due to the action of the flexor carpi ulnaris and flexor carpi radialis. The palmaris longus, when present, may also assist but its function as a prime mover is to tighten the palmar fascia during movements of the thenar and hypothenar muscles.

**Extension**

Extension or dorsiflexion is much more limited in range than volar flexion and varies between twenty five and forty five degrees. It is important to note the range of this movement as it has a great bearing on the function of the hand, for a strong grip is possible only when the wrist is dorsiflexed. The prime or principal movers in dorsiflexion are the extensor carpi ulnaris, the extensor carpi radialis longus and the extensor carpi radialis brevis.

**Lateral and Medial Movement**

Radial or lateral flexion is brought about by the two radial extensors of the carpus and by the abductor pollicis longus. The flexor carpi radialis does not take part in this action unless the wrist is also volar flexed. Ulnar deviation is due to the action of the flexor carpi ulnaris and the extensor carpi ulnaris. Circumduction may also be performed at the wrist. It is really a combination of the various movements.

**Movements of the Fingers**

The movements of the metacarpo-phalangeal joints are four, namely, flexion and extension, adduction and abduction. Flexion of these joints is brought about by the contraction of the lumbrical muscle. Extension is due to the action of the extensor communis digitorum. Abduction and adduction are brought about by the interosseous muscles. The dorsal interosseous muscles
produce abduction of the three lateral fingers from a line drawn longitudinally through the third finger, while the palmar interosseous muscles adduct the thumb, the second, fourth and fifth fingers towards the third finger. Extension of the interphalangeal joints is also the function of the interosseous muscles. Flexion of the interphalangeal joints is brought about by the action of the flexor profundus digitorum at the distal joints and by the flexor sublimus digitorum at the proximal joint.

When both ulnar and medial nerves are divided it is still possible to extend the interphalangeal joints, but the movement is abnormal in the respect that there is excessive dorsiflexion of the first phalanx. When the ulnar nerve is divided, extension of the index and middle finger can be carried out, but there is again excessive dorsiflexion of the third and fourth fingers at the metacarpo-phalangeal joints. Abduction and adduction of the fingers as described above are also lost. If the radial nerve be divided there is inability to dorsiflex and metacarpo-phalangeal joints, but the interphalangeal joints may still be extended. In the fingers the common extensor and interosseous muscles act simultaneously in extension of the interphalangeal joints.

**MOVEMENTS OF THE LITTLE FINGER**

Abduction of the little finger is performed by the abductor minimi digitii. The small flexor produces flexion at the metacarpo-phalangeal joint. The opponens minimi digiti does not bring about true opposition of the little finger to the thumb but its contraction has the effect of drawing into prominence the hypo-thenar eminence and thereby deepening the hollow of the hand.
### TABLE 2.613  PRINCIPAL MUSCLE GROUPS ACTING ON THE WRIST

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volar flexion</td>
<td>flexors carpi ulnaris and radialis; palmaris longus; the long thumb flexors</td>
</tr>
<tr>
<td>Dorsal flexion</td>
<td>extensors carpi radialis longus and brevis and carpi ulnaris; the digital extensors; the extensor pollicis longus</td>
</tr>
<tr>
<td>Ulnar flexion</td>
<td>flexor and extensor carpi ulnaris, extensor digiti minimi</td>
</tr>
<tr>
<td>Radial flexion</td>
<td>extensors pollicis longus and brevis; abductor pollicis longus</td>
</tr>
</tbody>
</table>

### TABLE 2.616  WRIST AND DIGITAL MUSCULATURE

<table>
<thead>
<tr>
<th>Motion</th>
<th>Muscles</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dorsal flexion</td>
<td>extensors carpi radialis longus and brevis and carpi ulnaris</td>
</tr>
<tr>
<td>Volar flexion</td>
<td>flexors carpi ulnaris and radialis; palmaris longus</td>
</tr>
<tr>
<td>Digital extension</td>
<td>extensor digitorum communis; lumbricalis acting on the distal phalanges</td>
</tr>
<tr>
<td>Digital flexion</td>
<td>flexors digitorum sublimus and profundus; lumbricalis acting upon the metacarpophalangeal joint</td>
</tr>
</tbody>
</table>
MOVEMENTS OF THE THUMB

Extension of the terminal phalanx is due to the action of the extensor longus pollicis. When this muscle is unopposed in complete lesions of the median nerve, it rotates the whole thumb so that the nail faces backwards. This produces the "flat hand" deformity. The extensor pollicis brevis extends the first phalanx of the thumb. Flexion and extension at the metacarpo-phalangeal joint of the thumb does not occur in the same plane as the movement in other fingers. Flexion of this joint draws the thumb so that it points forwards and medially. The abductor pollicis longus by its action on the metacarpal bone really becomes a principal mover in abduction of the hand. Flexion of the terminal phalanx of the thumb is produced by the action of the flexor pollicis longus.

THENAR GROUP OF INTRINSIC MUSCLES

The opponens pollicis rotates the metacarpal bone medially to cause the palmar surface of the thumb to face the palmar surface of the other fingers. The short abductor lifts the thumb anteriorly from the palm while the adduction is performed by the adductor pollicis. Flexion of the first phalanx is the action of the flexor pollicis brevis.

THE MUSCULATURE OF WRIST FLEXION AND EXTENSION is important in relation to prehension, where the volar and dorsal flexor groups must act to fix the wrist and hand for stable grasp. It is important to note that the lines of action of the various muscles contributing to wrist motions are seldom so placed that they can directly provide moments in the major flexion planes. Rather, obliquely placed muscles functionally combine their actions to give the desired flexions. Steindler (32) defines the functional groups given in Table 2.613.
FIGURE 2.614  
HAND PREHENSION PATTERNS  REFER KLOPSSTEG & WILSON (33)  
a, palmar prehension. Opposition of thumb with digits 11 and 111 in the near-closed position gives a three-jawed chuck prehension.  
b, palmar prehension. At openings of 1 to 3 in., opposition between thumb and digits 11 and 111 is in the manner of a pliers grip.  
c, tip prehension. Distal phalangeal segments of digits 11 and 111 and of thumb are strongly flexed to bring the finger and thumb tips into opposition.  
d, lateral prehension. Ball of thumb opposes lateral surface of digit 11, usually the second phalanx.  
e, hook prehension. Load supported by hooked terminal phalanges, thumb acting largely to prevent slipping volarly.  
f, spherical prehension. In addition to elements of group prehension, curving across the knuckle line permits conformity to spherical objects.  
g, grasp prehension. Digits curled about object, such as a handle; thumb curves and overlays finger tips to close the prehension "ring."
Schematic sections through digit 111 and the thumb show essential relations of muscles and bones in prehension. LG indicates the presence of ligamentous guides which channel the tendons of muscles originating in the forearm close to the wrist. Guide lines xx indicates relative position of carpal bases of thumb and digits.

Bones: R, radius; L, lunate; C, capitate; M, metacarpal; GM, greater multangular; FP, first phalanx; SP, second phalanx; TP, terminal phalanx.

Muscles: EDC, extensor digitorum communis; ECRL, ECRB, extensor carpi radialis, longus, brevis; FCU, FCR, flexor carpi ulnaris, radialis; FDP, FDS, flexor digitorum profundus, sublimis; PL, palmaris longus; I, interosseus; OP, opponens pollicis; L, lumbricalis; FPL, APL, flexor, abductor pollicis longus; APT, APÔ, adductor pollicis transversus, obliquus; EPB, EPL, extensor pollicis brevis, longus.
The prehension patterns are most easily understood by reference to the shape, size, and form of the objects which the hand is adapted to grasp. Combining the observations of Schlesinger (1919) (64) and of Keller (1947) (65), six major prehension patterns may be defined as seen in figure 2.614. For each, a mechanical analogue is also illustrated. The essential features of prehension can be seen in figure 2.615, which presents a schematic section through the sagittal plane of hand and wrist. There are six articulations in series, but only four are of major importance. The first of these is compounded of the radiocarpal and intracarpal joints, upon which the volar and dorsal flexions of the wrist take place. Motion between the carpal and metacarpal bones is relatively slight. The metacarpo-phalangeal joint and the interphalangeal joints are those concerned in finger flexions. Their angular excursions range typically from the straight (extended) position to about ninety degrees of flexion in full-grasp prehension. The musculature indicated in figure 2.615 can be considered in terms of the actions and muscle groups given in Table 2.616.

Considering these mechanisms separately from thumb opposition, two major and independent actions are possible. That is, with digital groups maintaining finger posture, the wrist may be dorsal or volar-flexed. Conversely, with the wrist groups stabilized, various patterns of digital position may be assumed. Besides, there are many interactions between these groups. For example, digital flexion requires stabilizing contraction of the dorsal wrist flexors in order to prevent volar wrist flexion from occurring simultaneously with digital flexion. This action may be
felt in the contracted musculature on the dorsal aspect of the forearm when strong grasp prehension is carried out. Interaction between the lumbricals and the digital flexors, with perhaps some partial digital-extensor activity, permits the varieties of digital posture seen in palmar, tip, and grasp prehension. (Refer figure 2.614). Another combination having an important mechanical effect is the use of the wrist dorsal flexors to pre-stretch the digital flexor muscles in order to augment their forces of grasp.

The essential opposing member in all prehensions is the thumb, of which the bone and joint pattern in prehension is shown in figure 2.615. The versatility of this digit lies in the fact that, although opposition and extension motions take place on a plane not unlike that of the other digits, it can select, over a considerable range, the plane in which its opposition-extension movements take place. In effect, the thumb is a digit which is capable of "wheeling" upon its saddle-shaped carpometacarpal articulation so that it can oppose any other digit, ranging from the lateral surface of digit II to the palmar surface of digit V. The muscular actions which effect this positioning can be carried out either in unison with or independently of the muscles which produce opposition and extension of its metacarpal and phalangeal segments.

The musculature is complicated, and it will suffice here to name only the few groups principally concerned in thumb prehension. The adductor and opponens groups, acting through their insertions on the thumb metacarpal, pull the thumb toward the hand. Of these, the adductors act chiefly when the thumb is in the palmar plane; at the opposite extreme, the opponens group acts when the
thumb is involved in spherical grasp. At all positions over this range, the flexor pollicis longus strengthens the opposition by acting on the distal segments of the thumb. Conversely, extension or abduction of the thumb is caused by the extensors pollicis longus and brevis and the abductor pollicis. Thus the thumb, because of the large number of muscle groups acting upon it, is capable of action patterns of great complexity and diversity.

Considering all the neuromuscular mechanisms available to give variety to natural prehension, it is little wonder that difficulty is found in classifying prehensions into the few categories shown in figure 2.614. The intrinsic hand muscles, singly or in groups, constitute independent control units which can array the skeletal segments of wrist and hand into manifold patterns of form and force.
DIAGRAM OF MOTOR UNIT

- Spinal Cord
- Spinal Nerve
- Anterior Horn Cell
- Nerve Fibre
- Muscle Fibres
2.61 **MUSCLE ELECTRIC ACTIVITY**

Muscles contract as a result of a pattern of electrical impulse brain commands sent to the muscle fibres. The "firing" of a number of muscle fibres at different frequencies throughout the muscle is responsible for the overall contraction. (Refer Section 2.3 for details). The brain command impulses are also responsible for the overall muscle (myo) electric activity being produced at the skin surface of the limb. This myoelectric activity is convenient to measure and record, and is directly proportional to the muscle effort involved.

Muscle cells are controlled by motor nerve cells in the spinal cord referred to an Anterior Horn cells as illustrated in Figure 2.617.

Each anterior horn cell sends a nerve fibre to the muscle where it divides into branches, and may control a large number of muscle fibres. The nerve cell, and the muscle fibres it controls, are called a "motor unit", and a muscle contains many such motor units.

The "command" is sent down the motor nerve as a small electrical impulse; as it reaches each muscle fibre, a similar brief impulse is triggered off in the muscle fibre. This travels along the muscle fibre, which a few milliseconds later gives a single mechanical twitch. A sustained contraction is obtained by the twitches of a large number of motor units "firing" at different frequencies throughout the muscle.

Both nerve and muscle fibres can be thought of as tubes, which maintain a "D.C." potential difference of a few hundred millivolts across their walls in the resting state. When an impulse travels along a muscle
FIGURE 2.618  COMMAND - CONTROL PATH FOR ARM
"Action Potential" for muscle fibre

FIGURE 2.619 ELECTRICAL SIGNAL REACHING A SINGLE MUSCLE FIBRE

E.M.G. Signal for several muscle fibres

FIGURE 2.6110 ELECTROMYOGRAPHIC (E.M.G.) SIGNAL
fibre, this potential difference collapses and the "depolarisation front" travels along the muscle at a speed of about 12 ft/sec. After the front has passed, the resting potential is restored chemically within a few milliseconds.

While it is not practicable to detect motor nerve impulses directly, it is possible to pick up the electrical activity from the skin surface as shown in figure 2.618. This myoelectric activity results from the signals sent along the residual nerve fibres.

It is a measure of the effort made by the subject to contract that particular muscle.

A pair of electrodes attached to an individual muscle cell detects the "action potential" as shown in figure 2.619.

Muscle potentials, termed electromyo-graphic signals (commonly EMG signals) can be detected at the surface of the body using small "flat" electrodes pressed against the skin, covering the desired muscle.

The complex waveform is a summation of the "action potentials" for a large number of muscle fibres.

A typical EMG signal, recorded during a sequence of strong and light muscle effort and relaxation, is shown in figure 2.6110, exhibiting the properties of a A.C. voltage, having frequency components ranging from a few cycles per second, to well into the audible range.

The EMG signal may be regarded as the stimulation reaching the muscle, and actually precedes the mechanical response.

The total electrical activity at a given time is randomly distributed along the muscle mass and only a sample is detected by the surface electrodes. For this reason,
EXPERIMENTAL ARRANGEMENT TO INVESTIGATE E.M.G. SIGNAL CHARACTERISTICS

Arm is supported at shoulder so that flexion and extension could be executed in a horizontal plane about a vertical axis through the elbow.
the detected signal is only roughly proportional to the contraction effort. Similarly, the EMG signal fluctuates (even at constant muscular tension) around a mean value as seen in figure 2.6110.

In general, muscles produce tension which increases with the increased stimulation (EMG) they receive, although the relationship is altered to a small extent by the length at which the muscle is operating, and whether it is extending or contracting. In a normal muscle, when a force is being exerted, very little EMG is required to produce a movement, and the velocity of movement is probably controlled by local feedback of nerve signals between the muscle and the spinal cord.

In conclusion, muscles affected by polio or other paralyzing disease produce EMG signals, even when the muscle is so weak that it cannot operate against gravity. Thus only mental concentration is required to produce EMG signals at the muscles.

In an effort to investigate EMG signal characteristics the following experiment was conducted on elbow flexor and extensor muscles.

The normal subject was seated comfortably with one arm supported, with the upper arm fixed in line with the shoulder, so that flexion and extension could be executed in a horizontal plane about a vertical axis through the elbow. (Refer figure 2.6111). In some experiments, known loads could be applied at right angles to the wrist, and in others a strain gauge was fixed to the wrist support so that the tension developed during a graded voluntary effort could be measured. A potentiometer on a spindle provided an electrical signal
FIGURE 2.6112
VARIATION OF E.M.G. SIGNALS DURING FLEXION
Relationship between the E.M.G. and the angle of flexion for a constant load of 4 kg. applied at right angles to the wrist.
proportional to the angle of flexion.

The forearm support carried a lamp, shining a spot of light on a semi-circular scale marked in degrees of flexion set up a few feet away, with the centre of the scale, coinciding with the axis of rotation of the elbow.

A second lamp was arranged to throw a larger guide spot on the scale, moved by an electric motor at a pre-determined angular speed. The subject was required to track the guide spot using the light spot attached to his forearm.

The electrodes were silver discs, 1 cm. in diameter, attached by adhesive tape to the skin, 3 cm. apart. The subjects' skin was prepared by gently rubbing with fine sandpaper and electrode jelly, until the resistance was less than 5000 ohms.

The two channels from the electrodes were amplified and passed through a band-pass filter and displayed directly on a monitor Cathode Ray Oscilloscope, and also were fed via a rectifying and averaging circuit (100 msec time-constant) to a photographic galvanometer recorder. A Servo system was used to drive the recording paper, using as the input signal either the potentiometer, with the abscissae representing the angle of flexion, or the strain gauge, where the abscissae represented the force developed.

Figure 2.6112 is a recording which shows the relationship between the EMG and the angle of flexion for a constant load of 4 Kg. or a moment of 100 Kg.cm. applied at right angles to the wrist. When the upper arm muscles are relaxed, the load pulled the arm into a hyper-extended
FIGURE 2.6113

E.M.G. SIGNAL/ANGLE OF FLEXION BICEPS & TRICEPS
FOR ZERO LOAD AT WRIST
position, so that the graph commences on the left of the arm - straight position. As the biceps muscle is contracted, it assists the elastic forces in the tissues, and the forearm begins its movement. The biceps EMG then increases and so presumably does the tension in the muscle. As the contribution of the elastic forces decreases, a peak is reached at about ten to twenty degrees of flexion. The EMG then begins to decrease until at eighty-ninety degrees it reaches a minimum. It then increases again and eventually, during the last few degrees of flexion, the tissues of the forearm and upper arm are being compressed. Apart from the effects of these elastic forces another factor which contributes to the variation in EMG is the change in the effective leverage of the muscle; the tendon lies a little closer to the axis of the elbow joint when the arm is fully extended than when it is held at ninety degrees.

The relationship of EMG for triceps and biceps for zero load at the wrist is shown in figure 2.6113. At small angles the triceps is active to straighten the arm against elastic forces, and at large angles the biceps is required to flex against elastic forces. In the intermediate range both muscles are almost completely relaxed and the signal approaches the level of the electrical "tissue noise" approximately 1.5 μV (rms).

Figure 2.6113 also shows the apparent simultaneous activity in both flexor and extensor groups. However it is very unlikely that both muscle groups are really active at the same time and it is interpreted as the effect of "Cross-talk" from one muscle to the electrodes placed over the other.

The magnitude of the cross-talk varies with
FIGURE 2.6114  BICEPS E.M.G./ANGLE OF FLEXION

FIGURE 2.6115  TRICEPS E.M.G./ANGLE OF FLEXION
FIGURE 2.6116  E.M.G./LOAD AT WRIST FOR GRADUALLY INCREASING EFFORT
Angle of flexion 135 degrees.
the siting of the electrodes and from one subject to another, but it is usually less than 1/5 of the genuine signal from the active muscle.

The effect of applying increasing loads at the wrist is shown in figures 2.6114 and 2.6115 for the biceps and triceps respectively.

For the biceps muscle the central portion of the range of movement at angles of flexion of sixty to one hundred degrees, the EMG signal goes through a flat minimum caused by the change in leverage of the muscle tension.

In this case of the triceps in figure 2.6115 the movement starts from the arm fully bent, and the effect of tissue compression can be seen at angles near fifteen degrees. The peaks in the upper three curves and reduction in EMG at zero degrees are attributed to a movement by the subject against his restraining shoulder harness, changing the true angle, to one, in which, it was easier to maintain the load. It is seen that the EMG changes relatively slowly with angles over the range of one hundred and twenty degrees to sixty degrees. However there is no minimum as in the case of biceps.

The relation between the EMG and the increasing load, with the strain gauge "driving" the recorder is shown in figure 2.6116. The subject gradually increased the effect to flex, starting with his muscles relaxed. This traced out the curves on the right, which shows the biceps activity with "cross-talk" to the triceps. The let down phase of the effort was not recorded.

The subject gradually increased effort to extend, so that the triceps become active. This traced out the similar curves on the left for the triceps activity with
FIGURE 2.6117  **TRICEPS HYSTERESIS EFFECT OF E.M.G.**
Curves for gradually increasing effort.

FIGURE 2.6118  **TRICEPS EFFECT OF BANDWIDTH ON E.M.G. SIGNALS**
Curves for gradually increasing effort.
"cross-talk" to the biceps.

Since only one of the opposing muscles is active at a time, the other is either "producing" a lower voltage cross-talk or a stabilising reverse force for control damping. An estimate of the real activity is obtained by using the difference in voltage between the two.

The above illustrates that the relationship between EMG and tension for the upper arm muscles is not linear, but convex to the tension axis. In addition, even after rectification and smoothing, there remains some apparently random variation in the resulting D.C. Voltage, even when the muscle tension remains constant. The random variation is probably due to the fact that a few muscle fibres around the electrodes have a predominant effect on the voltages obtained, so that the EMG represents only a "sample" of the total stimulation reaching the muscle. The voltage therefore shows sampling error, or random variation.

However, figure 2.6116 shows that the variation of EMG signal is roughly proportional to the mean signal level at any time. Measurement of EMG intensity against tension in a normal muscle is limited by the fact that the arm consists of a conducting mass covered by a poorly-conducting layer of skin. As skin has an impedance of some hundred thousand ohms per square cm., the EMG amplifiers used should have a correspondingly high input impedance to compensate this effect.

The effect of the direction of movement on EMG signals is shown in figure 2.6117. The EMG is less during "let down" when the movement is in the same direction as the load. This looping of the EMG angle
curve has been found to be almost independent of speed of movement up to twenty four deg/sec.

In using the EMG signals, difficulties were encountered with electrical interference arising from stray electromagnetic fields, especially from power lines, myoelectric tissue noise, involuntary contractions and electrode-skin contact instability.

By designing the input of the amplifier as a differential stage, spurious signals can be reduced. Any extraneous potential picked up by the two skin electrodes are in phase and if the amplifier is perfectly symmetrical, no output results. The desired out-of-phase signal is then passed to the transistorized amplifier which is in fact less subject to spurious-signal pick ups.

The input amplifier should have a front end impedance in excess of 1,000 ohms and should preferably match skin resistance that varies between 50,000 - 100,000 ohms when dry, to less than 5,000 - 10,000 ohms when lightly sandpapereed and coated with an electrode jelly.

Limiting the overall bandwidth increases the EMG signal to noise ratio. Preliminary tests show that beyond 1,000 cps., useful myoelectric signals were insignificant whereas, noise was added limiting the lower cut off frequency to 100 cps. results in information loss, but this condition may be tolerated under conditions of severe power-line noise. Thus the most significant harmonic content of the EMG signal falls between 100 - 1,000 cycles/sec.

Figure 2.6118 shows the results of experiments in which the low frequency response was restricted, and high frequency responses above 800 cps. was eliminated.
A second difficulty in using EMG signals from surface electrodes, is the variability of the integrated EMG during steady mechanical effort by the muscle. Figure 2.6116 shows that even after rectifying and smoothing, using a 100 msec time constant, considerable fluctuations remain and this effect increases as the force rises.

It is evident that the muscle EMG activity of the limb can be readily measured by fitting the subject with the appropriate electrodes on the skin surface. Such recorded information when plotted against the characteristic dynamic properties of the model study would automatically indicate the muscles effort contribution to the motion.
Figure 2.71: Six Degrees of Freedom for Any Object in Space
FIGURE 2.72  OPENING A DOOR--COMMAND/FEEDBACK
CONTROL OF LIMB MOVEMENTS

For a human hand, six degrees of freedom of motion exist to position an object; three to place it in space, as defined by the three familiar coordinates x, y and z, and three to orient the object itself in the attitudes known as Pitch, Roll and Yaw as illustrated in figure 2.71. In addition to this, there exist kinesthetic senses which are capable to detecting changes in force and position, large and small, and transmitting this information back to the brain. This complex activity can be seen in a detailed analysis of the elementary motions and factors involved in any sort of operation.

Consider, say, the seemingly simple operation of opening a door as shown in figure 2.72. The hand, actuated by the arm and directly controlled from the brain, grasps the doorknob and swings the door in an arc of a circle with the hinge axis as its centre. The hand pulling the door must follow an arc lying in the plane, at the level of the knob, parallel to the plane of the floor, and it must conform to the circumference of the circle defined by the distance from the knob to the hinge axis. In doing this, the hand, assisted by the human nervous system, is guided by the door's resistance to being pulled along any other path. In other words, the human motor system responds to a feedback of forces to the brain. If this feedback system did not exist the arm would be free to pull in any direction, and might easily pull the door off its hinges, instead of swinging it open.

It is also apparent that the grasping force of the hand exerted on the knob of the door is controlled to suit the resistance offered in opening the door. This force is
regulated to prevent slippage of the knob from the grasp of the hand. The regulation is dependent on the surface kinesthetic sensors (sensory receptors) of the hand, namely, those of touch, pressure, temperature and skin movements, which are registered in the brain. This allows corrective action to increase or decrease muscular forces in the limb to give the desired condition as required by the "conscious mind".

It is of note to realise that various degrees of conscious and subconscious "brain effort" are necessary for effective control of limb movements. While the factors affecting the amount of effort required are numerous it is apparent that certain fundamental relationships do exist. For instance, the first encounter of a particular manoeuvre requires a conscious effort to perform it and the refinement of control for this movement is crude. The action is said to be clumsy.

Repetition of this manoeuvre however, increases the sensitivity of control until a state is reached where conscious thought is only necessary "to define the task, but not to perform it". This condition can be explained by considering the mechanism involved. That is, the brain initiates the manoeuvre in the form of "motor impulses" to the muscles, which in turn produces muscular movement. This muscular movement is detected by special "sensory receptors" in the muscles and their associated "sensory impulses" are fed back to the brain where they are compared and evaluated. The relationship between command and response signals relates speed and degree of movement of the limb in terms of displacement and forces in the muscles.

In this way the brain has complete control over
the entire movement, especially when the above is considered in conjunction with impulses from the eyes (sight). In figure 2.72a, for example, the impulses from the eyes and the sensory muscle impulses are dominant in controlling the hand up till the instant the hand makes contact with the door knob. As the hand proceeds to grasp the knob, the skin sensory receptors as in figure 2.72b register the contact of the hand with the knob in terms of sensory impulses of touch, temperature, pressure, sliding and surface roughness, and in this way further control over the manoeuvre is available.

The important feature to realise here is that, for one particular movement the brain receives a fixed pattern of sensory impulses (neglecting differences in magnitude) of the same order and frequency. That is, the sensory impulses are a function of the movement. If the above is true, then the clumsiness of the first encounter of a particular movement and the refinement of the same movement after repetition is a direct result of the amount of control by the brain of the motor impulses producing the movement in both instances. Since the basic movement is the same, and results from the same type of motor impulses, it is concluded that the response of the brain, in interpreting the sensory impulses and correcting the motor impulses, increases as the "pattern" of sensory impulses associated with this particular movement is repeated.

The ease of interpretation, or response, is ever increasing from the first encounter of the movement, and it is relatively easy to accept the fact that with an increasing refinement of the movement there is an associated decreasing of the conscious effort by the brain. The
process however does not end there because the refinement of
the particular movement is "finite" while the conscious effort
will continue to decrease until finally no conscious effort
is necessary.

By virtue of this fact, that a particular movement
occurs, then there exists, associated with that movement,
motor and sensory impulses which are initiated and
terminated in the brain, and therefore brain effort exists,
even though no conscious evidence for its existence is
apparent. This brain activity is called the subconscious
effort of the brain.

