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Journal of the American Orthotic and Prosthetic Association

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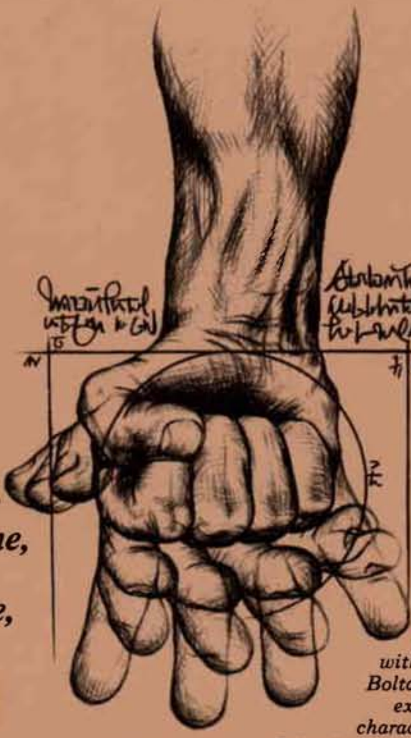

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A TERMINAL QUESTION

ROBERT RADOCY[1]
RONALD E. DICK[1]

Terminal devices for upper-limb prostheses, mechanically operated "hooks", have remained basically unchanged for more than 25 years. This lack of innovation has severely limited the options available to upper-limb amputees when seeking a terminal device suited to their particular set of needs. Up to now, the choice available to upper-limb amputees has been primarily variations of the voluntary-opening split-hook. This particular design, and its variations, has been prescribed often in an effort to meet all of the needs of the majority of upper-limb amputees, regardless of the level of the amputation.

Even the most superficial examination reveals how inadequate these prescriptions have met the needs of the patient. The more popular model of hooks provides the pinching action analogous to that of forceps, and the gripping strength is limited to the power provided by rubber bands or springs (approximately four lbs per rubber band). The currently available voluntary opening hooks have performed admirably in light duty applications for bilateral amputees, but, have proven less than adequate when used for vigorous activities, since tools and other objects tend to be forced out when pressure on the hook fingers exceeds the rather limited capacity of the rubber bands. Perhaps a voluntary-closing device might be better suited to the demands of strenuous work and recreational activities.

In spite of the fact that some rather sophisticated voluntary-closing designs have been offered in the past by both government supported and privately supported groups, voluntary-opening devices have been provided to an over-whelming majority of upper-limb amputees for many years. Perhaps it is time to evaluate scientifically the successes and failures, be as they may, of the terminal devices that have been available. It is the purpose of this paper to re-examine this very important issue in the light of the experiences of upper-limb amputees during the past 25 years, and re-introduce the debate that might be called "The Terminal Question."

First, it is necessary to understand just how the two systems operate. The voluntary-opening system is one in which the amputee, utilizing relative motion between parts of the human body through a harness-and-cable system opens the fingers of a mechanical terminal device by overcoming a closing, biased force. The voluntary-closing system is one in which the amputee, utilizing a harness and cable, closes the fingers of a mechanical terminal device by overcoming an opening biased force. The two systems thus are exactly opposite in operation.

The fundamental problem with the conclusions of the past debate over the "Terminal Question" was that, for many, the question predicated a single answer: voluntary-opening or voluntary-closing? The lessons the the past have made it apparent that it is more appropriate to evaluate the merits of each system in relationship to the needs and capabilities of the specific segments of the upper-limb amputee population rather than to design a single system which must be effective for all segments of the upper-limb amputee population.

Studies published in the 1970's have estimated the total upper-limb amputee population to be approximately 100,000 persons, (2, 3, 4, 6, 8). Of these, approximately three percent are bilateral and approximately 60 per cent are below-elbow unilateral amputees. These estimates are important inasmuch as they indicate that although unilateral below-elbow amputees represent the majority of the upper-limb amputee population they are, for the most part, wearing the same terminal device as the bilateral amputee or the above-elbow unilateral amputee.

Since the capabilities of bilateral above-elbow amputees, and below-elbow amputees are fundamentally different, the lack of a diverse offering of terminal devices forces amputees to rely on the same voluntary-opening "standard hook." For example, a below-elbow amputee retains the functions provided by the elbow joint and, thus, possesses considerably more "leverage" than the above-elbow amputee. However, the weak and ineffective gripping potential of the voluntary-opening split-hook equalizes the potentials of the two different types of amputations. That is, the below-elbow amputee has no more potential for gripping strength than the above-elbow amputee. Conversely, with a voluntary-closing terminal device, gripping strength increases with the amount of the residual limb. Thus, a wrist disarticulee has greater capability than a 4-inch below-elbow amputee, or an above-elbow amputee. This lack of innovation in terminal device design is as responsible for the degree of disability experienced by the majority of the upper-limb, unilateral below-elbow amputees as the nature of the amputation itself.

Advances have been made in externally powered terminal devices, especially those controlled by myoelectrical signals, but the age of bionics is still on the horizon and no realistic advances for the amputee interested in engaging in strenuous, vigorous activities can be expected in the near future. In fact, at this time, shoulder disarticulees and other patients with severe limb deficiencies can be expected to be the group that could derive the most benefit from externally powered prostheses. What is needed now is a useful option for the majority of the upper-limb amputee population, the unilateral below-elbow amputee. It is important to remember that disuse of the muscles of the residual limb causes atrophy. The greater the length of the residual limb, the greater the need for a muscle powered terminal device.

A literature review revealed that several committees, panels, and books have attempted to answer the "Terminal Question." In *Human Limbs and Their Substitutes* (5), printed in 1954, which is considered by many professionals and educators to be the most definitive text on the subject of artificial limbs, the following conclusions were made regarding the advantages and disadvantages of voluntary-opening and voluntary-closing terminal devices:

1. "Prehension, or the ability to grasp, is the primary function to be sought."
2. Voluntary-opening terminal devices have the advantages of simplicity and do not require a locking device to maintain grip, but voluntary-opening terminal devices have no continuous, progressive range of force controlled directly by the amputee. They are totally insensitive and lack neuromuscular control. Spring tension must be overcome in every operation, and they represent a direct opposite to the normal action of prehension. A living hand and arm does not relax to grasp and then contract to release.

In light of the above criticisms, one wonders why voluntary-opening terminal devices have enjoyed so much popularity and why other designs have not replaced it. The reason is that voluntary-closing devices of that period had problems of their own. However, objections centered around the poor engineering of the existing voluntary-closing terminal devices, and not the action itself. In spite of shortcomings in the existing voluntary-closing terminal devices, the authors concluded:

1. "Yet the voluntary-closing prothesis, if properly developed, offers the possiblity of active amputee control over the amount of grasping force exerted, of furnishing automatic locking of the grasp, and of accommodating the amputee with functional action of the kind found in the natural arm and hand."
2. Finally: "When weighing the considerations, it is apparent that the voluntary-closing terminal devices present the most desirable features, provided only that the engineering problems can be worked out satisfactorily."

The "Advisory Committee on Artificial Limbs" (5) was formed in 1947 to, among other objectives, analyze upper-limb prostheses and to propose solutions to existing engineering problems. The committee, an assembly of professionals, "decided to use the voluntary-closing action in searching for improvements in terminal devices." This committee accepted a set of design criteria which resulted in the development of the APRL hook which included a cam-quadrant clutch, and a two-position thumb. Unfortunately, this new terminal device was unreliable, clumsy to operate, and difficult to maintain in the production model. The failure of these terminal devices is overshadowed by the failure of this committee to analyze and evaluate their mistakes and failure to continue development of voluntary-closing devices. Virtually all research and development in mechanically operated terminal devices ceased at this time and has remained so until recently.

It is important to recognize that the past failures in the design of voluntary-closing terminal devices had been *due to* engineering problems resulting from a conventional set of design criteria and subsequent perception of performance, and not due to the action itself. So, the "Terminal Question" is broader in scope and much more complex than voluntary-opening vs. voluntary-closing. In order to answer the "Question," we must re-evaluate accepted criteria of terminal device design with regard to the specific needs and capabilities of specific segments of the upper-limb amputee population.

In the past, many designs for complete mechanical hands have been proposed. A lack of structural integrity, extreme complexity, and low reliability made these unfit to meet the demands of an active lifestyle. The V.C. APRL hand, the V.C. Miracle hand, the Pecorella V.C. hand, the Becker, and the Trautman V.C. hand are notable examples.

Patent drawings of some of these early mechanical hands such as the Lohmann hand of the 1950's, and the Pecorella of 1950, illustrate the various systems and structural variations designers have used. However, the most predominant design of V.O. and V.C. hooks has been the split hook. The split hook is illustrated by Hosmer-Dorrance hooks, the APRL hook, the V.O. Northrop, the V.O. David, the V.O. Thornton, and the Trautman devices. Since a primary consideration in the design of terminal devices is prehension, it would seem reasonable to consider other hook designs that may represent improvements over the conventional split-hook. For example, the L.A. Caron hook, 1913, and the D.C. Mollenhour, 1947, both attempt to emulate the action of the human forefinger and thumb as opposed to the forceps action of the split hook, and therefore merit consideration.

Two other more exotic designs are the Multiprise hook and the Bottomley Four-bar Link hook. Past evaluations of these devices stated that they had the advantage of prehension over the existing V.O. terminal devices and that their unusual structure was due to an attempt to improve lateral strength characteristics, (5).

With the benefit of this historical perspective, it is to be expected that basic design criteria, and the direction for future development should be readily apparent. But, conventional wisdom and tradition have a way of hanging on in spite of recommendations to the contrary, (5). The

Panel on Upper-limb Prosthetics, 1977 (1), a panel of professionals, met and concluded to perpetuate some of the past mistaken assumptions regarding the design of upper-limb prosthetics. The following is a review and critique of a few of these conventional assumptions:

First and foremost, it is paramount that exclusion of input by the general upper-limb amputee population from initial design considerations be stopped. How can terminal devices be adequately designed without first consulting each specific segment of the upper-limb amputee population with respect to their needs and capabilities? Traditionally, the devices have been designed and prototyped and then the amputees have been asked to evaluate them or a few so-called representative examples of amputees have acted as consultants during the design process and the subsequent evaluation. This represents a fundamental error in research methodology. Finally, too much effort has been invested in trying to discover the panacea of terminal devices, the one and only best terminal device of all. Consideration must be directed toward the specific needs and capabilities of each segment of the upper-limb amputee population. Our review and critique will proceed from this perspective.

1. The highest priority recommendation by the 1977 "Panel on Upper-limb Prosthetics" (1) was: "It is strongly recommended that the delivery of available technology and techniques (e.g., below-elbow myoelectric prostheses) be promoted actively." This is a perfect example of the result of excluding the input of the general upper-limb amputee population from these deliberations, and the subsequently wasteful and expensive "barking up the wrong tree" development program.
   Our interviews with below-elbow amputees have revealed strong opposition to this recommendation, due to the inability of myoelectrics to withstand the elements, the rigors of the vigorous physical activities that below-elbow amputees are capable of, lack of feedback, and the inconvenience of the battery pack on extended hunting and fishing trips.
2. Weight is an obvious consideration. Conventional wisdom tells us that a prosthetic terminal device should be as light as possible. Perhaps, a better set of criteria would include optimum weights for artificial limbs and terminal devices. For example, an above-elbow amputee might require a lightweight device to prevent fatigue, but a below-elbow amputee might require the therapueutic aid of a heavier terminal device in order to restore and maintain the tone of upper-arm musculature, and to provide balance bilaterally to prevent spinal misalignment, (7).
3. Overall size criteria, in the past, have led to the development of terminal devices that are smaller than the normal human hand. Is it possible that the small size lacks the support of amputees? The small size of the "standard hook" limits the size of objects that can be handled, and the bilateral asymmetry and vestigal nature of the abnormally small size may be psychologically demeaning to the wearer. These are questions that need to be put to the amputees.
4. It has been commonly assumed for many years that any "properly" designed V.C. terminal devices should include some sort of automatic locking device. Since a normal human hand cannot lock in place, why should a terminal device? The cam-quadrant lock of the APRL V.C. was rejected by the general amputee population due to frequency and costs of maintenance, lack of durability and reliability, due to poor quality control, high costs, and because it tended to hang up on hard objects since some compression of the fingers was necessary to release the cam-quadrant clutch. Perhaps this criterion requiring a lock should be re-examined. It appears that the belief that all voluntary closing hooks "needed" a locking

device orginated during the time that cineplasties were popular. Genevieve V. Reilly's paper in Physical Therapy Review in May, 1951 stated "The prosthesis must be constructed to provide for special acts of strength far beyond the power of the plastic "motor" itself. This problem is solved through the medium of a lock on the hand." Since conventional figure-eight and figure-nine harnesses do not have the limitations of the cineplasty, these so-called special acts of strength can be accomplished without a lock. A voluntary closing terminal device can easily provide a grasping strength in excess of a normal human hand. Conscious effort in grasping can increase sensitivity and improve muscle tone. However, a manual locking system could provide convenience when prolonged tool use or carrying is necessary, without having the disadvantage of eliminating the rapid release reflex characteristic of automatic locks. Consideration should be redirected toward the use of safe, reliable, and convenient, manually-operated locks.

5. Cosmesis will always be an important factor to consider in prosthetic design. Maybe too much emphasis has been placed on imitative cosmesis in attempts to create a living likeness of a human hand. The smooth surface of the split-hook and its balanced appearance has had much to do with its success. It is not necessary to sacrifice function for cosmesis if a terminal device is designed to be pleasing to the eye like any other precision tool.
6. Versatility to function in a wide range of activities is of utmost importance. Emphasis should be placed on the elimination of the use of special adaptors. Tasks involving complex sequence of events are simply impractical if the amputee has to change special adaptors when the use of a new tool is called for.
7. Finally, the most basic criterion is reliability. A terminal device that does not stand up to shock, torque and abuse from elements is worse than useless; it is a source of frustration and danger. This very important factor is a significant reason to renew consideration of mechanically-operated terminal devices.

There are, of course, many more factors to consider in developing design and performance criteria for prosthetic terminal devices. These have been examples to demonstrate that it is time to face the fact that we do not know all there is to know about designing mechanically-operated prostheses, and that our best source of input about needed changes will come from the amputees themselves. All we need to do is ask.

SUMMARY

We must recognize that the needs and capabilities of upper-limb amputees vary, and that due to a lack of innovation during the last 25 years no successful alternatives have been developed to satisfy the special needs of each segment of the upper-limb amputee population. Outdated design criteria persist in spite of amputee dissatisfaction with the performance of available terminal devices. Since general amputee input has been excluded from the initial design process, it is imperative that a representative sample of the upper-limb amputee population be subjects in a research program designed to establish valid and grounded criteria for the design and development of upper-limb prostheses. This much needed information will finally initiate the development of specific devices to satisfy the needs of specific segments of the upper-limb amputee population, rather than to continue the past practice of trying to develop a panacea for all amputees. Such research will not only correct a long standing and fundamental error in the research process, it will begin the process of designing and developing prostheses that will encourage upper-limb amputees to live active, independent and more productive lives.

FOOTNOTES

[1]Therapeutic Recreation Systems, 2860 Pennsylvania Avenue, Boulder, Colorado, 80303

The "Terminal Questions," by Robert Radocy and Ronn E. Dick was sponsored by Therapeutic Recreation Systems, Inc.

