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Artificial
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A Review of
Current Developments

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National Academy of Sciences

National Research Council

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Artificial Limbs

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Prelude, Prophecy, and Promise

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AT HAS become almost axiomatic to say that the effective and rational treatment of human ills is dependent upon a knowledge of the normal structure and function of the human body and its response to the abnormal. But application of this dictum to the development of prosthetic apparatus as a substitute for loss of an extremity has been long delayed, and consequently improvement in artificial limbs has been extraordinarily slow. Almost to the immediate present, design of artificial limbs has been characterized largely by empirical developments and has depended little on fundamental investigation.

The reasons for this state of affairs are many and complex. Among them might be mentioned the late appearance, in the nineteenth century, of the humanitarian movement which was to provide much of the stimulus and the monetary support from private philanthropist and, later, from the public purse. Tradition too has had its influence, for initially the development of prostheses lay in the hands of the armorer, an association reminiscent of the relationship between amputation and warfare, and thence it passed to his lineal heirs, the surgical-instrument maker and the skilled artisan, who, however ingenious, had no background either in anatomy or in physiology and little knowledge of mathematics or of engineering.

Replacement of hand and arm, the tool through which the highest endowments of the human mind have been expressed, offered no great possibility of complete success. This circumstance influenced the surgeon toward an extreme conservatism in upper-extremity amputation, and failure to achieve perfection in a prosthesis brought no greater disappointment than there were expectations. From the point of view of amputee rehabilitation, furthermore, it was recognized that in the unilateral arm amputee the left hand could be taught to perform the functions of the right, and vice versa, so that a partial restoration of function in supplying stability was all that was sought. Limitations in expectation provided limitations in objective.

In the lower extremity, the problem of restoration seemed, on superficial analysis, to be infinitely more simple. As might have been expected, initial concepts of replacement were in terms of support only, to be followed by development of a jointed support in mimicry of the human leg as a static rather

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than a dynamic mechanism. The degree of disappointment and measure of failure in these simple objectives, without change in fundamental concepts, is to be seen in the countless empirical modifications of initial designs which bestrew the literature on artificial limbs over the past hundred years and more.

Earlier optimisms were gradually replaced by indifference and the inertia of failure, as is well known to those associated with the problem of the amputee after World War I. Locomotion, as we ordinarily understand it, is impossible on a single extremity. But it was realized insufficiently that, unlike the upper extremities, the two lower limbs together constitute but a single organ—the organ of locomotion. Consequently, the complexity of locomotion in relationship to prosthetics design never really was understood, and even where designs were in question the available information was inadequate to support newer developments of principle.

Preliminary efforts in the study of human locomotion are to be found in the work, *De Motu Animalium,* of the Neapolitan mathematician and physician, Giovanni Borelli (1608-1679). As a pupil of Galileo, he was stimulated to take a mechanistic view of bodily function and to study locomotion as a problem in leverage, but his theories and those of his followers soon were reduced to absurdity in the attempt to apply the same mechanistic principles to the whole of medical practice. Continuation of Borelli's approach had to await the nineteenth century and the advent of the Weber brothers, Edward (1806-1871) and Wilhelm (1804-1891), physician and physicist respectively, who with primitive electrical apparatus made the first accurate measurements of gait and undertook its mathematical analysis. The development of photography as a method of recording enabled Etienne-Jules Marey (1830-1904) to avoid previous errors and to correct earlier ideas, and further improvements in photography led to the classical work of Christian Braune and Otto Fischer, *Der Gang des Menschen* (1895), which has constituted the main source in the formulation of principles for the construction of artificial legs, as in the well-known books of H. von Recklinghausen (1920) and Frederich Mommsen (1932). Over more than a decade (1933-1945) Elftman published the results of extensive locomotion studies. To these and many others we owe a great debt.

Despite all these investigations, at the end of World War II our knowledge of human locomotion was still quite incomplete, and such knowledge as existed was only poorly understood. Thus it was that, when approached in September of 1945 by the then Committee on Artificial Limbs of the National Research Council, the representatives of the College of Engineering and of the Medical School of the University of California could point to the necessity of the adoption of a long-term outlook which envisioned the study of the fundamentals of human locomotion, of the amputee who must wear a lower-extremity prosthesis, and of the prosthesis itself. It could be shown that the experience of 400 years in trial-and-error techniques had offered little and that a firm basis

for progress could be established only by a systematic approach. It was predicted that at least seven years of study would be required to collect the fundamental data necessary for improved design of artificial legs.