The process of writing is a good example of the
above, considering of course only the motor and sensory
impulses associated with limb movements and neglecting the
more complex activities of spelling, grammar and content.

The first task to be met in this process is to
grasp the writing instrument in the hand so as to
facilitate its use in writing. This process initially
requires conscious thought to position the "pencil", as
instructed. In addition a constant pressure must be
applied to the sides of the pencil to achieve the required
grasp necessary when writing. Once this has been achieved
the pencil is then directed to "slide" over the paper
surface, and the friction between the paper and the "lead"
of the pencil results in a mark being made as some of the
lead is rubbed off.

The first encounter associated with writing is to
learn to control the movement of the pencil over the paper
to produce a predetermined pattern as required by the
subconscious mind. This takes the form of copying
singularly, in turn, the recognised shape of the
alphabetic characters "aa to zz". On the first encounter usually the initial process of holding the pencil has been mastered and now most of the conscious effort is related to "copying" of the alphabetic characters. However in spite of this concentrated conscious brain effort the limb is unable to achieve "free flowing" movement and hence smooth curved alphabetic characters. The "staccato" movement results from the brain's inability to respond quickly enough to the motor-sensory impulses it receives, and the initial motor impulses initiating the movement result in an over exerted movement of the pencil before the sensory impulses of sight and limb movement can correct the motor impulses continuing the movement. This delay is termed the response, which in the human machine increases as the frequency of operation increases.

Once this second task is mastered, this limb activity is extended into "running writing", where the alphabetic characters are joined together to form, at first small, then later on lengthy combinations of these characters - called words. This final stage requires that the limb be further controlled to move in a longitudinal direction while forming characters or letters; and of course, this is logically extended to forming, in continuous sequences of letters, sentences. By tradition, this varying sequence of letters, called words, repeated over and over, lie in parallel straight lines for convenience of vision. However the important feature to note is that as the familiarity of the words increases, then the familiarity of the limb movements involved to produce the words increases and the lesser is the conscious brain effort required to produce them. Even though little or no conscious brain
effort is required during writing, the limb movements only occur as a direct result of the correct motor and sensory impulse patterns.

Finally, when the process of writing is repeated to the "nth" degree, a stage is reached where no conscious brain effort is necessary to guide the limb through its complicated movements to form the necessary words. In the limit, writing as far as the conscious brain is concerned is a process of putting thoughts onto paper while all associated limb movements are guided by the subconscious brain effort.

This process of expressing thoughts on paper can occur at extremely fast rates, but the mechanism of limb operations is sometimes too slow to execute all the motions initiated in the brain, so that words, and even phrases are left unwritten. This occurrence can be explained by considering the method by which limb movements are achieved.

The motor impulses transmitted from the brain via the nerves are impulses with a maximum deviation time of approximately one thousandth of a second, and this in turn is followed by a "absolute refractory" period or "recovery" time of two to five thousandths of a second. In this recovery time the nerve regenerates itself in readiness to transmit the next motor impulse. By virtue of this fact, there exists a physical limit to the number of motor impulses that can be transmitted to any nerve ending throughout the body in a given period of time. Thus if any particular process initiated by the brain requires that motor impulses be transmitted from the brain at rates exceeding the above, then full recovery of the nerves does not occur and some motor impulses even though initiated in the subconscious
mind to be executed, cannot be executed as they strike an impossible barrier of the absolute refractory period of the nerves (an infinite resistance to impulse passage). However as soon as the refractory period of the nerves has passed, the subsequent motor impulses can flow unrestricted; and the continuation of these impulses depends on the interval between impulses being large enough to allow nerve recovery. Here we are ignoring skeletal muscle fatigue which also results in movements being left unexecuted even though initiated by the brain.

In this way the subconscious mind has full control over the movements of the limb during movements and it is only when "saturation" of the nervous system occurs, that the visual "feedback loop" is used (takes over control of the limb motion) to decrease the rate of conscious effort to within acceptable limits. It is evident that for any manoeuvre by the limbs, that there is firstly, great conscious effort to control the movement, followed by decreasing brain effort and increasing movement control with increasing frequency of encountering the movement.

This decrease in brain activity, suggests that the conscious mind is, as it were, an "automatic programmer" for the brain as a whole. In other words the human senses of the conscious mind, motivated by its senses and instincts, sets in action an instinctive motion of the muscles of the body resulting in sensory impulses to the brain, further adding to the movements control. For a particular repeated movement, the brain "learns" to associate the fixed pattern of motor and sensory impulses with the movement and in so doing, lessens the conscious effort by the brain. This suggests that the brain impulse
pattern loses its resistance and requires less brain effort to sustain the limb movement. This also implies that the conscious mind is only that portion of the whole brain effort at any particular instant which requires some motivation of the impulse patterns it is associated with. As soon as these patterns are learnt by the brain, then the "power" of brain activity is diverted to other tasks requiring more motivation. Therefore the conscious brain effort is only related to "programming" the brain to a stage where it is self sufficient and overridden by another conscious task. This means that the conscious mind has in its power the ability to bring into action any of the subconscious brain impulse patterns as part of its conscious effort and need only initiate the task.

This decrease of brain effort is justified by observations made by Claude A. Villee in "Biology" 4th ed. W.B. Saunders Company London P368-238 Thoughts, Memory and Learning, (35).

**SELF PROTECTION CONTROL**

Two special types of "self protection" control exist for the human limb, namely, (i) Reflex activity, and (ii) Reciprocal innervations of muscles. The reflex action is so named because of the "turning back" of the sensory impulses concerned. When a receptor is stimulated, as for example, when the finger touches a hot stove - an impulse is carried to the cell body in the central nervous system and makes contact with a motor nerve cell, or motor neuron, which, in turn, sends an impulse down to the muscles to lift the burned finger off the stove. The impulse is thus turned back to a place near where it started. Thus reflexes are involuntary, and often unconscious.
Reciprocal innervation of muscles is related to the coordination of muscle activity during movements. Consider; when the arm is bent (flexed) at the elbow by the action of the biceps muscle, the triceps muscle, which normally straightens (extends) the arm at the elbow, must relax, or the two muscles would be pulling against each other and nothing would be accomplished. The reverse must be true when an attempt is made to extend the arm.

This is accomplished by a process called reciprocal innervation, in which cooperation is brought about between the nerves supplying any antagonistic pair of muscles.

In addition to the above there are many other similar coordinating systems between motor and sensory receptors, and although little is at present understood concerning them, the effect upon limb control is unmistakeable.
SIMULATED MODEL THEORY
CHAPTER 3 SIMULATION MODEL THEORY

3.1 THEORETICAL

The dynamic analysis of the upper extremity limb is likened to the dynamics of a rigid linked, frictionless jointed, mechanism in free space. This analogy is valid, provided, the rigid bone structures and surrounding muscle and tissues behave as a rigid body during the analysis. The shoulder, elbow and wrist joints must also behave as perfect ball and socket or hinged joints.

The physical characteristics of the upper extremity limb as detailed in Chapter 2, allows a number of simplifying assumptions to be made. Additionally, the selection of a suitable simulation model experimental task further enhanced the validity of these simplifying assumptions.

In the overall sense, the upper extremity limb consists of three basic members - namely (1) upperarm (2) forearm and (3) the hand. The upperarm has a single rigid bone, the humerus which is attached to the shoulder by a ball and socket joint (refer Section 4.31 figure 4.311 photos 1 to 4) and is attached to the forearm by a pure hinge joint at the elbow. The humerus is surrounded by a mass of muscle, skin and tissue which varies its relative position, mass distribution and viscous drag and inertia forces during any motion. The effect of this tissue, muscle and skin movements is common to all three basic members of the limb.

The forearm consists of two rigid bones the ulna and the radius surrounded by muscles, tissue membrane and skin. At the elbow, the ulna is attached directly as a hinged joint to the humerus of the upperarm, (refer
Section 4.31 figure 4.311 photos 2 to 9) while the radius rotates in a fixed arc around the ulna, allowing limited axial rotation of the wrist with respect to the fixed elbow hinge joint. (Refer figure 3.11).

The ulna and radius rigid bones are attached to the hand via rolling ball joints at the wrist. (Refer Section 4.31 figure 4.311 photos 5 and 6). In three dimensional studies, the rotational movement of the forearm is extremely important, particularly when considering the dynamic analysis of say, opening a door catch or using a screw driver. Since the ulna and radius bone joints are separately attached at the hand, then the "compass" effect of this forearm bone structure can be measured by studying the movements of the limb external skin protuberances produced by the underlying bone structures.

The technique of identifying known bone reference points by the external skin contours and skin marks they produce is used in studying the limb motions. The exact relative positions of the external skin marks with respect to the underlying rigid bone structures and joint positions are located by either superimposing photos of the limb and similar skeleton subjects, or by direct X ray measurements of the subject.

The hand with its twenty nine bones and complex muscle and joint structures is extremely difficult to analyse in terms of individual rigid links, even though each bone forms either a sliding, hinged or rolling type joint with the other bones in the hand. For example, the first and second joints of the fingers act as pure hinge joints while the third joint, a ball joint, allows three dimensional movement. All these limb movements are
FIGURE 3.11  BONE AND JOINT SIMPLIFIED DETAILS OF THE UPPER EXTREMITY LIMB
Clavicle
Humerus
Triceps
Radius
Hand
Ulna
Pivot Point

DOWNWARD STROKE

Biceps

UPWARD STROKE

FIGURE 3.12 SIMPLIFIED DIAGRAM OF MUSCLE ACTIVITY AND MUSCLE-BONE MECHANICS
subject to the usual physical constraints of observed degrees of movement. (refer Section 2.6 - Movements of arm with associated muscle activity).

The problem of the complex configuration of bones and tissue in the hand can be overcome by considering a simulation model task which uses the entire hand mass as a single rigid body during the complete study cycle.

The upper extremity limb therefore closely approximates a rigid linked mechanism whose physiological characteristics can be grouped as follows:

(a) The rigid member boned frame is linked by rolling, hinged or sliding joints. (refer figure 3.11).
(b) Muscle motors attached to the rigid bone frame, in combination, produce lever type action. (Refer figure 3.12 for a simplified diagram of muscle activity and muscle-bone mechanics.).
(c) Movement control nerves on both the muscle motors (power demand) and kinesthetic sensors (force and position measurement) operate together during movement with an overriding visual and motivating control action. (Refer Section 2.7 - Control of limb movements).
(d) A membrane type external skin covering changes shape and contour as the underlying tissue, bone and muscles move.

The degree of sophistication of the dynamic analysis model in terms of the above is directly responsible for the accuracy of the analysis. For instance, a simple model involves only a three member rigid linked mechanism with all dynamic effects being attributed to these pivoted rigid bodies. A more exact model would involve calculation
of the joint frictional forces including the viscous and joint tissue effects. The rolling and sliding nature of each joint in dynamic terms can be obtained by studying either X ray measurements of subjects or by skeleton simulation.

The inclusion of the dynamic analysis of muscle activity and muscle-bone lever mechanics would complete the basic physiological characteristics of the model simulation. The physical consideration of the muscle-bone attachments and lever line of action of muscle forces on these bone attachments can be obtained from the human subject itself. Change in muscle tissue mass distribution is harder to assess, since the degree of muscle contraction varies with the limb position and the duration and degree of effect involved. However, the movement of muscle tissue is evident during the motion by the change in skin contours. In addition, muscle activity can be monitored electrically through the nervous control system characteristics.

When muscles contract they supply muscle power through their attachments on the pivoted bone structures. At the same time the opposing muscle relaxes and a correspondence of electrical muscle activity versus particular movements for known muscle activity can be obtained. (Refer Section 2.61 figure 2.61 biceps/triceps EMG Signals.). Thus by knowing the associated muscle activity to produce motion, and then studying the associated skin surface contours during the motion, the exact duration and degree of muscle activity is obtained.

The degree of muscle contraction and "bulging" can also be measured and correlated to the physical dimension change required to produce the limb movement on a purely bone-muscle mechanism basis.

The change in muscle tissue shape also gives rise
to tissue inertia and viscous drag forces. These forces would not alter the model mechanics of the applied bone forces, but would change the actual physical muscle effort or power dissipation by the muscle during the motion. The viscous drag forces could be calculated on the basis of the physical characteristics of the tissue and the resistance of the skin tissue to distortion and stretching.

A simulation model incorporating the above characteristics requires the following experimental data associated with the limb movements in free space. Two high speed cameras are required mounted at right angles to each other to record the details of the physical characteristics of the limb during its motion. The subject's skin surface requires marks to identify the important underlying bone positions and to obtain a measurement of the rotational movement of the limb members. Concurrent measurement and recording of muscle nervous and control operations define the muscle activity responsible for producing the movements.

Accurate timing and scaling of muscle and associated limb characteristics is an essential requirement for this method of external dynamic analysis.

The high speed film study produces a finite step by step study of the movement. The actual step and limb displacements and shapes are measured on an enlarged scale using ordinary photographic projection and recording techniques. The desired accuracy is obtained by firstly increasing or decreasing the number of film frames per reference step, and secondly, by choosing a suitable enlargement size that allows detailed measurement of the limb positions and surface conditions.

Finally for parts of the study cycle which
require a greater accuracy, for example, during periods of sudden acceleration changes, the number of reference frames can be varied to suit.
3.2 SELECTION OF A MODEL SIMULATION TASK

The upper extremity limb task was chosen after considering a number of everyday movements and their adaptability to the simulation study. The simple task of hammering a 2" nail into a piece of Oregon wood was selected on this basis for the simulation.

The hand and hammer combination behaves as one rigid structure during this hammering motion, while the analysis of the forearm, (ulna and radius bones) was simplified by the absence of forearm rotation. Hence the forearm became a two boned rigid structure which restricted motion of the arm to that in one plane - the XY plane containing the humerus - ulna, elbow hinged bones. The wrist joint also became a better approximation to a simple rolling joint under these conditions of planar motion. The shoulder joint is a pure ball and socket joint.

The high speed camera study was filmed at right angles to the plane formed by the upperarm and forearm members, with the hand and hammer rigid members remaining naturally restrained in this same reference plane.

A separate film study was done on the rotational motion of the forearm and the finger and wrist movements while using a screwdriver to tighten up a screw. However the dynamics of the hand, and in particular the finger movement dynamics were too time consuming to be pursued at that time, and it was decided to restrict the dynamic analysis to the planar motion study of a rigid linked mechanism.
3.3 SIMPLIFYING ASSUMPTIONS

The model simulation task of hammering a nail into a piece of wood simplified the complex nature of the upper extremity limb movements in terms of the inherent model assumptions it permitted.

The hammering planar motion, without forearm or wrist rotation, meant that the shoulder, elbow and wrist joints behave as simple pivot points during the motion. The bones at the shoulder joint, for instance, form a ball and socket joint, which involves frictional joint forces. The analysis assumes that all joints are perfect frictionless pin joints. The presence of the synovial eiling membrane on the bone joint surfaces (refer Section 2.12 Mechanism of Joints) and the relative size of these joints further enhances the assumption that the joint frictional effects are negligible in comparison to the actual rigid member dynamics. This assumption is supported by Wright and Johns (1960) (27).

The limb members consist of an underlying rigid bone structure surrounded by flesh and muscle tissue which is viscous in nature. For this analysis the entire member construction of bones, muscles, nerves and skin covering is considered to behave as one rigid body. By observing the overall member behaviour during the motion the actual reference step velocities, accelerations, mass distribution and geometrical details used for the dynamic analysis are self compensating for the muscle, tissue and skin effects.

In adopting the basic three member pin-jointed frame structure it is assumed that the structure is pin-jointed instantaneously at the reference step and the
FIGURE 3.31 PROPERTIES OF THE HUMAN LIMB ASSUMED AS INPUT DATA FOR THE COMPUTER ANALYSIS
FIGURE 3.32

PROPERTIES OF THE HUMAN LIMB CALCULATED AS OUTPUT RESULTS FROM THE COMPUTER ANALYSIS.
member links allow "full positioning" of the limb members within a hemisphere of revolution.

A planar motion study was finally used after considering the suitability of the various coordinate systems. It was considered that the dynamics of a "free-moving" limb can be evaluated by vectorially summing the dynamics of the limb in more than one plane. This is particularly important when considering the limb movement "data", since a simple method of limb analysis is of little value, if the necessary "input data" for the analysis becomes too complicated to obtain.

On this basis, two planes at right angles were considered for the analysis, using two high speed cameras mounted perpendicular to each other (one viewing vertically downwards and a second viewing horizontally at the front or rear of the subject). Filming a complete limb motion at high speeds, defines most variables associated with any movement of the human limb.

The following computer programme (refer Appendix 8) was written to evaluate the planar dynamics of this particular pin-jointed frame structure, made up of three rigid links free to rotate in the plane of reference as shown in figures 3.31 and 3.32. These figures show the velocities, accelerations, forces, torques, arbitrary applied load at the hand, and their angles of application which have been assumed to act on this particular "frame".

For ease of analysis, the following data obtained by methods previously described, has been assumed. Individual links are designated (1), (2), and (3) with lengths SL1, SL2, and SL3 (feet), and the angles that these links make to the downward vertical axis are designated.
TH₁, TH₂, and TH₃ (radians). Links are shown to have angular velocities THV₁, THV₂, and THV₃ (radians/second) and angular accelerations THA₁, THA₂, and THA₃ (radians/second squared), with all inertia forces calculated using results of the above data and masses SM₁, SM₂, and SM₃ (pounds mass) at centre of gravity distances from their respective pivotal points (A, BC, & DE) of SLG₁, SLG₂, and SLG₃ (feet). Radius of gyrations of these links are given to be SKG₁, SKG₂, and SKG₃ (feet).

The final assumption was to apply an arbitrary load "W" (pounds force) to the hand (at F), at an angle TH₄ (radians) to the downward vertical axis, so as to generalise this analysis.
In general, the simplifying assumptions as listed in Section 3.3 reduce the human limb analysis to a planar study of a three bodied, rigid link structure, with three fixed and frictionless pivot points; these pivots corresponding to the shoulder, elbow and wrist joints. The dynamic analysis of such a structure, is obtained by considering each rigid link as an identity, restrained at its pivot points.

Velocities and accelerations are obtained by relative motion techniques, while forces and torques are calculated by considering the dynamics of each link, knowing that at the common pivot points, these forces are equal. Because of the nature of the variables involved, horizontal and vertical components of velocities, accelerations and forces are calculated throughout, simplifying vectorial summation of these quantities at all joints. This procedure was adopted to comply with the photographic studies of the input data, as described earlier.

The following computer analysis has been developed in three distinct categories; (a) Velocities, (b) Accelerations and (c) Forces and Torques. The procedures involved in each are discussed below:-

(a) Velocities
The elbow, wrist and finger tip velocities, relative to the shoulder joint, are of interest in this analysis since they indicate the speed of movement of the individual parts of the limb. For convenience, the angular velocity of each link was assumed, and this multiplied by the radial distance from the point of application of the velocity to the link pivot point, gives the linear velocity of such a point, relative to the pivot point. For example,
FIGURE 3.33 VELOCITIES ACTING ON LINK (1) - UPPERARM
FIGURE 3.34 VELOCITIES ACTING ON LINKS (2) AND (3) - FOREARM AND HAND
figure 3.33 shows link (1) with pivot point A, about which link (1) is rotating with angular velocity \( \theta_{V1} \) (radians/second) in the clockwise direction, and with an initial displacement angle of \( \theta_{1} \) (radians) to the downward vertical.

The elbow velocity \( V_{BA} \) (feet/second) is obtained by multiplying the angular velocity \( \theta_{V1} \) by the link length \( L_{1} \) (feet). From the geometry of the diagram, the velocity \( V_{BA} \) acts at an angle \( \Delta \theta_{1} \) (radians) equal to the angle \( \theta_{1} \) (radians) plus ninety degrees, or \( \pi/2 \) (radians), as shown in figure 3.33. For later calculations, the vertical and horizontal components of the velocity \( V_{BA} \) are determined; vertical component \( V_{BAV} \) (feet/second) equals \( V_{BA} \) times the SINE of the angle \( \theta_{1} \), and the horizontal component \( V_{BAH} \) (feet/second) equals \( V_{BA} \) times the COSINE of the angle \( \theta_{1} \). This procedure is repeated for links (2) and (3). The wrist velocity relative to the elbow \( V_{DC} \) (feet/second) equals the link length \( L_{2} \) (feet) times the angular velocity \( \theta_{V2} \) (radians/second), and acting at an angle to the downward vertical of \( \Delta \theta_{21} \) (radians) = \( \theta_{2} + 3.1416/2 \) as shown in the attached figure 3.34. Vertical and horizontal components are given by \( V_{DCV} = V_{DC} \times \text{SINE} (\theta_{2}) \) and \( V_{DCH} = V_{DC} \times \text{COSINE} (\theta_{2}) \). Likewise the finger tip velocity relative to the wrist \( V_{FE} \) (feet/second) = \( L_{3} \) (feet) \times \text{angular velocity} \( \theta_{V3} \) (radians/second squared) at an angle of \( \Delta \theta_{31} \) = \( \theta_{3} + 3.1416/2 \) radians, as shown on the attached figure 3.34. Vertical and horizontal components of velocity are given by \( V_{FEV} = V_{FE} \times \text{SINE} (\theta_{3}) \) and \( V_{FEH} = V_{FE} \times \text{COSINE} (\theta_{3}) \).

The velocity of the wrist, relative to the shoulder, is evaluated by vectorially summing the velocities of the elbow, relative to the shoulder; and the wrist velocity
relative to the elbow, as illustrated in the velocity diagram figure 3.34. Here it is a matter of adding the vertical and horizontal components of each of the above velocities, and using Pythagoras' theorem, to find the resultant velocity of these two summed components. That is, vertical and horizontal components of the wrist velocity, relative to the shoulder, VDA are given by:

\[ V_{DAV} \text{(feet/second)} = (V_{BAV} + V_{DCV}) \]
\[ V_{DAH} \text{(feet/second)} = (V_{BAH} + V_{DCH}) \]

with \( V_{DA} = \sqrt{(V_{DAV})^2 + (V_{DAH})^2} \). The angle of application of the velocity \( V_{DA} \) is given as \( \Delta \theta_2 = \phi_1 + 3.1416/2 \) (radians) where \( \phi_1 = \tan^{-1}(V_{DAV}/V_{DAH}) \). Similarly from figure 3.34 the velocity of the finger tips relative to the shoulder \( V_{FA} \) (feet/second) is obtained. \( V_{FA} = (V_{DA} + V_{FE}) \) and

\[ V_{FAH} = (V_{DAH} + V_{FEH}) \]

with \( V_{FA} = \sqrt{(V_{FAV})^2 + (V_{FAH})^2} \). The angle \( \Delta \theta_3 = \phi_2 + 3.1416/2 \) (radians) where \( \phi_2 = \tan^{-1}(W_{AV}/W_{AH}) \).

(b) Accelerations

Angular accelerations and velocities assumed acting on each link of the rigid bodies, pivoted at one end, result in two distinct accelerations; a tangential acceleration due to the links angular acceleration and a normal acceleration resulting from the angular velocity of rotation of the link about its pivot point.

Consider link (1) as shown in figure 3.35, with angular velocity \( \theta_{V1} \) (radians/second) and angular acceleration \( \theta_{AI} \) (radians/second squared). The tangential acceleration of B with respect to A, \( ABAT \) (feet/second squared) equals links length \( L_1 \) (feet) times the angular acceleration \( \theta_{AI} \). Vertical and horizontal components of \( ABAT \) are given by:

\[ ABATV = ABAT \times \sin \]
FIGURE 3.35  ACCELERATIONS ACTING ON LIMBS (1) AND (2)
the angle $\theta_1$ and $AB_{ATH}$ equals $AB_A$ times the \text{COSINE} of the angle $\theta_1$. The normal acceleration of $B$ with respect to $A$, $AB_{AN}$ (foot/second squared) equals the link length $L_1$ (feet) times the square of the angular velocity $\theta_{V1}$ (radians/second$^2$). Horizontal and vertical components of $AB_{AN}$ are by $AB_{AN_V}$ equals $AB_{AN}$ times the \text{SINE} of the angle $\theta_1$ and $AB_{AN_H}$ equals $AB_{AN}$ times the \text{COSINE} of the angle $\theta_1$.

The actual acceleration of point $B$ with respect to $A$, which in this case is the acceleration of the elbow relative to the shoulder, is evaluated by vectorially summing the normal and tangential accelerations as illustrated in the attached figure 3.35. This is a matter of adding the vertical and horizontal components of each of the above velocities and using Pythagoras' theorem to find the resulting acceleration. Thus, vertical and horizontal components of elbow acceleration $ABA$ become $AB_{AV}$ equals $AB_{ATV}$ minus $AB_{ANV}$ and $AB_{AH}$ equals $AB_{ATH}$ plus $AB_{ANH}$, with $ABA$ equal to $((AB_{AV})^2 + (AB_{AH})^2)^{\frac{1}{2}}$. The angle of application of the acceleration $ABA$ is given as $\phi_1 = \phi_3 + 3.1416/2$ (radians) where $\phi_3 = \text{ATAN} (AB_{AV}/AB_{AH})$.

This procedure is repeated for links (2) and (3). The wrist acceleration relative to the elbow $ADC$ (foot/second squared) equals the vectorial sum of the tangential and normal accelerations $AD_{CT}$ and $AD_{CN}$, as shown in figure 3.35. The tangential acceleration $AD_{CT} = L_2 \times \theta_{A2}$ has vertical and horizontal components $AD_{CT_V} = AD_{CT} \times \text{SINE} (\theta_2)$ and $AD_{CT_H} = AD_{CT} \times \text{COSINE} (\theta_2)$. The normal acceleration $AD_{CN} = L_2 \times (\theta_{V2})^2$ has vertical and horizontal components $AD_{CN_V} = AD_{CN} \times \text{COSINE} (\theta_2)$ and $AD_{CN_H} = AD_{CN} \times \text{SINE} (\theta_2)$. 


FIGURE 3.36 ACCELERATIONS ACTING ON LINK (3) AND FINGER TIPS
The angle of application of the acceleration ADC is given as $RH_2 = PH_4 + 3.1416/2$ (radians) where $PH_4 = ATAN (ADCV/ADCH)$. The vertical and horizontal components of the wrist acceleration ADC are given by:

$$ADCV = (ADCTV - ADCNV)$$ and $$ADCH = (ADCTH + ADCNH)$$ and

$$ADC = ((ADCV)^2 + (ADCH)^2)^{1/2}.$$

The finger tip acceleration, relative to the wrist, $AFE$ (feet/second squared) equals the vectorial sum of the tangential and normal accelerations $AFET$ and $AFEN$ as shown in the figure 3.36. The tangential acceleration $AFET = L_3 * TH_3$ has vertical and horizontal components $AFETV = AFET * SINE (TH_3)$ and $AFETH = AFET * COSINE (TH_3)$ and the normal acceleration $AFEN = L_3 * (THV_3)^2$ has vertical and horizontal components $AFENV = AFEN * COSINE (TH_3)$ and $AFENH = AFEN * SINE (TH_3)$.