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Orthotics and Prosthetics, Vol. 35, No. 1, pp. 7-28, March 1981

The Biomechanics of Control in Upper-Extremity Prostheses

CRAIG L. TAYLOR, Ph.D.[1]

This editor, while consulting the work of Craig Taylor and his associates in the course of developing material for the training of orthopedic surgery residents, was so struck by the fact that, although published 25 years ago, the principles and philosphy expressed in the following article are applicable in every sense today, and because of the clarity of the article, felt that it should be republished, if for no other reason than to help all of us in retaining the persepctive needed to meeting the requirements of the upper-limb amputee. This article appeared originally the the September 1955 issue of "ARTIFICIAL LIMBS."

In the rehabilitation of the upper-extremity amputee, structural replacement by prosthetic arm and hand is an obvious requirement, and it poses a comparatively easy task; functional replacement by remote control and by substitute mechanical apparatus is more elusive and hence infinitely harder. For the purposes of functional utility, remaining movements of upper arm, shoulder, and torso must be harnessed, and use must be made of a variety of mechanical devices which amplify remaining resources by alternators, springs, locks, and switching arrangements. The facility of control attained through this apparatus is the key to its ultimate value.

The future of upper-extremity prosthetics depends upon an ever-increasing understanding of the mechanics of the human body by all who minister to the amputee—prosthetist, surgeon, and therapist alike. It must always be stressed that the final goal is an amputee who can function. Too often there is a tendency to put undue faith in the marvels of mechanism alone, when in fact it is the man-machine combination that determines performance. It is in this broad frame of reference that the biomechanical basis of upper-extremity control must be approached.

Prosthetics Anthropometry Surface Landmarks

If successful control is to be obtained, the various components of the prosthesis must be positioned with a good degree of accuracy. To do so requires reference points on the body, of which the most satisfactory are certain bony landmarks. Most of these skeletal prominences protrude to such an extent that location is easily possible by eye. Others require palpation, and this method should be used to verify observation in every case. The bones most concerned in upper-extremity anthropometry are the clavicle, the scapula, the humerus, the ulna, and the seventh cervical vertebra. Surface indications of protuberances, angles, or other features of these bones constitute the landmarks, the locations and definitions being given in Figure 1.

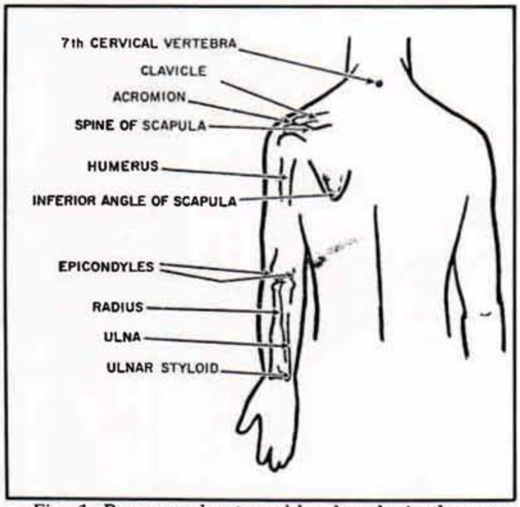

Fig. 1. Bones and external landmarks in the upper extremity. Definitions: *seventh cervical vertebra,* most prominent vertebra in the neck region; *acromion,* extreme lateral edge of the bony shelf of the shoulder; *inferior angle of scapula,* lowest point on shoulder blade; *epicondyles,* lateral and medial bony points at the pivot of the elbow; *ulnar styloid,* projecting point on little-finger side of wrist.

Arm and Trunk Measurements

The typical male torso and upper extremity are shown in Figure 2, which, together with Table 1, was derived from average measurements on Army personnel (16). Such an average form serves to establish harness patterns and control paths. The arm, forearm, and epicondyle-thumb lengths[2] constitute the basis of sizing prostheses (2). Arm length places the artificial elbow; forearm length locates the terminal device. The epicondyle-thumb length is an important over-all sizing reference because in the unilateral arm amputee it is customary to match hook length (and, in the case of the artificial hand, thumb length) to the length of the natural thumb (Fig. 3). The bilateral arm amputee can be sized from body height by means of the Carlyle formulas (3), which employ factors derived from average body proportions.

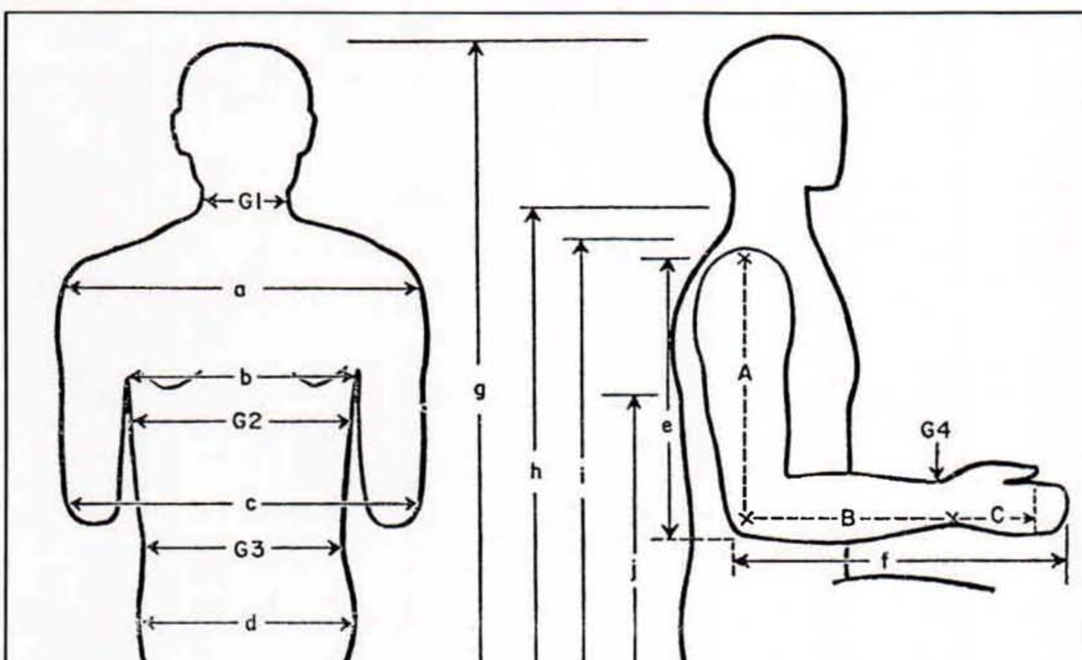

Fig. 2. Basic anthropometry of the male torso and upper extremity. See Table 1.

Table 1

AVERAGE BODY MEASUREMENTS
(See Figure 2)

Lengths	
a, bideltoid	17.9 in.
b, cross-back width	14.8
c, elbow breadth	17.5
d, hip breadth	14.0
e, shoulder-elbow length	14.3
f, forearm-hand length	18.7
g, sitting height	35.8
h, cervical height	26.1
i, supersternale height	23.0
j, height of inferior angle of scapula (approximately at midpoint of sitting height)	
Girths	
G1, neck	14.5
G2, chest	36.3
G3, waist	30.7
G4, wrist	6.7
Prosthetic Dimensions	
A, acromion-epicondyle length	13.2
B, epicondyle-styloid length	9.9
B + *C*, epicondyle-thumb length	14.4

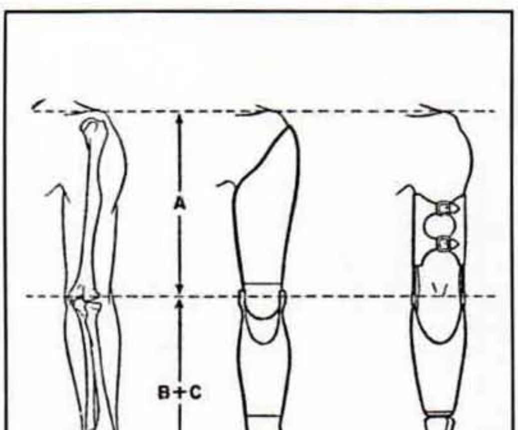

Fig. 3. Correct lengths for upper-extremity prostheses. In the unilateral case, hook length is made to coincide with normal thumb length, as is also the thumb length of the artificial hand. For bilateral arm amputees, $A = 0.19 \times$ (body height); $B + C = 0.21 \times$ (body height). After carlyle (3).

Functional Anatomy

The human torso, shoulder, and upper extremity are exceedingly complex structures. In any dealing with these elements of anatomy, therefore, it is desirable to sort out from the mass of detail those features important to the particular area of study and application. Where prosthetic controls are concerned, the mechanism of movement is the central subject of consideration. This functional anatomy treats of the aspects of bone, joint, and muscle structure that together determine the modes and ranges of motion of the parts. It is a descriptive science, and while to escape dependence upon nomenclature is therefore impossible, the purpose here is to convey a basic understanding of the operation of the upper-extremity mechanisms without undue use of specialized terminology. In any case, the reader should have abailabe basic anatomical references such as *Gray's Anatomy* (13) or kinesiology texts such as those of Steindler (17) and of Hollinshead (9).

Elementary Motions of the Upper Extremity

The geometry of each joint is complex, and most movements involve an interaction of two or more joints. Consequently, a motion nomenclature based on joint movements would be unnessarily complicated. More simply, the motion of each part upon its proximal joint may be described with respect to the principal planes which intersect at that joint. In this system, moreover, one may define a standard position in which the trunk is erect, the arms hang with their axes vertical, the elbows are flexed to 90 deg., and the wrist planes are vertical to assume the "shake-hands" position.

Figure 4 presents the angular movements possible in the three planes of space. The shoulder-on-chest, arm-on-shoulder, and hand-on-wrist actions take place through two angles, as if moving about a universal joint. Geometrically, the arm motions are more precisely defined by a

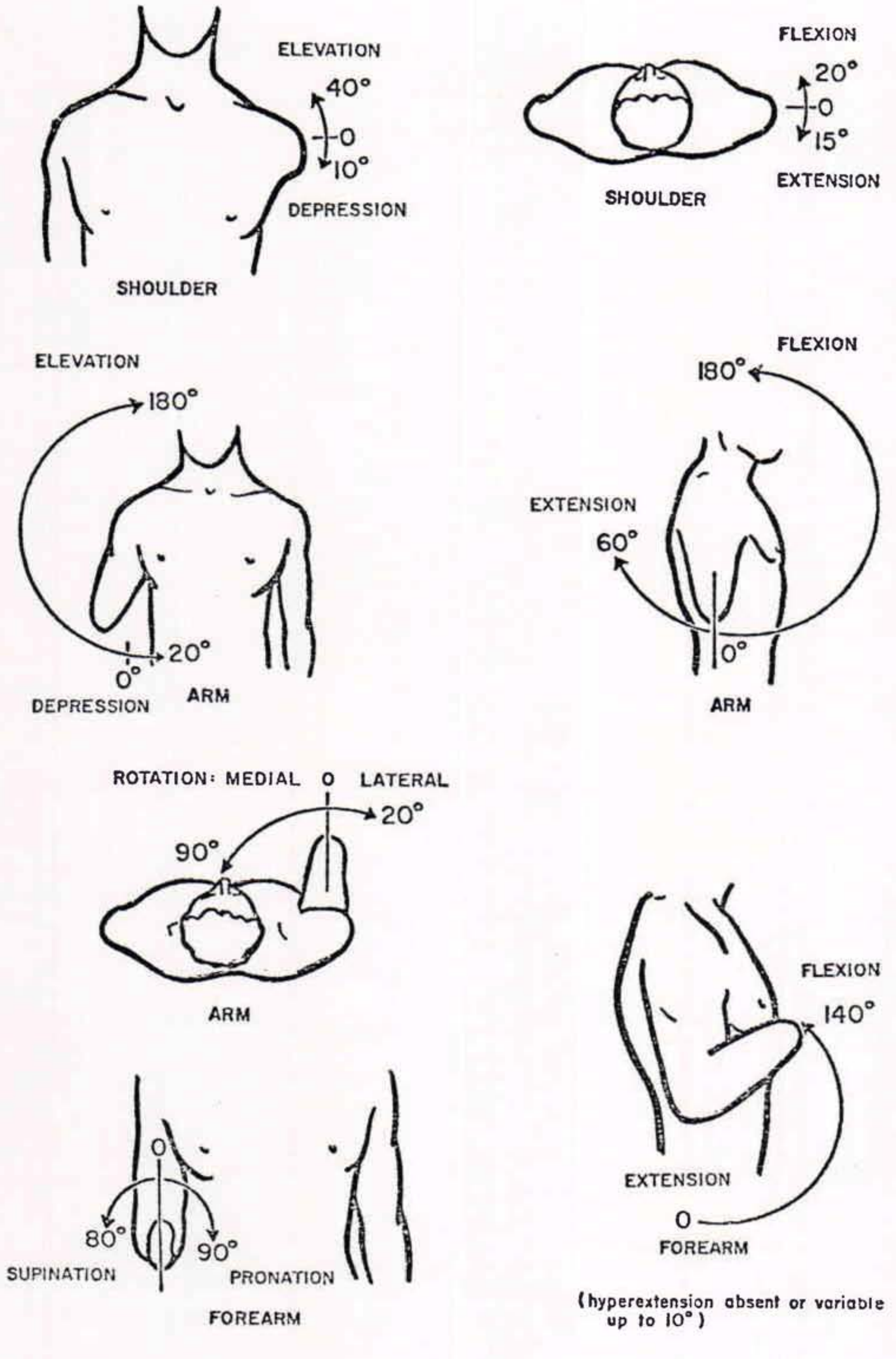

Fig. 4. Simplified movement system in the upper extremity. Wrist flexion is omitted since ordinarily it is not involved in upper-extremity controls.

spherical coordinate system where the segment position is given by longitude and colatitude angles. For descriptive purposes, however, the anatomical nomenclature is commonly used. It should be recognized that, for 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 simplified movement system depicted in Figure 4 is incomplete in many ways. Not included are such movements as twisting of the shoulder due to various scapular movements, anterior-posterior swings of the arm in positions of partial elevation, and the slightly conical surface of revolution of forearm flexion.[3] These details may, however, be ignored in the interest of the simplicity of description that is adequate for the purposes of upper-extremity prosthetics.

The Shoulder Girdle

Skeletal members and Joints

The scapula and clavicle are the chief bones making up the shoulder girdle. Secondarily, the proximal protion of the humerus may be included, since the close interarticulation of all three bones at the shoulder joint gives a considerable degree of coordinated activity among them and also extends to the complex as a whole the actions of many of the muscles inserting on the individual members.

Details of the skeletal anatomy involved are shown in Figure 5. There are in the system two joints and one pseudo joint. In the sterno-clavicular joint, the clavicle articulates with the sternum in a somewhat saddle-shaped juncture recessed in a concavity within the sternum. The biaxial surfaces permit movements in two planes. Ligaments crossing the joint prevent displacement of the clavicle anteriorly and laterally. The elevation-depression range is 50 to 60 deg., the flexion-extension range from 25 to 35 deg.

In the acromioclavicular joint, the distal end of the clavicle articulates with the scapula in an elliptical juncture which permits a ball-and-socket type of action The

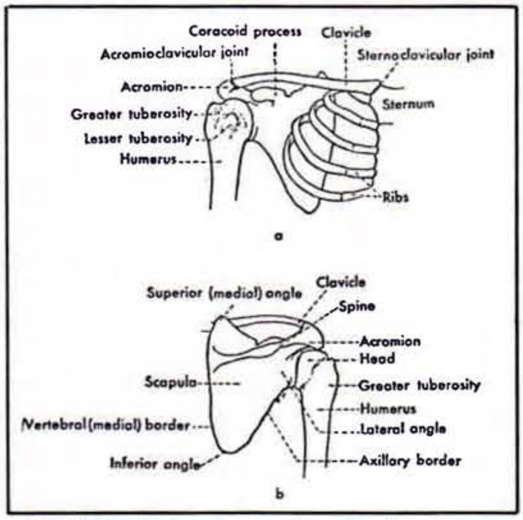

Fig. 5. Skeletal anatomy of the shoulder region. *a*, Anterior view. *b*, Posterior view.

acromio-clavicular ligaments bind the joint directly. Strong ligaments from the clavicle to the coracoid process give important additonal stabilization. The range of movement is small, being only about 10 deg. in the frontal and sagittal planes.