That that prophecy was not needlessly pessimistic is revealed in the fact that only today can it be said with a degree of confidence that we are about to enter a period of practical development in the evolution of a truly satisfactory lower-extremity prosthesis. Within the next two or three years we should see the appearance of sound improvements based upon the preceding nine years of pioneering work.

But the problems of the leg amputee are not wholly "prosthetic." Such a patient presents a clinical picture of considerable significance. The whole being the sum of its parts, the amputee can scarcely be looked upon as normal in the medical sense, however good general health may be. He is, indeed, quite abnormal, for from amputation of an extremity come changes in skeletal, muscular, and circulatory systems to be dealt with in the design and application of the prosthetic replacement. Complications of pain, real and phantom, and of skin disorders are other matters needing the skills and experience of the medical profession.

Taking cognizance of this situation, the Advisory Committee on Artificial Limbs, in the spring of 1953, recommended that the University of California initiate an extensive clinical program to be integrated with the work already under way in the fundamentals of locomotion and in the techniques of lower-extremity fit and alignment. Utilizing space and services afforded by the U. S. Naval Hospital at Oakland and personnel from the University of California Medical and Engineering Schools, the Clinical Study aims to apply to the practical problems of difficult amputee cases the results of the earlier work on the Berkeley Campus.

This issue of ARTIFICIAL LIMBS is concerned with two major factors in the management of the lower-extremity amputee—the solution of medical problems associated with the amputated state, and the proper application of the prosthetic replacement on the basis of established biomechanical considerations. In the first of two articles, an orthopedic surgeon and an engineer collaborate in describing the origin, observations, and objectives of the Lower-Extremity Clinical Study. In the second, an engineer develops the principles of alignment and socket fit so indispensable to comfort and function, and hence to the success, of the above-knee artificial leg. In this cooperative effort is reflected the whole basic philosophy of the Artificial Limb Program in approaching the problems of the amputee.

The Lower-Extremity Clinical Study—Its Background and Objectives[1]

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HOWARD D. EBERHART, M.S.[3]

IF IT may be postulated correctly that the most satisfactory artificial leg is the one which most nearly simulates the static and dynamic behavior of the natural limb it replaces, the successful practice of lower-extremity prosthetics poses a twofold requirement. The first is an intimate and detailed knowledge of the characteristics of the normal leg in all common activities, and the second is the ability to reproduce as nearly as possible, by a combination of design and fit of the substitute limb, the kinetic and kinematic features essential to normal locomotion. In the Artificial Limb Program, principal responsibility for fundamental studies in normal and amputee gait and in lower-extremity prosthetics has, since 1945, resided in the Prosthetic Devices Research Project at the University of California, Berkeley Campus.

But the problems facing the leg amputee are not wholly prosthetic. Many, indeed, are clearly medical. For the amputee, being no longer the whole normal individual, manifests gross structural and physiological changes to be dealt with successfully only by the physician.

The Lower-Extremity Clinical Study being conducted jointly by the Department of Engineering, University of California, Berkeley, and the University of California Medical School, San Francisco, and in cooperation with the U. S. Naval Hospital, Oakland, has as its chief objectives the analysis of medical problems inherent in the amputated state and the application of fundamental knowledge to practical problems in the management of lower-extremity amputees. Current techniques and practices in the fitting of leg amputees still are so varied from place to place and from prosthetist to prosthetist that some orderly means has been wanting for establishing what is, everything considered, the best prosthetics practice in the lower extremity. Designed to close the gap between basic work in the laboratory and work in the field, the Clinical Study is an outgrowth of the fundamental research in locomotion conducted earlier by the Berkeley Project.

THE BACKGROUND

For a number of years during World War II a group at the University had been conducting research in the field of biomechanics and had published data relating to the behavior of the upper extremity. In the autumn of 1945, therefore, the University was approached by a representative of Northrop Aircraft, Inc., a company which at that time was already engaged in prosthetics research *(70)* under contract with the then Committee on Artificial Limbs of the National Academy of Sciences— National Research Council. It was requested that the University group undertake an investigation aimed at providing information that

could be utilized in the design and construction of lower-extremity prostheses.