The vertical and horizontal components of the wrist acceleration $AFE$ are given by:

$$AFEV = AFETV - AFENV$$ and $$AFEH = AFETH + AFENH$$ and

$$AFE = ((AFEV)^2 + (AFEH)^2)^{1/2}.$$

The angle of application of the acceleration $AFE$ is given as $RH_4 = PH_6 + 3.1416/2$ (radians), where $PH_6 = ATAN (AFEV/AFEH)$.

The accelerations of the wrist and finger tips, relative to the shoulder, are evaluated by vectorially summing the appropriate accelerations. Refer figure 3.36 for example, which illustrates the acceleration of the wrist relative to the shoulder $ADA$ (feet/second squared), obtained by vectorially summing the two component accelerations of the wrist relative to the elbow $ADC$ and the elbow relative to the shoulder $ABA$. This is done by adding the vertical and horizontal components of each of the above accelerations and finding their resultant. Vertical and horizontal components of the wrist
Figure 3.37
Centre of Gravity Accelerations of Links (1), (2) and (3)
acceleration \( \text{ADA} \) become \( \text{ADAV} = \text{ADCV} + \text{ABA} \) and \( \text{ADA} = \sqrt{((\text{ADAV})^2 + (\text{ADA})^2)} \). The angle of application of the acceleration \( \text{ADA} \) is given as \( \text{RH3} = \text{PH5} + 3.1416/2 \) (radians) where \( \text{PH5} = \text{ATAN}(\text{ADAV}/\text{ADAH}) \).

Similarly, the acceleration of the finger tips, relative to the shoulder, \( \text{AFA} \) is obtained by vectorily summing the two component accelerations of the finger tips, relative to the wrist, \( \text{AFE} \) and the wrist, relative to the shoulder \( \text{ADA} \), as shown in figure 3.36. Vertical and horizontal components of each of the above accelerations are added and the wrist acceleration \( \text{AFA} \) becomes \( \text{AFA} = \sqrt{((\text{AFA})^2 + (\text{AFA})^2)} \) where \( \text{AFA} = \text{AFE} + \text{ADAV} \) and \( \text{AFA} = \text{AFE} + \text{ADAH} \). The angle of application of the acceleration \( \text{AFA} \) is given as \( \text{RH5} = \text{PH7} + 3.1416/2 \) (radians) where \( \text{PH7} = \text{ATAN}(\text{AFA}/\text{AFA}) \).

Finally, the accelerations of the centre of gravity of each link, relative to its pivot point, is calculated for use in determining link inertia forces. The accelerations calculated for each link \( \text{ABA}, \text{ADC} \) and \( \text{AFE} \) result from the vectorial summing of the normal and tangential accelerations, whose magnitude depends, linearly, on the link length; that is, \( \text{SL1}, \text{SL2} \) and \( \text{SL3} \) (feet). Thus the centre of gravity accelerations, of links whose lengths are \( \text{SLG1}, \text{SLG2} \) and \( \text{SLG3} \), are proportionately \( \text{SLG1}/\text{SL1}, \text{SLG2}/\text{SL2} \) and \( \text{SLG3}/\text{SL3} \) of the accelerations \( \text{ABA}, \text{ADC} \) and \( \text{AFE} \), as shown in figure 3.37. That is, \( \text{AG1} = \text{SLG1}/\text{SL1} \ast \text{ABA} \), \( \text{AG2} = \text{SLG2}/\text{SL2} \ast \text{ADC} \) and \( \text{AG3} = \text{SLG3}/\text{SL3} \ast \text{AFE} \), where \( \text{AG1}, \text{AG2} \) and \( \text{AG3} \) (feet/second squared) are the link centre of gravity accelerations.

(c) **Forces and Torques**

This analysis is obtained by considering the
dynamic equilibrium of the linked structure at any instant, for that particular instant, of operation. For dynamic equilibrium, the following conditions must be satisfied; the algebraic sum of all vertical forces is zero, the algebraic sum of all the horizontal forces is zero, and the algebraic sum of all the moments about any point on the link or structure is zero. In addition to the above, the following technique was used to determine unknown forces acting within, and or/ on the links, where varying procedures were used, depending upon the number of unknown values involved. For instance, if there are only three unknowns, then a solution is obtained by applying the three equations for dynamic equilibrium to each isolated link. If there are more than three unknown quantities for a single link, then it is necessary to obtain additional information by considering the adjoining links. In this way, the unknowns are reduced in number to three for each link, and hence solved.

In this particular case, each link, when isolated has four unknown forces, so that by taking moments of forces about the link's pivot points, the perpendicular component of the pivot force is obtained at the link extremity. This procedure is repeated, to find the corresponding perpendicular component of the pivot force on the adjoining link. The vectorial summation of these two component forces, yields the actual force acting at that particular pivot point.

In this way the shoulder, elbow and wrist joint forces are calculated. Torques, acting on limbs, are obtained by considering properties of the links; with torque variations during movements automatically compensated for by changes in the link's angular accelerations.

The forces and torques acting within and on the
links have been evaluated by assuming the following information (as well as that already included in velocity and acceleration calculations); (a) Angular accelerations, (b) Mass of the link, (c) Radius of gyration of the links, (d) Location of the link centre of gravity from its pivot point. Torques are found by multiplying the mass of the link (say SM1) by its radius of gyration squared (SKG1)^2, times the angular acceleration of the link (THA1). In this way the torques T1, T2 and T3 were obtained, with T1 = SM1 * (SKG1)^2 * THA1 (pounds feet),
T2 = SM2 * (SKG2)^2 * THA2 and T3 = SM3 * (SKG3)^2 * THA3.

Link inertia forces (FI1 say) are obtained by multiplying the mass of each link (SM1) times its linear acceleration at its centre of gravity (AG1). Thus FI1 = SM1 * AG1 (pounds force), FI2 = SM2 * AG2 and FI3 = SM3 * AG3.

The weight of each link acts through the link's centre of gravity, where for link 1, the weight (pounds force) equals the mass of the link (SM1 pounds mass) times the gravitational acceleration G (feet/second squared). Using the above, and assumptions imposed on this analysis, it is evident that the only unknown quantities are the forces acting at the link pivot points, A, B, C, D and E. However, points B and C, D and E are common frictionless pivot points, and therefore forces applied by connecting links at these points are equal and opposite in magnitude.

The force analysis is obtained by considering each link separately, and replacing the pivot pin forces by vertical and horizontal component forces of unknown value. Thus, by taking moments of the forces and torques about the appropriate point on the link, some of these unknown
SHOULDER JOINT

ELBOW JOINT

FIGURE 3.38 FORCES AND TORQUES ACTING ON LINK (1)
FIGURE 3.39 (CONT.)  FORCES AND TORQUES ACTING ON LINKS (2) AND (3)
**FIGURE 3.39** FORCES AND TORQUES ACTING ON LINKS (2) AND (3)
forces can be obtained in terms of the known quantities involved. The final magnitude of forces acting at pivot points is found by vectorially summing the component forces of the two links at the common pivot points.

Consider for example, link (1) as shown in figure 3.38 with torque $T_1$ acting around A and forces $F_{BC}$ and $F_{AO}$ acting at points $A$ and $B$ respectively. The link weight $S_{M1} \times G$ acts through its centre of gravity and inertia force $F_{I1}$ through the centre of percussion of the link. Therefore taking moments about the point $A$, gives the vertical component of force $F_{BC}$ at the point $B$;

$$F_{BCV} \text{ (pounds force)} = \left( F_{I1} \times S_{LQ1} - T_1 \right) + \left( S_{M1} \times G \times X_{LG1} \right) \text{ divided throughout by the link length } S_{L1},$$

where $F_{I1} = S_{M1} \times A_{G1}$, $S_{LQ1} = (B_{I1} + B_{G1})$, $B_{I1} = \left( (S_{KG1})^2 \times \theta_{A1} \right) / A_{G1}$, $B_{G1} = S_{LG1} \times S_{INE}(G_{A1})$, $G_{A1} = (R_{H1} - \theta_{H1})$,

$$T_1 = S_{M1} \left( S_{KG1} \right)^2 \times \theta_{A1} \text{ and } X_{LG1} = S_{LG1} \times S_{INE}(\theta_{H1}).$$

Taking moments about point $D$ for link (2), as shown in figure 3.39, gives $F_{CBV} = (T_2 + (S_{L2} \times G \times X_{LG2})) / S_{L2}$ where $T_2 = S_{L2} \left( S_{KG2} \right)^2 \times \theta_{A2}$,

$$X_{LG2} = (X_{L2} - X_{LG2})$$

$S_{L2} = S_{LG2} \times S_{INE}(\theta_{H2})$,

$G_{A2} = (R_{H2} - \theta_{H2})$, $X_{LG2} = S_{LG2} \times S_{INE}(\theta_{H2})$,

$F_{I2} = S_{L2} \times A_{G2}$, $Y_{LQ2} = (Y_{L2} - Y_{LQ2})$, $Y_{L2} = S_{L2} \times S_{INE}(G_{A2})$,

$S_{LQ2} = (B_{I2} + B_{G2})$, $B_{I2} = \left( (S_{KG2})^2 \times \theta_{A2} \right) / A_{G2}$ and $B_{G2} = S_{LG2} \times S_{INE}(G_{A2})$.

The actual pivot pin force $F_{CB}$ (and $F_{BC}$) is obtained by vectorially summing the two component forces $F_{BCV}$ and $F_{CBV}$. This is obtained by taking vertical and horizontal components of the above component forces; adding them, and finding their resultant. (Refer figure 3.39). That is, $F_{CB} = \left( (F_{CBV})^2 + \right.$
where \(FCBXV = (FCBV + FBC7V), FCBV = FCBV * \sin (TH2), FBCV = FBCV * \sin (TH1), FCBXH = (FCBVH + FBCVH), FCBVH = FCBV * \cos \theta (TH2) \) and \(FBCVH = FBCV * \cos \theta (TH1).\) The angle of application of the force \(FCB\) is given by \(DEL4 = (PH8 + 0.5 * 3.1416)\) radians, where \(PH8 = \arctan (FCBXV / FCBXH).\)

Similarly, by considering the forces acting on links (3) and (2), the wrist joint force \(FDE\) is obtained. However, because of the applied load \(W\) (pounds force) at point \(F\) on link (3), the torque \(T3\) is obtained separately.

Consider link (3) as shown in figure 3.39, where the effect of the applied load \(W\) is to reduce the effective torque \(T3\) which would otherwise be applied due to the link's angular acceleration. Therefore the new \(T3 = (T3 - W * SL3')\), where \(SL3W = SL3 * \cos \theta (TH4)\) and \(T3 = SM3 * (SKG3) * THA3.\)

This new value of \(T3\) is then used in the calculation of the moments about \(F\) for link (3) as shown in figure 3.39. This gives, \(FEDV = (T3 + (FI3 * YLQ3) + SL3 * G * XLG33) / SL3,\) where \(T3\) is defined above, \(FI3 = SL3 * AG3, YLQ3 = (YL3 - SLQ3), YL3 = SL3 * \sin (GA3), GA3 = (RH4 - TH3), SLQ3 = (BI3 + BG3), BI3 = ((SKG3)2 * \theta A3) / AG3, BG3 = SLG3 * \sin (GA3), XLG33 = (XL3 - XLG3), XL3 = SL3 * \sin (TH3) \) and \(XLG3 = SLG3 * \sin (TH3).\)

Taking moments about the point \(C\) for link (2), as shown in figure 3.39, gives \(FDEV = ((FI2 * SLQ2) + (SM2 * G * XLG2) - T2) / SL2,\) where all quantities have been previously defined in calculations for \(FCBV.\)

The wrist joint force \(FDE\) (and \(FED\)) is obtained by vectorally summing the two component forces \(FEDV\) and \(FDEV.\) This is obtained by taking vertical and horizontal
FIGURE 3.310 FORCES AND TORQUES ACTING ON LINK (1) (CONT.)
Figure 3.310 Forces and Torques Acting on Link (1)
components of the above component forces; adding them, and finding their resultant. (Refer figure 3.310). That is

\[ F_{DE} = \left( (F_{DEXV})^2 + (F_{DEXH})^2 \right)^{\frac{1}{2}} \]

where \( F_{DEXV} = (F_{EDW} + F_{DEW}) \),

\( F_{EDW} = F_{ED} \times \sin(\theta_2) \),

\( F_{DEW} = F_{DEV} \times \sin(\theta_3) \),

\( F_{DEXH} = F_{EDVH} + F_{DEVH} \),

\( F_{EDVH} = F_{EDV} \times \cos(\theta_2) \) and

\( F_{DEVH} = F_{DEV} \times \cos(\theta_3) \). The angle of application of this force \( F_{DE} \) is given by \( \Delta_5 = (\phi_{11} + 0.5 \times 3.1416) \) radians, where \( \phi_{11} = \arctan \left( \frac{F_{DEXV}}{F_{DEXH}} \right) \).

By considering the forces acting on links (1) and the results previously obtained, the shoulder joint force \( F_{AO} \) is obtained. The vertical component of this force \( F_{AOV} \) is obtained by taking moments about the point \( B \) as shown in figure 3.310, giving

\[ F_{AOV} = \left( (F_{I1} \times Y_{LQ1}) + \left( S_{M1} \times G \times X_{LG11} \right) + T_1 \right) / S_{L1}, \]

where \( F_{I1} = S_{M1} \times A_{G1}, \)

\( Y_{LQ1} = (Y_{L1} - S_{LQ1}), \)

\( Y_{L1} = S_{L1} \times \sin(\alpha_{1}), \)

\( \alpha_{1} = (R_{H1} - \theta_{1}), \)

\( S_{LQ1} = (B_{I1} + B_{G1}), \)

\( B_{I1} = ((S_{KG1})^2 \times \theta_{A1}) / A_{G1}, \)

\( B_{G1} = S_{LG1} \times \sin(\alpha_{1}), \)

\( X_{LG11} = (X_{L1} - X_{LG1}), \)

\( X_{L1} = S_{L1} \times \sin(\theta_{1}), \)

\( X_{LG1} = S_{LG1} \times \sin(\theta_{1}) \) and

\( T_1 = S_{M1} \times (S_{KG1})^2 \theta_{A1} \). The horizontal component of the shoulder force \( F_{AOH} \) is obtained by considering the axial forces in link (1). The axial force at the elbow \( F_{BCH} \) is found by taking the component of the elbow joint force \( F_{OB} \) in the direction of link (1), as illustrated in figure 3.310. That is, \( F_{BCH} = F_{CB} \times \cos(\phi_{10}) \), where \( \phi_{10} = (\theta_{2} - \phi_{8}) \). The axial force acting at the shoulder \( F_{AOH} \) is obtained by applying the dynamic equilibrium condition, that the algebraic sum of all the horizontal forces is zero.

Forces acting are: elbow force \( F_{BCH} \), inertia force \( F_{I1} \), link weight \( S_{M1} \times G \) and shoulder force \( F_{AOH} \);
of which $\text{FAOH}$ is the only unknown. (Refer figure 3.310). The shoulder force $\text{FAOH} = \text{FBCH} + \text{FHI}_1 - \text{SMH}_1$, where the horizontal forces $\text{SMH}_1 = \text{SM}^1 * G * \cos(\text{TH}_1)$ and $\text{FHI}_1 - \text{FI}^1 * \cos(\text{GA}^1)$. The shoulder joint force $\text{FAO}$ is then obtained by vectorially summing its horizontal and vertical forces.

Thus, $\text{FAO} = \sqrt{(\text{FAOH})^2 + (\text{FAOV})^2}$. The angle of application of this force is given by $\text{DEL} = \text{TH}_1 + \text{PH}_12$ where $\text{PH}_12 = \arctan(\text{FAOV}/\text{FAOH})$. 
3.5 **BASIC COMPUTER PROGRAMME**

The basic computer programme was written in FORTRAN for a IBM 1620 Electronic Digital Computer. Appendix 8 contains the FORTRAN computer language programme listing for this particular dynamic analysis of the upper extremity limb.

3.6 **COMPUTER PROGRAMME SAMPLE CALCULATIONS**

The computer programme was used to calculate the human limb dynamics. Computer input data was selected using rough estimates of the upper extremity limb behaviour. Details of input data and calculated output values are given in Appendix 8.

There are two calculations illustrated; each for the same subject, but with the difference that Case (a) has no load applied at the hand, while Case (b) has an applied load of 100 pounds force acting vertically downwards.
3.7 DISCUSSION OF RESULTS

The results listed as computer output (refer Appendix 8) indicate the type of dynamic relationships involved in this "mathematical model" and illustrate the magnitude of the variables involved. All values are quoted in British Standard Units of pounds (mass), feet, seconds, with all angles in radians. For example, in case (a) results, the elbow velocity equals 7.00 feet per second at an angle of 1.57 radians to the downward vertical and measured in a clockwise direction.

Case (a) and (b) studies differ only by the applied load at the hand; with case (a) there is no load, while case (b) has a 100 pounds (force) load applied vertically downwards. The results obtained from these two calculations show that only the wrist torques and wrist joint forces change, all other results remaining the same. That is, in case (a) the wrist torque of 0.02 pounds (force) feet and wrist joint force of 2.8 pounds (force) change in case (b) to -32.98 pounds (force) feet and 97.20 pounds (force).

The absence of change in the velocities, accelerations and forces for cases (a) and (b) results from the choice of computer input data. While it was assumed that the angular accelerations and velocities of the "links" were the same for both cases, any sluggishness, inherent in the human limb movement and resulting from the applied load in case (b), is not considered, and is therefore automatically cancelled by it. However, if studies were made of human limb movements of one person, to obtain the necessary input data under the conditions of case (a), and then under these same conditions of case (b) (with the 100 pounds force applied), a much greater effort would be
required by the limb muscles to obtain such velocities as in case (b). It is more probable that much smaller accelerations and velocities would be obtained, requiring more muscle power for their execution.

In this way the above computer programme results are already compensated; relying only on accurate photographic studies of limb movements to furnish the appropriate input data.
3.8 FUTURE PROGRAMME REFINEMENTS

The ideal mathematical model, embracing all the properties, quantities and facets of the human limb dynamics, will define the human limb and its characteristic movements as we have learnt to recognise them. The major portion of such a mathematical model is represented in the type of elementary model as described in Sections 3.3 to 3.7 of the above. Such a model would supply the necessary information related to limb movements; including the human limb’s limitations and capabilities, and the essential "data" for the ultimate design of "real life" artificial limbs.

The nature and type of variables involved in this study of limb motion, makes it ideally suited for computer analysis. This is especially the case when considering the sequencing of large amounts of information and automatic "output" plotting available in computer analysis techniques. It is therefore considered most fruitful to develop a model in accordance with the above.

The basic background of knowledge required for the development of the ideal mathematical model will be obtained from the detailed study of the following:-

(i) The complete tabulation of the physical (outward) appearances of the limb, is required for individual subjects, including cross section and shape changes, observed movements, limb dimensions, and skin surface distortions, to mention a few.

(ii) X-ray studies to record bone and tissue movements, joint action and "mass flow" during movements of the limb is required.

(iii) The behaviour during motion of bone structures and their joint attachments, including methods of
lubrication and deflection effects of bones requires tabulation.

(iv) Muscle structure, mechanics and attachment details, including changes in muscle shape, muscle forces (both total and distributed) and frictional effects of muscles moving within the limb is required. The latter includes consideration of "flesh flow", (and its energy effects in motion) skin distortions, and blood vessel flow.

(v) Environmental temperature effects on motion with particular reference to skin shrinkage, viscosity changes, blood flow, frictional drag forces, and the body temperature effect on efficiency of muscle power is required.

(vi) Measurement of nerve neuro and muscle electromyographic signals, in conjunction with external photography of particular limb activities is required. This should give a pattern of muscle, bone and skin behaviour, and their characteristics for particular limb movements.

(vii) Detailed study of the senses of the skin, including especially the pressure and touch "receptors" is required to understand the limb control systems.

From the above considerations, the following logical steps in the development of this study have been listed:-

(a) Develop an elementary mathematical model containing as many major characteristics as possible; (b) Development and tabulation of all properties related to the limb; (c) Refinement of the elementary mathematical model so that complete motions can be obtained synthetically in terms of "sensory and motor nerve impulses"; (d) Development of
an artificial limb incorporating all the refinements possible, and including the "sense of touch", without which upper extremity limbs (artificial) are dispensible. The latter explains the reason why "hooks", using visual feedback to the brain, partly compensating for the sense of touch, are more widely used for upper extremity artificial limbs than cosmetic mechanical hands, which because of their bulk, obscure the visual feedback, and thus they lack the "touch sense"; (e) Design of "real life" artificial limbs. Any development of a real life artificial limb is required to replace all the essential movements of the human limb, and at the same time, maintain the dynamic and nerve impulse body balance in its original state.

As a summary of the general types of variables involved, the following list of quantities of immediate interest in future studies has been prepared. The list includes many variables which can readily be obtained by visual limb studies, as well as other quantities which can be analysed within fixed limits and tolerances, depending upon how many of the above variables are neglected in the analysis; (f) Refinement of the mathematical model to include joint and muscle mechanics. This involves the measurement of muscle electromyographic signals as a means of supplementing data obtained from the high speed camera study.

Quantities of interest are:-

(1) Displacements of limb members from a prescribed reference.
(2) Angular velocities and accelerations of members.
(3) Torques and axial rotations of members.
(4) Linear velocities and accelerations of the shoulder, elbow, wrist and individual fingers relative to one another and to the trunk of the body.

(5) Type and nature of forces acting in joints.

(6) Forces within bones (axial forces, direct and bending shear forces).

(7) Muscle forces and their distribution throughout the muscle bodies and tendons; velocity of shortening, dimension changes, mass distribution change, viscous drag on muscle body on its outer tissues, and inertia forces acting throughout.

(8) Skin and tissue movements, change in shape, section change, inertia effects and mass movements.

The tabulation of these quantities during any movement is essential to give the kind of knowledge needed for future studies of this kind. Thus in computer analysis techniques, the human limb movements are broken up into stages, say one hundredth of a second apart, and quantities are evaluated at each successive stage. The complete set of results are then graphed or plotted automatically for any one or more quantities over the entire movement, giving a visual and continuous account of all quantities as required. In this way, maximum and minimum values, movement characteristics, and limb component behaviour can be studied in the manner required.
CHAPTER 4

EXPERIMENTAL PROCEDURE
CHAPTER 4  EXPERIMENTAL PROCEDURE

The procedure adopted for the dynamic analysis of the upper extremity limb during hammering a nail into a piece of wood are detailed below.

The simulation model details are contained in Chapter 3.

The experimental procedures involved:
(a) High speed camera filming of the subject during the motion.
(b) Plotting of the film step by step displacements.
(c) Skeleton studies and limb measurements to define physical properties.
(d) Tabulated calculations of reference step data.
(e) Computer simulation analysis of the motion dynamics.

The dynamic analysis results are detailed in Chapter 6.
PHOTO 4.11  HIGH SPEED CAMERA EXPERIMENTAL FILMING
ARRANGEMENT OF SUBJECT, CAMERA AND LIGHTING
4.1 **HIGH SPEED CAMERA STUDY EXPERIMENTS**

A high speed film was taken of the motion of the right arm during the hammering of a 2" nail into a 3" x 2" Oregon timber block.

A number of trial films were made to determine the best filming speed for this motion. Difficulty was experienced in obtaining sufficient light to yield the correct film exposure and subject clarity at the required film speed. The film speed was determined by considering the fact that of the one hundred feet of film on the reel, an allowance had to be made for speeding up of the film to the required speed prior to the start of the motion cycle, and that sufficient film was available to cover a full one and one half cycles of the study motion. Additionally in order to film the motion, the subject had to be in the process of hammering successive blows to the nail, and the film was started to capture the motion on the appropriate film position.

Using ILFORD 100 feet reels of HP S 16 mm Panchromatic Safety negative cine film, the average film speed during the study was 1800 frames per second for a hammer cycle time of about 1.5 seconds. For this filming, the camera was positioned to frame the subject throughout the motion and to record details of a complete downward strike, impact and upward stroke of the hammering motion. The axis of the camera lens was positioned at right angles to the surface plane formed by the upperarm and forearm. Photo 4.11 shows the high speed camera experimental filming arrangement of the subject, camera and lighting positions. The camera is clearly shown to point downwards by about fifteen degrees from the horizontal axis. This
PHOTO 4.12  TYPICAL SERIES OF HIGH SPEED FILM SEGMENTS
Inclination angle corresponds to the inclination of the plane of the arm from the vertical axis of the body during the motion. The camera used was a standard 16 mm high speed camera with adjustable speed ranges and selected timing mark facilities.

Before proceeding with further filming a study of the limb bone structure was undertaken to determine the behaviour of the rigid bone structure during the motion. A number of bone protuberances were identified by the characteristic skin contours they produce. The skin was marked at a number of these points with \( \frac{1}{4} \)" diameter circles using a black felt pen. Reliable measurements of bone structure movement results from selecting the skin reference points on the same rigid bones, and carefully positioning these marks to avoid relative displacements of the skin with respect to the underlying bone protuberances.

After setting up the equipment, filming speeds and exposure times, the film was made of the hammering motion. A series of segments of the film is shown in Photo 4.12 as being typical of the details captured on film. This film was viewed to verify that sufficient detail of the motion was available for the dynamic analysis.
PHOTO 4.13

HIGH SPEED FILM PROJECTOR AND TRACING TABLE
ARRANGEMENT FOR VIEWING AND RECORDING
FILM DETAILS
4.2 FILM EVALUATION

A 16 mm film projector fitted with a remote control switch was used to project the film image onto a specially built tracing table. (Refer Photo 4.13 for details). The film image is projected onto a mirror at the base of the table and is reflected onto the glass top. A sheet of semi opaque tracing paper was placed on the glass top for viewing and recording purposes.

Standing astride of the projector image beam, the film image was viewed on the tracing paper and accurate step by step identification and plotting of details of the movements was recorded.

After viewing the film in this manner the size of projected image was adjusted to a suitable size and scale as required. Secondly the number of film frames per reference step was chosen to obtain the desired accuracy in the plotted data, namely - 40 frames per reference step resulted in an average angular displacement of three degrees during the majority of the motion. Thirdly the film speed timing mark track was checked to verify that the camera was up to speed at the start of, and during the actual motion.

To record and plot the upper extremity limb motion details the film was set at a selected starting point, such as, reference frame 00 on Drawing LIMBO 3 corresponding to a point in the motion preceding the actual start of the downward swing of the hammer. At this and successive reference points the projected image outline and skin mark positions were plotted and labelled. Using the remote projector control switch, the film was then
stepped through a single frame at a time from reference step to reference step by counting the number of frames per reference step - in this case 40 frames per step. At each successive reference step position the outline and skin mark details are recorded. Care was taken to guarantee that the photo frame had the same fixed datum reference to eliminate errors resulting from the film image and projector image displacements.