The pseudo joint, the scapulothoracic, is a muscular suspension which holds the scapula against the thoracic wall but which at the same time permits translatory and rotatory movements. A large factor in maintaining this joint in position is barometric pressure, which is estimated to act upon it with a force of 170 lb.

Muscles and Movements

The complex arrangement of bony elements is rivaled by the involved nature of the muscles of the shoulder girdle and by the intricate ways in which they act upon it. The schematic view of Figure 6 presents the fundamentals. Elevation of the shoulder is seen to be brought about principally by elevators and downward rotators of the scapula, such as the upper trapezius, the levator scapulae, and the rhomboids. Although the rhomboids assist in elevation, they do not contribute to upward rotation. Depression of the shoulder is mediated by muscles inserted on the scapula, the clavicle, and the proximal end of the humerus. Anteriorly the lower fibers of the pectoralis major, the pectoralis minor, and the subclavius, and posteriorly the lower trapezius and latissimus, act as depressors.

Rotation of the scapula upward (i.e., right scapula, viewed from the rear, rotates counterclockwise) or downward (i.e., right scapula, viewed from the rear, rotates clockwise) is brought about by a special combination of the elevators and depressors. As shown in Figure 6, two portions of the trapezius, together with the serratus, cause upward rotation. Conversely, the pectorals, the latissimus, and the rhomboids cooperate to cause downward rotation. As will be seen later (page 13), the mechanical principle of the couple applies in these rotatory actions upon the scapula.

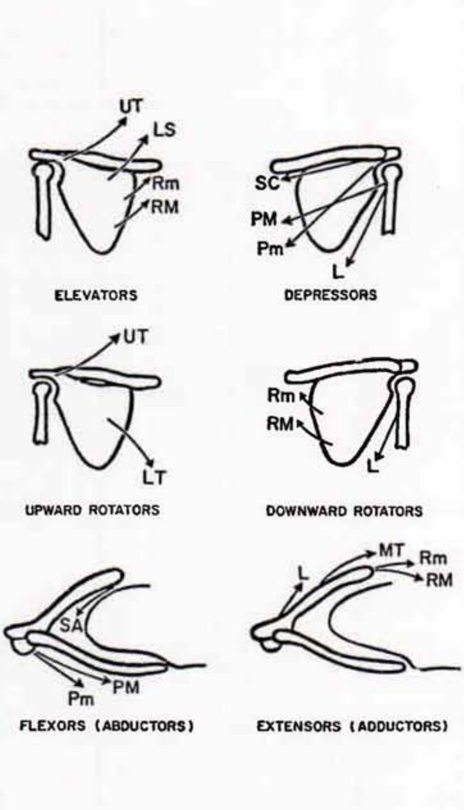

Fig. 6. Schematic kinesiology of the shoulder girdle. *L*, latissimus; *LS*, levator scapulae; *LT*, lower trapezius; *MT*, medial trapezius; *PM*, pectoralis major; *Pm*, pectoralis minor, *RM*, rhomboid major; *Rm*, rhomboid minor; *SA*, serratus anterior; *SC*, subclavius; *UT*, upper trapezius.

Flexion and extension of the shoulder involve as principal elements the abduction and adduction, respectively, of the scapula. The flexor muscles acting on the shoulder complex are the pectoralis major and minor, which swing the clavicle and acromion forward. The serratus anterior aids strongly by abducting the scapula. The extensors, placed posteriorly, include the latissimus, which pulls posteriorly and medially on the humerus, and the trapezius and rhomboids, which pull medially on the scapula.

The forward and backward shrugging of the shoulders with abduction and adduction, together with some upward and downward rotation of the scapulae, constitutes a major control source. Even in above-elbow amputees who use humeral flexion for forearm lift and for terminal-device operation at low elbow angles (page 22), scapular abduction is utilized for terminal-device operation at large angles of elbow flexion (e.g., when the terminal device is near the mouth). In shoulder amputees, both these operations depend wholly upon scapular abduction augmented by upward rotation.

THE ARM

The Humerus and the Glenohumeral Joint

The humerus, together with its joint at the shoulder, comprises the skeletal machinery of the arm. As noted in Figure 4, it is capable of flexion-extension, elevation-depression, and rotation upon its proximal joint. The glenoid cavity, a lateral process on the scapula, receives the spherical surface of the humeral head. The glenohumeral articulation is therefore of true ball-and-socket character. The fibrous joint capsule is remarkable in that it envelops the humeral head and the glenoid margins in complete but rather loose fashion, so that a wide range of movement is possible. To some extent barometric pressure, but to larger extent the musculature spanning the joint, is responsible for keeping the articular surfaces together in all angular positions. A group of muscles including the subscapularis, the supraspinatus, and the infraspinatus function principally in this holding action.

Muscles and Movements

The kinesiology of the arm is closely associated with that of the shoulder girdle, nearly all natural movements involving a coordinated movement between arm and shoulder. It is helpful, however, first to describe the pure movements of the arm. Schematics of the muscles acting upon the arm are presented in Figure 7. Elevation is effected by the lateral deltoid and the supraspinatus, depression by the latissimus, the pectoralis major, the long head of the triceps, and the teres major. In both actions, the contributions of individual muscles differ according to the angle of the arm. And it should be noted that, with insertions near the pivot point of the humeral head, the rotatory moments are proportionately small, thus accounting for the large number of muscles necessary to give adequate joint torques.

Arm flexion and extension are brought about by two groups of muscles. The biceps, the coraco-brachialis, the anterior deltoid, and the clavicular fibers of the pectoralis major mediate flexion, while the posterior deltoid, the long head of the triceps, the latissimus, and the teres major effect extension. Rotation of the arm depends upon muscles that insert on the surface of the humerus and then pass anteriorly or posteriorly around it to impart medial or lateral torsion. As would be expected, rotational forces are greatest when the arm hangs at the side; torque is reduced drastically when the arm is elevated over the head and the twisting angles of the muscles tend to disappear.

Combined Arm and Shoulder Movements

In most natural arm movements, such as arm elevation, arm flexion, forward reaching, and to-and-fro swings of the partially elevated arm, both arm and shoulder girdle participate. In full arm elevation of 180 deg., for example, 120 deg. are contributed by rotation of the arm on the glenohumeral joint, 60 deg. are contributed by upward rotation of the scapula (17,) In forward reaching, involving partial arm flexion, the shoulder flexes and the scapula abducts and rotates slightly. Properly managed, this motion, the common flexion control motion of both the above- and the below-elbow amputee (pages 19-22) can give marked gracefulness to prosthetic operation.

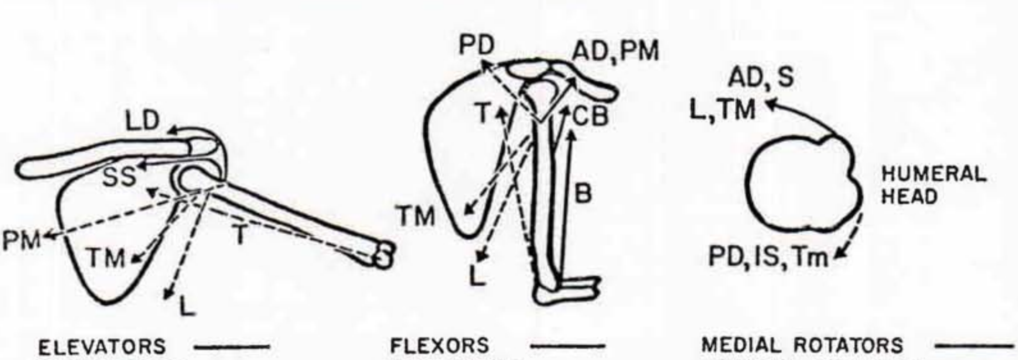

Fig. 7 Schematic kinesiology of the arm. *AD*, anterior deltoid; *B*, biceps; *CB*, coracobrachialis; *IS*, *infraspinatus*; *L*, latissimus; *ID*, lateral deltoid; *PD*, posterior deltoid; *PM*, pectoralis major; *S*, subscapularis; *SS*, supraspinatus; *T*, triceps; *TM*, teres major; *Tm*, teres minor.

The Forearm

Skeletal Members

The radius and ulna together constitute a forearm lever which can rotate about the elbow axis. By virtue of the arrangement at the proximal head of the radius and at the distal end of the ulna, the forearm can also carry out torsion about its longitudinal axis to produce wrist rotation. With the aid of the mobility at the shoulder and at the wrist, it is possible to place the hand in space in an almost unlimited number of positions. The skeletal anatomy of the elbow is shown in Figure 8, the articulations being the ulnohumeral and the radiohumeral. Participating in forearm rotation is the radioulnar joint at the wrist.

The ulnohumeral joint has an unusual stucture. The complex surfaces of articulation between ulna and humerus are such that the axis of rotation of the forearm is not normal to the long axis of the humerus. As the elbow is flexed or extended, therefore, the forearm does not describe a plane. Instead, the ulna swings laterally as the elbow is extended, until at full extension the cubital angle is about 170 deg. Nevertheless, only a small error is involved in considering the motion to be essentially that of a simple hinge with an axis of rotation perpendicular to ulna and humerus and allowing the ulna to swing through about 140 deg. of flexion.

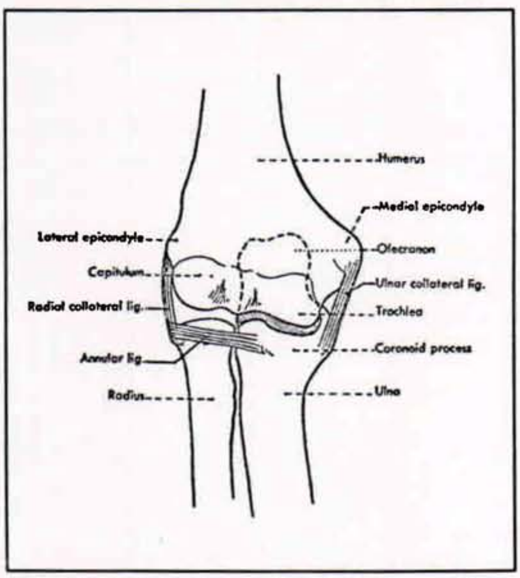

Fig. 8 The right elbow joint, viewed from in front. The thin capsular ligament is not shown. Note that the ulna, with its posteriorly projecting olecranon, forms a hinge joint with the humerus, while the head of the radius is free to rotate within the annular ligament.

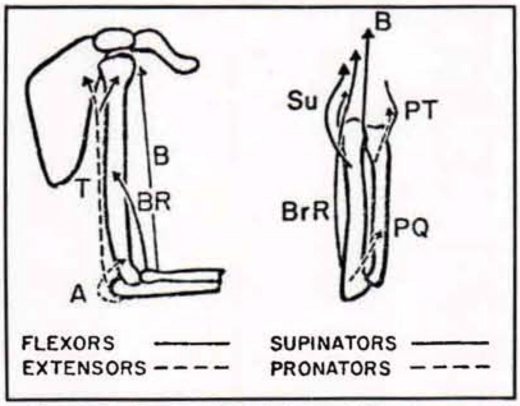

Fig. 9. Schematic kinesiology of the forearm. *A*, anconcus; *B*, biceps; *BR*, brachialis; *BrR*, brachioradialis; *PT*, pronator teres; *PQ*, pronator quadratus; *Su*, supinator; *T*, triceps.

In the radiohumeral joint, the slightly concave proximal end of the radius articulates with the hemispherical capitulum placed somewhat laterally on the anterior surface of the distal end of the humerus. The radius is free to move with the ulna through the complete range of flexion and, in addition, to rotate with forearm pronation and supination.

In the radioulnar joint, the distal end of the ulna forms a curved surface against which the radius opposes an articulating concavity. As the forearm goes through a pronation-supination range of about 170 deg., the radius "swings like a gate" about the distal end of the ulna.

Muscles and Movements

As shown in Figure 9, the musculature for providing forearm flexion and extension is comparatively simple, while that for pronation-supination is somewhat more involved. Flexion of the forearm is effected principally by the biceps, originating on the scapula and inserting on the radius, and by the brachialis, spanning the elbow from humerus to ulna. Secondarily, the brachioradialis and other muscles, originating distally on the humerus and coursing down the forearm, contribute to flexion. Extension is largely the function of the triceps, originating on both the scapula and humerus and inserting on the leverlike olecranon process of the ulna. A small extensor action is added by the anconeus.

Rotation of the forearm is a function of many muscles. Some, such as the supinator, evidently are designed for the purpose, while others, as for example the finger flexors, have different principal functions, the contribution to forearm rotation being only incidental. Figure 9 presents the major rotatory muscles only. Supination is mediated by the brachioradialis, the supinator brevis, and the biceps, pronation by the pronators quadratus and teres. Of great importance to upper-extremity prosthetics is the fact that rotation of the forearm is a function of total forearm length. With successively shorter stumps, not only are the rotation limits of the radius and ulna reduced, but also the contributions of muscles are eliminated as their insertions are sectioned.

Musculoskeletal Mechanisms

The upper extremity having been considered from the standpoint of functional and descriptive anatomy, attention may now be turned to a more mechanical view of its operations. Typical elements of mechanism in the upper extremity include joints (bearing surfaces), joint-lining secretions (lubricants), bones (levers and couple members), tendons (transmission cables), and muscles (motors). The arrangement of these elements makes up a complex machinery capable of such diverse activities as precise orientation in space, performance of external work, fine digital manipulations, and so on.

Typical Joint Mechanics

The elbow joint embodies the essential structures of diarthrodial joints. The bearing surfaces are covered with a thin layer of articular cartilage that is continuous with the synovial membrane lining the whole joint capsule. Subsynovial pads of fat serve to fill up the changing spaces that occur during movement of the joint (Fig. 10). It is believed that these fatty deposits serve as "pad oilers" to maintain the continuous film of synovial fluid over the articular surfaces (4). This fluid contains mucin (a glycoprotein which serves as a lubricant for the joint) and other material constituting a nutritional medium for the articular cartilage. Considerable uncertainty exists concerning the method of formation and distribution of the fluid to the joint, but its mechanical function is clear and the normal joint performs as a well-oiled bearing.

BONES AND THEIR MECHANICAL FUNCTION

The bones of the upper extremity, besides forming a support for soft tissue, provide a system of levers which makes the arm an important mechanism for the performance of gross work, such as lifting, slinging, and thrusting. The arm bones serve further as positioners of the hand, in which other, finer bones constitute the intricate articulated framework of the

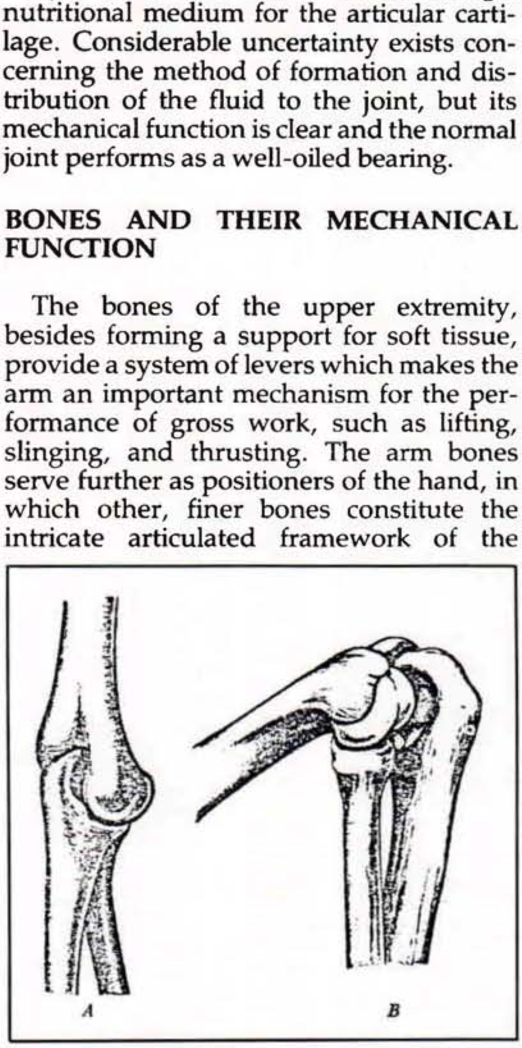

Fig. 10. Typical change in joint spaces with flexion extension, as revealed by the elbow. Redrawn from Steindler (17), after Fick. *A*, Gap of the medial border of the olecranon surface with elbow in extreme extension. *B*, Gap of the lateral border of the olecranon in extreme flexion.

manipulative mechanism. Two main features of bones merit discussion here—their internal composition and construction and their external shape and adaptations that permit them to serve as members of mechanical systems.

Internal Structure

There is much evidence that the gross internal structure of bone is eminently suited to withstand the mechanical stresses placed upon it by the compressive loads of weight-bearing, by the tensions of tendons and ligaments, and by the lateral pressures of adjacent tissues (4). The nature and orientation of the trabeculae in cancellous bone have, for example, long been held, in theory, to provide the maximum strength along the lines of major stresses. This idea, originally suggested by von Meyer, has been championed by many, including Koch, who carried out a stress analysis on the femur (12). Objections to the von Meyer theory have dealt largely with the frequent and incautious extension of the concept. It is now believed that genetic and growth factors determine the essential form and dimensions of bone. Mechanical stresses serve secondarily to mold and modify it to give added strength where stresses are greatest. One must grant from even a superficial examination of the internal structure of bone that Nature has done an admirable job of designing for maximum strength with minimum weight.

Members of Mechanical Systems

The second prinicpal feature of bones, that of serving as rigid members in a complex of mechanical systems, is the one that has engaged the most attention. it is surprising that the simple lever concepts of Archimedes have persisted in anatomy and kinesiology texts to the present day. Thus, the forearm-flexor system is said to act as a third-class lever, the extensor system as a first-class lever. Although these assertions are of course true, both of these systems are, in the more complete language of Newtonian mechanics, parts of force-couple systems in which equal and opposite components of force are transmitted

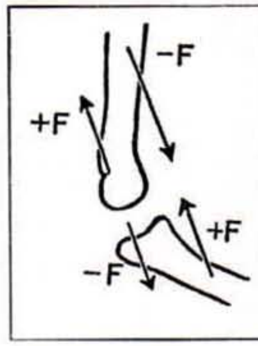

Fig. 11. Force couples at the elbow. Tensile forces in biceps and brachialis are associated with equal, opposite, and parallel forces through the joint.

through the bones and joints (Fig. 11). Elftman (7) has emphasized this view. The magnitude of the couple is given by the product of the force (either of the equal but opposite forces) and the distance between them, which also is numerically equal to the torque of the muscle force. The concept of the couple calls attention to the existence of the equal and opposite forces in joints and emphasizes the loads placed upon them by muscular work.

Another and more complicated application of the couple is seen in scapular rotation. Here, as described by Inman *et al.* (11) and as shown in Figure 12, the pull of the

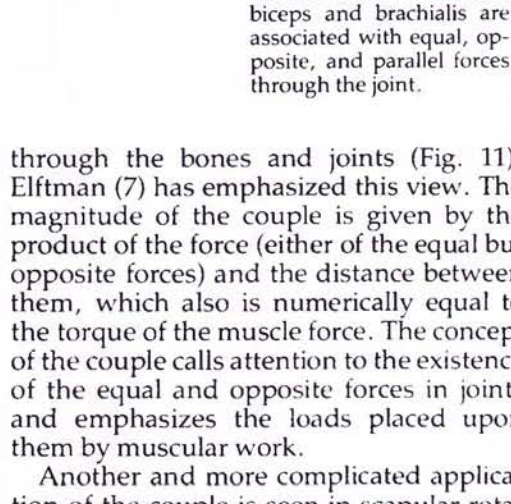

Fig. 12. Muscle forces acting on the shoulder, anterior view. The trapezius, acting diagonally, gives a supportive component, F_{y}, and a horizontal component, F_x, which together with the opposite force from the serratus, *S*, comprise an upward rotatory force couple on the scapula.

lower fibers of the serratus anterior upon the scapula is such as to give it upward rotation, while the thrust of the clavicle, acting through the acromioclavicular joint, holds a pivot for the rotation. Simultaneously, the pull of the upper trapezius fibers causes the clavicle to undergo angular rotation about the sternoclavicular joint. The result is that, at least through the first 90 deg. of arm elevation, the motion is shared by coordinated angular rotations of scapula, clavicle, and humerus. As a basic part of this rotatory action, the scapula acts as the moment arm of a force couple, the trapezius and serratus providing components of force which are equal and opposite.

TENDONS AND MUSCLES

The specific functions of tendons are to concentrate the pull of a muscle within a small transverse area, to allow muscles to act from a distance, and in some instances to transmit the pull of a muscle through a changed pathway. The mechanical importance of this tissue is nowhere more evident than in the arm, where a large degree of versatility of motion in the segment distal to each joint is preserved by "remoting" the action of muscles through slender, cablelike tendons over joints. By this means lines of pull are brought near the joint axes, thus providing a lever arm consistent with the tensile force of the muscle at all joint angles and also giving at low joint angles an increased angular motion for a given linear contraction. Other advantages of remoting the muscles are seen in the forearm and hand. In order to afford the variety and complexity of interdigital movements, many independent muscle units are necessary, and critical space problems are avoided because muscles such as the common flexors and extensors of the fingers are placed at some distance up the forearm.

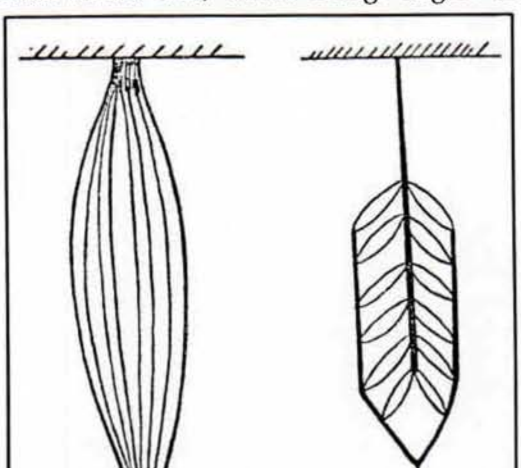

Fig. 13. Muscle fiber patterns. *A*, Fusiform. *B*, Bipinniform.

The predominant function of tendon as a tension member in series with muscle, which is a tension motor, is seen in early growth stages. An undifferentiated cellular reticulum of connective tissue is everywhere found in embryonic tissue. The parent cells are fibroblasts; they elaborate and extrude the collagenous material of which white fibers are made (4). At this point the presence of mechanical tensions in the tissue influences the rate, amount, and direction of the resultant fiber formation. At maturity the tendon is composed almost entirely of white collagen fibers, closely packed in parallel bundles, to form a cablelike strand. It is contained within a sheath which forms a loose covering lubricated continuously by a mucinous fluid to reduce friction with surrounding tissues.

Mutual adjustment of the characteristics of muscle and tendon is shown in many respects. The musculotendinous juncture varies with the arrangement of the muscle fiber. It shows a simple series arrangement for fusiform muscles like the biceps, or it comprises a distributed attachment zone by continuation of the tendon into intramuscular septa where pinniform fibers may insert (Fig. 13). In some unexplained way the relative lengths of muscle and associated tendon are so composed that the shortening range of the muscle is that necessary to move the segment distal to the joint through its maximum range (8). The

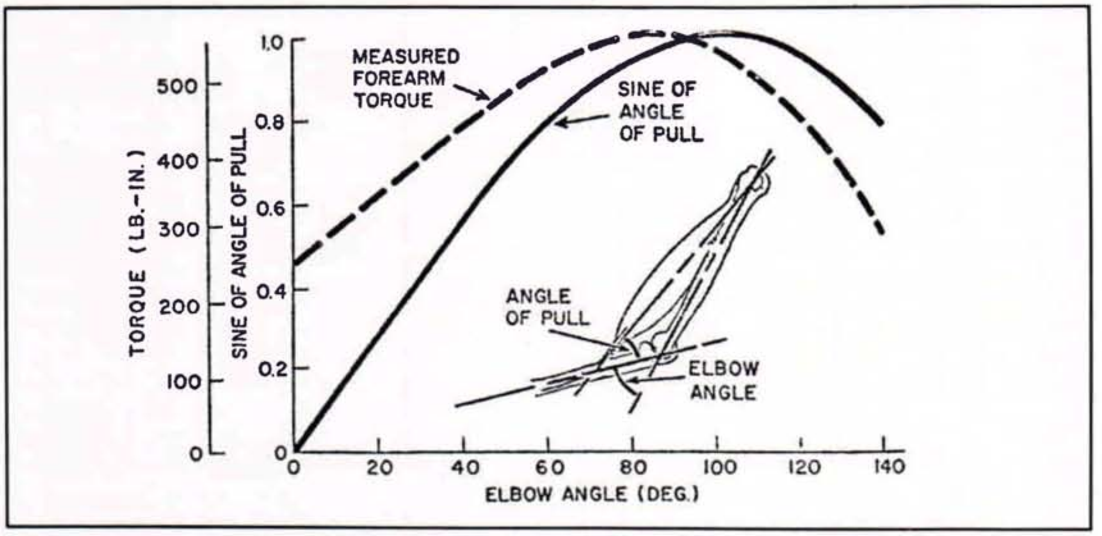

Fig. 14. Forearm-flexor mechanics. Insert gives the geometry of the idealized flexor system.

capacity to adapt the ratio of muscle length to tendon length has been demonstrated in an experiment in which the pathway of the tibialis anterior tendon in the rabbit was shortened. The result was that the tendon shortened while the muscle lengthened to regain the normal joint range (4).

The relative strengths of muscle and of tendon also show an approximate compatibility, the tensile strength of tendon, measured at from 8700 to 18,000 lb. per sq. in. (6), being greater than that for muscle. Strength tests of excised muscle-tendon systems show that failure commonly occurs in the belly of the muscle, or at the musculotendinous juncture, or at the bone-tendon juncture, but never exclusively in the tendon itself. Analysis of clinical cases indicates that muscle is still the site of failure even when it is maximally tensed (14). It is clear, then, that of the muscle-tendon combination the tendon is normally always the stronger.

FOREARM-FLEXOR MECHANICS

The forearm-flexor system is well suited to serve as an example of biomechanics because the bone-joint system comprises a simple uniaxial hinge while the flexor muscles, though five in number, can be reduced to a single equivalent muscle whose geometry and dynamics can be specified from measurment data. Figure 14 illustrates the lever system on which the equivalent muscle acts. The angle between the axis of the muscle and that of the forearm bones, *i.e.*, the "angle of pull," theoretically ranges from 0 deg. at full extension to 90 deg. at 100 deg. of elbow angle, and since the moment arm is continuously proportional to the sine of the angle of pull the mechanical advantage of the lever also is proportional to it.

There are of course departures from this idealized geometry. For one thing, the angle of pull and the elbow angle are not exactly equal. Moreover, at small elbow angles the torque component does not actually drop to zero because the muscles must always pass over the elbow joint at some finite distance from its center. Finally, the force-length curve (10) of the equivalent muscle must also be taken into account in expressing the effective torque. For these and other reasons, actual torque measurements take precedence over theoretical calculations, and the composite curve of Figure 14 has been plotted from the results of a number of investigators. Whereas the moment arm peaks at an elbow angle of 100 deg., the muscle force is declining throughout the elbow-flexion range, and the net effect, as reported by Miller (15), is a maximum torque of about 625 lb.-in. at from 80 to 90 deg. Clarke and Bailey (5) found a peak of about 400 lb.-in. at between 70 and 80 deg., and the author has obtained 550 lb.-in. just under 90 deg. in a group of subjects. Wilkie's data give a value of about 525 lb.-in. at 80 deg., measured on himself (22). These variations can be explained as resulting from the effect of a limited sampling of an inherently variable characteristic. Greater consistency probably could be obtained in a larger series of measurements.

MAXIMUM TORQUES IN MAJOR ACTIONS

Because they express the fundamental output characteristics, and because they are most easily measured, the muscle torques about the major joints represent the most significant and practical aspects of the statics and dynamics of the musculoskeletal system. Not only is muscular power a concept of uncertain validity but also it is very difficult to measure. The combined effect of muscle and lever, however, can easily be measured in many subjects, so that statistical stability can be achieved in the results. Because muscle agonists change length with joint angle, and because they are thus caused to work on different parts of their length-tension diagrams, joint torques vary as a function of joint angle. As demonstrated by Clarke (5), this phenomenon, shown in Figure 14 for the forearm-flexor system, holds more or less for all major actions about the joints. But these details may be neglected in summarizing the maximum torques throughout the upper-extremity system (Table 2).

THE FUNCTIONAL ROLE OF SOCKETS

The socket is the foundation of the upper-extremity prosthesis. It obtains purchase upon the most distal segment of the remaining member and should be stable, though comfortable, in its fit with this member. The socket must bear weight both axially and in all lateral directions. It is the attachment member for mechanical components and for control guides and retainer points. Hence the socket must be a sound structural member as well as a custom-fit, body-mating part. Finally, the socket extends the control function of the member to which it is fitted, giving movement and direction to the prosthesis. In any discussion of prosthetic controls, therefore, the starting point is the socket.

The requirement of formability and strength in sockets has been met satisfactorily by the introduction of polyester laminates (3, 20). These materials permit close matching of the stump impression, and

Table 2

MAXIMUM TORQUES IN THE MAJOR ACTIONS ABOUT JOINTS OF THE UPPER EXTREMITY

Joint Motion	Action	Torque (lb.-in.)	Conditions
Arm-on-shoulder (*5*)	Flexion Extension	470[a] 470[a]	Subjects reclining with arm at side. N = 64
Forearm-on-elbow (*5*)	Flexion Extension	420[b] 280[b]	Subjects reclining with forearm flexed 75 deg. N = 64
Forearm rotation (*21*)	Pronation Supination	110 115	Subjects standing, torques of wrist cuff, midposition. N = 20
Hand-on-wrist (*21*)	Volar flexion Dorsal flexion Ulnar flexion Radial flexion	200 135 150 120	Subjects seated, torques measured at the metacarpophalangeal line with hand axial to forearm. N = 15

[a]Lever arm of 6 in. assumed for computation.

[b]Lever arm of 5 in. assumed for computation.

variations in strength can be introduced by increasing the number of laminate layers. The double-wall construction (3) provides a stump-fitted inner wall, with an outer wall that can be designed to structural uniformity and cosmetic requirement. Sizing to achieve this aim has now been reduced to standard practice (20). Finally, the texture and coloring of the plastic laminate can be controlled to achieve satisfactory cosmetic results.