Because of the number of lines drawn for each reference step the motion was broken up into two parts - the downward and upward strokes. Drawing LIMBO 3 shows the overall limb reference step displacements and skin marks during the downward stroke and Drawing LIMBO 4 shows the corresponding details of the impact and upward stroke. Each Drawing LIMBO 3 and LIMBO 4 also contains the details of the start of the film speed timing track marks on the right hand edge of the film frame.

The timing track is electrically produced at the time of filming by turning a light on and off at selected time intervals. This light is registered as a black strip on the edge of the film - beside the photo frames. The time interval between the leading edge of two successive black strips in this case was chosen at 100 milliseconds. In stepping the film through the motion these timing marks randomly occur. The film is considered to be a continuous strip with each film frame being allocated a number between 0 and 39 (40 frames per reference step) and each frame subdivided further into ten parts along its length. The leading edge of the timing marks are not sharply defined but the closest one tenth frame division is taken to represent the timing reference mark. As an example of this, consider
Drawing LIMBO 4 where on the right hand side of the frame there are ten divisions of the frame length or height. Between the graduations (1) and (2) a timing mark leading edge started on the thirty fifth frame after the reference step (17). This timing mark then became identified as (17) + 35.1 film frames. This process was repeated to record the film length positions corresponding to each fixed time interval mark. Knowing the film distance moved in a finite time interval enables calculation of filming speed and accurate computation of limb displacements.

The film details plotted on Drawings LIMBO 3 and LIMBO 4 were double checked for correct timing marks and reference step positions for the entire motion. The details of these two drawings were converted into more meaningful information by eliminating the limb outline and considering only the rigid bone structure and joint displacements. The simulation model (refer Chapter 3) is based on a rigid linked mechanism and requires data of the true bone positions for the analysis. The skin marks in Drawings LIMBO 3 and LIMBO 4 were specially chosen to locate two points on the one rigid bone for the assumed three member rigid bone structure. The stick diagrams Drawings LIMBO 5 and LIMBO 6 were produced directly from the skin marks details on Drawings LIMBO 3 and LIMBO 4 and the lines or sticks result from joining points on the same rigid bones. Consider the reference step point (8) on Drawing LIMBO 5 as a typical example. At the shoulder, point (8) is clearly identified as being a point on the top end of the humerus bone of the upperarm. At the elbow two reference points (8) appear — (a) one corresponding to the peak on the elbow and of the humerus bone and (b) one corresponding to the point of
the elbow or the elbow end of the forearm ulna bone. (Refer Chapter 4.31 for details of Bones and known bone protuberances). Similarly at the wrist the lower point (8) corresponds to a known bone peak on the wrist end of the ulna, while the higher point (8) is a point on the forearm radius bone. Similarly marks on the knuckles and the hammer face define two points on the hand rigid member. In this way the joining of points on the same rigid bone produces the stick diagrams which record the relative displacements of the rigid bones at reference step positions during the motion. The angles formed by this stick diagram were measured and summarised in tabular form on the right hand side of the film frame outline. Typically for reference step (12) the angle of inclination \( (\theta_1)_{12} \) of the line joining the two points on the upperarm humerus bone (member 1) to the vertical axis is measured at 17.7 degrees. Similarly the forearm ulna angle on member 2, \( (\theta_2)_{12} \) is ninety one degrees and the hand–hammer angle, \( (\theta_3)_{12} \) is ninety six degrees.

The hammer impact position and nail movement into the wood is marked on Drawing LIMBO 5.
4.3 SKELETON EXPERIMENTS

The idea of using a human skeleton was employed for three reasons, namely (1) to overcome the difficulty of identifying and recording details of the underlying bone structure location and characteristics of subjects, (2) to enable a close study of the bone joint details and bone characteristics as a means of analysing the credibility of joint assumptions and of bone protuberances and skin mark relationships, and (3) to enable physical measurements of bone and joint details to be made so as to correct film data to exact simulated limb conditions.
4.31 SKELETON BONE AND JOINT CHARACTERISTICS

The mathematical model for the study of the upper extremity limb is based on a structure made up of three rigid bodies, pivoted at the shoulder, elbow and wrist joints. A detailed study of the physiological structure of the limb is contained in Section 2 and a specific study of the bone structure and joint details is discussed here to validate this approach. The procedure used was to examine and photograph a human skeleton. This was done in three parts, namely, (1) overall limb and hammer photographs, (2) location of bone versus skin protuberances, and (3) study of joint details as pivot points.

The fact that the skeleton bone lengths were within 1/32 inch of the subject's bones further enhanced this procedure.

OVERALL LIMB AND HAMMER PHOTOGRAPHS were taken to correspond to a high speed camera photo frame of the subject's arm motion. Superimposing one photo onto the other located the bone structure inside the overall limb outline and fixed the joint "pivot points". This procedure is discussed in Section 4.32 and is illustrated on Drawing LIMBO 7.

LOCATION OF BONE VERSUS SKIN PROTUBERANCES

By examining the human arm, certain underlying bone protuberances are evident on the skin. The actual position of the underlying bone structure of the arm can be accurately located by feeling and identifying these areas with respect to the bone shapes and contours as existing on a human skeleton. Additionally bone and skin reference points are selected to correspond to two points on the same rigid bone.
With this in mind a series of bone and joint photographs were taken as shown in figure 4.311 and figure 4.321.

In figure 4.311 Photos 1 and 2, for example, the shoulder has two distinct bone protuberances—namely (a) point P1 corresponding to the sharp edge of the scapula and (b) a prominent peak point P2 on the upperarm humerus bone. The former acts as a fixed reference position for the shoulder joints and torso, and the latter identifies one end of the humerus bone. To locate the actual humerus position a second point on this bone point P3 on photo (3) and (4) was selected at the elbow end. Because of the hinge nature of the elbow joint, point P3 is also used to locate the true elbow pivot point, since the forearm ulna and radius bones rotate about the humerus.

The two points P2 and P3 on each end of the humerus bone can be quickly located and marked by the skin protuberances they produce. These markings indicate the underlying rigid member position during high speed filming.

Similarly points P4 and P5 on Photo (3) and (4) were used for the forearm ulna bone location. The elbow point P4 is readily visible at all times since the actual end of the ulna bone is covered only by a thin skin and tissue layer. At the wrist end, two skin protuberances P5 and P6 on Photo (5) were marked corresponding to the prominent peaks on the ulna and radius bones.

Due to the complex bone structure of the hand only the knuckles of the hand were used to locate the wrist pivot point of this rigid body.

Identification of at least two points on each rigid member in the upper extremity limb, enabled location
of the actual instantaneous pivot points for the assumed three member rigid body pivoted structure analysis. Furthermore exact location of bones with respect to skin marks can be obtained by X-ray photographs. This was not considered necessary for this analysis in this instance.

The choice of reference points was also influenced by the degree of movement that occurs between the skin marks and the bone protuberances, and the bone joint pivots.

The biggest source of error in the location of bones and joint pivots exists in the wrist and shoulder, where muscle, tendon and tissue movements allow the bones to "float" in the joint sockets. The elbow joint however is much more restricted because the muscle and tendon attachments lie outside the joint and the bones themselves are self retaining during motion.

**JOINT CHARACTERISTICS**

The upper extremity limb is made up of a rigid bone structure as shown in figure 4.321. The assumption that the upper extremity limb is made up of three rigid members pivoted at the shoulder, elbow and wrist joints therefore depends upon the muscle and bone actions and the joint characteristics.

The major motions of the upper extremity system are shown in figure 2.4. The range of movements possible for each joint is discussed in Section 2.6 and the examination of the joint bone contours provides positive evidence of "mechanical stops" being an inherent part of the joint characteristics.

For the degrees of limb movement possible the shoulder joint acts on a ball and socket joint, the elbow
as a simple hinge between the humerus and the ulna bones, while the wrist is a double ball and socket (or double hinge) joint.

The upperarm bone - the humerus - is a single rigid member attached at the shoulder and elbow joints. The forearm ulna bone forms a rigid member connecting the hand bones at the wrist; Moreover a second bone - the radius - locates and stabilizes the hand at the wrist and also has a rolling joint at its elbow end where it rolls on the ulna during forearm rotation. If forearm rotation is restrained during the motion, then the ulna and radius act as one rigid member hinged at the elbow and pivoted at the wrist. Likewise if the bones of the hand are held in a fixed relative position, the hand acts as a single rigid member.
PHOTO 4.321

SKELETON WITH HAMMER IN PREARRANGED POSITION
SIMULATED SKELETON BONE PHOTOGRAPHY

The actual bone and joint pivot details with respect to the known and filmed bone reference skin marks of the subject were obtained by studying and photographing a human skeleton. The skeleton arm was set up and photographed from the same relative position as the high speed film. (Refer Photo 4.11). The standard 35 mm still photograph (refer Photo 4.321) shows the skeleton bone structure and hammer in a prearranged position corresponding to reference step (11) position on Drawing LIMBO 3. The exact relationship between skeleton bone positions and subjects high speed film step (11) position was obtained by superimposing the projected skeleton Photo 4.321 onto the plotted outline of the limb and skin marks of Drawing LIMBO 3 using the special tracing table.

The skin reference marks correspond to known bone protuberances, and alignment of these points correctly locates the bone joint and skin positions.

The resulting Drawing LIMBO 7 shows the true relationship of the bones, joints and skin reference marks with respect to the overall limb outline.
4.4 **DETAILING LIMB BONE DIMENSIONS**

The dynamic analysis model requires data for a rigid, three member, pin jointed structure. Therefore the subject film data displacements require corrections to include for the bone joint assumed pivot point displacements being different to the bone protuberances - skin mark displacements.

The skeleton bones were measured and compared to the subjects actual external bone measurements. An ideal one to one correspondence between the subject and skeleton simplified this procedure - eliminating the need to scale the photographs and measurements.

The locations of the true centres of the rounded bone joints were computed by considering the nature of the joint, bone joint shapes and the degree of rotation of the joints involved in the study motion. The resulting bone joint pivot points appear on Drawing LIMBO 7 as A, B and E corresponding to the shoulder, elbow and wrist joints respectively. Difficulty was experienced in locating the wrist pivot point because of the complex bone structure at the wrist. However by studying the skeleton ulna and radius bones at the wrist joint and by observing and feeling the underlying bone movements of the subjects wrist - the wrist pivot point E was obtained.

The actual distances between pivot points A B, B E, and E H appears as distances SL1, SL2 and SL3 on Drawing LIMBO 7.

The correction of bone protuberances displacement angles to pivot point member displacements are also shown on this Drawing. Typically the upperarm displacement angle correction is three degrees thirty minutes. In
this way the actual observed high speed film data of a rigid boned limb is related to the dynamic analysis model data requirements for a pin jointed rigid frame structure.
4.5 **TABULATION OF SIMULATION MODEL DATA**

The simulation model data is obtained from the observed high speed film displacements summarised in tables on Drawings LIMBO 5 and LIMBO 6. These angles were firstly corrected by an amount equal to the difference between the pivot point angles and the skin mark reference angles. Secondly the instantaneous reference step velocities and accelerations are computed on the basis of reference step displacement angles and time intervals. The reference step velocity is considered to be the change in step displacement divided by the time interval between reference steps, while the acceleration is the change in reference step velocity divided by the time interval between reference steps. (Refer Appendix 9 for details of the incremental step velocity and acceleration analysis).

Drawing LIMBO 8 contains the tabulated method of computation of the simulated data from the observed high speed film displacements. This Drawing is divided across the page into three groups - the upperarm LINK 1, - the forearm LINK 2 and the Hand and Hammer combination LINK 3. Each of these groups is further divided into subgroups of displacements, time intervals, velocities and accelerations. Intermediate subgroups exist to simplify the tabular calculations.

The entire high speed camera study motion is in tabular form with respect to the reference steps 00 through to 31, drawn in rows down the page. Reference frames 0 to 11 are for the downward hammer swing, frames 11 to 13 for the impact phase and frames 13 to 31 for the upward hammer swing.

The tabulated computation depends on data in
the columns entitled "Skin Mark Angle" and "Reference Step time intervals" being supplied from the high speed camera study. The figures contained in columns entitled "Skin Mark Angles" STH 1 (A) say, are obtained directly from the bone protuberance and skin mark angles as tabulated in Drawings LIMBO 5 and LIMBO 6. The figures for the time intervals between reference steps, entitled "Reference Step time intervals", DTIM (A, B) say, are obtained directly from Appendix 5 - Table TIME which is produced from the high speed film timing marks as shown on Drawings LIMBO 3 and LIMBO 4. Sample calculations at the bottom of each column outlines the method of computation for corrected bone angles, velocities and accelerations for each of the reference steps starting at the left hand side column and moving across the page to the right. The sample calculations are for reference step number 29.

The columns entitled "Bone Angle", say TH1 (A), "Angular Velocity", say THV1 (A) and "Angular Accelerations", say THA1 (A) for the three LINKS 1, 2 and 3 contains the dynamic analysis simulation model computer input data for the hammering motion.
4.6 Determination of the Limb Mass and Inertia Data

The rigid member pin jointed frame dynamic analysis relies on instantaneous limb displacements, velocities, and accelerations being known for the motion. Additionally, the limb physical properties of pivot point lengths, centre of gravity positions, masses and radius of gyration of these members are required. The summary of this data is shown on Drawing LIMBO 7; limb lengths are shown as SL1, SL2 and SL3, while the centre of gravity lengths are SLG 1, SLG 2 and SLG 3 and the radii of gyration are designated SKG 1, SKG 2 and SKG 3. The member masses are shown as SM 1, SM 2 and SM 3 acting at the centre of gravity positions.

The calculation of centre of gravity of each link is performed by assuming that each limb has a uniform mass distribution of bone, muscle and tissue in the form of a tapered cylinder of similar dimensions to the limb members.

Detailed calculations of the centre of gravity distances SLG 1, SLG 2 and SLG 3, the masses SM 1, SM 2 and SM 3, and the radius of gyration SKG 1, SKG 2 and SKG 3 are contained in Appendix 4.

The mass of each member is an important property since it is used directly in the dynamic analysis and indirectly involved in the determination of the member radius of gyration.

The mass or weight of the members was obtained by two independent methods in order to reduce the overall error. Ideally, the limb mass could be measured exactly by weighing disjointed human limb members of exact proportions to the subject's limb. However, this was
impractical for this thesis and the following procedure adopted.

Firstly the limb was simply weighed on a spring scale. For this the subject was comfortably seated with the trunk of the body in the upright position. The arm was rested on the spring scales in such a way that the entire forearm and upperarm weight was transferred onto the scales. The other hand was used under the armpit to relax the muscles around the shoulder joint to minimise the weight of the limb supported by the shoulder joint. This method yielded an overall limb weight of 8.5 lbs approximately.

The second method involved the estimation of the overall limb density based on the assumption that the bone, bone marrow, muscle, blood, skin and fat tissues are uniformly distributed according to limb volume.

The volume of a pound of T Bone steak of equivalent bone and flesh content to the limb was measured at 29.1 cu. ins. which yield a density of 59.5 lb/cu.ft. - a surprising result. It was later realised that this bone and meat had dried out considerably before the weight was taken.

A pigs trotter was selected for the next sample since it consisted of a large mass of bone, skin, hair and an equivalent volume of muscle and flesh. In this sample the meat was pumped with "brine" so that it was wet and probably a good equivalent to the limb with "blood" flowing through it. The pigs trotter density of 73.5 lb/cu.ft. was calculated from its weight of 0.875 lbs. and a displacement volume of 21 cu. ins.

The calculation of member masses and inertia
effects is then based on an assumed uniform bulk density of 73.5 lb/cu.ft. and the overall limb member volumes. Limb members are assumed to have a volume equal to that of an equivalent tapered cylinder. Details of the comparison of actual subject measured member volumes and calculated volumes are shown in Appendix 4.

Similarly the moment of inertia 'I' and radius of gyration 'k' calculations are based on a uniform flesh and bone density of 73.5 lb/cu.ft. for a volume of approximate circular pyramidal shape. (Refer Appendix 4 for these calculations). The radius of gyration is important in the dynamic analysis in that it is at this distance or radius from the pivot point that the total inertia force of the limb member mass can be considered to act on an equivalent basis. In reality, the member mass and inertia forces are distributed along the entire member.

The radius of gyration $k = \sqrt{\frac{I}{m}}$ where 'm' is the member mass, or the weight divided by the acceleration due to gravity.

The physical properties of length, mass and radius of gyration are detailed on Drawing LIMBO 7. This data then becomes the fixed variables data from the High Speed Camera Study and is contained in Appendix 2 in that form.
4.7 COMPUTER SIMULATION

The experimental procedures described above supply the necessary computer simulated data in two parts, namely (a) Physical properties of the limb as shown on Drawing LIMBO 7 and (b) Reference step data of the limb motion as detailed on Drawing LIMBO 8. The simulated model computer programme uses these values in the form of paper tape formatted data. Refer Appendix 2 for a source listing of this computer data tape.

The limb dynamics analysis was performed by the computer programme as detailed in the source listing Appendix 1 and as described in Chapter 5. The computed values of limb forces and torques are contained in Appendix 3. A summary and discussion of results of the limb analysis technique is contained in Chapter 6. Results are graphed against limb displacements and a comparison is made between these results and values obtained by other techniques.
4.8 EXPERIMENTAL PROCEDURE SUMMARISED IN STEP FORM

Step 1  Evaluate test procedure for subject filming
Step 2  Carry out high speed filming of simulation task movements
Step 3  Process film
Step 4  View film for noteworthy features
Step 5  Select number of frames per reference step to give the desired accuracy to the analysis
Step 6  Produce downward swing hammer stroke, step by step displacement Drawing LIMBO 3 including progressive film timing marks
Step 7  Produce upward swing hammer stroke, step by step displacement Drawing LIMBO 4 including progressive film timing marks
Step 8  Produce hammer cycle skin reference stick Drawings LIMBO 5 and LIMBO 6 of the step by step analysis including Tables of reference step displacements
Step 9  Photograph Skeleton bones
Step 10  Superimpose Skeleton photos on projected High Speed film to obtain bone positions and joint locations with respect to the skin marks. Drawing LIMBO 7 shows the relationship between the actual bone positions and the skin mark displacement angles
Step 11  Produce tabulated calculation of time intervals between reference steps from the timing marks on Drawings LIMBO 3 and LIMBO 4 (Refer Appendix 5)
Step 12: Establish limb physical properties such as lengths, centre of gravity, centre of percussion and limb mass distribution (Refer Appendix 4 for details of centre of gravity, mass and movement of inertia)

Step 13: Produce tabulated calculations Drawing LIMBO 8 of angular displacements, velocities and accelerations at the reference steps of the camera study. Corrected bone displacements from Drawings LIMBO 5 and LIMBO 6 and timing data from Table Time Appendix 5 supplies the data for these calculations.

Step 14: Detail the physical dimensions of the limb members such as lengths, masses, centre of gravity on Drawing LIMBO 7 (Refer Appendix 4).

Step 15: Summarise the simulation model input data in tabulated form - (a) Physical properties as shown on Drawing LIMBO 7 and (b) Reference step data as calculated on Drawing LIMBO 8.

Step 16: Prepare computer paper tape of simulation model data as in Appendix 2.

Step 17: Compute Dynamic analysis of limb using the computer programme detailed in Chapter 3.5 and Appendix 1.

Step 18: Graph results of dynamic analysis and compare these to results obtained by similar methods of limb analysis.
CHAPTER 5

COMPUTER PROGRAMME
FIGURE 5.1 (CONT.)

A

LOAD & RUN
LINKING LOADER (V3A)
& I/O MONITOR SYSTEM
SOFTWARE

RUN
COMPUTER
PROGRAMME

READ FIXED
DATA PAPER
TAPE

READ STEP
DATA PAPER
TAPE & COMPUTE
DYNAMIC ANALYSIS

TTY

TYPED RESULTS

ITERATE FOR
NEXT STEP

END

B

COMMANDS

FORTRAN
SYSTEM LIBRARY
PAPER TAPES

READ STEP
DATA PAPER
TAPE & COMPUTE
DYNAMIC ANALYSIS

TYPED
OUTPUT
RESULTS

PAPER TAPES

RUN
COMPUTER
PROGRAMME

FORTRAN
SYSTEM LIBRARY

LOAD & RUN
LINKING LOADER (V3A)
& I/O MONITOR SYSTEM
SOFTWARE

READ FIXED
DATA PAPER
TAPE

READ STEP
DATA PAPER
TAPE & COMPUTE
DYNAMIC ANALYSIS

TTY

TYPED RESULTS

ITERATE FOR
NEXT STEP

END
FIGURE 5.1 PDP-9 COMPUTER PROGRAMME OPERATIONAL FLOW DIAGRAM

START

WRITE FORTRAN IV PROGRAMME

LOAD & RUN EDITOR (V3A) SOFTWARE

COMMANDS & TYPING

SOURCE TAPE

8 BIT ASCII

PAPER TAPE

LOAD & RUN FORTRAN IV (V3A) 2 PASS COMPILER SOFTWARE

VARIABLE DATA

8 BIT ASCII

PAPER TAPE

COMMANDS

OBJECT TAPE

MACRO-9

MACHINE LANGUAGE

PAPER TAPE
CHAPTER 5 COMPUTER PROGRAMME

The dynamic analysis of the upper extremity limb was performed by a FORTRAN IV computer programme. The mathematical model used to simulate the dynamics of the arm has been previously detailed in Chapter 3.4.

The computer programme essentially computes the hammer force, wrist, elbow and shoulder velocities, accelerations, forces, and bone torques for each step position, during the hammering motion. The computer input data results from high speed filming of a typical hammering motion of the upper extremity limb. The instantaneous angular displacements, velocities, and accelerations at each step are obtained from tabulated calculations based on observed film data. (Refer Drawing LIMBO 8).

The computer programme calculates forces and torques for each reference step of the movement, and these values are then graphed separately to show the cyclic variation during that motion.

A standard Digital equipment PDP - 9 Computer with Input/output Monitor and 16K Paper Tape oriented system was used for the calculations. The original computer programme as described in Chapter 3.5 used an IBM 1620 computer with FORTRAN oriented (FAP) language.

The procedures involved in operating the PDP - 9 Computer system are shown in block diagram form in figure 5.1.

The computer programme was first typed up as a standard 8 bit "ASCII" source tape using the system EDITOR software. The attached Appendix 1 is a source listing of the FORTRAN IV computer programme used. Throughout this programme appropriate comments have been
READ MEASURED FIXED DATA FOR 3 LINKS - LENGTHS, MASSES C.G. AND RAD. OF GYRATION

READ REFERENCE STEP DATA FOR 3 LINKS - ANGULAR DISPLACEMENTS, VELOCITIES AND ACCELERATION

REPEAT FOR NEXT STEP DATA

COMPUTE STEP LINEAR VELOCITY AND, ACCELERATIONS & ANGLES

COMPUTE STEP SHOULDER, ELBOW & WRIST FORCES, ANGLES & TORQUES

WRITE STEP INPUT DATA & CALCULATED VARIABLES FOR THE 3 LINKS

END

FIGURE 5.2  COMPUTER PROGRAMME FLOW DIAGRAM
inserted to enable identification of the programme operation. The source tape was then compiled into a machine language (object) tape using the FORTRAN IV software compiler. The FDP - 9 Input/output Monitor controlled LINKING LOADER software was used to load the object tape and library subroutines into the computer memory for programme execution.

At execution time, the computer reads in the first part of a separately produced, 8 bit ASCII data tape (refer Appendix 2) and begins to calculate the results for the first reference step data. The input data is in two sections - namely - (1) the physical dimensions, such as bone lengths, masses, and centre of gravity positions which have been assumed constant throughout the entire motion, and (2) the single reference step data of instantaneous angular displacements, velocities and accelerations.

The programme involves the direct computation of results for each single reference step. The programme data for each step is contained on a continuous paper tape, so that after one complete pass through the programme, the computer returns to the start to iterate the calculation for the next step data. (Refer figure 5.2 for the Computer Programme Flow Diagram).

The details of computer results are contained in Chapter 6.
CHAPTER 6

EXPERIMENTAL MODEL RESULTS
6.1 Experimental Results

The technique of using high speed photography and digital computing facilities proved satisfactory in studying the dynamics of the upper extremity limb during motion. Section 4.8 summarises in step form the Experimental Procedures used.

The calculated forces and torques as in Appendix 3 and Section 6.3 are in general agreement with results as recorded by other researchers.

A preliminary analysis of the muscle forces and limb leverage effects as shown on Drawing LIMBO 9 was carried out using the computer analysis forces and torques.

The model experimental procedure consisted of filming and plotting of the upper limb during the motion of hammering. Further filming and plotting of skeleton bone details and measurement of limb physical properties was carried out.

The technique of high speed filming was successful for the analysis undertaken. The experimental procedure as described in Chapter 4 was straightforward and inexpensive. Some difficulty was experienced in obtaining sufficient light to yield the correct film exposure and subject clarity at the required film speed. (Refer Typical Film Segment Photo 4.12). This film showed clearly the bone reference points - the skin marks and the displacement of the nail and hammer impact effects. Limb tissue movements were clearly visible during the predictable times of change in muscle activity, that is, the start of the upward swing when the biceps takes over from the relaxed triceps to raise the arm.
Although the tissue flow effects on inertia are assumed constant in this study, they were clearly visible in operation on the film.

The use of high speed film is restricted by the film emulsion speeds (sensitivity to light) available. However, the use of special film would enable extremely high speeds and therefore small time interval framed films to be taken. The possible use of video recording and display equipment instead of filming may give the better result. Video recording with its contrast adjustments and instant replay displays is less sensitive to light problems than filming. Additionally if details of sections of the limb are required then a second track of video recording close up could be concurrently recorded. Plotting of results would be similar to filming but would be better suited to computer techniques. For the type of overall limb analysis dealt with here the high speed film proved extremely successful.

The results depend entirely on locating the relative position of the underlying rigid bone structure with respect to the overall limb surface. This task required precise and accurate understanding of the bone shapes and characteristics. In particular bone protuberances are externally identified by the characteristic skin contours and they are understood best by skeleton comparisons. The skeleton photo images are superimposed onto the high speed film images and the enlargement is adjusted to get sizes and limb proportions to coincide. Skin Temperature pattern (Thermographic) displays could also be used to locate these bone protuberances since the increase in skin tension in these areas would lower blood flow and therefore change skin temperature.
The procedure of filming (or video recording) the skeleton bones and joint details proved adequate in locating the underlying bone structure of the limb. Additionally, the details of joint types, sizes, sliding surface contours, and degree of movement restrictions were easily understood by this approach. The use of X-ray photography of the subject is another procedure available to exactly locate the rigid bone structure. While this was not considered necessary for this analysis it does offer a means of checking the basic assumptions that the skeleton bone proportions are directly related to the subject's bone dimensions.

It is clear from the results of experiments that the filming, plotting and tabulation of limb data during the study motion is an extremely time-consuming process. Therefore the future of this method is dependent upon simplifying and or speeding up this process.