THE BELOW-ELBOW SOCKET

The peculiar feature of the forearm, that pronation-supination is a function of the whole forearm length, places a special limitation on the below-elbow socket. Although for stability in flexion the whole remaining forearm stump is best sheathed in the socket, to do so prohibits forearm rotation. In the case of the longer below-elbow stumps, therefore, some sacrifice in stability can be afforded in the interest of retaining forearm rotation. The proximal portion of the socket is fitted loosely to give freedom for forearm rotation while the distal portion is fitted snugly to provide a stable grip. Figure 15 shows the amount of forearm rotation available at various levels of the natural forearm and that remaining in below-elbow amputees of various types. Because of torsion of the flesh, however, and because of slippage between the skin and the socket, effective socket rotation is lost in stumps which are only 50 percent of forearm length. The effective socket rotation remaining in the wrist-disarticulation case is only about 90 deg.

Further adaptations of below-elbow sockets to suit the functional requirements at the various levels are shown in Figure 16. In the long below-elbow stump, the elliptical cross-section of the forearm near the wrist permits a "screw-driver" fit of the socket to yield the maximum in rotational stability. With the shorter stumps, the possibility of effective rotation is reduced and is lost completely at about 50 percent of forearm length. At this level, the problem of forearm rotation is outweighted by that of providing flexion stability. Dependence upon a rigid or semirigid hinge system is necessary in the short below-elbow stump, and finally, in the very short stump, effective forearm flexion is so reduced that a split socket with step-up hinge becomes a necessity.

The goal of below-elbow socket design is to regain as completely as possible the control function of the forearm, which includes (a) positioning of the hand by forearm flexion and (b) hand rotation by means of pronation-supination. In the bleow-elbow prosthesis, adequate forearm flexion is obtained rather easily; rotation is limited to the potential available in the longer stumps. Manual wrist rotation, of course,

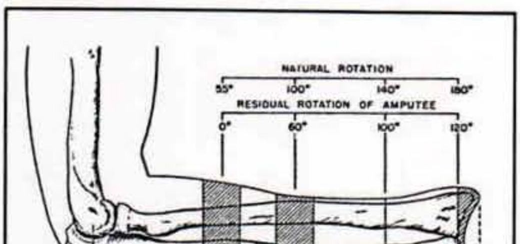

Fig. 15. Below-elbow amputee types, based on average forearm length, epicondyle to styloid. After Taylor (*18*).

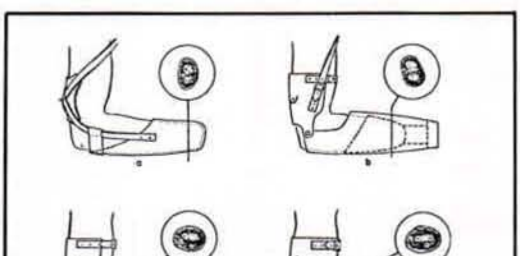

Fig. 16. Schematics of below-elbow prostheses. For each type, an insert gives the cross-sectional anatomy 1 in. from the end of the stump. Sections are taken from the normal anatomy of the forearm. Sockets, hinges, cuffs, and suspensions are for *a*, single socket; *b*, rotation type; *c*, double-wall socket; and *d*, split socket. After Taylor (*18*).

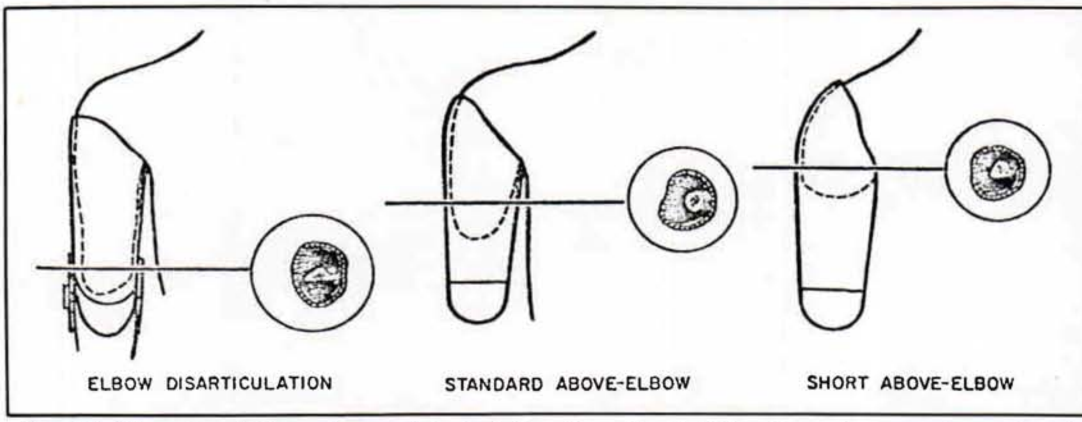

Fig. 17. Schematics of above-elbow sockets, including elbow disarticulation. For each type, an insert gives the cross-sectional anatomy at the indicated level. Dashed lines show stump contour and inner wall of the socket. Standard and short above-elbow cases have a double-wall socket.

supplements the remaining natural rotation. In the below-elbow prosthesis, then, control of the terminal device in space depends in fair measure upon the role of the socket in preserving the residual flexion and rotation of the below-elbow stump.

THE ABOVE-ELBOW SOCKET

Unlike the below-elbow case, the above-elbow stump presents no problem of diminishing rotation with diminishing stump length because arm rotation is confined wholly to the glenohumeral joint. Socket design for the above-elbow case is therefore related principally to the requirement of fitting the stump closely so that the humeral lever can be fully effective in controlling the prosthesis. Figure 17 shows the minor variations corresponding to above-elbow type, including the elbow disarticulation. Sockets for the latter must take account of the bulbous end of the stump. They must provide snug fit around the epicondyle projections but maintain sufficient room in the region just above, where the stump cross-section is reduced, to permit insertion of the stump in the socket. In both the elbow-disarticulation and the standard above-elbow cases, the upper margin of the socket is terminated below the acromion for freedom of movement at the shoulder. In the short above-elbow case, the socket is carried up over the acromion to obtain additional stabilization and suspension from the shoulder, as required by the very limited stump area.

The control function of the above-elbow socket is two-fold. As in the below-elbow case, the socket extends the stump to the next more distal joint and thus gives range and direction to this component upon which the positioning of the still more distal segments depends. But in addition to this feature, the above-elbow socket also has a power function. Through its attachments to shoulders and torso, it provides the forces and displacements needed to produce forearm flexion,, terminal-device operation, and elbow lock. To fulfill these functions, the socket must have stable purchase on the stump in both flexion and extension. Hence, for elbow-disarticulation and above-elbow types, the socket should continue to the axillary level; for short-above-elbow amputees, it should come up over the acromion (Fig. 17). Finally, medical and lateral rotation of the socket are necessary for further functional positioning. Close fit and good suspension are required to give stability in these actions.

THE SHOULDER SOCKET

In the range of amputation sites from transection of the humeral neck to complete removal of the shoulder girdle, the socket form changes from shoulder cap to thoracic saddle. As displayed in Figure 18, the bearing area increases as the remaining shoulder elements are reduced; similarly, the amount of "buildout" needed to preserve shoulder outline increases with increasing amputation loss. With disarticulations and all more extreme losses, sectional plates may be introduced at the axillary parasagittal plane. This arrangement makes it possible to fabricate the prosthesis in two sections, a matter of considerable advantage to the limbmaker, and it also affords the functional advantage of a preposition swivel of the humeral section upon the saddle section to simulate flexion-extension of the arm.

The functional aspects of the shoulder socket are to some extent secondary to the structural; yet there are certain definite functional ends to be served. Shoulder and scapular mobility in elevation, flexion, and extension should be preserved to the highest possible degree. In humeral-neck and shoulder-disarticulation cases, aid can be given to the shrug control (biscapular abduction), and at least a small range of motion can be given to the elbow, but of course no such function can be expected in forequarter or partial-forequarter amputees.

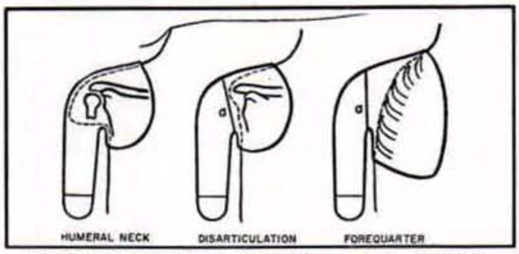

Fig. 18. Schematics of shoulder sockets. Solid lines show residual bony structure, dashed lines the body contour and inner wall of the socket. Disarticulation and forequarter sockets may be two-piece with sectional plates at *a*.

MAJOR ARM AND SHOULDER CONTROLS

The common method of operation of upper-extremity prostheses is by means of shoulder harness which provides suspension and which also transmits force and excursion for control motions. In this manner such operations as forearm flexion-extension, terminal-device operation, and elbow lock are managed. Figure 19 presents the essential features of the major harness controls. In principle, each effective control must begin with a point stabilized on shoulder or torso, pass over a voluntarily movable shoulder or arm part, and thus provide relative motions with respect to the origin. At the movable point, the control cable enters the Bowden-type housing, which transmits the relative motion independent of movements of the distal segments. Controls may be used singly or in combination, depending upon the level of amputation, amputee preference, and other practical considerations.

Besides the relative motions between various segments of the human body, still another source of energy for operation of upper-extremity prostheses can be made available by the surgical procedure known as cineplasty (1, 19), in which a skin-lined tunnel is fashioned in the belly of a muscle group. In various experimental programs conducted both here and abroad, muscle tunnels have been made in the forearm flexors, the forearm extensors, the biceps, the triceps, and the pectoralis major.

Of all the various combinations tried, the biceps tunnel in below-elbow amputees has proved to be the most successful. Failure of other cineplasty systems has been due in some cases to inability of designers to overcome the mechanical problems involved in harnessing the energy thus provided and in other cases to the inherent properties of the particular muscle group concerned. In the below-elbow case, use of the biceps tunnel eliminates the need for shoulder harness and permits operation of the prosthesis with the stump in any position. It has given excellent results in many instances and has been made available to those beneficiaries of the Veterans Administration who can make effective use of the procedure.

The cineplasty tunnel in the biceps of the average male will provide sufficient force and excursion to operate modern terminal devices—an average maximum force of 50 lb. and 1½ in. of useful excursion. It is not unusual for some individuals to be able to build up the force available to a value in excess pf 100 lb., but such a high force normally is not required.

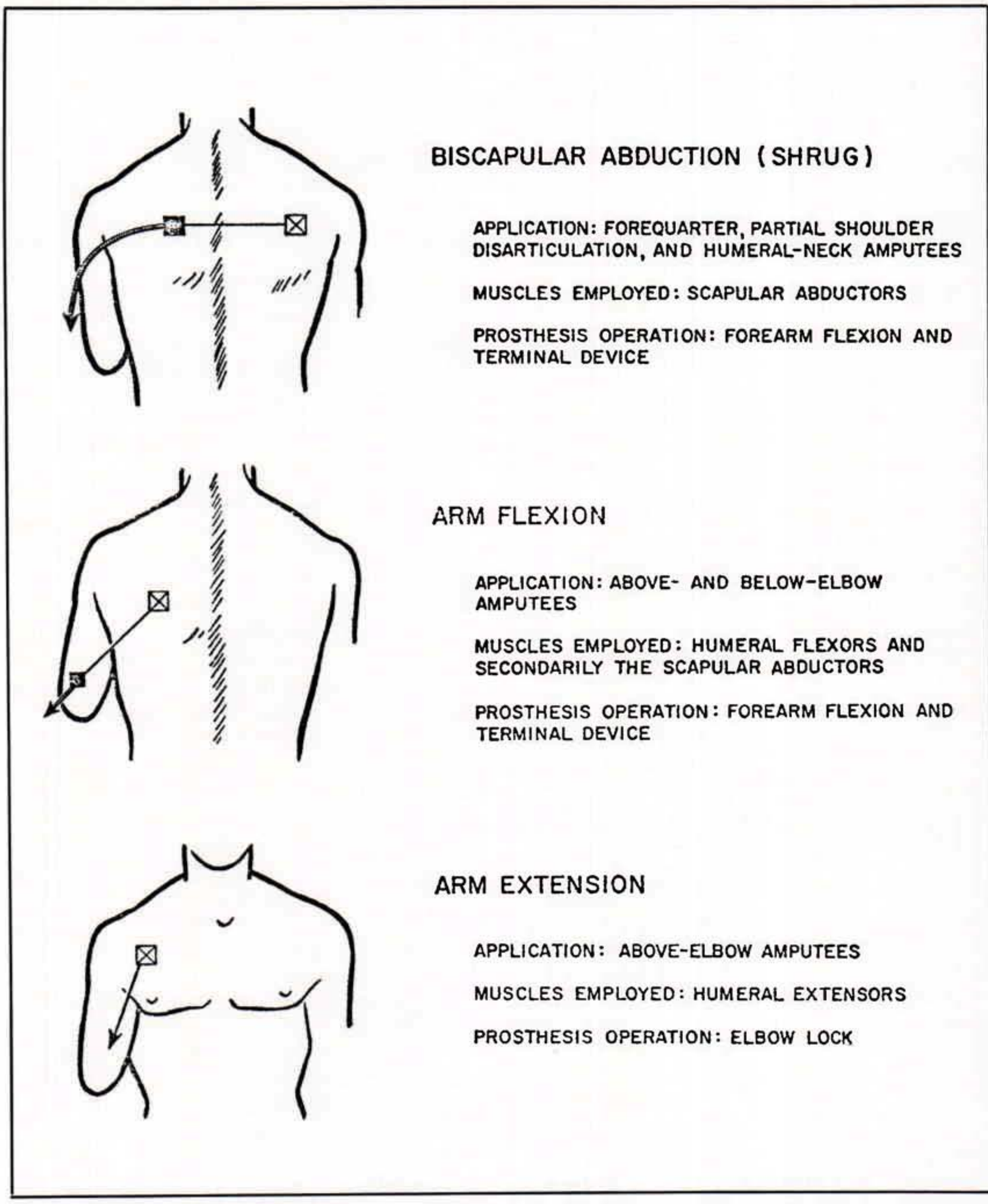

Fig. 19. Major harness controls. The points stabilized by harness (☒) are beginning points for the control cable, which passes into a Bowden-type housing at movable points (■). The relative motion is transmitted via the Bowden cable (▬▬) to distal points on the prosthesis.

The Nature and Operation of Control Systems

The Below-Elbow Single-Control System

The single control for the below-elbow amputee is powered by arm flexion to provide terminal-device operation. This control motion, used by the above-elbow amputee also, depends upon a coordinated flexion of the humerus and abduction of the scapula on the amputated side; little shoulder activity is required on the sound side. It is substantially the same motion as that used in normal unilateral reaching. The displacements of humerus and scapula are additive, so that the resulting motion is quite natural. With full Bowden-cable transmissions of power from arm cuff to forearm socket, there is no influence of elbow angle, and the operation is mastered easily by all amputees with stumps of 35 percent or more of normal forearm length.