In terms of the overall analysis the time spent on experimental testing was very short compared to the plotting and tabulation time. The majority of time was taken up in relating the limb mechanics and physiological details in terms of the measured data to the actual joint pivot details.

Fortunately, the filming, plotting and tabulation process lends itself to computer processing. Typically if the high-speed film was displayed on a CRT display and a light pen used to trace out the image characteristics - the actual film outline would be automatically recorded. In this way the entire film could be computer recorded or by video-computer recording techniques this could be accomplished as part of the "filming". The tabulation of results is readily computer programmable as shown in
Appendix (7).

Similarly the calculations of limb parameters such as moments of inertia, radii of gyration, weight distributions and centre of gravity can be performed by the computer. The approximations of uniform and symmetrical distribution of bone, and tissue for these calculations (refer Appendix 4) was sufficient for the present requirements but future development of this model would entail more accurate calculations. The use of synthetic limbs such as composite cork and linoleum discs as used by Pearson, McGinley and Butzel (1963 (26) for actual measurement of these values is noteworthy.

It is interesting to note, that when viewing the high speed film, the hammer actually strikes the nail with two successive impacts in the one downward swing, and that the first impact did all the work in driving in the nail. Therefore it is concluded that the hammer imparts an impact load and energy to the nail and then is itself rebounded by the recoil energy of the impact. This reversal of energy is more efficient than the direct rigid structure impact since the hammer is acting on an impact tool instead of a driving weight. The sudden change in energy is typical of large accelerations as shown in figure 6.33.

Typically a camera analysis at 1100 frames/second by Borland - R.A.F. Institute of Aviation Medicine - on the hand movements of a boxer showed that the fist accelerated over 5.5 inches, at up to Sixty (60) times the force of gravity, reaching a velocity at impact of 30 miles per hour and lasting only forty eight thousandths of a second.
6.2 **Computer Dynamic Analysis**

The basic dynamic analysis model defines a planar motion study of a three bodied, rigid link structure with three fixed and frictionless pivot points.

The model details are described in Section 3.4 and the computer model source listing is contained in Appendix 1. Applying this basic model to the dynamics of the upper extremity limb required a deep appreciation and understanding of the physiological and engineering features of the human limb (Refer Chapter 2).

The analogy between the rigid links of the model and the limb members is acceptable within engineering limits when a number of basic limb details are examined. Such a discussion is contained in Chapter 3.

A number of references have been made to simplifying assumptions for the simulation model - these include Section 3.1 - Simulation Model Theory, Section 3.2 - Selection of Model Simulation Task and Section 3.3 - Simplifying Assumptions. The selection of the hammering experimental task was instrumental in simplifying the simulation model to a planar motion study.

The plane in which motion occurs is described in Section 4.1 and Photo 4.11 and is defined by the plane set at an angle (fifteen degrees) formed by the elbow hinge joint and the elevation of the upperarm at the shoulder joint. Note that this is not the standard engineering orthogonal position as defined in Figure 2.2 nor is it the anatomical reference position as defined in Figures 2.4 and 2.5.

For instance the forearm ulna and radius bones are supinated ninety degrees from the anatomical position which
brings the wrist joints into the plane of the motion. Anatomically there is no medial or lateral rotation of the upperarm, no supination or pronation of the forearm, and no dorsal or volar flexion of the wrist. Therefore the planar motion is defined by the plane passing through the longitudinal axes of the humerus, ulna-radius and hand-hammer combination since all other movements outside this plane are not present.

The anatomical position is defined in terms of the muscles providing the driving force for the bone structure. (Refer Section 2.6). Typically the forearm flexion results from brachialis - then biceps muscle contraction, and extension results from lattissimus dorsi and the triceps muscles. By referring to figure 2.4 it is evident that the upperarm motion in the filming reference place is a combination of flexion and elevation, extension and depression.

The humerus-ulna elbow joint is capable of rotation (flexion and extension only) in the manner of a simple hinge. (Refer Section 2.11). However the anatomical centre lines of arm and forearm are not coaxial as shown in figure 2.112. Across the full range of hinge movement the elbow axis is ten degrees off the orthogonal frontal plane and only one hundred and seventy degrees cubital angle instead of one hundred and eighty degrees for bones in line. From this it is evident that the measured fifteen degrees tilt filming reference plane makes a compromise of the above. Since the upperarm planar arc movement is only eighteen degrees and the forearm elbow arc movement is only seventy two degrees then this assumption is justified.
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Moreover should the analysis require greater accuracy, then the relative plane reference angles could be so adjusted in the step by step computer data to compensate for the changes in axes rotation and joint surface contour across the range of movement involved. It should be noted that no attempt has been made to correct for the fifteen degrees tilt of the planar motion forces. Experimental data is obtained at right angles to this tilted plane but the simulation model analysis assumes this plane is vertical with respect to gravity forces. Correction to the acceleration due to gravity constant can correct for this effect.

The simplifying assumptions made have been incorporated to enable the human limb to be treated as a rigid member structure. Therefore, deformation of soft tissue, blood displacements, movements of muscle tissue and joint membrane effects on mass distribution, member symmetry and inertia forces have been neglected. Typically inertia forces for symmetrically formed members are approximately fourteen per cent of the overall member forces acting (refer Section 6.5 and LIMBO 9) and therefore the change in these inertia forces due to the above mentioned effect will be only a fraction of that amount — estimated two per cent.

Joint friction and viscous effects are considered negligible within the accuracy of this analysis (refer Wright and Johns (1960) (27)).

Limb member velocities and accelerations were obtained by finite difference analysis of the step by step filmed data. (Refer Appendix 9). A computer programme for calculation of data as in tables LIMBO 8 is contained in Appendix 7. These and other calculated physical
parameters such as centre of gravity and radius of gyration were used in the simulation model to calculate the joint forces and member torques at each reference step throughout the motion.

The output of the computer is in tabular form (refer Appendix 3). Computer programming effort (refer Chapter 5) was minimal since it involved only straightforward calculations as detailed in Section 3.4.

The overall estimated error of approximately ten per cent is expected from this first order analysis. The majority of this error is attributed to the finite difference method of analysis — specifically the film definition and clarity of subject reference points from one reference frame to the next with respect to an overall torso reference. Where film data values are squared or multiplied then subsequently the errors are increased. Typically if velocity and distance errors are two per cent, then acceleration and related forces are five per cent and torques at approximately ten per cent. Because of the small number of reference frames per step in the region of the hammer impact points the estimated error in these values could be as high as fifty per cent.

The second major factor in errors relates to the joint and tissue frictional and viscous effects on inertia and mass distribution. The overall limb movement observed data automatically compensates for these effects.

The computer analysis results were plotted initially in terms of the reference step numbers. Some specific comparisons were made as a means of relating these results to other recorded works. Reference should be made to the following Section 6.3 for details.
6.3 **Dynamic Analysis Graphed Results**

The computer analysis of the experimental hammering planar motion produced calculated velocities, accelerations, forces and torques in a tabular form as in Appendix 3.

Initially these results were graphed and they compared favourably to similar experimental results by Pearson, McGinley and Butzel (1963) (26).

Secondly, the analysis results were compared to measurements of human limb dynamic characteristics as described in Section 2.33 - Some Typical Muscle Forces, and Section 2.34 - Cineplasty and Muscle Mechanics.

It is evident from the latter comparison that the overall limb dynamics are a result of muscular effort, position control and bone mechanics, such as leverage and joint frictional effects. For this reason it is not surprising that conclusive relationships are not evident between the overall limb dynamics and the specific muscle activity. However where specific results are available they are compared to the analysis graphed results obtained.

A separate comparison of muscle and bone lever effectiveness and muscle dynamics is contained in Section 6.5 entitled Typical Tabulation of Limb Bone and Muscle Details. In Drawing LIMBO 8 the biceps muscle dynamic analysis is shown as an example of the future requirements for this type of study.

The following discussion of graphed results has been prepared from the computer analysis print out as shown in Appendix 3.
ANGULAR DISPLACEMENT (RADIANS) REFERRED TO VERTICAL AXIS

FIGURE 6.31 REFERENCE FRAME NUMBER VERSUS ANGULAR DISPLACEMENT
Figure 6.31  Reference Frame Number versus Angular Displacement

Here the angular displacement (corrected) computer data (refer Drawing LIMBO 8 and Appendix 2) is plotted against reference frame number in X - Y coordinates. The appropriate scales and units are shown on the two axes. The curves are relatively smooth, indicating that the experimental filming procedures were of a suitable accuracy. The cyclic nature of the motion is obvious.

The upperarm angular displacement remains constant from just prior to impact until just after the second impact has occurred. Therefore the upperarm is stationary during the impact phase of the motion and the forearm, hand and hammer members transfer their energy to the nail on impact. Additionally, the movement of the upperarm during the downward motion also aids the forearm and hand-hammer acceleration by the lever effect it creates. (Refer figure 6.33 for details of these accelerations).

The second nail impact probably results from the recoil energy of the first impact and the delayed response of the brain and limb control system in detecting the first impact. The duration of the first impact phase is 0.0029 seconds and the time from impact to the start of the upward stroke is 0.0205 seconds.

This response takes into account the control-muscle deadband and hysteresis at changeover from relaxed to contracted muscle activity. Therefore, it is not surprising that the upperarm which is at rest during impact begins to raise the arm while the inertia of hammer, hand and forearm combines to bend the wrist downwards under the effect of this vertical loading.
FIGURE 6.32  REFERENCE FRAME NUMBER VERSUS ANGULAR VELOCITY
Figure 6.33  Reference Frame Number Versus Angular Acceleration
FIGURE 6.34  PEARSON'S (26) POSITION VERSUS ANGULAR VELOCITY AND ANGULAR ACCELERATION
Here the angular velocities and accelerations of the upperarm, forearm and hand-hammer combination are separately plotted against the reference frame number in X - Y coordinates. The changing pattern of these curves in general indicates the presence of controlled muscular activity during the motion. Angular accelerations and velocities as plotted in figures 6.32 and 6.33 are calculated in tabular form in Drawing LIMBO 8 and listed as computer input in Appendix 2. This data is used directly for the calculation of forces and torques.

Figures 6.32 and 6.33 compare favourably with results by Pearson (1963) (26) as plotted in figure 6.34 even though the magnitudes of values in this study are much smaller and are for upward motion only. It is noted that the positions marked on figure 6.34 correspond to a forearm flexion of 136 degrees and do not correspond to the reference frame numbers of this study which is for 43 degrees elbow flexion.

These observations suggest that the actual process of hammering is not as strenuous an activity as the arm motion studied in figure 6.34. It also suggests that the peculiar acceleration-deceleration patterns result from the limb control action rather than the limb muscle bone leverage effects. (Refer also figure 6.39 for the effects of acceleration on member torques). This also supports the view that the sudden changes in accelerations
FIGURE 6.35  REFERENCE FRAME NUMBER VERSUS JOINT FORCES
FIGURE 6.36  REFERENCE FRAME NUMBER VERSUS JOINT TORQUES
Figure 6.37 Pearson's (26) Position versus joint forces and joint torques.
FIGURE 6.38  ANGULAR ACCELERATION VERSUS JOINT TORQUES
result from the relatively low (percentage stimulation) muscle activity required for this motion as compared to the large isometric muscle forces they are capable of producing. Additionally, the limb sensory feedback and brain control activity in guiding the hammer into impact with the nail would modulate muscle activity (and therefore accelerations) to achieve the desired motion.

Figure 6.35  Reference Frame Number versus Joint Forces
Figure 6.36  Reference Frame Number versus Joint Torques

The variations in forces and torques at the shoulder, elbow and wrist joints are shown in figures 6.35 and 6.36.

The irregular force and torque curve patterns indicate the presence of inertia force changes resulting from increased and decreased muscle activity (applied moments) and the associated acceleration changes they produce. The applied moment or muscle activity is directly related to the muscle developed force times the moment arm distance, and this explains the apparent straight line relationship between Accelerations and Torques as shown in figure 6.38. However the pattern is indicative of the flexion and extension type of limb movement and seems independent of the range of that movement. For example, the same dynamic pattern exists in figure 6.37 by Pearson (1963) (26) even though the elbow angle of movement is larger. This suggests that the muscle dynamics and in particular the observed acceleration-deceleration pattern is a function of the muscle control activity of the limb and not the range of its movement. Further, if differences in acceleration and velocity patterns are considered in respect to the torque
**Figure 6.39**  Elbow Angle Versus Forearm Torque
Figure 6.310  Wrist Velocity Versus Elbow Torque
and force curves, then the results compare very favourably.

**Figure 6.39   Elbow Angle versus Forearm Torque**

When this torque curve is compared to the acceleration curve in figure 6.33 it is evident that the change in torque is related directly to the rate of change of acceleration. The initial period of clear muscle activity corresponding to Reference Frames 17 to 20 is shown as an acceleration phase - but this is immediately followed by a deceleration phase of either reduced or "coasting" muscle activity. If the deceleration portions are removed and an allowance made for muscle activity (percentage stimulation), then a similarity exists between these results and the expected parabolic forearm torque versus elbow angle curve as shown in figure 2.338.

Even though the magnitude of forces and torques are small, the muscle-bone leverage effects or Sine of angle of pull effects as shown in figure 2.338 are evident in the forearm torque curves of figure 6.39. However it is also possible that this result is only a coincidence and the torque falls off with reduced elbow angle simply because the limb muscle activity is required initially to start flexion and as the inertia increases the muscle activity falls off. A repeat of this forearm flexion torque dynamic analysis study for a motion of operating an adjustable resistance ratchet lever would be one way of verifying these effects and relating the muscle stimulation activity to torque magnitude.

**Figure 6.310  Wrist Velocity versus Elbow Torque**

A characteristic exponential relationship is known to exist between force and velocity applied at the
hand for the muscle group causing elbow flexion. This is shown in figures 2.3411 and 2.3414.

Figure 6.310 is a curve drawn for wrist velocity versus elbow torque for the hammering motion. While the slope of the line is correct for the initial elbow flexion Reference Frames 17-21, no conclusive result could be drawn from this comparison because of the relatively small variations in muscle force.

Finally it should be noted that calculations of joint forces were done on an overall basis and joint frictional forces and viscous drag forces were therefore included. However when individual muscle forces are being studied such as in Section 6.5 then these frictional forces must be added to the observed data forces since these additional internal limb forces must be overcome in producing the observed motion.
6.4 DISCUSSION OF ANALYSIS FUTURE REFINEMENTS

The analysis as described in Section 3.4 entitled MODEL DETAILS is a basic dynamic model designed to obtain an overall appreciation of the system using the minimum simplifying assumptions. Additionally, this basic model allows for future refinements as discussed in Section 3.8. Since this stage of development however, there has been a rapid change in computer technology, which if fully utilized will enhance and support this type of kinematic analysis. Therefore this section describes the areas of improvement of this analysis and the future use of computer and electronic technology related to this type of study.

It is evident from this thesis that the step up from a simple planar motion to a three dimensional study is of secondary importance to the inclusion of muscle-bone dynamics during the motion. A natural progression of six stages of development is put forward as a means of obtaining a suitable three dimensional limb model. This approach allows successive models to be compared for accuracy, advantages, disadvantages, ease of analysis, degree of effort, time and cost involved, limitation and many other important factors.

The development is as follows:

Stage 1 - Basic Planar three rigid member frictionless pin jointed analysis - refer Section 3.4.

Stage 2 - Planar Three member rigid bone structure with instantaneous pin jointed analysis - refer Section 4.5.

Stage 3 - As in Stage 2 but including muscle group activity and bone joint dynamics - refer Section 6.5.

Stage 4 - As in Stage 3 but including measured muscle Electro Myographic (EMG) signals versus muscle group activity - refer Section 2.61 for EMG effects.
Stage 5 - Repeat of Stage 4 for Maximum muscle effort conditions to allow comparison with exoskeletal measured results and characteristics as in Section 2.3.

Stage 6 - Three dimensional limb study using observed data (filming) and EMG measurements for a full dynamic analysis.

A suitable approach is to select a motion conducive to all six models and to carry out a complete set of model studies. In this way the peculiarities of each model, its ease of analysis, its related accuracy and overall ability to model the dynamic characteristics can be evaluated.

The first three models are the subject of this thesis and the remaining models have been put forward as a natural extension of this work following the assessment of the results obtained in Section 6.3 and the physiological and experimental work as described in Sections 2.33 and 2.34.

Listed below for reference purposes are a number of suggested refinements, and discussion of current technological advances which relate to this thesis.

6.41 NATURE OF MODEL REFINEMENTS

In future studies, experimental correlation of engineering data could be performed directly by a computer and it is envisaged that future studies of limb dynamics could be restricted to external observations of limb movements. The assumption being that the movements of the limb members are the direct result of specific muscle, tendon, joint bone and skin movements. The magnitude of the limb dynamics will be evidenced both in their velocity and acceleration patterns, and the muscle EMG or muscle percentage stimulation patterns with respect to limb position and leverage effects.

Techniques such as measurements of amputated sections of the limb and experimental model studies of tissues are
required to obtain both overall and specific engineering data of the limb tissues and mechanics. In particular the joint properties and mechanics are fundamental of this model study. A great deal depends on the future ability to incorporate such variables as joint shape, size, types of lubrication, frictional effects, motion types, sliding surface contours, and other bone and tissue constraints in the analysis.

Moreover the effect of muscle tendon attachments and sheathed tendons bone provisions on the model dynamics is of utmost importance.

It is evident from the mechanics of the muscle-bone structures of the limbs that certain muscles or muscle groups provide the driving force for the motion. (Refer Section 6.5). It is also apparent from Sections 6.5 and 2.61 that muscles operate in antagonist/agonist pairs which are responsible for muscle tone control, characteristic muscle EMG signals and muscle to muscle EMG crosstalk. The effect of muscle velocity of shortening, and muscle force as shown in Section 2.34 for a defined motion is evidenced by the fixed pattern of signals involved. Hence either computer pattern recognition of the EMG signals with limb displacements or conversely by knowing the motion and monitoring the EMG patterns, the level of muscle activity can be calculated as described by Lawrence and Lin (1972) (23).

A much more difficult approach involves the monitoring of muscle motor (power demand) and kinesthetic sensory signals (force and position measurement) which operate together and involve an overriding visual and motivating control action. Measurement of nerve neuro and muscle electromyographic signals, in conjunction with external photography or video recording of particular limb motions is required to obtain the pattern
relationships of muscle and limb dynamics characteristics.

The internal movement of bone and muscles can be externally assessed and measured in terms of the skin contour changes that result. With the additional tabulated knowledge of observed muscle activity and EMG signal patterns required to produce these movements, combined with the skin surface contours changes that occur during the motion, the exact duration and muscle dynamic activity, movement and leverage effects can be obtained. Additionally the environmental temperature effects on motion with particular reference to skin shrinkage, viscosity changes, blood flow, frictional drag forces and body temperature effects on muscle power efficiency is required.

6.42 ENGINEERING DATA

It is evident from this study that the dynamic characteristics and properties of limb tissues are not adequately available and it is therefore necessary to tabulate and record the following types of engineering data for future reference.

(a) Overall limb outlines, shapes, sectional areas, surface contours, limb sizes and proportionate genetic size relationship.

(b) Bone locations and externally identifiable skin-bone protuberances.

(c) Muscle-bone relationships including muscle attachment positions, leverage effects and muscle-bone efficiency factors, muscle tissue distributions and muscles used to produce specific movements.

(d) Muscle tissue sizes, shape, volumes, weight distribution, density and other similar characteristics including muscle dynamics and force factors, both overall and distributed, muscle free and rest length positions.
during motions, relationship of muscle body volumes and lengths to tendon lengths.

(e) Muscle Electromyographic (EMG) signals versus muscle movement, velocity of shortening, muscle body size and percentage stimulation (neuro signals), and muscle control and tone effects. This includes tabulation of limb movements versus EMG patterns for overall muscle groups and individual muscles.

(f) Properties such as densities, elasticity, viscosity, and friction factors for the various tissue types such as bones, bone marrow, muscles overall, muscle bundles, tendons, blood, blood vessels, skin surface, skin sublayers, fatty tissues, joint membranes and synovial fluid properties.

(g) Moments of Inertia forces of tissues during motion, particularly the non rigid body effects and the component moment of inertias, and instantaneous Radius of Gyrations incorporating modelling of viscous effects.

(h) Frictional and viscous drag effects of muscles moving within the limb, flesh flow, skin distortions, blood vessel flow and joint tissue forces for the complete range of movements.

6.43 SUMMARY

It is indeed possible to set up an on-line computer system fitted with suitable peripheral equipment to automatically record and graph such data as video recorded movement and limb physical data, muscle EMG signals and step by step movement velocity and acceleration patterns. After pattern recognition, pre-programmed movement control is then possible. Additionally the complex changes in limb outline and the possible
pre-programming of muscle leverage effects, muscle volume changes and tissue displacements for specifically positioned limbs involves tabulation of muscles acting to produce these motions in terms of position dependent data. This would enable on-line calculation of radius of gyration, centre of gravity and dynamic inertia effects, and muscle rest length position.

It is also an accepted fact that bone dimensional relationships exist (refer Figures 1.2 and 1.3) such that by measurement of one externally accessible bone size, the full set of bone dimensions can be calculated for different genetic group sizings.

In this way future dynamic studies are envisaged to entail only externally recorded video-computer analysis techniques.
Dynamic analysis of the human body depends almost entirely upon the understanding of the predominant physiological mechanisms involved. Typically the human body consists of (a) a rigid skeletal bone structure, (b) skeletal muscles to produce bone movements, (c) body control organs such as the brain and spinal cord, motor and sensory nerves, (d) outer skin covering and (e) special purpose organs such as the heart, lungs, kidneys, intestines and many others. Besides understanding the terminology and functional characteristics of the various parts of the body such as joint lubrication, mechanisms of muscle contraction and muscle activity, it is necessary that this type of information be tabulated and documented in a suitable engineering fashion.

Chapter 2 and Drawing LIMBO 9 documents engineering and biomechanical aspects of the upper extremity limb locomotion. It should be noted that the collection of this information was extensive and that a wealth of information has been documented in the Aerospace and Defense research industries, but this is not freely available.

One large obstacle in modelling the dynamics of the body is its inherent flexibility in overcoming defective and damaged tissue effects. Therefore the tabulation of specific muscle effort relating to a specific motion should be done on the basis that muscles and movements are normal. This is particularly important because limb motion results from muscle group rather than single muscle activity and except during maximum effort conditions, the role of a defective muscle is usually taken over by another muscle in the group without any appreciable change in the resultant motion. Similarly, muscle tone is important, since a desk clerk using a hammer would be less representative than a carpenter, but more
representive in the respect that the majority of people are amateur carpenters. Therefore a dynamic analysis is best performed on a subject to subject basis, with the subject chosen to represent a specific type of group. With the above considerations in mind, the muscle dynamic analysis as shown on Drawing LIMBO 9 and Appendix 10 was carried out for the hammering motion forearm flexion.

Firstly this involved defining the motion reference position, defining the muscles and muscle group activity to produce the motion, and detailing the upper extremity limb skeletal bone and muscles in relation to the plane of the filmed hammering motion. Cross sections of the limb members serve to illustrate the muscle bone distributions and validates the assumptions that the overall limb mass distribution is reasonably symmetrical and volume dependent.

Secondly the recorded bioengineering studies of Sections 2.33 and 2.34 relating to muscle and limb dynamics are shown in relation to the limb members involved. Furthermore the application of muscle modelling theories as described in Section 2.35 becomes important in defining the observed limb and muscle system control characteristics.

Thirdly, the relationship between limb movement, muscle activity (single and group), muscle-bone attachment details, leverage effects and joint positions are shown on Drawing LIMBO 9. The actual hammering motion upperarm range of movement is only eighteen degrees and this validates the assumptions relating to negligible muscle leverage effects and insignificant changes in joint instantaneous centres as used in Chapter 3. The percentage contribution of muscles and muscle groups producing a motion and the muscle bone leverage and joint frictional effects are required to complete
this understanding. It should also be noted that while certain muscles are active in providing the driving force for a motion, other muscles are held in "tone" to position the limb and support the joint actions. While these muscles in "tone" do not actively produce motion they do however produce the necessary antagonist forces which are necessary to allow control over the motion to be obtained.

Fourthly a typical stage 3 muscle bone dynamic analysis is shown for a period corresponding to brachialis and biceps muscle activity about the elbow joint for reference step No. 20 on Drawing LIMBO 6 or Reference Frame No. 24 in Appendix 3. Details of the dynamic analysis is contained also in Appendix 10. Consider this upward stroke of the hammer under predominant biceps control where the wrist is ulnar flexed under the inertia forces of the hammer. The forearm, hand and hammer combination behaves almost as a single rigid body during this period. In fact the film study data shows that the wrist angle does not alter significantly until a flexion elbow angle of 86.4 degrees is reached. The forearm flexion also starts to accelerate the limb upwards at this point indicating a coordinated effort between the biceps and the forearm flexors muscles in raising the limb.

At Reference Frame No. 24 the elbow angle is 86.4 degrees and the forearm at the wrist is moving upwards at a velocity of 2.86 ft/sec. and an acceleration of 14.51 ft/sec². The simplified limb muscle system dynamic details are shown on Drawing LIMBO 9 Detail A. In this case the muscle shortening is almost directly related to the forearm velocity since the muscle attachment is pulling at approximately right angles to the forearm ulnar bone. The muscle force of 27 lbs. is required to produce the forearm torque at this
Figure 6.51 Force-velocity curve for biceps muscle during forearm flexion.
instant. This force is calculated directly from the dynamic analysis forearm torque and thus the actual muscle force is larger than this value because the muscle viscous drag and joint frictional effects require an additional muscle force before the observed motion forces are produced. It is evident from the above that a more thorough knowledge of the limb leverage systems, the muscle bone insertion dimensions and characteristics, and the effective muscle moment arm distances is needed to complete this study. Actual muscle lengths and volume changes are required because some muscles are long and contract over long distances, while others are short with larger cross sectional areas. The study and preliminary tabulation of muscles, leverage effects and their associated movements is called kinesiology and is partly described in Section 2.3.

The biceps muscle dynamics are documented throughout Sections 2.33, 2.34 and 2.35. In particular the Force-Velocity curves as in Figures 2.3414 and 2.352 relate directly to the biceps dynamic analysis as on Drawing LIMBO 9 and Appendix 10. The action of the biceps is to apply a flexion torque to the forearm which is proportional to the applied muscle force and velocity of shortening (Refer figure 6.51). It is also evident from Figure 6.52 that the Force-Length curve for the biceps is approximately linear during this acceleration phase and is similar to that shown on Figure 2.346 for isometric contraction. Moreover the forearm flexion joint torque versus elbow angle and bone leverage effects which are known to be parabolic (refer Figure 2.338) are not directly evident in these results. However it should be noted that during the hammering motion the muscle forces are not at maximum effort but are only sufficient to perform the hammering task and to provide the
necessary guidance control to the limb. For this reason the computer results in Section 6.3 and Drawing LIMBO 9 differ greatly from the exoskeletal results of Section 2.3.

A close similarity exists between the Force-Velocity diagram for the Biceps as in Figures 2.352 and 6.51. The analysis by Vickers (1968) (51) was for two identical muscles acting on a simple limb member and allowed for the equation relating the muscle force F to the velocity of shortening V. The equation $F = (Z)F_0(1-(1+1/n)V/V_0)$ also allows for the percentage muscle stimulation Z. Applying this equation to Figure 6.51 for the biceps/triceps muscle pair, the equation gives $V_0 = 1\text{ft/sec.}$ and $F = 380 \text{ lbs.}$ force yielding $Z = 0.2$ by comparison with Figure 2.352.

Similar agreement exists with work by Harrison (1963) (50) in Figure 2.351 where the elbow flexion contraction time of 0.3 seconds is compared to 0.4 seconds recorded in this study. Additionally a recorded elbow flexion muscle group time constant of 0.03 seconds compares favourably with the selected film time interval between reference steps of 0.022 seconds. Considering that there are forty film frames per reference step, then the time rate of change of motion is adequately covered.

Finally, Drawing LIMBO 9 shows the dynamic analysis results for the initial hammer-nail impact. It is of note that the upperarm is stationary at the time of impact and the forearm continues to move down after impact while the hammer rebounds. Furthermore, the weight of the hammer-hand combination and forearm are similar and tends to suggest that the selection of a hammer for good balance and grip also includes an inbuilt ability to balance the inertia loads at the wrist.
6.6  **Summary of Results**

The dynamic analysis results for the hammering motion show clearly that the muscle activity involved is relatively similar to the isometric forces they are capable of producing (refer Sections 2.33, 2.34, 6.3 and 6.5). Moreover, the muscle activity required for limb control, coordination, timing and movement is so small, that during periods of the motion, limb inertia is responsible for the continuation of the motion. This on-off muscle activity is responsible for the peculiar acceleration-deceleration patterns which appear in Figures 6.33 and 6.39. For this reason alone, no conclusive relationship can be established between muscle forces, joint torques and muscle bone leverage effects for reasons outlined in Section 6.5.

The exact nature of muscle activity is best described in terms of the muscle electromyographic signals or neuromuscular action potentials. A suitable neuromuscular model as detailed in Section 2.35 is already available to allow the present model to be extended to a stage four model as described in Section 6.4.

In an overall sense the graphed results compare favourably with actual subject external force measurements obtained by Miller (1942) (38) and Klopsteg and Wilson (1954) (33). Likewise the results of Pearson, McGinley and Butzel (1963) (26) exhibit similar characteristics to those contained in Section 6.3.
CHAPTER 7

CONCLUSIONS
Chapter 7  CONCLUSIONS

A practical engineering approach to the measurement of the dynamic properties of the upper extremity human limb is presented in this thesis. This involved the successful use of a high speed camera data input procedure and a computer analysis simulation model.

The dynamic analysis model consists of a planar rigid linked, three member system pivoted at the shoulder joint. The computer programme accepts the subject's step by step time lapsed data for the motion involved and computes the magnitudes and directions of torques and reaction forces at the wrist, elbow and shoulder joints. The plotting of displacements, angular velocities, accelerations, forces and torques for each member of the limb, describes the motion.

The model analysis procedure is a comparatively fast method of obtaining dynamic patterns for motions of the human limbs. This model is easily adapted to act as a predictive motion simulator. This model may also be used on any real process equipment such as saws, looms, controllers and actuators, reciprocating the fast moving equipment, and the like, and where the evaluation of inertia and rigid member forces are required.

Although it is elementary in its present form, sufficient research has been recorded in Chapters 3 and 4 - Overall dynamic analysis, Section 6.5 - Individual muscle dynamic analysis, Section 2.35 - Limb and muscle system control modelling characteristics, and Section 2.61 Muscle Electric activity, to show that the observed filming technique is capable of supplying data for most future
model requirements.

The main advantage of this procedure over other methods is that it is a simple real time model using the actual observed unencumbered data of distances, velocities, and accelerations. Other methods such as exoskeletal studies involve the attachment to the limb itself of measurement and recording devices, which by their very nature, change or inhibit either physically or psychologically the limb movement characteristics.
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NOMENCLATURE
The physiological details of the upper extremity limb are discussed at length in this thesis. Reference is made to certain summary figures and nomenclature of the relevant parts involved as a means of defining the terminology used.

UPPER EXTREMITY LIMB STANDARD POSITION
NOMENCLATURE

Page 10 Section 2 Details of the Arm.
Terms: Reach, Grip, Grasp, Sagittal, Horizontal, Frontal, Flexion, Extension, Elevation, Depression, Supination, Pronation, Lateral and Medial Rotation, Radial, Ulnar, Dorsal and Volar Flexion, Parasagittal Flexion, and Parasagittal Hyperextension.
Figures: 2.1 to 2.5 Major motions of the Arm.

Page 88 Section 2.6 Movements of the Upper extremity limb.
Terms: Flexion, Extension, Rotation, Abduction, Adduction.
Figures: 2.61, 2.62, 2.63 and 2.64

Page 111 Section 2.7 Control of Limb Movements.
Terms: Forward, Upwards and Sideways, Pitch, Roll and Yaw.
Figure 2.71: Six degrees of freedom of any object in space.
**BONE NOMENCLATURE**

Page 12

Section 2.1  Bone Structure of the arm.
Terms:  Fossa, Facet, Process, Shaft, Neck, Head.
Figure:  2.11:  Bone Terminology.

Page 12

Terms:  Clavicle, Scapula, Humerus, Ulna, Radius, Carpals, Metacarpals, Phalanges.
Figure:  2.117:  Upper extremity limb bone and joint summary.

**JOINT NOMENCLATURE**

Page 24

Terms:  Synovial Joints, Articular Cartilage, Synovial Membrane and Fluid, Ligaments.
Figure:  2.121:  Schematic diagram of a human joint.

**MUSCLE NOMENCLATURE**

Page 28

Section 2.2  Skeletal Muscles and motion produced on the arm.
Terms:  Flexors, Biceps, Pectoral, Extensors, Triceps, Deltoid, Trapezius and Latissimus Dorsi.
Figures:  2.21 and 2.22  Skeletal muscles of the upper extremity limb.

Page 30

Section 2.3  Details of Muscles.
Terms:  Origin, Tendons, Belly, Insertion, Muscle Spindle and Fibrils.
Figures:  2.311 and 2.312  Muscle and Bone arrangements.
NERVE NOMENCLATURE

Section 2.4 Nerve Structure of the arm.

Brain, Spinal Column, Central Canal,
Cervical Enlargement, Thoracic Section, Lumbar
Enlargement, White and Grey Matter, Posterior and
Anterior Horns. Posterior Sensory Fibres,
Cerebro-spinal nerves.

Figure: 2.41 Spinal Cord and Cerebro-spinal nerves.

Section 2.4 Nerve Structure of the arm.

Thought - Each instant of awareness could be
defined as a thought. The
comprehension of a visual scene at a
given instant, would be a single
thought.

A thought probably results from the
momentary "pattern" of stimulation
of many different parts of the
nervous system at the same time,
probably involving most importantly
the cerebral cortex, the thalamus,
the rhinencephalon, and the upper
vehicular formation of the brain
stem.

Consciousness - The awareness of a thought can be
defined as consciousness.

Memory - This is the capability of the nervous
system to store memories.

Will - Control exercised over impulse.

Reason - The intellectual faculty by which
things are judged. Reason is to
form or try to reach conclusion by
corrected thought.
SKIN NOMENCLATURE

Page 84  Section 2.5  Skin Structure of the arm.
Terms:  Horny and Living Layers, Epidermis, Dermis, Subcutaneous Tissue and Carpuscles.
Figure: 2.51  Typical skin details.
LIST OF DRAWINGS

LIMBO 3  -  PHOTO STUDY 2 POSITION PLOT - HAMMER DOWNWARD STROKE
LIMBO 4  -  PHOTO STUDY 2 POSITION PLOT - HAMMER UPWARD STROKE
LIMBO 5  -  STICK DIAGRAM FOR DOWNWARD STROKE BASED ON SKIN REFERENCE MARK LOCATIONS
LIMBO 6  -  STICK DIAGRAM FOR UPWARD STROKE BASED ON SKIN REFERENCE MARK LOCATIONS
LIMBO 7  -  SKELETAL BONE AND JOINT LOCATION DETAILS, LIMB DIMENSIONS AND RIGID MEMBER PROPERTIES
LIMBO 8  -  TABULATED CALCULATION OF UPPER EXTREMITY LIMB HAMMER STUDY - COMPUTER DATA
LIMBO 9  -  UPPER EXTREMITY LIMB BONE, LEVERAGE AND MUSCLE ACTIVITY FOR HAMMER MOTION
APPENDIX 1

SOURCE LISTING OF COMPUTER PROGRAMME "LIMBO"
START

TITLE LIMBO
DIGITAL EQUIPMENT PDP-9 COMPUTER (16K)
STANDARD DEC. FORTRAN V5A

COMPUTER PROGRAMME DATA IS FORMATTED PAPER TAPE

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DYNAMIC ANALYSIS OF UPPER EXTREMITY LIMB

PREPARED BY* N.T. HODKINSON BSc. MECH. ENG.
UNIVERSITY OF N.S.W.

DATE * 30/9/69

THIS PROGRAMME CALCULATES ARM POSITIONS, VELOCITIES
ACCELERATIONS, TORQUES, FORCES AND ASSOCIATED ANGLES
FOR THE SHOULDER, ELBOW, WRIST AND HAMMER FACE POSITIONS
DURING ANY TWO DIMENSIONAL PLANE MOTIONS

SPECIFICALLY FOR A HAMMER STUDY, INPUT DATA HAS
BEEN PREPARED USING A HIGH SPEED CAMERA STUDY OF
THE UPPER EXTREMITY LIMB DURING THE MOTION OF HAMMERING
A 2" NAIL INTO A SAMPLE 3" X 2" OREGAN TIMBER

LIMB LINK LENGTHS, MASSES, CENTER OF GRAVITY POSITIONS
RADIUS OF GYRATIONS AND ACCELERATION DUE TO GRAVITY
ARE HELD CONSTANT FOR THE ENTIRE STUDY

ANGULAR POSITIONS, VELOCITIES AND ACCELERATIONS ARE
COMPUTED ELSEWHERE AND IS IN TABULATED FORM THE REFERENCE
FRAME PAPER TAPE INPUT DATA
A1/3

READ(3,10) SL1,SL2,SL3,SM1,SM2,SM3,SKG1,SKG2,SKG3
READ(3,11) SLG1,SLG2,SLG3,W,TH4,G

N = N+1

IF(N-2)101,102, IU2

WRITE(1,70)
FORMAT(2H ///)
WRITE(1,71)
FORMAT(62H HIGH SPEED CAMERA STUDY OF UPPER EXI

2 REM ITY LIMB///)
WRITE(1,72)N
FORMAT(36H DYNAMIC ANALYSIS AT REFERENCE FRAME ,13//)
WRITE(1,91)
FORMAT(32H HAMMER STUDY CONSTANT VARIABLES)
WRITE(1,81)
FORMAT(66H Lengths)
2 RADIUS OF GYRATION)
WRITE(1,61)
WRITE(1,62)
WRITE(1,63)
FORM 56H FEET 2 FEET )
WRITE(1,64)
WRITE(1,65)
FORMAT(71H SL1 SL2 SL3 SM1 SM2 SM3 2KG1 SKG2 SKG3 )
WRITE(1,10)SL1,SL2,SL3,SM1,SM2,SM3,SKG1,SKG2,SKG3
WRITE(1,75)
WRITE(1,76)
WRITE(1,73)
WRITE(1,74)
WRITE(1,77)
FORMAT(59H CENTER OF GRAVITY)
2 GRAVITY)
WRITE(1,80)
FORMAT(62H FEET 2 FEET )
WRITE(1,81)
WRITE(1,82)
FORMAT(47H SLG1 SLG2 SLG3 W,TH4,G)
WRITE(1,13)SLG1,SLG2,SLG3,W,TH4,G
FORMAT(5F8.3,4X,PE.3)
WRITE(1,78)
FORMAT(27H REFERENCE FRAME INPUT DATA)
14 FORMAT(70H 1 UPPERARM 2 FOREARM
2 3 HAND / HAMMER//)
WRITE(1,79)
5 FORMAT(63H POSITION
2 VELOCITY
2 ACCELERATION)
WRITE(1,76)
6 FORMAT(66H RADIANS
2 RAD PER SEC
2 RAD PER SEC SEC)
A1/4

WRITE(1,77)
FORMAT(7H TH1 TH2 TH3 THV1 THV2 THV3 THA1
2 THA2 THA3 )
READ(3,12) TH1,TH2,TH3,THV1,THV2,THV3,THA1,THA2,THA3
10 FORMAT(9F4)
11 FORMAT(6F8.3)
12 FORMAT(6F7.3,3F9.3)
IF(N-2)103,104,104
104 WRITE(1,72) N
WRITE(1,77)
103 WRITE(1,12) TH1,TH2,TH3,THV1,THV2,THV3,THA1,THA2,THA3
WRITE(1,78)
78 FORMAT(2H )

C DYNAMIC CALCULATION FOLLOWS
C
C ARC LENGTH DISPLACEMENTS DISP
DISPL1=(SL1*TH1)/(2.*3.1415)
DISPL2=(SL2*TH2)/(2.*3.1415)
DISPL3=(SL3*TH3)/(2.*3.1415)
VDA=SL1*THV1
VBAV=VDA*Sin(TH1)
VBAH=VBA*Cos(TH1)
DEL1=(TH1+0.5*3.1415)
VDC=SL2*THV2
VDCV=VDC*Sin(TH2)
VDCH=VDC*Cos(TH2)
DEL21=(TH2+0.5*3.1415)
VDAV=(VBAV+VDCV)
VDAH=(VBAH+VDCH)
VDAY=((VDAV*VDAV)+(VDAH*VDAH))
IF(VDAX)32,30,31
30 VDA=0.5
GO TO 32
31 VDA=(VDA**0.5
IF(VDAV X,32,32
1 VDA=VDA
32 PHI=ATAN(VDAV/VDAH)
DEL2=(PHI+0.5*3.1415)
VFE=SL3*THV3
VF=E=VFE*Sin(TH3)
$VF_{EH} = VFE \times \cos(TH3)$

$DEL31 = (TH3 + 0.5 \times 3.1416)$

$VF_{AV} = (VF_{AV} + VF_{EV})$

$VF_{AH} = (VF_{AH} + VF_{EH})$

$VF_{AX} = ((VF_{AV} \times VF_{AH}) + (VF_{AH} \times VF_{AH}))$

IF($VF_{AX}$) $33, 33, 34$

$VFA = 0.0$

GO TO $35$

$VFA = (VF_{AX})^{0.5}$

IF($VF_{AV}$) $2, 35, 35$

$VFA = -VFA$

$PH2 = ATAN(VF_{AV} / VF_{AH})$

$DEL32 = (TH2 + 0.5 \times 3.1416)$

$ABAT = SL1 \times THA1$

$ABATV = ABAT \times \sin(TH1)$

$ABATH = ABAT \times \cos(TH1)$

$ABAN = SL1 \times (THV1 \times THV1)$

$ABANV = ABAN \times \cos(TH1)$

$ABANH = ABAN \times \sin(TH1)$

$ABAV = (ABAVF - ABANV)$

$ABA = (ABAT + ABAN)$

$ABAV = ((ABAV \times ABAV) + (ABAN \times ABAN))$

IF($ABAV$) $36, 36, 37$

$AB = 0.0$

GO TO $38$

$AB = (ABAV) \times 0.5$

IF($ABAV$) $3, 38, 38$

$ABA = -ABA$

$PH3 = ATAN(ABAV / ABAH)$

$RH1 = (PH3 + 0.5 \times 3.1416)$

$AGI = (SLG1 / SL1) \times ABA$

$ADC = SL2 \times THA2$

$ADCV = ADC \times \sin(TH2)$

$ADCH = ADC \times \cos(TH2)$

$ADCN = SL2 \times (THV2 \times THV2)$

$ADCV = ADCN \times \cos(TH2)$

$ADCV = (ADCV - ADCV)$

$ADCN = (ADCN + ADCH)$

$ADCX = ((ADCV \times 4) (ADCV) + (ADCH \times ADCH))$

IF($ADCX$) $40, 40, 41$

$ADC = 0.0$

GO TO $42$

$ADC = (ADCX) \times 0.5$

IF($ADCV$) $4, 42, 42$

$ADC = -ADC$
42  \( \text{PH4} = \text{ATAN} (\text{ADCV/ADCH}) \)
43  \( \text{RH2} = (\text{PH4} + 0.5 * 3.1415) \)
44  \( \text{ADAV} = (\text{ADCV} + \text{ADAV}) \)
45  \( \text{ADAH} = (\text{ADCH} + \text{ADAH}) \)
46  \( \text{ADAX} = ((\text{ADAV} * \text{ADAV}) + (\text{ADAH} * \text{ADAH})) \)
47  \( \text{IF} (\text{ADAX}) 43, 43, 44 \)
48  \( \text{ADA} = 0.6 \)
49  \( \text{GO TO 45} \)
50  \( \text{ADA} = (\text{ADAX}) * * 0.5 \)
51  \( \text{IF} (\text{ADAV}) 5, 45, 45 \)
52  \( \text{ADAH} = -\text{ADAH} \)
53  \( \text{PH5} = \text{ATAN} (\text{ADAV}/\text{ADAH}) \)
54  \( \text{RH3} = (\text{PH5} + 0.5 * 3.1416) \)
55  \( \text{AFET} = \text{SL3} * \text{THA3} \)
56  \( \text{AFETV} = \text{AFET} * \sin (\text{TH3}) \)
57  \( \text{AFETH} = \text{AFET} * \cos (\text{TH3}) \)
58  \( \text{AFEN} = \text{SL2} * (\text{THV3} * \text{THV3}) \)
59  \( \text{AG2} = (\text{SLG2}/\text{SL2}) * \text{ADC} \)
60  \( \text{AFENV} = \text{AFEN} * \cos (\text{TH3}) \)
61  \( \text{AFENH} = \text{AFEN} * \sin (\text{TH3}) \)
62  \( \text{AFEV} = (\text{AFETV} - \text{AFENV}) \)
63  \( \text{AFEX} = (\text{AFEV} * \text{AFEV}) + (\text{AFEN} * \text{AFEN}) \)
64  \( \text{IF} (\text{AFEX}) 45, 45, 45 \)
65  \( \text{AFV} = 0.3 \)
66  \( \text{GO TO 48} \)
67  \( \text{AFV} = (\text{AFEX}) * * 0.5 \)
68  \( \text{IF} (\text{AFEV} 5, 48, 48 \)
69  \( \text{AFZ} = -\text{AFZ} \)
70  \( \text{PHS} = \text{ATAN} (\text{AFEV}/\text{AFEN}) \)
71  \( \text{RH4} = (\text{PHS} + 0.5 * 3.1416) \)
72  \( \text{AFAV} = \text{AFEV} + \text{ADAV} \)
73  \( \text{AFAH} = \text{AFEN} + \text{ADAH} \)
74  \( \text{AFAX} = ((\text{AFAV} * \text{AFAV}) + (\text{AFAH} * \text{AFAH})) \)
75  \( \text{IF} (\text{AFAX}) 110, 110, 111 \)
76  \( \text{AFA} = 0.6 \)
77  \( \text{GO TO 112} \)
78  \( \text{AFA} = (\text{AFAX}) * * 0.5 \)
79  \( \text{IF} (\text{AFAV}) 7, 112, 112 \)
80  \( \text{AFA} = -\text{AFA} \)
81  \( \text{PH7} = \text{ATAN} (\text{AFAV}/\text{AFAH}) \)
82  \( \text{RH5} = (\text{PH7} + 0.5 * 3.1416) \)
83  \( \text{TI1} = \text{SI1} * (\text{SKGL} * \text{SKG1}) * \text{THA1} \)
84  \( \text{FI1} = \text{SI1} * \text{AG1} \)
85  \( \text{XLG1} = \text{SLG1} * \sin (\text{TH1}) \)
86  \( \text{BI1} = ((\text{SKGL} * \text{SKG1}) * \text{THA1}) / \text{AG1} \)
GA1 = (RH1 - IH1)
BG1 = SLG1 * SIN(GA1)
SLQ1 = (BI1 + BG1)
FBCV = (((FI1 * SLQ1) - T1 + (SM1 * G * XLG1)) / SL1)
I2 = SM2 * (SKG2 * SKG2) * THA2
FI2 = SM2 * AG2
XLG2 = SLG2 * SIN(TH2)
BI2 = ((SKG2 * SKG2) * THA2) / AG2
GA2 = (RM2 - IH2)
BG2 = SLG2 * SIN(GA2)
SLQ2 = (BI2 + BG2)
YL2 = SL2 * SIN(FA2)
YLQ2 = (YL2 - SLQ2)
XL2 = SL2 * SIN(TH2)
XLG22 = (XL2 - XLG2)
FBCV = (I2 + (SM2 * G * XLG22) + (FI2 * YLQ2)) / SL2
FBCVH = FBCV * COS(TH1)
FBCVH = FBCV * SIN(TH1)
FCBVH = FCBV * COS(TH2)
FBCV = FBCV * SIN(TH2)
FCBVV = (FCAV + FSCAV)
FCAV = (FCAVH + FBCVH)
FCAV = ((FCAV * FCBV) + (FCB * FCBVH))
IF(FCAV) 113, 113, 114

FCB = 0.8
GO TO 115

FCB = (FCBX) ** 0.5
IF(FCBXV) 8, 115, 115

PH8 = ATAN(FCBXV / FCBXH)
DEL4 = (PH8 + 0.5 * 3.1415)
PH9 = (DEL1 - PH8)
FCH = FCB * COS(PH9)
PH10 = (IH2 - PH8)
FCHH = FCB * COS(PHI2)
YL1 = SL1 * SIN(GA1)
YLQ1 = (YL1 - SLQ1)
XL1 = SL1 * SIN(TH1)
XLQ1 = (XL1 - XLQ1)
FAOV = (((FI1 * YLQ1) + (SM1 * G * XLG11)) / SL1)
T3 = SM3 * (SKG3 * SKG3) * THA3
IF(318) 32, 50, 50

SL3 = SL3 * COS(TH4)
T3 = (T3 - (318 * SL3))

F13 = SM3 * AG3
XLG3 = SLG3 * SI N(TH3)
SI3 = ((SKG3 * SKG3) * THA3) / AG3
GA3 = (RH4 - TH3)
BG3 = SLG3 * SI N(GA3)
SLG3 = (SI3 + BG3)
YL3 = SLG3 * SI N(GA3)
YLq3 = (YL3 - SLG3)
XL3 = SLG3 * SI N(TH3)
XLG3 = SLG3 * SI N(TH3)
XLG3 = (XL3 - XLG3)

FEDV= (I3 + (F13 * YLq3) + (C13 * G * XLG3)) / SL3
FEDV= (F12 * SLQ2) + (C12 * G * XLG2 - T2) / SL2
FEDVH = FEDV * COS(TH2)
FEDVH = FEDV * SIN(TH2)
FEDVH = FEDV * COS(TH3)
FEDVH = FEDV * SIN(TH3)
FDEXV = FEDVH + FDEVV
FDEXV = FEDVH + FDEVV
FDEXH = FEDVH + FDEVV
FDEX = (FDEXV * FDEXV) + (FDEXH * FDEXH)

IF (FDEX) 116, 116, 117

116
FDE = 0.6
GO TO 112

117
FDE = (FDEX) * 0.5
IF (FDEXV(S, 118, 118)

FA0 = -FDE

118
PH11 = ATAN(FDEXV/FDEXH)
DEL5 = (PH11 + 0.5 * 3.1416)
FH11 = FI1 * COS(GA1)
SMH1 = SM1 * G * COS(TH1)
FAOH = ((FBCH) + (FH11) - SMH1)
FAOX = ((FAOH * FAOH) + (FAOV * FAOV) )
IF (FAOX) 63, 63, 64

63
FA0 = 0.6
GO TO 61

64
FA0 = (FAOX) * 0.5
IF (FAOV) 60, 61, 61

60
FA0 = -FA0

61
PH12 = ATAN(FAOV/FAOH)
DEL5 = (TH1 + PH12)
GO TO 120
WRITE(1,50)
FORMAT(27H JOINT DISPLACEMENTS FEET)
WRITE(1,51)
FORMAT(26H JOINT DISPLACEMENTS
WRITE(1,52)
FORMAT(31H JOINT VELOCITIES FEET PER SEC)
WRITE(1,53)
FORMAT(36H JOINT ACCELERATIONS FEET PER SEC SEC)
WRITE(1,54)
FORMAT(25H ARM TORQUES LBS WI FEET)
WRITE(1,55)
FORMAT(48H ANGLES ARE CLOCKWISE POSITIVE FEET DOWN VERTICAL)
WRITE(1,56)
FORMAT(22H JOINT FORCES LBS WI)
WRITE(1,57)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,58)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,59)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,60)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,61)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,62)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,63)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,64)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,65)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,66)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,67)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,68)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,69)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,70)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,71)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,72)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,73)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,74)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,75)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,76)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,77)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,78)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,79)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,80)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,81)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,82)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,83)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,84)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,85)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,86)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,87)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,88)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,89)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,90)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,91)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,92)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,93)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,94)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,95)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,96)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,97)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,98)
FORMAT(26H JOINT FORCES LBS WI)
WRITE(1,99)
FORMAT(26H JOINT TORQUES LBS WI FEET)
WRITE(1,100)
GO TO 100
TEST PRINTOUT OF ALL COMPUTED VALUES

```
WRITE(1,65) VDC, DEL21, VFE, DEL31, ABAT, ABAN, PH3, AG1
WRITE(1,65) ADCT, ADCN, ADC, PH4, RH2, PH5, AFZT, AFZEN
WRITE(1,65) AG2, AFE2, PH5, RH4, PH7, F11, B11, GA1
WRITE(1,65) F3CV, A3, F12, B12, GA2, FCBV, FCB, PH8
WRITE(1,65) PH9, FAOV, F13, B13, GA3, FEDV, FDEV, PHI1
WRITE(1,65) PHI1, SMH1, FAOH, PHI2
FORMAT(8F8.4)
```

END
APPENDIX 2

COMPUTED INPUT DATA

PDP - 9

COMPUTER PROGRAMME

"LIMBO"
THE FOLLOWING DATA HAS BEEN COMPILED ELSEWHERE AND IS IN TWO PARTS (1) FIXED (2) VARIABLE. (REFER A2/3 FOR FORMAT)

1. DATA HELD CONSTANT THROUGH THE ENTIRE HAMMER SWING

<table>
<thead>
<tr>
<th>1. DATA HELD CONSTANT</th>
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<tr>
<td>0.9580 0.9026 1.0820 4.0300 1.9650 2.9300</td>
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<td>0.407 0.403 0.604 6.006</td>
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2. STEP REFERENCE DATA