The Below-Elbow Dual-Control System[4]

In harnessing below-elbow stumps shorter than 35 percent of normal forearm length, it generally is necessary to use an auxiliary type of lift to help the amputee flex the forearm. This procedure is applicable to a split-socket type of prosthesis. It merely is an adaptation of the above-elbow dual-control system (page 22) using a lever loop positioned on the forearm section so that arm flexion may be utilized to assist in forearm lift. The cable housing is split and assembled so that when the arm is flexed the elbow will flex. The elbow hinge has no locking mechanism, the short below-elbow stump being used to stabilize the forearm. Normally, sufficient torque is available about the elbow axis to give adequate stability in all usable ranges.

In prescribing for a new amputee with this level of amputation, it might be advisable first to have the amputee try a split-type prosthesis without the below-elbow dual-control system. If, at time of initial checkout, the amputee cannot lift his forearm, or if he complains of painful contact with his stump, then of course the dual system is indicated. After the assist lift has been worn for some time, the remaining muscles of the stump may have hypertrophied, in which case the amputee might be able to discard the dual system and convert to the below-elbow single control.

The Below-Elbow Biceps-Cineplasty System

Force and excursion provided by the biceps muscle tunnel are harnessed by inserting into the tunnel a cylindrical pin of a nontoxic material and attaching a cable to each end of the pin. As in the other types of control systems, the Bowden-cable principle is employed to maintain a constant effective distance between the source of energy and the mechanism to be operated, regardless of relative motions occurring between body segments. In order that conventional terminal devices may be employed, it is necessary to join the two cables before attachment to the mechanism. Several devices for making this coupling are available commercially.

Suspension of the socket is provided by an arm cuff, which is attached to the socket by any of the various hinges normally used in fabrication of below-elbow prostheses. The arm cuff is fashioned in such a manner that forces tending to pull the prosthesis from the stump are absorbed by the condyles of the elbow rather than by the muscle tunnel.

The Above-Elbow Dual-Control System

In above-elbow amputees, the humeral stump furnishes the motive power for the three operations of the prosthesis—flexion of the forearm, operation of the terminal device, and management of the elbow lock. The first two operations are so linked mechanically that a single control motion, arm flexion, produces either terminal-device operation or forearm flexion, depending on whether the elbow is locked or unlocked (Fig. 20). Although the control motion by arm flexion in the above-elbow case is similar to that described for the below-elbow amputee, there are several differences. Because the cable passes through a lever loop on the forearm to give torque about the elbow, it is affected by

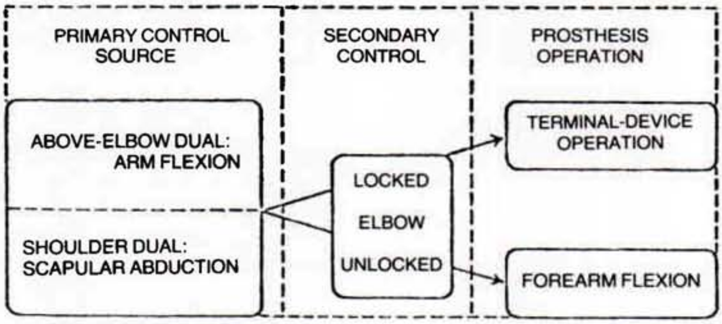

Fig. 20. Operation of above-elbow and shoulder dual controls

elbow position. As the forearm is flexed, arm-flexion excursion is used up, and the excursion needed to operate the terminal device must come from scapular abduction (shrug), as in shoulder cases. Typically, the above-elbow amputee manages a full range of free forearm flexion by a normal arm-flexion movement. But in the elbow-angle range of from 90 to 135 deg., with elbow locked for terminal-device operation, he must call upon supplementary excursions from biscapular abduction. With the terminal device at the mouth, practically all operation depends upon shoulder shrug.

In the above-elbow dual-control system, operation of the elbow lock depends upon humeral extension and associated coordinations. When the forearm has been flexed to the position desired, the elbow lock is engaged by the arm-extension movement. Skill is needed to maintain tension on the arm-flexion cable so that the arm does not drop during the locking control motion. Well-trained amputees elevate the arm moderately to compensate for the humeral extension and thus maintain the elbow angle. The extension control motion is complex. The humerus is simultaneously extended and elevated so that it moves obliquely to the side. During this phase, the point of the shoulder must be stabilized, or even moved forward, and the trapezius is bulged by downward rotation of the scapula (Fig. 21).

The Above-Elbow Triple-Control System

The triple-control system has been devised to separate terminal-device operation from forearm lift. When the dual-control system is used, the amputee must select, by the use of the elbow lock, either terminal-device operation or forearm lifting. By separating forearm flexion and terminal-device operation, the triple control makes it possible for the terminal device to be controlled by an independent body motion. Although in general an above-elbow amputee fitted with triple control has an elbow lock, a few such cases are able to separate prehension from forearm flexion without use of the lock.

A control cable from the terminal device is so attached and positioned that biscapular abduction or merely shoulder shrug will operate the terminal device through its full range of prehension. To lift the forearm the amputee uses arm flexion. Elbow-lock operation is accomplished in the same manner as in the dual-control system, that is, by arm extension.

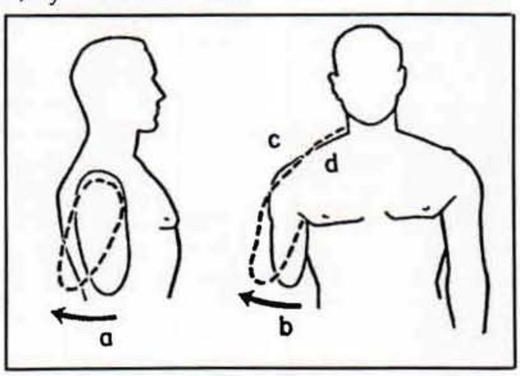

Fig. 21. Coordinated control motions for elbow lock. Simultaneously the humerus is both extended (*a*) and abducted (*b*) while the shoulder is depressed (*c*) and the trapezius is bulged (*d*) by downward rotation of the scapula.

It is apparent that this arrangement will work best with a comparatively stable socket and a relatively long above-elbow stump. The chief advantage of the triple-control system is that at full forearm flexion the terminal device may still be operated through its complete range.

The Shoulder Dual-Control System

In the absence of the humeral lever, the shoulder becomes the major power source, biscapular abduction controlling both forearm and terminal device in the dual-control system. The control path courses horizontally across the scapulae, and either opposite-axilla loop or basic chest-strap harness (page 46) captures the action satisfactorily. The combination afforded by the dual principle also is illustrated in Figure 20.

The shoulder amputee has a special difficulty in obtaining the combination of full forearm flexion and terminal-device operation because, unlike the above-elbow amputee, who can add the excursions of humeral flexion and scapular abduction, he must obtain all movement from biscapular abduction. Shoulder amputees with broad shoulders and wide chests usually achieve this action satisfactorily; others must accept the limitation of partial terminal-device operation at full forearm flexion. Partial-shoulder and forequarter amputees must depend upon the sound shoulder entirely, and in this case the action range of the terminal device typically is limited to not more than 90 deg. of forearm flexion.

In shoulder amputees, operation of the elbow lock must be managed by various special arrangements. The waist control, utilizing shoulder elevation; the perineal strap, based on relative motion between shoulders and pelvis; the nudge control, requiring either manual or chin operation; extreme shoulder flexion on the sound side; and extension of the shoulder on the amputated side complete the array of known feasible possibilities. It is evident that with this class of amputees control motions will be slower and deliberately sequential. They are therefore necessarily more noticeable and awkward.

The Shoulder Triple-Control System

The harness required for the triple-control shoulder-disarticulation system consists of a chest strap for forearm flexion, a waist strap to operate the elbow lock, and an opposite-shoulder loop for prehension. The amputee must have excellent scapular abduction and must be able to separate it from extreme opposite-shoulder shrug, and he must have available good shoulder elevation on the amputated side. The chief advantage of the triple control in the shoulder-disarticulation case is identical to that of the triple control in the above-elbow case, namely, that the terminal device may be operated fully in the vicinity of the mouth. To operate the prosthesis from an extended position, the amputee first produces biscapular abduction, thus raising the forearm. Then, with the forearm held in place, he elevates the shoulder on the amputated side to lock the elbow. To operate the terminal device, he then flexes the sound shoulder. Excursion for terminal-device operation is thus unaffectd by forearm flexion.

Unfortunately this system must be restriced to humeral-neck and shoulder-disarticulation cases. For lack of sufficient excurision on the amputated side, it is unlikely that a forequarter amputee would be able to use triple control

Mechanical Application of the Major Controls

To elucidate practical amputee biomechanics, it is necessary to refer to several aspects of the connecting mechanism between amputee and prosthesis in the power-transmission system. Of first importance are the proximal retainers, which are located at the point where the cable from the shoulder harness enters the cable housing. These retainers are the beginning points of the transmission systems indicated in Figure 19. In both below- and above-elbow cases, the proximal retainer is positioned in accordance with the ratios shown in Figure 22. For all above-elbow stumps of greater than 50 percent of acromion-to-epicondyle length, the proximal retainer point is placed slightly lower than half way down the arm, the reason being that the control passes naturally through this point in its course from opposite shoulder, across the scapula, and thence to the lever loop on the forearm shell. The humeral lever power is quite adequate at this point (Table 3), and no practical advantage is gained by a lower placement. With above-elbow stumps less than 50 percent as long as the normal arm length, acromion to epicondyle, the proximal retainers must be placed at the level of the stump end in order to prevent undue tipping of the socket, as would occur if forces developed beyond the end of the stump.

Table 3

PROSTHETIC CONTROL FORCES AND DISPLACEMENTS[a]

(Average Measurements from 50 Normal Subjects)

Control Source	Force (lb.)	Displacement (in.)
Arm flexion	63	2.1
Shrug	61	2.2
Arm extension	56	2.3

[a] From Taylor (*18*).

In shoulder cases, the control path is directed horizontally at approximately the midscapular level and brought to the arm section at the axilla. The control motion is purely biscapular abduction, and consequently the proximal retainer is placed on the prosthesis at the midscapular level. The resulting force and excursion are given in Table 3.

Arm-extension forces are potentially quite high, as also shown in Table 3. Because only 2 to 6 lb. of force and ½ in. of excursion are required to operate an elbow lock, normally there is a generous power excess. The principal concern in harnessing arm-extension control is to obtain operation with minimal movement and thus to avoid awkwardness.

Conclusion

The central purpose of this article has been to outline the biomechanical basis of control in upper-extremity prostheses. Consequently, emphasis has been placed upon the normal and residual functional anatomy and kinesiology underlying this service. The particularized biomechanics of prosthesis control has been defined, and the limitations incurred in amputations at high levels have been stressed. The major

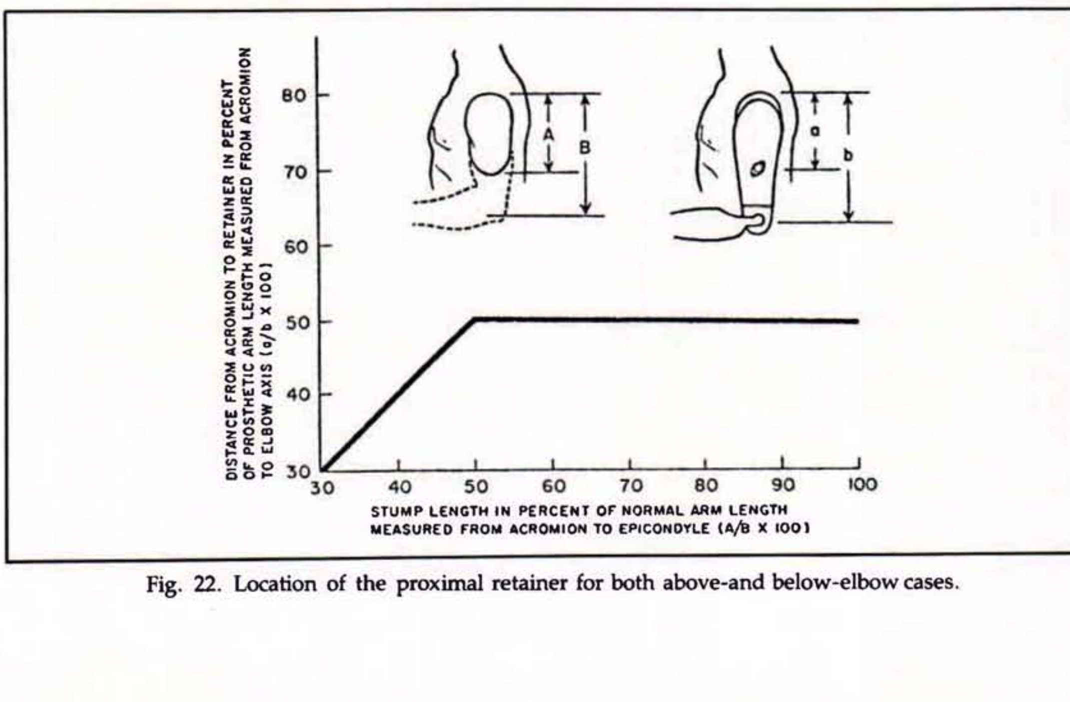

Fig. 22. Location of the proximal retainer for both above-and below-elbow cases.

message is that a thorough understanding of the motions of control available to each type of patient is necessary to the proper prescription, fitting, and training of the upper-extremity amputee. Thus only can full advantage be taken of the improved functional features to be found in modern arm components.

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[2]In everyday language the word "arm" is of course taken to mean the entire upper extremity, or at least that portion between shoulder and wrist. In anatomical terms, "arm" is reserved specifically for the segment between shoulder and elbow, that between elbow and wrist being the "forearm." Although in the lower extremeity the word "leg" commonly means the entire lower limb, whereas anatomically the "leg" is that segment between knee and ankle, confusion is easily avoided because we have the special word "shank." No such spare word is available to describe the humeral segment of the upper limb.—Ed.

[3]It deserves to be noted here that, taken literally, expressions such as "forearm flexion-extension," represent questionable nomenclature. To "flex" means to "bend." Limb segments do no bend very readily without breaking. Joints are *designed* for flexion. In the lower extremity, for example, one speaks not of "shank flexion" but of "knee flexion," not of "thigh flexion" but of "hip flexion." That is, one uses "flexion" or "extension" not with reference to motion of the distal segment but with reference to the more proximal joint. Although Webster accepts the expression "to flex the arm," he obviously uses the word "arm" in the everyday sense of meaning the entire upper extremity, or at least that portion between shoulder and wrist. Because this loose terminology in the upper extremeity is so widely established, not only among workers in prosthetics, it is used throughout this issue of ARTIFICIAL LIMBS, with the understanding that "forearm flexion" means "elbow flexion," "arm flexion" and "humeral flexion" mean "flexion of the glenohumeral joint (and associated structures)" See page 9 *et seq.*—ED.

[4]Although the terminology commonly used to describe the several control systems could well afford to be better systematized, it is adopted here because it is now so well established throughout the field of prosthetics. One *may* think of "dual control" as meaning that two control sources are involved in the provision of all necessary functions, but according to convention it means that two functions, specifically elbow flexion and terminal-device operation, are provided by a single control source, the third function, elbow lock, if needed, being managed by an additional control source. Yet "triple control" (page 22) in the accepted sense means not that three functions are furnished by a single control source but that three control sources are used to provide three functions, one for each.—ED.

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Additions to the Vertical Fabrication Machine

CHARLES H. PRITHAM, C.P.O.[1]

The Vertical Fabrication Machine (Fig. 1) is a device intended to facilitate the bench alignment and transfer of below-knee and above-knee prostheses. As such the clamps and locking fixtures are designed to permit a component to be moved readily and returned to the same position or shifted to a new position accurately. With the intention of accurately maintaining alignment, the central vertical column and horizontal arms are keyed with longitudinal slots to which toggles of the clamps mate. The use of these features is absolutely essential to ensure precise duplication of alignment in the transfer process, but in the bench alignment process the lack of flexibility in positioning components is a hindrance. In order to overcome this problem and to extend the function of the vertical fabrication machine, the following subsidiary components have been developed.

Universal Positioning Arm

The Universal positioning arm (Fig. 2) when substituted for the top arm (Fig. 3) of the Vertical Fabrication Machine enables a pipe or rod clamped in it (Fig. 4) to be rotated in three planes and moved linearly in two. Linear motion in the third plane is accomplished by using the other features of the Vertical Fabrication Machine enables a pipe or rod clamped in it (Fig.4) to be rotated in three planes and moved linearly in two. Linear motion in the third plane is accomplished by using the other features of the Vertical Fabrication Machine. The Universal Positioning Arm was originally devised for the setup of upper-limb prostheses, and in use it is intended that the wrist unit or elbow turntable be mounted on the bottom arm (normally used to hold the ankle block) or affixed to the table top by a bead of clay. The Universal Positioning

Fig. 1. Vertical Fabrication Machine.

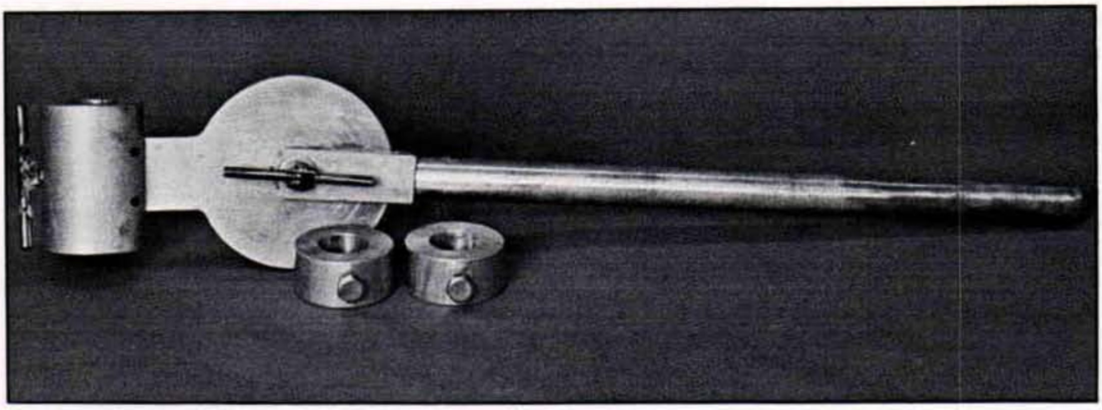

Fig. 2. Universal Positioning Arm. Two locking rings are shown in the foreground.

Arm is then used to position the socket relative to the wrist unit or elbow turntable, a plastic or paper cone is taped in place between them and wax or urethane foam poured to complete the set-up. The device has also been used to facilitate the bench alignment of lower-limb prostheses (Fig. 4) and the modification of plaster-of-Paris models with forefoot buildups for partial foot prostheses. In this context it can be used in conjunction with either the flat table top or with a casting board to duplicate the contour of a shoe. In either instance, the surfaces should be covered with plastic wrap to act as a separator. If desired flat slabs of modeling clay can be used to create a form into which plaster of Paris can be poured.

Hans Richard Lehneis has described a similar device (2) for use in bench alignment of below-knee prostheses with PVC pylons. While undoubtedly simpler to construct, in this author's opinion it suffers from the disadvantage that one adjustment can not be made independent of another.

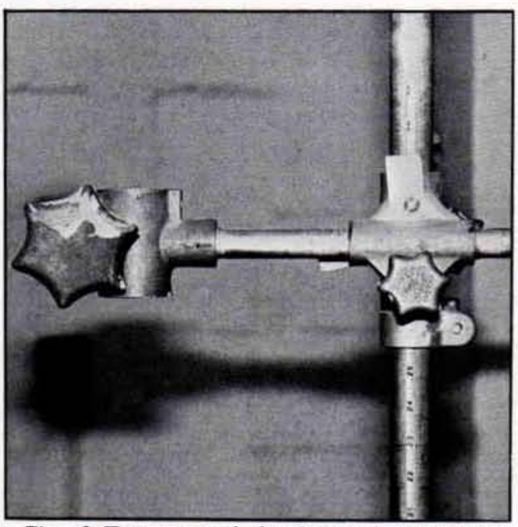

Fig. 3 Top arm of the Vertical Fabrication Machine.

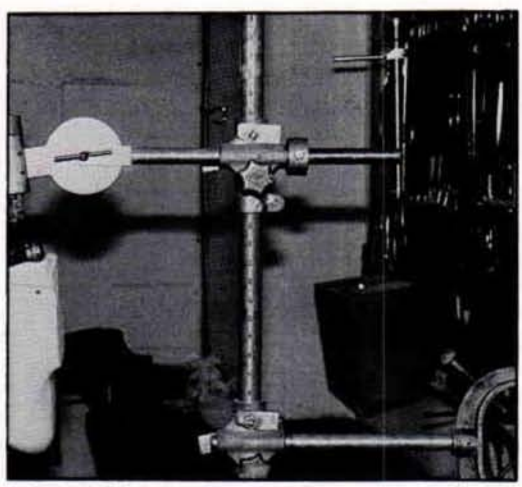

Fig. 4 Universal Positioning Arm installed in the Vertical Fabrication Machine with a below-knee socket model held in place as for establishment of bench alignment and distal buildup prior to vacuum forming. The two locking rings insure that changes in rotation will not lead to inadvertent linear changes.

Knee Joint Alignment Jig

A U.S.M.C. alignment jig (2T402) (Fig. 5) intended for use in fracture bracing has been modified to accept the various adapters of the Otto Bock alignment jig (743R4) (Fig. 6) and, in addition, other adapters have been made to accommodate the English thread sizes of American knee joints. To complete the process, modifications have been made to both the alignment jig and the Vertical Fabrication Machine to permit the alignment jig to be used instead of the regular knee bolt clamp (Fig. 7). This was done originally so that the Vertical Fabrication Machine could be used to apply the buildups (Fig. 8) necessary for the fabrication of the V.A.P.C. Genucentric Knee Orthosis (1). A special pair of discs with central pointers (Fig. 9) were fabricated to fit the alignment jig for this purpose and the Universal Positioning Arm is used to align the plaster model. It is, of course, possible to use these various modifications to position a knee orthosis or a knee-ankle-foot orthosis model and contour and align the uprights of conventional knee joints. Although cumbersome and awkward, it has the advantage of enabling the orthotist to duplicate the functional position of the

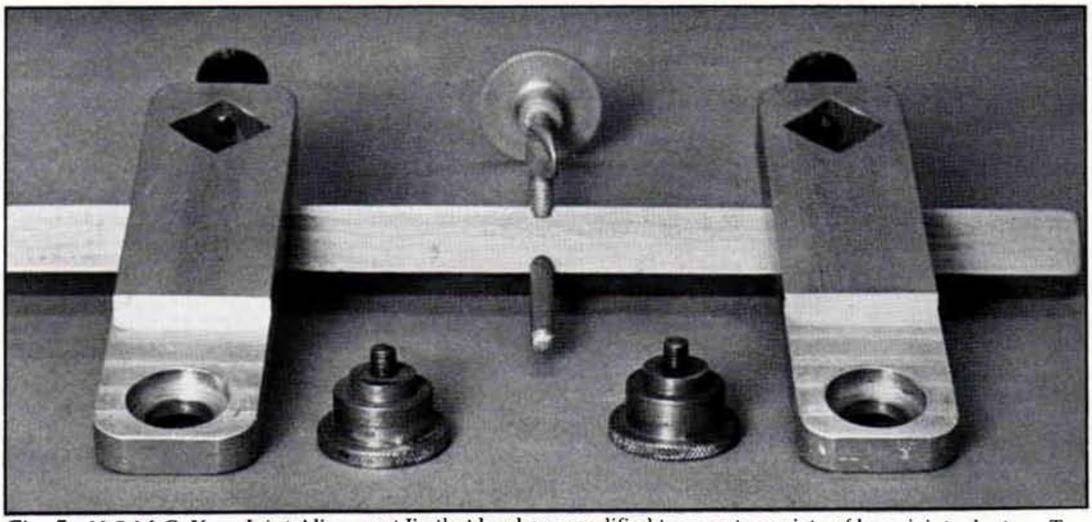

Fig. 5. U.S.M.C. Knee Joint Alignment Jig that has been modified to accept a variety of knee joint adapters. Two of these are seen in the foreground.

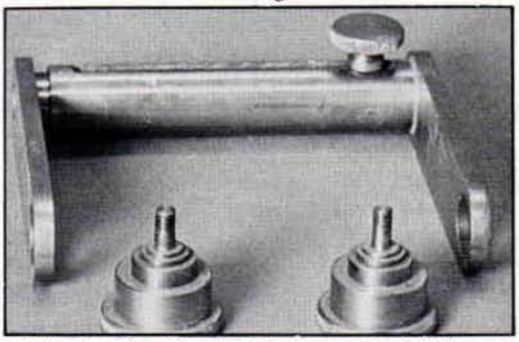

Fig. 6. Otto Block Alignment Jig 743R4 and two of the adapters that accompany it. These adapters are fully compatible with the modified alignment jig shown in Figure 5.

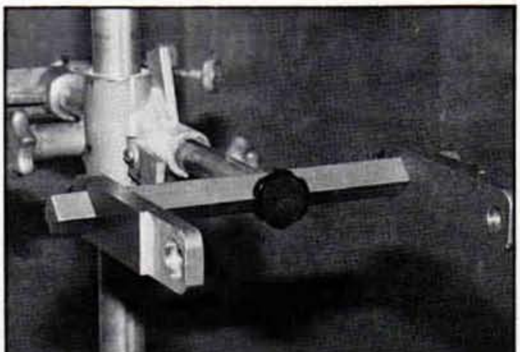

Fig. 7. Modified alignment jig mounted in the Vertical Fabrication Machine.

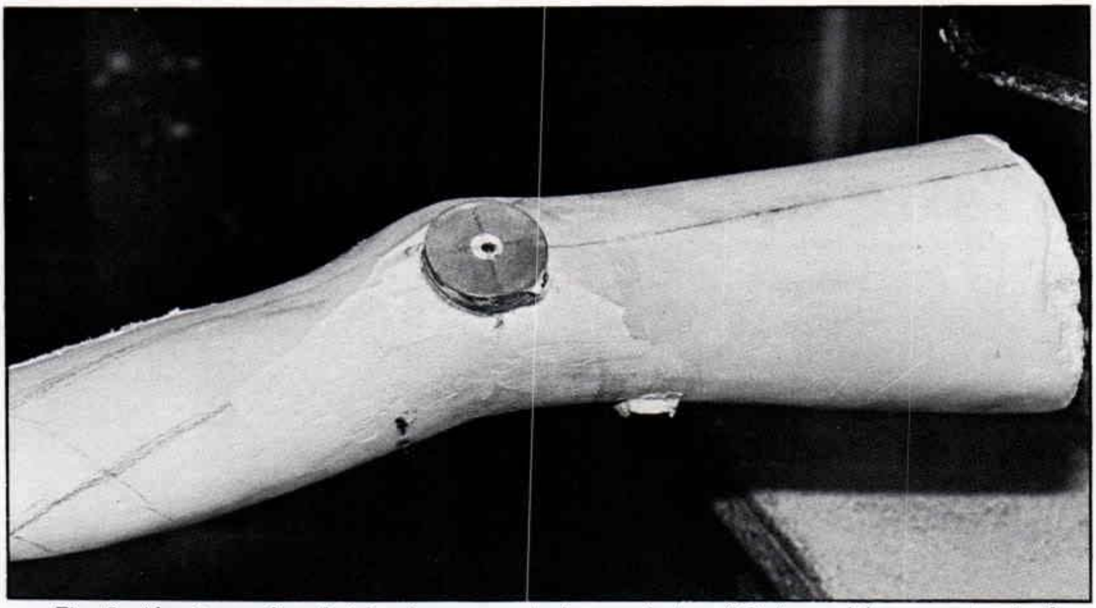

Fig. 8. Aluminum disc that has been mounted on a plaster of Paris model as one step in the fabrication of a VAPC Genucentric Knee Orthosis. In use, plaster is removed from the underlying area so as to maintain the proper medial-lateral diameter, and alignment is established in the Vertical Fabrication Machine. Polyester resin is then used to secure the disc in place, the mounting fixture is removed, and plaster of Paris is used to create the proper contours about the disc. After fabrication of the orthosis the disc can be recovered for further use.

Fig. 9. Brass mounting fixtures mounted in the alignment jig with the appropriate adapters. These mounting fixtures are turned down to points on the ends towards the model. A threaded aluminum disc, as described in Figure 8, is mounted on the right-hand mounting fixture. The point of the fixture is discernible in the center of the disc.

limb, and thus accurately visualize the alignment of components. In a similar fashion with appropriately modified alignment jigs, it would be possible to align other orthotic joints on models and, in addition, the knee joints of a below-knee prosthesis that required joints and corset. In this latter instance, it would be desirable when at all possible to first perform dynamic alignment so that once applied the joints would be horizontal.

Summary

The use of a Universal Positioning Arm and Knee Joint Alignment fixture to extend the versatility of the Vertical Fabrication Machine in prosthetics and orthotics has been described. This includes situations with which the author has had personal experience as well as some purely hypothetical ones.

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MYOELECTRIC CONTROL SYSTEMS —A SELECTED BIBLIOGRAPHY

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Note: This bibliography with an author index was prepared in 1979 in the course of developing a report on external power. Because of the great current interest in the subject of myoelectric controls and because the existence of this bibliography is not widely known, we felt it appropriate to include it in this issue of "Orthotics and Prosthetics". While the title contains the modifier "selected" to indicate that it is not complete, very little of importance has not been included.

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Orthotics and Prosthetics, Vol. 35, No. 1, pp. 46-48, March 1981

NEW PUBLICATIONS

***THE MILWAUKEE BRACE* by Walter P. Blount and John H. Moe; second edition, Williams and Wilkins, Baltimore, Maryland 1980: 252 pages, 117 pages**

The first edition of "The Milwaukee Brace" was issued in 1973. This authorative, well written, and well illustrated text was, as would have been expected, extremely well received. An exhaustion of the supply of the first edition demanded reprinting and changes in the procedure that usually evolve during the application of complex techniques such as the "Milwaukee Brace", demanded a revision.

Like its predecessor the second edition has been prepared for the entire treatment team. Of especial interest to orthotists is the addition of methods of fabrication of thermoplastic girdles, a procedure that was still being developed when the first edition was published.

This new edition should be added to the library of every clinician that is involved in the treatment of scoliosis, idopathic or otherwise.

A. Bennett Wilson, Jr.

Bulletin of Prosthetics Research (BPR 10-33, Spring 1980)

The Office of Technology Transfer of Veterans Administration's Rehabilitative Engineering Research and Development Service announces the availability of the Spring 1980 issue of the Bulletin of Prosthetics Research (BPR 10-33)

The Spring 1980 issue which contains 259 pages is the first to use a full-size (8½ x 11 in.) two-column format. Among the new features is an easy-to-use Index of Progress Reports by subject, institution, and investigator, and a new section containing progress reports from research groups in prosthetics and allied field supported by the National Science Foundation.

BPR-33 is available from the Superintendent of Documents, Government Printing Office, Washington, D.C. 20402. The price is $6.00.