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<tr>
<td>0.592 2.496 3.177 1.996 3.593 1.531</td>
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<tr>
<td>0.552 2.365 3.098 2.311 4.055 4.252</td>
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<tr>
<td>0.472 2.264 2.956 2.314 4.627 9.920</td>
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<tr>
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<tr>
<td>0.370 1.564 1.567 0.000 2.660 6.932</td>
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<tr>
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<tr>
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<tr>
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<tr>
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HIGH SPEED CAMERA STUDY OF UPPER EXTREMITY LILB

DYNAMIC ANALYSIS AT REFERENCE FRAME 1

HAMMER STUDY CONSTANT VARIABLES

LENGTHS
FEET
SL1  SL2  SL3  SM1  SM2  SM3
0.5660 0.9020 1.0630 4.0300 1.9600 2.9300

MASSES
LBS MASS
SM1  SM2  SM3  SKG1  SKG2  SKG3
0.5660 0.5170 0.6440

RADIUS OF GYRATION
FEET

REFERENCE FRAME INPUT DATA

CENTER OF GRAVITY
FEET
SLG1  SLG2  SLG3
0.407 0.403 0.605

LINK 3 APPLIED FORCE
LBS WI RADS
W  TH4
0.000 0.000

VELOCITY
RAD PER SEC
TH1  TH2  TH3  THV1  THV2  THV3
0.665 2.609 3.063 -0.037 -0.148 -3.706

ACCELERATION
RAD PER SEC SEC
THA1  THA2  THA3
20.627 45.148 60.240

DYNAMIC ANALYSIS AT REFERENCE FRAME 2

TH1  TH2  TH3  THV1  THV2  THV3
0.665 2.609 3.142 0.533 0.857 -2.000

DYNAMIC ANALYSIS AT REFERENCE FRAME 3

TH1  TH2  TH3  THV1  THV2  THV3
0.637 2.554 3.168 1.396 1.973 -1.670

DYNAMIC ANALYSIS AT REFERENCE FRAME 4

TH1  TH2  TH3  THV1  THV2  THV3
1.592 2.496 3.177 1.996 3.593 1.331

YUAMC ANALYSIS AT REFERENCE FRAME 1
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<th>TH2</th>
<th>TH3</th>
<th>THV1</th>
<th>THV2</th>
<th>THV3</th>
<th>TAH1</th>
<th>TAH2</th>
<th>TAH3</th>
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<td>3.096</td>
<td>2.311</td>
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<td>2.956</td>
<td>2.314</td>
<td>4.627</td>
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<td>1.624</td>
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<td>11.050</td>
<td>-28.438</td>
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<tr>
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<td>1.955</td>
<td>2.400</td>
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APPENDIX 3

COMPUTER RESULTS

PDP - 9

COMPUTER PROGRAMME

"LIMBO"
High Speed Camera Study of Upper Extremity Limb

Dynamic Analysis at Reference Frame

Hammer Study Constant Variables

Lengths

Feet

Sl1 Sl2 Sl3 Sn1 Sn2 Sn3
0.9580 0.9020 1.0830 4.0300 1.9600 2.9300

Masses

Lbs Mass

Sk1 Sk2 Sk3 Sg1 Sg2 Sg3
0.5580 0.5170 0.5440

Radii of Gyration

Feet

Sa1 Sa2 Sa3

Reference Frame Input Data

1 Upperarm

2 Forearm

3 Hand / Hammer

Center of Gravity

Lbs Wt

Feet

Lbs Mass

Gravity

Feet

Per Sec Sec

Link 3 Applied Force

Lbs Wt

Rad

Per Sec

Position

Link 3 Applied Force

Gravity

Feet

Per Sec Sec

Joint Displacements

Feet

Joint Velocities

Feet Per Sec

Joint Accelerations

Feet Per Sec Sec

Rm Torques

Lbs Wt

Feet

Angles

Rad

Lbs Wt

Angles

Rad

Rm Torques

Lbs Wt

Angles

Rad

Note: Angles are clockwise positive from downward vertical.
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**Joint Displacements (Feet)**

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WKB TONGUES LB3 WT FEET

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**Joint Displacements (Feet):**

- **Elbow:** 0.06
- **Wrist:** 6.22

**Hammer Face:** 0.24

**Joint Velocities (Feet per Sec):**

- **Elbow:** 0.00
- **Wrist:** 1.46

- **Angle (Rad):** 1.94

**Hammer Face:** 11.26

**Joint Accelerations (Feet per Sec per Sec):**

- **Elbow:** 0.00
- **Wrist:** -28.19

- **Angle (Rad):** 1.57

**Hammer Face:** 385.81

**Arm Torques (Lbs-Wt Feet):**

- **Shoulder:** 0.00
- **Elbow:** -16.36

- **Wrist:** 450.58

**Joint Forces (Lbs-Wt):**

- **Shoulder:** 104.79
- **Elbow:** 34.12

- **Wrist:** 351.30

- **Angle (Rad):** 0.11

- **Wrist:** 2.55

- **Wrist:** 3.12
Dynamic Analysis at Reference Frame

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<th>Value (Feet)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ELBOW</td>
<td>0.66</td>
</tr>
<tr>
<td>WRIST</td>
<td>0.22</td>
</tr>
<tr>
<td>FACE</td>
<td>0.25</td>
</tr>
</tbody>
</table>

### JOINT VELOCITIES FEET PER SEC

<table>
<thead>
<tr>
<th>Joint</th>
<th>Value (Feet/Sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ELBOW</td>
<td>0.66</td>
</tr>
<tr>
<td>WRIST</td>
<td>-1.71</td>
</tr>
<tr>
<td>ANGLE RAD</td>
<td>1.94</td>
</tr>
<tr>
<td>FACE</td>
<td>-25.83</td>
</tr>
</tbody>
</table>

### JOINT ACCELERATIONS FEET PER SEC SEC

<table>
<thead>
<tr>
<th>Joint</th>
<th>Value (Feet/Sec^2)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ELBOW</td>
<td>3.08</td>
</tr>
<tr>
<td>WRIST</td>
<td>-187.73</td>
</tr>
<tr>
<td>ANGLE RAD</td>
<td>1.57</td>
</tr>
<tr>
<td>FACE</td>
<td>2235.83</td>
</tr>
</tbody>
</table>

### AN TORQUES LBS WT FEET

<table>
<thead>
<tr>
<th>Joint</th>
<th>Value (Lbs-Wt Feet)</th>
</tr>
</thead>
<tbody>
<tr>
<td>HOULDER</td>
<td>0.00</td>
</tr>
<tr>
<td>ELBOW</td>
<td>-109.02</td>
</tr>
<tr>
<td>WRIST</td>
<td>-2204.51</td>
</tr>
</tbody>
</table>

### JOINT FORCES LBS WT

<table>
<thead>
<tr>
<th>Joint</th>
<th>Value (Lbs-Wt)</th>
</tr>
</thead>
<tbody>
<tr>
<td>HOULDER</td>
<td>73.98</td>
</tr>
<tr>
<td>ELBOW</td>
<td>-52.11</td>
</tr>
<tr>
<td>WRIST</td>
<td>1432.84</td>
</tr>
<tr>
<td>ANGLE RAD</td>
<td>6.35</td>
</tr>
</tbody>
</table>
### Dynamic Analysis at Reference Frame

<table>
<thead>
<tr>
<th>Joint</th>
<th>Displacements</th>
<th>Velocities</th>
<th>Accelerations</th>
<th>Torques</th>
<th>Forces</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Feet</td>
<td>Feet/Sec</td>
<td>Feet/Sec/Sec</td>
<td>Lbs*In</td>
<td>Lbs</td>
</tr>
<tr>
<td>Elbow</td>
<td>0.06</td>
<td>0.00</td>
<td>1.94</td>
<td>96.11</td>
<td>149.72</td>
</tr>
<tr>
<td>Wrist</td>
<td>2.22</td>
<td>6.86</td>
<td>1.57</td>
<td>252.84</td>
<td>3434.82</td>
</tr>
</tbody>
</table>

*Note: The table values are approximate and illustrate the analysis results.
<table>
<thead>
<tr>
<th>Joint</th>
<th>Angle (Degrees)</th>
<th>Angle (Radian)</th>
<th>RISST</th>
<th>Flap</th>
<th>Shoulders Loss</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.47</td>
<td>2.63</td>
<td>2.62</td>
<td>0.15</td>
<td>33.90</td>
<td>9.93</td>
</tr>
<tr>
<td>1.67</td>
<td>1.51</td>
<td>1.45</td>
<td>4.36</td>
<td>30.90</td>
<td>6.92</td>
</tr>
<tr>
<td>4.54</td>
<td>2.63</td>
<td>2.62</td>
<td>0.15</td>
<td>33.90</td>
<td>9.93</td>
</tr>
<tr>
<td>8.17</td>
<td>2.31</td>
<td>2.31</td>
<td>0.11</td>
<td>3.10</td>
<td>6.55</td>
</tr>
</tbody>
</table>

Displacement Field: 0.47 in. The GM 5.57 - 3.11 - 5.55 - 1.79 - 3.11 - 5.55 - 3.11 - 5.55

Dynamic Analyses at Reference Plane: 24
DYNAMIC ANALYSIS AT REFERENCE FRAME 28

<table>
<thead>
<tr>
<th>TH1</th>
<th>TH2</th>
<th>TH3</th>
<th>THV1</th>
<th>THV2</th>
<th>THV3</th>
<th>THA1</th>
<th>THA2</th>
<th>THA3</th>
</tr>
</thead>
</table>

**JOINT DISPLACEMENTS FEET**

<table>
<thead>
<tr>
<th>Elbow</th>
<th>Wrist</th>
<th>Hammer Face</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.88</td>
<td>0.34</td>
<td>0.03</td>
</tr>
</tbody>
</table>

**JOINT VELOCITIES FEET PER SEC**

<table>
<thead>
<tr>
<th>Elbow</th>
<th>Wrist</th>
<th>Hammer Face</th>
</tr>
</thead>
<tbody>
<tr>
<td>-1.52</td>
<td>-3.41</td>
<td>-10.35</td>
</tr>
</tbody>
</table>

**JOINT ACCELERATIONS FEET PER SEC SEC**

<table>
<thead>
<tr>
<th>Elbow</th>
<th>Wrist</th>
<th>Hammer Face</th>
</tr>
</thead>
<tbody>
<tr>
<td>-45.63</td>
<td>55.55</td>
<td>75.55</td>
</tr>
</tbody>
</table>

**RATIOMQUES LBS WT FEET**

<table>
<thead>
<tr>
<th>Shoulder</th>
<th>Elbow</th>
<th>Wrist</th>
</tr>
</thead>
<tbody>
<tr>
<td>-59.55</td>
<td>15.11</td>
<td>23.67</td>
</tr>
</tbody>
</table>

**JOINT FORCES LBS WT**

<table>
<thead>
<tr>
<th>Shoulder</th>
<th>Elbow</th>
<th>Wrist</th>
</tr>
</thead>
<tbody>
<tr>
<td>-115.63</td>
<td>12.20</td>
<td>40.77</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Angle Rad</th>
<th>Angle Rad</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.16</td>
<td>1.31</td>
</tr>
<tr>
<td>Joint</td>
<td>Measurements</td>
</tr>
<tr>
<td>-------</td>
<td>--------------</td>
</tr>
<tr>
<td>Elbow</td>
<td>6.89 feet, -2.54 rad, 17.69 ft/sec, 1.72 rad/sec</td>
</tr>
<tr>
<td>Wrist</td>
<td>2.35 feet, -2.81 rad, 3.61 ft/sec, 3.53 rad/sec</td>
</tr>
<tr>
<td>Hammer Face</td>
<td>6.49 feet, -7.87 rad, 4.87 ft/sec, 3.65 rad/sec</td>
</tr>
</tbody>
</table>

**Joint Torques (lbs-ft)**

<table>
<thead>
<tr>
<th>Joint</th>
<th>Torque</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>21.71 lbs-ft</td>
</tr>
<tr>
<td>Elbow</td>
<td>33.61 lbs-ft</td>
</tr>
<tr>
<td>Wrist</td>
<td>71.34 lbs-ft</td>
</tr>
</tbody>
</table>

**Joint Forces (lbs)**

<table>
<thead>
<tr>
<th>Joint</th>
<th>Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>117.91 lbs</td>
</tr>
<tr>
<td>Elbow</td>
<td>45.19 lbs</td>
</tr>
<tr>
<td>Wrist</td>
<td>-39.11 lbs</td>
</tr>
<tr>
<td>Wrist Rad</td>
<td>2.62 lbs</td>
</tr>
</tbody>
</table>

**Dynamic Analysis at Reference Frame**

<table>
<thead>
<tr>
<th>Joint</th>
<th>TH1, TH2, TH3, THV1, THV2, THV3, THA1, THA2, THA3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Values</td>
<td>6.593, 2.438, 2.835, -2.551, -2.512, -5.392, 17.322, 54.125, 58.442</td>
</tr>
</tbody>
</table>
APPENDIX 4

LIMB VOLUMES AND MASSES

AND RADIUS OF GYRATION

CALCULATIONS
DENSITY OF LIMB

Method used was to calculate the density from a volume and weight measurement of a similar bone-muscle combination.

Example (1) Teebone Steak - 1 lb. and 29.1 cu. ins.

or 59.5 lb./cu. ft.

Since this density is lower than that of water at 62.3 lb./cu. ft. it was assumed that the steak had lost a considerable quantity of water.

Example (2) Pig's Trotter - \( \frac{7}{16} \) lb. and 10.3 cu. ins

or 73.5 lb./cu. ft.

This density was used for the calculations of member weights.
VOLUME AND WEIGHT OF LIMB MEMBERS

The volume of each member of the arm was measured by the water displacement method.

Using the bone-muscle density of 73.5 lb./cu. ft. the corresponding member weights were calculated.

<table>
<thead>
<tr>
<th>Member</th>
<th>Volume (cu. ins.)</th>
<th>Weight (lbs.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upperarm</td>
<td>94</td>
<td>4.0</td>
</tr>
<tr>
<td>Forearm</td>
<td>46</td>
<td>1.96</td>
</tr>
<tr>
<td>Hand</td>
<td>22</td>
<td>0.94</td>
</tr>
</tbody>
</table>
The centre of gravity of the upperarm and forearm was calculated assuming that the bone and muscle material has a constant mass distribution proportional to volume, and the shape approximated by a part circular cone.

The hand, centre of gravity was estimated by studying the bone and tissue distribution.

Consider the segment of the Cone below:

\[
X \cdot \rho \cdot V_{\text{tot}} = \rho \cdot V_{\text{cyl}} \cdot \frac{h}{2} + \rho \cdot V_{\text{tri}} \left( \frac{h}{1 + a/b - a} \right)
\]

Where Volume total = \( \frac{\pi \cdot h}{3} \left( b^2 + \left( \frac{a}{b-a} \right) (b^2 - a^2) \right) \)

V cylinder = \( \pi a^2 b \)

V triangloid = \( \frac{1}{3} \left( \pi \cdot b^2 (h + \frac{ah}{b-a}) - \pi a^2 \left( \frac{a}{b-a} \right) h \right) \)

(or V total - V cylinder)

where \( \rho \) is the density
Using measurements made of the subject the following values were obtained.

<table>
<thead>
<tr>
<th>Member</th>
<th>a</th>
<th>b</th>
<th>h</th>
<th>Calculated Volume</th>
<th>Measured Volume</th>
<th>X inches centre of Gravity Distance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upperarm</td>
<td>1.5&quot;</td>
<td>2.0&quot;</td>
<td>11.5&quot;</td>
<td>111.5</td>
<td>94in.³</td>
<td>4.88</td>
</tr>
<tr>
<td>Forearm</td>
<td>1.0&quot;</td>
<td>1.5&quot;</td>
<td>11&quot;</td>
<td>54.75</td>
<td>46in.³</td>
<td>4.83</td>
</tr>
<tr>
<td>Hand</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>22in.³</td>
<td>1.5&quot;</td>
</tr>
<tr>
<td>Hammer</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>10&quot;</td>
</tr>
<tr>
<td>Hand &amp; Hammer</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>7.27&quot;</td>
</tr>
</tbody>
</table>
RADIUS OF GYRATION CALCULATIONS

The moment of inertia (I) can be expressed in terms of the radius of gyration (k) and mass (m) of the pivoted members; namely:

\[ I = k^2 m \]

or \[ k = \sqrt{\frac{I}{m}} \]

Using the dimensions as listed in the table for centre of gravity calculations Page A4/5 the upperarm \( k = 6.7 \text{ inches} \) and forearm \( k = 6.23 \text{ inches} \) was obtained.

The hand and hammer combination was assumed equivalent to a composite set of cylinders.
Moment of inertia \( I_{xx} = I_0 + d^2 m \) for each mass and \( I_1 + I_2 + I_3 + I_4 = k^2 m \) for the combination, resulting in \( k = 5 \) inches.
The moment of inertia equation was developed for the following part cone by considering an elemental mass "dm".

The elemental inertia $dI = dm \, r^2$ is integrated, and

Substituting $\frac{\rho}{\epsilon} \cdot dA = dm$

\[
x = a + f (b - y) \\
f = \frac{b - a}{h}
\]

yields $I = \frac{\rho \cdot r}{\epsilon} \left( \frac{h^3}{5} + \frac{ab}{10} + \frac{b^2}{30} \right)$

The moment of inertia is calculated to obtain the Radius of gyration of the limb member segments.
APPENDIX 5

TABLE TIME

<table>
<thead>
<tr>
<th>HIGH SPEED FILM TIMING MARKS</th>
<th>DETAILS</th>
</tr>
</thead>
</table>


## APPENDIX 5 TABLE TIME

**HIGH SPEED FILM TIMING MARKS DETAILS**

### DOWNWARD SWING

<table>
<thead>
<tr>
<th>Step A</th>
<th>No. of Frames</th>
<th>Time(Secs)</th>
<th>Step A</th>
<th>No. of Frames</th>
<th>Time(Secs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>00/0</td>
<td>14.53</td>
<td>0.0275</td>
<td>12/13</td>
<td>16.9</td>
<td>0.0237</td>
</tr>
<tr>
<td>0/1</td>
<td>14.73</td>
<td>0.0272</td>
<td>13/14</td>
<td>16.95</td>
<td>0.0236</td>
</tr>
<tr>
<td>1/2</td>
<td>14.8</td>
<td>0.0270</td>
<td>14/15</td>
<td>17.0</td>
<td>0.0235</td>
</tr>
<tr>
<td>2/3</td>
<td>15.67</td>
<td>0.0255</td>
<td>15/16</td>
<td>17.2</td>
<td>0.0233</td>
</tr>
<tr>
<td>3/4</td>
<td>14.98</td>
<td>0.0267</td>
<td>16/17</td>
<td>17.35</td>
<td>0.0231</td>
</tr>
<tr>
<td>4/5</td>
<td>15.47</td>
<td>0.0259</td>
<td>17/18</td>
<td>17.6</td>
<td>0.0227</td>
</tr>
<tr>
<td>5/6</td>
<td>15.6</td>
<td>0.0256</td>
<td>18/19</td>
<td>17.8</td>
<td>0.0225</td>
</tr>
<tr>
<td>6/7</td>
<td>15.73</td>
<td>0.0254</td>
<td>19/20</td>
<td>17.87</td>
<td>0.0224</td>
</tr>
<tr>
<td>7/8</td>
<td>15.9</td>
<td>0.0251</td>
<td>20/21</td>
<td>17.9</td>
<td>0.0223</td>
</tr>
<tr>
<td>8/9</td>
<td>16.0</td>
<td>0.0250</td>
<td>21/22</td>
<td>18.15</td>
<td>0.0220</td>
</tr>
<tr>
<td>9/10</td>
<td>16.3</td>
<td>0.0245</td>
<td>22/23</td>
<td>18.27</td>
<td>0.0219</td>
</tr>
<tr>
<td>10/11</td>
<td>16.57</td>
<td>0.0241</td>
<td>23/24</td>
<td>18.4</td>
<td>0.0217</td>
</tr>
<tr>
<td>11/I1</td>
<td>16.8</td>
<td>0.0208</td>
<td>24/25</td>
<td>18.53</td>
<td>0.0216</td>
</tr>
<tr>
<td>I1/12</td>
<td>5</td>
<td>0.0029</td>
<td>25/26</td>
<td>18.67</td>
<td>0.0214</td>
</tr>
<tr>
<td>11/12</td>
<td>16.75</td>
<td>0.0239</td>
<td>26/27</td>
<td>18.65</td>
<td>0.0214</td>
</tr>
<tr>
<td>12/12a</td>
<td>5</td>
<td>0.0029</td>
<td>27/28</td>
<td>18.8</td>
<td>0.0213</td>
</tr>
<tr>
<td>12a/12b</td>
<td>5</td>
<td>0.0029</td>
<td>28/29</td>
<td>19.1</td>
<td>0.0209</td>
</tr>
<tr>
<td>12b/I2</td>
<td>10</td>
<td>0.0059</td>
<td>29/30</td>
<td>18.63</td>
<td>0.0215</td>
</tr>
<tr>
<td>I2/13</td>
<td>16.8</td>
<td>0.0119</td>
<td>30/31</td>
<td>18.93</td>
<td>0.0211</td>
</tr>
</tbody>
</table>

### UPWARD SWING
Where $A$ is Reference Frame Numbers
$B$ is Number of Timing marks between reference frames
$C$ is Time interval between reference frames

Typically with forty (40) Photo frames between reference frames and 0.01 seconds per timing mark

$$C = \frac{40 \times 0.01}{B} \text{ Seconds.}$$
APPENDIX 6

FIGURE 2.45 - TABLE -

MUSCULAR SUPPLY OF PERIPHERAL NERVES

(Motor and Sensory)
FIGURE 2.45 MUSCULAR SUPPLY OF PERIPHERAL (MOTOR & SENSORY) NERVES

(A) Plexus Cervicalis (C1 - C4)

Nervi cervicales Musculi profundi colli

Mm. scaleni

Nervi phrenicus Diaphragma

(B) Plexus Brachialis (C5 - D2) (D2 = T2)

Nervi Thoracic ant. Musculi pect. maj. et. min.

Nervi Thoracic long.

Nervi dorsalis scap.

Nervi suprascap

Nervi subscapul

Musculi levator scapul

Mm. rhomboidei

Musculi supraspinatus

Musculi infraspinatus

Musculi latissimus dorsi

Musculi teres major

Flexion, extension and rotation of the neck

Elevation of ribs (inspiration)

Inspiration

Adduction and forward depression of the arm

Fixation of the scapula during elevation of the arm

Elevation of the scapula

Elevation and drawing inward of the scapula

Elevation and external rotation of the arm

External rotation of the arm

Internal rotation and dorsal adduction of the arm
N. axillaris s. circumflexus

Musculi subscapulavis

N. musculocut

M. deltoideus

M. teres minor

M. biceps brach.

M. coraco-brachialis

M. brachialis ant.

M. pronator teres

M. flexor carpi rad.

M. palm long.

M. flex. digit.

M. flex. poll. long.

M. flex. dig. prof.

M. abduct. poll. brev.

N. medianus

Internal rotation of the arm

Elevation of the arm to the horizontal

External rotation of the arm

Flexion and supination of the forearm

Flexion and adduction of the forearm

Flexion of the forearm

Pronation

Flexion and radial flexion of the hand

Flexion of the hand

Flexion of the middle phalanges of the fingers

Flexion of the terminal phalanx of the thumb

Flexion of the terminal phalanges of the index and middle fingers

Abduction of the first metacarp
<table>
<thead>
<tr>
<th>Nervi ulnaris</th>
<th>Nervus radialis</th>
</tr>
</thead>
<tbody>
<tr>
<td>M. flex. poll. brev.</td>
<td>M. triceps brach.</td>
</tr>
<tr>
<td>M. opponens poll.</td>
<td>M. supin. longus</td>
</tr>
<tr>
<td>M. flexor carpi uln.</td>
<td>M. extensor carpi rad.</td>
</tr>
<tr>
<td>M. flex. digit. prof. (ulnar</td>
<td>M. extensor digit. comm.</td>
</tr>
<tr>
<td>portion)</td>
<td></td>
</tr>
</tbody>
</table>

Flexion of the first phalanx of the thumb
Opposition of the first metacarp
Flexion and ulnar flexion of the hand
Flexion of the terminal phalanges of the ring and little fingers
Adduction of the first metacarp
Abduction, opposition, and flexion of the little finger
Flexion of the first phalanges, extension of the others
The same, in addition, abduction and adduction of the fingers
Extension of the forearm
Flexion of the forearm
Extension and radial flexion of the hand
Extension of the first phalanges of the fingers
M. extensor digit. V prop.

M. extensor carpi uln.

M. supinator brevis

M. abduct. poll. longus

M. extensor poll. brevis

M. extensor poll. longus

M. extensor index. prop.

(C) Nervi Thoracales (D3 - D11) or (T3 - T11)

Nervi thoracales Mm. thoracici et abdominales

(E) Plexus Sacralis (L5 - S5)

Extension of the first phalanx of the little finger

Extension and ulnar flexion of the hand

Supination of the forearm

Abduction of the first metacarp

Extension of the first phalanx of the thumb

Abduction of the first metacarpal and extension of the terminal phalanx of the thumb

Extension of the first phalanx of the index finger

Elevation of the ribs, expiration, compression of abdominal viscera relating to lower trunk and lower extremities - of no interent here
APPENDIX 7

SOURCE LISTING OF PHOTO STUDY

COMPUTER DATA PREPARATION PROGRAMME
PHOTO-STUDY OF THE UPPER EXTREMITY LIMB

PROGRAMME TO PREPARE THE NECESSARY DATA FOR THE DYNAMIC ANALYSIS ROUTINE TITLED "LIMBO" -

OUTPUT PAPER TAPE FROM THIS PROGRAMME IS USED DIRECTLY FOR THE LIMBO DATA INPUT TAPE

LIST OF PHOTO-STUDY DATA *

CONST ----- THIS FIXED POINT NUMBER EQUALS ZERO IF PHYSICAL PROPORTION OF THE LIMB ARE ASSUMED CONST. THAT IS, JOINT ROLLING IS IGNORED - ASSUME PURE PIVOT ACTION AT JOINTS

+ OR - (IVE) VALUE REQUIRES STEP INPUT DATA

FRSTEP ----- THIS FIXED POINT NUMBER EQUALS THE NUMBER OF SINGLE PHOTO FRAMES BETWEEN TWO CONSECUTIVE STATIONS, OR STEPPING FRAMES. THIS VALUE SHOULD ALWAYS BE POSITIVE

ACCELERATION DUE TO GRAVITY (FLOATING POINT)

TH1, TH2, TH3 THESE THREE ANGLES HAVE BEEN CHOOSEN AS THE REFERENCE ANGLES OF THE REFERENCE STATION FOR THE ENTIRE CYCLE OF THE MOTION. THIS IS A FLOATING POINT NUMBER AND ANGLES ARE EXPRESSED IN RADIANS. THE THREE ANGLES ARE RESPECTIVELY RTH1 UPPER ARM DISPLACEMENT ANGLE FROM THE DOWN VERTICAL, RTH2 FOREARM AND RTH3 HAND DISPLACEMENT ANGLES
A7/3

DIMENSION DELSTP(50), DELTIM(50)
DIMENSION TH1(50), TH2(50), TH3(50), SL1(50), SL2(50)
DIMENSION SM1(50), SM2(50), SM3(50), SL3(50)
DIMENSION SKG1(50), SLG2(50), SKG3(50)
DIMENSION SLG1(50), SLG2(50), SLG3(50)
DIMENSION W(50), TH4(50), TIMF(150)

READ(3,201) CONST, FRSTFP, C, RTH1, KTH2, RTH3

FORMAT(213, 4F9.4)

C FRSTFP IS THE NUMBER OF FRAMES PER STEPP...MAX. OF 999
C CONST EQUALS ZERO FOR SL1, SM1, ECT. = CONSTANT AND
C + OR - HAS DISCRETE STEPP VALUES
IF(CONST) 202, 203, 202

READ(3,204) SL1, SL2, SL3, SM1, SM2, SM3
READ(3,204) SLG1, SLG2, SLG3, SKG1, SKG2, SKG3
DO 208 KB = 1, 50
READ(3,207) KB, (TH1(K), TH2(K), TH3(K), K=1, KB)

FORMAT(3F9.4)

IF(TH1(K) - 9999,9999) 208,209,210

C 209 STARTS TIMING MARKS READ & NOTIFIES END OF STEPP POSITION
CONTINUE
STOP

FORMAT(6F9.4)

DO 210 JA = 1, 150
READ(3,211) JA, (TIME(J), J=1, JA)

FORMAT(F9.4)

IF(TIME(J) - 9999,9999) 210,213,210

CONTINUE

C 213 STARTS PROGRAMME CALCS. & NOTIFIES END OF TIMING DATA
STOP

DO 216 KB = 1, 50
READ(3,214) KB, (TH1(K), TH2(K), TH3(K), K=1, KB)
READ(3,214) KB, (SM1(K), SM2(K), SM3(K), K=1, KB)
READ(3,214) KB, (SKG1(K), SKG2(K), SKG3(K), K=1, KB)
READ(3,214) KB, (SLG1(K), SLG2(K), SLG3(K), K=1, KB)
READ(3,214) KB, (TH4(K), G(K), K=1, KB)

FORMAT(3F9.4)

IF(TH1(K) - 9999,9999) 216,209,216

CONTINUE
STOP

JLTIME = (JA-1)
KFRAME = (KB-1)

C JLTIME IS NUMBER OF TIMING MARKS FOR STUDY CYCLE
C KFRAME IS NUMBER OF STEPPING FRAMES PER STUDY CYCLE
DO 217 JX = 1, JFRAME
FRSTFP = FRSTFP * JX
DO 217 JX = 1, JLTIME
IF(TIME(JX) - FRSTFP) 217, 218, 218
C DETERMINES THE NEXT HIGHEST TIMING MARK ABOVE THE REFERENCE FRAMES
C 218 -- DETERMINES TIMES
C 217 CONTINUE
STOP
DO 221 FORMAT(43H ERROR IN DETERMINING TIMING MARKS.....STOPS)
WRITE(4,221)
C START OF VELOCITY & ACCELERATION CALCS.
C
C DIFFTH1= (RTH1-TH1(1))
C DIFFTH2= (RTH2-TH2(1))
C DIFFTH3= (RTH3-TH3(1))

C DIFFTH1-3 IS THE ANGULAR DIFFERENCE BETWEEN BONE INCLINATIONS
C AND PHOTO STUDY SKIN ANGLES FOR REFERENCE FRAME (1)

C DISPLACEMENT ANGLES OF EACH LINK IS GIVEN WITH REFERENCE
C TO ANGLES RTH1-3, THE TRUE LINK PIVOT POINT ANGLES

C CORRECTED ANGLES ARE CTH1-3 AND EQUAl TH1-3 + DIFFTH1-3
DO 222 KC=1,50
CTH1(KC)=TH1(KC)+DIFFTH1
CTH2(KC)=TH2(KC)+DIFFTH2
CTH3(KC)=TH3(KC)+DIFFTH3
CONTINUE
DO 223 KV=2,49
THV1(KV)=((CTH1(KV-1)-CTH1(KV+1))/(DELTM(KV)+DELTM(KV+1)))
THV2(KV)=((CTH2(KV-1)-CTH2(KV+1))/(DELTM(KV)+DELTM(KV+1)))
THV3(KV)=((CTH3(KV-1)-CTH3(KV+1))/(DELTM(KV)+DELTM(KV+1)))
CONTINUE
DO 224 KA=3,48
THA1(KA)=((THV1(KA-1)-THV1(KA+1))/(DELTM(KA)+DELTM(KA+1)))
THA2(KA)=((THV2(KA-1)-THV2(KA+1))/(DELTM(KA)+DELTM(KA+1)))
THA3(KA)=((THV3(KA-1)-THV3(KA+1))/(DELTM(KA)+DELTM(KA+1)))
CONTINUE

C ANGULAR VELOCITIES THV1-3 & ACCELERATIONS THA1-3 ARE IN ARRAYS

SPECIAL REFERENCE FRAMES PROPERTIES FOLLOW
FRAMES 1 TO 4 HAVE BEEN ALLOCATED FOR THIS PURPOSE

C WRITE STATEMENTS FOLLOW
END
APPENDIX 8

SOURCE LISTING

OF

BASIC COMPUTER PROGRAMME

FORTRAN 2 FAP. IBM - 1620

INCLUDING TEST INPUT AND OUTPUT
PROGRAMME: LIMP No1 (TH1 = TH2 = TH3 = 90 degrees)
CODED BY: N. HODKINSON
DATE - 10.10.66
STATEMENT NUMBERS
1...567...12......22............37..............52.................72
   CALL FAP
      XEQ
6      READ, TH1, TH2, TH3, THV1, THV2, THV3, THA1, THA2, THA3
      READ, SL1, SL2, SL3, SM1, SM2, SM3, SKG1, SKG2, SKG3
      READ, SLG1, SLG2, SLG3, W, TH4, G
VBA = SL1 * THV1
VBAV = VBA * SINF (TH1)
VBAH = VBA * C0SF (TH1)
DEL1 = (TH1 + 0.5 * 3.1416)
VDC = SL2 * TH2
VDCV = VDC * SINF (TH2)
VDCH = VDC * C0SF (TH2)
DEL21 = (TH2 + 0.5 * 3.1416)
VDAV = (VBAV + VDCV)
VDAH = (VBAH + VDCH)
VDA = ((VDCV)**2 + (VDAH)**2)**0.5
PH1 = ATANF (VDAV / VDAH)
DEL2 = (PH1 + 0.5 * 3.1416)
VFE = SL3 * THV3
VFEV = VFE * SINF (TH3)
VFEH = VFE * C0SF (TH3)
DEL31 = (TH3 + 0.5 * 3.1416)
VFAV = (VDAV + VFEV)
VFAH = (VDAH + VFEH)
VFA = ((VFAV)**2 + (VFAH)**2)**0.5
PH2 = ATANF (VFAV / VFAH)
DEL3 = (TH2 + 0.5 * 3.1416)
ABAT = SL1 * THA1
ABATV = ABAT * SINF (TH1)
ABATH = ABAT * C0SF (TH1)
ABAN = SL1 * (THV1)**2
ABANV = ABAN * C0SF (TH1)
ABANH = ABAN * SINF (TH1)
ABA = (ABATV - ABANV)
ABAH = (ABATH + ABANH)
ABA = ((ABA)**2 + (ABAH)**2)**0.5
A8/4

\[
\begin{align*}
R_{H1} &= \text{ATANF}(A_BAV/A_BAH) \\
R_{H2} &= (PH_3 + 0.5*3.1416) \\
AG_1 &= (SL_2/SL_1)*ABA \\
ADCT &= SL_2*THA_2 \\
ADCTV &= ADCT*SINF(TH_2) \\
ADCH &= ADCT*COSF(TH_2) \\
ADCH &= SL_2*(THV_2)**2 \\
ADCV &= ADCH*COSF(TH_2) \\
ADCNH &= ADCH*SINF(TH_2) \\
ADCV &= (ADCTV-ADCV) \\
ADCH &= (ADCTH+ADCNH) \\
ADG &= ((ADCV)**2+(ADCH)**2)**0.5 \\
FH_4 &= \text{ATANF}(ADCV/ADCF) \\
RH_2 &= (FH_4 + 0.5*3.1416) \\
ADAV &= (ADCV+ABA_V) \\
ADAH &= (ADCH+ABA_H) \\
ADA &= ((ADA)**2+(ADA)**2)**0.5 \\
FH_5 &= \text{ATANF}(ADAV/ADAH) \\
RH_3 &= (FH_5 + 0.5*3.1416) \\
AFET &= SL_3*THA_3 \\
AFETV &= AFET*SINF(TH_3) \\
AFETH &= AFET*COSF(TH_3) \\
AFEN &= SL_2*(THV_3)**2 \\
AG_2 &= (SLG_2/SL_2)*ADC \\
AFENV &= AFEN*COSF(TH_3) \\
AFENH &= AFEN*SINF(TH_3) \\
APSV &= (AFETV-AFENV) \\
APSH &= (AFETH+AFENH) \\
APT &= ((AFTV)**2+(APCH)**2)**0.5 \\
TH_6 &= \text{ATANF}(APSV/APTH) \\
RH_4 &= (PH_6 + 0.5*3.1416) \\
ADFV &= AFTEV/ADAV \\
ADFV &= ADTH+ADAH
\end{align*}
\]
\[ \begin{align*} 
AFA &= ((AFAV)^*+2+(AFAH)^*+2)^*0.5 \\
PH7 &= ATANF(AFAV/AFAH) \\
RH5 &= ((PH7+0.5*3.1416)^*3.1416) \\
T1 &= SM1*((SKG1)^*+2)^*TH1 \\
FT1 &= SM1*AG1 \\
XLG1 &= SLG1*SINF(TH1) \\
BI1 &= (((SKG1)^*+2)^*THA1)/AG1 \\
GA1 &= (RH1-TH1) \\
BC1 &= SLG1*SINF(GA1) \\
SLG1 &= (BT1+BG1) \\
FBCV &= ((FT1*SLQ1)-T1+(SM1*G*XLG1))/SL1 \\
AG3 &= (SLG3/SL3)^*AFE \\
T2 &= SM2((SKG2)^*+2)^*THA2 \\
FT2 &= SM2*AG2 \\
XLG2 &= SLG2*SINF(TH2) \\
BI2 &= (((SKG2)^*+2)^*THA2)/AG2 \\
GA2 &= (RH2-TH2) \\
BG2 &= SLG2*SINF(GA2) \\
SLQ2 &= (BT2+BG2) \\
YL2 &= SL2*SINF(GA2) \\
YL2 &= (YL2-SL2) \\
XL2 &= SL2*SINF(TH2) \\
XLG2 &= (XL2-XL2) \\
FBCV &= (T2+(SM2*G*XLG2)+(FT2+YL2))/SL2 \\
FBCVH &= FBCV*OSF(TH1) \\
FBCV &= FBCV*SINF(TH1) \\
FBCVH &= FBCV*OSF(TH2) \\
FBCV &= FBCV*SINF(TH2) \\
FBCV &= (FBCVH+FBCVH) \\
FBCV &= (FBCVH+FBCVH) \\
FBCV &= (((FBCV)^*+2+(FBCVH)^*+2)*0.5 \\
PHG &= ATANF(FBCV/FBCVH) \\
DEL4 &= (PHG+0.5*3.1416) \\
PHG &= (DEL^*+PHG) 
\end{align*} \]
\begin{verbatim}
1...567

FBCH = FCB * C0SF(PH9)
PH10 = (TH2 - TH8)
FCBH = FCB * C0SF(PH10)
YL1 = SL1 * SINF(GA1)
YLQ1 = (YL1 - SLQ1)
XL1 = SL1 * SINF(TH1)
XLG11 = (XL1 - XLG1)
FA9V = ((FI1 * YLQ1) + (SM1 * G * XLG11) + T1) / SL1
T3 = SM3 * ((SKG3) ** 2) * THA3
IF(W) 80, 90, 80
80 SL3W = SL3 * C0SF(TH4)
T3 = (T3 - (W * SL3W))
90 FI3 = SM3 * AG3
XLG3 = SLG3 * SINF(TH3)
BI3 = (((SKG3) ** 2) * THA3) / AG3
GA3 = (RH4 - TH3)
BG3 = SLG3 * SINF(GA3)
SLQ3 = (BI3 + BG3)
T3X = T3
YL3 = SL3 * SINF(GA3)
YLQ3 = (YL3 - SLQ3)
XL3 = SL3 * SINF(TH3)
XLG3 = SLG3 * SINF(TH3)
XLG33 = (XL3 - XLG3)
FEDV = (T3 + (FI3 * YLQ3) + (SM3 * G * XLG33)) / SL3
FDEV = ((YI2 * SLQ2) + (SM2 * G * XLG2) - T2) / SL2
FEDVE = FEDV * C0SF(TH2)
FEDV = FEDV * SINF(TH2)
FDBVF = FDEV * C0SF(TH3)
FDEVV = FDEV * SINF(TH3)
FDEXV = FEDV + FDEVV
FDEXH = FEDV + FDEV
FD = ((FDEXV) ** 2 + (FDEXH) ** 2) ** 0.5
\end{verbatim}
$\Phi_{H1} = \text{ATANF} \left( \frac{F_{DExV}}{F_{DExH}} \right)$

$\Phi_{H1} = \Phi_{I1} \cdot \text{COS} \left( \gamma \right)$

$\text{SMH1} = \text{SM1} \cdot \text{COS} \left( \text{SF} \left( \theta_{1} \right) \right)$

$\Phi_{\alpha H} = \left( \frac{\text{PBCH}}{\Phi_{H1}} \right) + \left( \Phi_{H1} \right) - \text{SMH1}$

$\Phi_{\alpha} = \left( \left( \Phi_{\alpha H} \right)^{*} \times 2 + \left( \Phi_{\alpha V} \right)^{*} \times 2 \right) \times 0.5$

$\Phi_{H12} = \text{ATANF} \left( \frac{\Phi_{\alpha V}}{\Phi_{\alpha H}} \right)$

$\text{DEL6} = \left( \theta_{1} + \Phi_{H12} \right)$

PUNCH 50, VBA

PUNCH 60, DEL1

PUNCH 51, VDA

PUNCH 60, DEL2

PUNCH 52, VFA

PUNCH 60, DEL3

PUNCH 53, ABA

PUNCH 60, RH1

PUNCH 54, ADA

PUNCH 60, RH3

PUNCH 55, AFA

PUNCH 60, RH5

PUNCH 56, T1

PUNCH 57, T2

PUNCH 58, T3X

PUNCH 59, FA\alpha

PUNCH 60, DEL6

PUNCH 61, FCB

PUNCH 60, DEL4

PUNCH 62, FDE

PUNCH 60, DEL5

PUNCH 63, W

PUNCH 60, TH4

PUNCH 64

50 \text{ FORMAT} \left( \text{26HELCWVELS\_CITY\_b\_b\_b\_b\_b\_b\_b\_b\_b\_b\_F7.2} \right)
60 FORMAT(26HANGLE=bbbbb=bb=F7.2)
51 FORMAT(26HWRIStbVELOCITY=bbbbb=F7.2)
52 FORMAT(26HFINGERbtIPVELOCITY=bb=F7.2)
53 FORMAT(26HELBwACCELERATIONbbb=F7.2)
54 FORMAT(26HWRIStBACELERATIONnb=F7.2)
55 FORMAT(26HFINGERbACCELERATIONnb=F7.2)
56 FORMAT(26HSH=ULDERbTORQUEb=F7.2)
57 FORMAT(26HELb7bTOR.UE=bbb=F7.2)
58 FORMAT(26HWRIStbTORQUEbb=F7.2)
59 FORMAT(26HSH=ULDERbINTbFORCEbb=F7.2)
61 FORMAT(26HELBwJINTbFORCEbbb=F7.2)
62 FORMAT(26HWRIStbJINTbFORCEbbb=F7.2)
63 FORMAT(26HWIGHTbAPPLIEDb=THAND=F7.2)
64 FORMAT(36HNOTEbALbANGLESbAREbCLOCKWISEbPOSITbEbFROMbDOWNbVERTICAL)
   IF(SENSSE SWITCH1)7,8
7 G0 T0 6
8 PRINT 91
91 FORMAT(45HSENSEbSWITCHb1bSENSEbREADb=DATAbPRESSbSTART)
   PRINT 92
92 FORMAT(28HSENSEbSWITCHb2bOFFbTbFINISH)
   ST0P
   IF(SENSSE SWITCH 2)6,4
4 ST0P
END

INPUT DATA CASE(A)
0.0 0.8029 0.8029 6.997 14.53 14.53 60.85 33.8 33.8
1.0 0.792 0.33 0.14 0.13 0.07 0.127 0.106 0.10
0.458 0.33 0.167 0.0 0.0 32.2

INPUT DATA CASE(B)
0.2 0.8029 0.8029 6.997 14.53 14.53 60.85 33.8 33.8
1.0 0.792 0.33 0.14 0.13 0.07 0.127 0.106 0.10
0.458 0.33 0.167 100.0 0.0 32.2
INPUT DATA

This input data appears on three separate computer cards, designated here as I, II and III.

I (i) Angles of links to the downward vertical (radians)
(a)&(b) $\theta_1=0.0$ $\theta_2=0.8029$ $\theta_3=0.8029$

(ii) Angular velocity of links (radians/second)
(a)&(b) $\omega_1=6.997$ $\omega_2=14.53$ $\omega_3=14.53$

(iii) Angular acceleration of links (radians/second squared)
(a)&(b) $\alpha_1=60.85$ $\alpha_2=33.8$ $\alpha_3=33.8$

II (i) Link lengths (feet)
(a)&(b) $l_1=1.0$ $l_2=0.792$ $l_3=0.33$

(ii) Link masses (pounds mass)
(a)&(b) $m_1=0.14$ $m_2=0.13$ $m_3=0.07$

(iii) Radius of gyration of links (feet squared)
(a)&(b) $k_{G1}=0.127$ $k_{G2}=0.106$ $k_{G3}=0.10$

III (i) Distance of link centre of gravity from its pivot point (feet)
(a)&(b) $l_{G1}=0.458$ $l_{G2}=0.33$ $l_{G3}=0.167$

(ii) Load at hand (pounds force)
(a) $W=0.0$ (b) $W=100.0$

(iii) Angle of application of load at hand (radians)
(a)&(b) $\theta_4=0.0$

(iv) Acceleration due to gravity (feet/second squared)
(a)&(b) $g=32.2$
**COMPUTER OUTPUT**

The computer output was in the form of punched cards which are then "listed", giving the following results:

**CASE (A)**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow Velocity</td>
<td>7.00</td>
</tr>
<tr>
<td>Angle</td>
<td>1.57</td>
</tr>
<tr>
<td>Wrist Velocity</td>
<td>7.45</td>
</tr>
<tr>
<td>Angle</td>
<td>1.63</td>
</tr>
<tr>
<td>Finger Tip Velocity</td>
<td>11.46</td>
</tr>
<tr>
<td>Angle</td>
<td>2.37</td>
</tr>
<tr>
<td>Elbow Acceleration</td>
<td>78.10</td>
</tr>
<tr>
<td>Angle</td>
<td>0.89</td>
</tr>
<tr>
<td>Wrist Acceleration</td>
<td>247.31</td>
</tr>
<tr>
<td>Angle</td>
<td>0.94</td>
</tr>
<tr>
<td>Finger Tip Acceleration</td>
<td>411.38</td>
</tr>
<tr>
<td>Angle</td>
<td>0.91</td>
</tr>
<tr>
<td>Shoulder Torque</td>
<td>0.14</td>
</tr>
<tr>
<td>Elbow Torque</td>
<td>0.05</td>
</tr>
<tr>
<td>Wrist Torque</td>
<td>0.02</td>
</tr>
<tr>
<td>Shoulder Joint Force</td>
<td>2.17</td>
</tr>
<tr>
<td>Angle</td>
<td>1.34</td>
</tr>
<tr>
<td>Elbow Joint Force</td>
<td>4.05</td>
</tr>
<tr>
<td>Angle</td>
<td>2.05</td>
</tr>
<tr>
<td>Wrist Joint Force</td>
<td>2.80</td>
</tr>
<tr>
<td>Angle</td>
<td>2.37</td>
</tr>
<tr>
<td>Weight Applied at Hand</td>
<td>0.0</td>
</tr>
<tr>
<td>Angle</td>
<td>0.0</td>
</tr>
</tbody>
</table>

NOTE ALL ANGLES ARE CLOCKWISE POSITIVE FROM DOWNVERTICAL
<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow Velocity Angle</td>
<td>1.57</td>
</tr>
<tr>
<td>Wrist Velocity Angle</td>
<td>1.63</td>
</tr>
<tr>
<td>Finger Tip Velocity Angle</td>
<td>2.37</td>
</tr>
<tr>
<td>Elbow Acceleration Angle</td>
<td>0.89</td>
</tr>
<tr>
<td>Wrist Acceleration Angle</td>
<td>0.94</td>
</tr>
<tr>
<td>Finger Tip Acceleration Angle</td>
<td>0.91</td>
</tr>
<tr>
<td>Shoulder Torque</td>
<td>0.14</td>
</tr>
<tr>
<td>Elbow Torque</td>
<td>0.05</td>
</tr>
<tr>
<td>Wrist Torque</td>
<td>-32.98</td>
</tr>
<tr>
<td>Shoulder Joint Force Angle</td>
<td>1.34</td>
</tr>
<tr>
<td>Elbow Joint Force Angle</td>
<td>2.05</td>
</tr>
<tr>
<td>Wrist Joint Force Angle</td>
<td>2.37</td>
</tr>
<tr>
<td>Weight Applied at Hand Angle</td>
<td>100.00</td>
</tr>
</tbody>
</table>

Note: All angles are clockwise positive from down vertical.
INCREMENTAL STEP DISPLACEMENT,

VELOCITY AND ACCELERATION ANALYSIS
INCREMENTAL STEP DISPLACEMENT, VELOCITY AND ACCELERATION ANALYSIS

Consider a typical segment of the high speed camera film during the hammering motion. The time interval between reference steps is taken in the limit to correspond to an incremental step change in displacement, velocity and acceleration during the motion as shown below:

The velocity at station "n" is given by the rate of change in distance around n with respect to the time interval to travel this distance.

Velocity

\[ V_n = \frac{S_n + a - S_n - a}{2 \Delta t} \]  

(1)
The acceleration at station $n$ is given by the rate of change in velocity around $n$ with respect to time.

$$\text{Acceleration } A_n = \frac{V_n - a - V_n + a}{2 \, dt}$$

$$= \frac{(S_n + b - S_n) - (S_n - S_n - b)}{2 \, dt} \frac{2 \, dt}{2 \, dt}$$

$$A_n = \frac{S_n + b - 2S_n - S_n - b}{4 \, dt^2} \text{ \quad \text{(2)}}$$
APPENDIX 10

TYPICAL FOREARM FLEXION

MUSCLE GROUP

DYNAMIC ANALYSIS
CALCULATIONS USED IN FOREARM FLEXION DYNAMIC ANALYSIS

For this analysis of forearm flexion it is assumed that the biceps muscle alone is producing the limb elbow movement.

During the analysis the biceps muscle force, muscle lengths and muscle velocities of shortening are calculated. These results were then used to produce the Biceps Muscle Velocity-Force and Force-length curves as in Figures 6.51 and 6.52 respectively.

CALCULATION OF BICEPS MUSCLE FORCE FOR REFERENCE FRAME 24 in APPENDIX 3 OR STEP NO. 20 in DRAWING LIMBO 6.

The Muscle velocity of shortening is computed directly from the wrist velocity (relative to elbow joint) and the respective lever arm distance from the elbow joint. Similarly the Muscle force is computed directly from the forearm torque divided by the lever arm distance of the muscle force from the elbow joint.
Velocity of Shortening $V_{BICEPS} = \frac{SLB}{SL1} \cdot V_{DC}$

(approximate method)

$$= \frac{2}{11} \cdot 2.86 \times 12$$

$$= 6.25 \text{ in/sec.}$$

Forearm Torque $T2 = \text{Biceps Force} \times (FBICEPS) \times SXMB$

Therefore $FBICEPS = 4.5\phi \times \frac{12}{2}$

$$= 27 \text{ lbs.}$$
Accurate calculation of Muscle Velocity of Shortening involving calculation of the angle between the line of action of the Biceps force $V_{BICEPS}$ and the forearm angular velocity component $VELMV$ as shown below.

\[ \text{THELB} + \text{THB1} + 90 - X = 180^\circ \]
\[ X = \text{THELB} + \text{THB1} - 90 \]

\[ VELMV = \text{THV2} \times \text{SXMB} \]

\[ V_{BICEPS} = VELMV \times \cos(X) \]
CALCULATION OF BICEPS MUSCLE LENGTHS

Biceps muscle lengths were measured using graphical methods. This involved the plotting of elbow angle from a minimum 25 degrees to a maximum 170 degrees range of forearm flexion, and included the intermediate upward motion step positions.

SHOULDER - (A)

MINIMUM LENGTH
BICEPS 9.75"

SL1 = 11.5"

MAX LENGTH
BICEPS 13.5"

12.5 ASSUMED BICEPS REST LENGTH

117° ELBOW ANGLE

LINE OF ACTION OF BICEPS FORCE ON FOREARM
The biceps muscle length SLMB (including tendons) has an estimated rest length of 12.5 inches by comparison with results in Figure 2.338 and this corresponds to the anatomical forearm rest position of 117 degrees elbow angle.

This allows an additional 1 inch stretching to achieve 170 degrees of elbow flexion. Thus the 9.75 inch minimum muscle length allows for up to 2.75 inches of biceps shortening.
### TABLE OF UPWARD HAMMER SWING BICEPS MUSCLE CALCULATIONS

<table>
<thead>
<tr>
<th>REF.FRAME NO.</th>
<th>THV2 (RADIANS)</th>
<th>THELB (DEGREES)</th>
<th>VBICEPS INS/SEC</th>
<th>SLMB (INS)</th>
<th>T2 (FT-LBS)</th>
<th>FBICEPS (LBS)</th>
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</table>

Where

- **THV2** is Forearm Velocity relative to elbow joint
- **THELB** is Elbow (flexion) angle
- **VBICEPS** is Biceps Velocity of shortening
- **FBICEPS** is Biceps Muscle force
- **T2** is Forearm Torque about elbow joint
- **SLMB** is Biceps muscle length including tendons obtained by graphical methods.