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Articles

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Max Cleland

Comments on the Article: "Development of Test Methods for Evaluation of Wheelchair Cushions."

Paul W. Brand

A Comparison of Sitting Pressures on Wheelchair Cushions as Measured by "Air Cell" Transducers and Miniature Electronic Pressure Transducers

V.R. Palmieri
G.T. Haalen
George V.B. Cochran

Development of Test Methods for Evaluation of Wheelchair Cushions

George V.B. Cochran
Vincent Palmiari

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NIHR—National Insitute of Handicapped Research, Division of Rehabilitation Engineering

VA RER&DS—Veterans Administration Rehabilitative Engineering Research and Development Service

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Orthotics and Prosthetics, Vol. 35, No. 1, pp. 49, March 1981

Letter To the Editor

Dear Editor:

As we discussed in our telephone conversation yesterday, I suspect that crucial passages are missing in two of the articles in the September, 1980 issue of *Orthotics and Prosthetics*. Correction notes in the next issue would seem to be in order.

The specific areas of concern are:

1. The first complete paragraph on page 25. Something is missing or very garbled.
2. On page 45 a paragraph or two describing the silicone elastomer as well as how the two sections of the mold were mounted relative to each other seem to be missing.
3. A minor point, Figure 7, page 9, seems to have been switched with Figure 9, page 11.

I feel these corrections should be made as both articles would seem to be of considerable relevance.

Very truly yours.

Charles H. Pritham

Mr. Pritham is quite right concerning the paragraph on page 25. Two lines were omitted. The paragraph should read as follows:

Some problems soon became evident, however. When only a chest strap harness was used, the shoulder cap migrated posteriorly when the prosthesis was activated, the motion being caused by the forces transmitted from the control attached to the shoulder cap, and the other end attached to the above-elbow attachment strap (Fig. 5). The problem was solved by application of a spring steel strut, (Fig. 6), with one end being attached to the shoulder cap, and the other end attached to the above-elbow socket. It should be noted that the point of attachment on the socket is critical, since it must be located as near as possible to the center of rotation of the shoulder when viewed from the transverse plane.

I agree that more detail on page 45 would be preferable; and, yes the Figure 7 and Figure 9 on pages 9 and 11 are transposed.

We appreciate having these errors called to our attention by Mr. Pritham and trust other readers will do the same when appropriate. ED.

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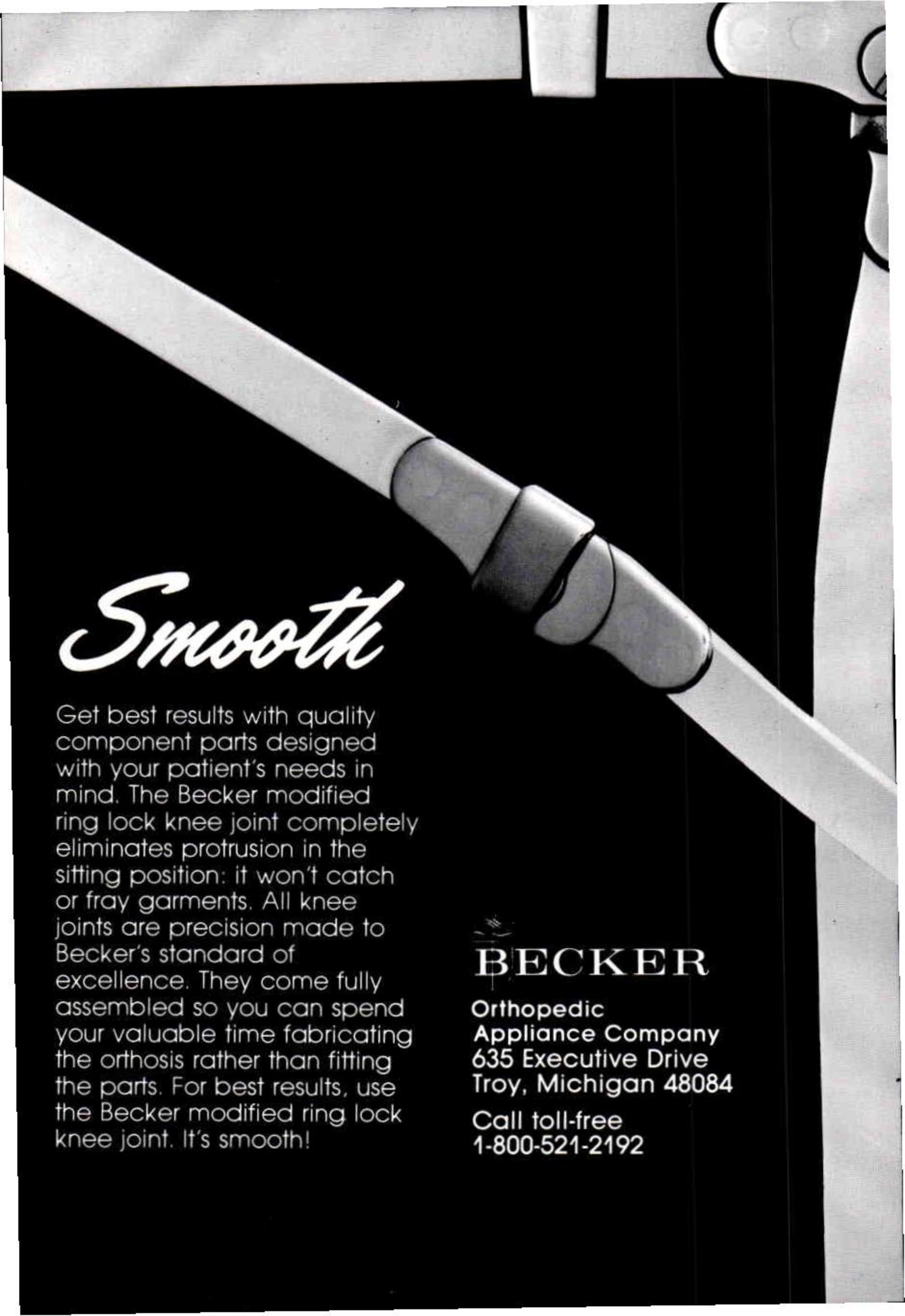
Scientific Papers

Three-page papers, prepared on an author's kit, are being requested in the general area of Rehabilitation Engineering. Prepared papers are due no later than April 1, 1981. The theme for the Conference of 1981 (The International Year of Disabled Persons) is "Technology that Enables".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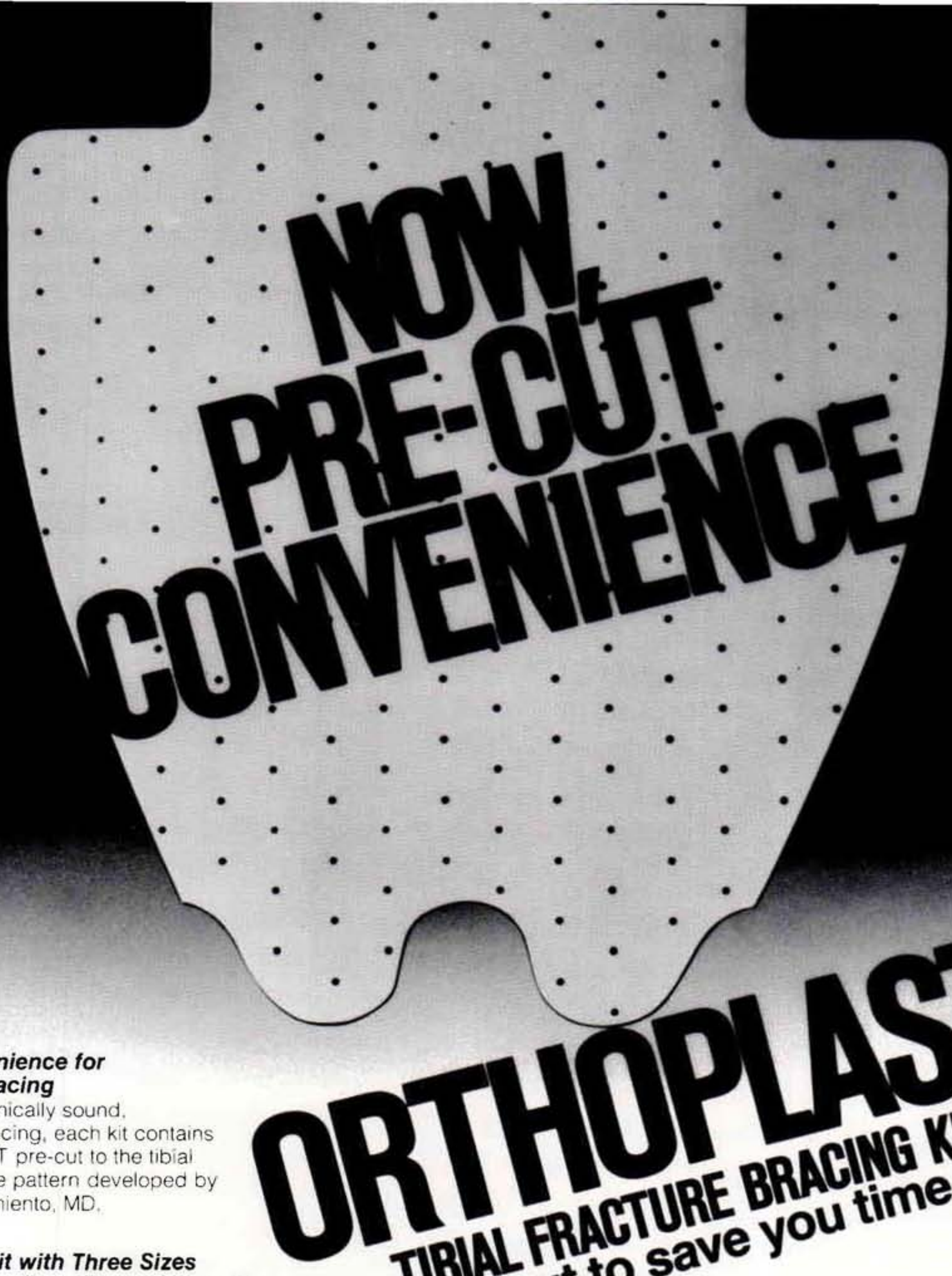

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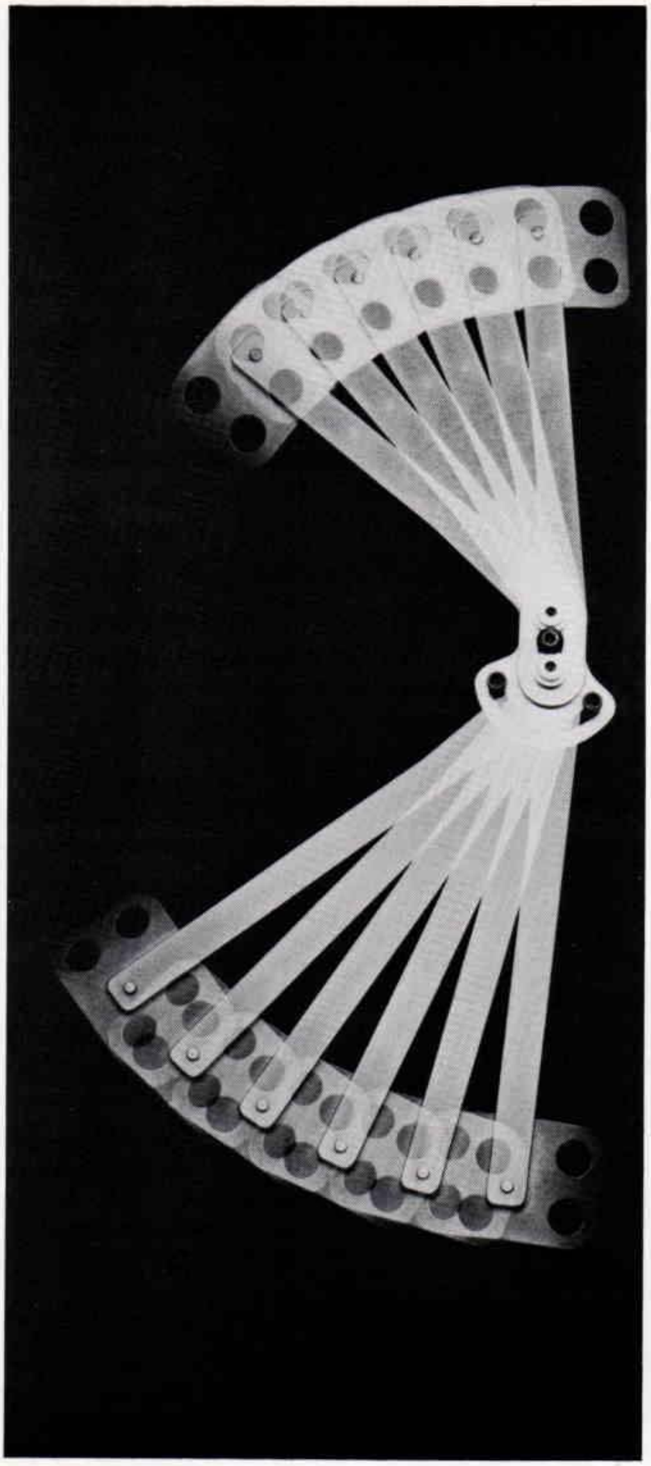

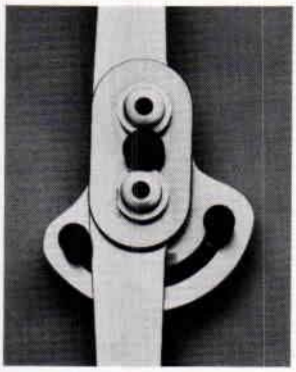

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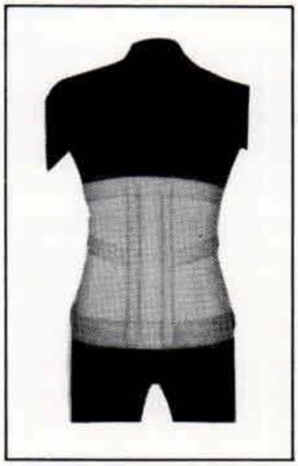

2-S1500
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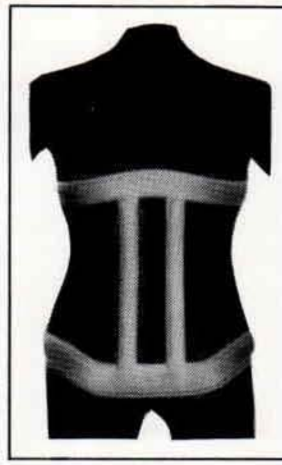

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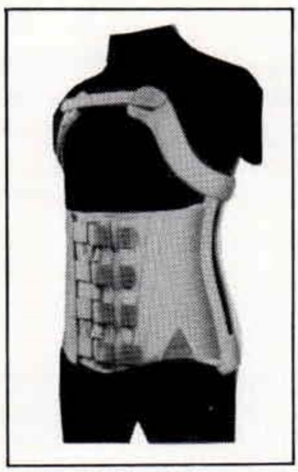

2-S1300
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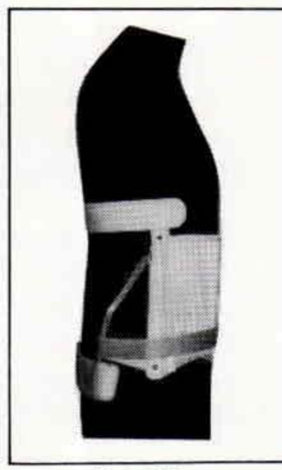

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