The suggestion having been taken under advisement, the entire Committee on Artificial Limbs met at the University shortly thereafter to consider the proposal and to evolve details of contractual arrangement. Out of this meeting came two basic observations. One was that, inasmuch as the financial support for the work was to come from public funds, any information derived from the contract would have to be shared with all other contractors participating in the Artificial Limb Program as well as with the general public. The other was that, in the opinion of the conferees, between five and seven years of study would be required before sufficient detailed and quantitative information could be accumulated to effect substantial improvement in lower-extremity prostheses *[113]*. At the outset, the University group insisted that it be kept free of the task of developing prosthetic devices—that it simply be permitted to investigate normal human locomotion and to furnish the collected data for others to use. The original concept of the scope of the project—as a program of basic research in human locomotion—has been adhered to up to the present time, the only deviations having involved development of experimental devices *(24,25,80,81,82,95,102, 112)* needed to assist in the locomotion studies.

The early years, then, were spent in working out techniques suitable for recording objectively the motions and the forces involved in the gait of man *(22)*. Of course, the investigators took advantage of all the previous work in this field, not only that done by other contractors *(1,12,49,51,67,71)* participating in the Artificial Limb Program but also that contained in material, particularly that of Elftman*(26,27,28,29,30,31,32,33,34,35,36,37,38,39, 42,43,44)*, published in the United States and in foreign countries over a period of many years. By 1947, enough data had been accumulated to publish a comprehensive report *(102)* on the walking pattern of normals and of leg amputees.[4]

Attempts to translate the results of basic research into criteria for the improvement of prosthetic devices led to the second phase of the project, that is, to developmental research, an area that involves engineering and prosthetics technology. During the last few years, this phase of the project has been conducted on a relatively small scale. As devices were prepared for trials by amputees, the problem of fit and alignment had to be attacked, and hence fundamental studies were undertaken in this area in order to establish a set of basic principles and techniques *(103,104,106,108,110)*. Because fitting and alignment contribute most to the comfort and therefore to the success of any artificial leg, the validation of these principles and techniques formed the basis for embarking on the third phase of the project, the Lower-Extremity Clinical Study, an activity that provides a laboratory where medical and prosthetic problems can be handled under controlled conditions. It offers an opportunity to see how individual solutions may be obtained by applying a set of general principles based on biomechanical considerations. Until recently, the study group has been concentrating on the problems of the above-knee amputee because that case appeared to offer neither the most difficult nor simplest set of circumstances.

THE LOCOMOTION STUDIES

Muscle Physiology

When the Prosthetic Devices Research Project first was organized, man was viewed as a machine, the object being to measure the displacements, accelerations, and forces required in human locomotion *(4,7,14,15,18,19,21,47, 48,96,105)*. But man is more than a single machine. He is powered by a complicated system of many internal engines served by muscles. Accordingly, the study was broadened to include the field of muscle physiology *(73,74,75,76,77,92,100)*. Investigation of the behavior of the musculature during normal locomotion (Fig. 1) revealed the basic action of the various muscles involved *(8,52,111)*. It was

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[4] The 1947 report *(102)* contains an extensive bibliography of earlier work, mostly German, on the mechanism of human locomotion and on related matters. It is available, either in photostat form or on microfilm, from the U. S. Armed Forces Medical Library, 7th Street and Independence Ave., S. W., Washington 25, D. C

shown that in locomotion each muscle acts when it is near its rest length but that it acts for a very short period of time in each walking cycle *(60)*. This action makes the contraction essentially isometric and limits the activity of each muscle fiber to a few twitches. Under these conditions the muscle works with minimal energy and maximum tension, which helps to explain why a person can walk considerable distances without tiring.

Upon working out the speed of contraction, it was found that, if muscles are halved, their contractile velocities likewise are halved (Fig. 2). Utilizing a profile electromyographic recording (electromyogram rectified and dampened to give a relatively smooth line), and taking the maximum amplitude in a given cycle as 100 percent, the average durations with an amplitude greater than 75, 50, or 25 percent are approximately 0.04, 0.1, and 0.2 second, respectively *{55,56,58,61,62,89, 116)*. Since it seems probable that the profile electromyographic amplitude largely indicates relative numbers of active motor units, it would appear that most of the units participating in this phasic action are active during bursts of 0.1 to 0.2 second only. According to Weddell *{115)*, at a repetition rate of 20 per second or less most motor units would fire in each cycle one to four times only. In such a case, any temporal summation taking place at neuromuscular junctions would not be effective fully, and the action of a motor unit, at least in a normal phasic pattern like locomotion, would not have the character of a sustained tetanus.
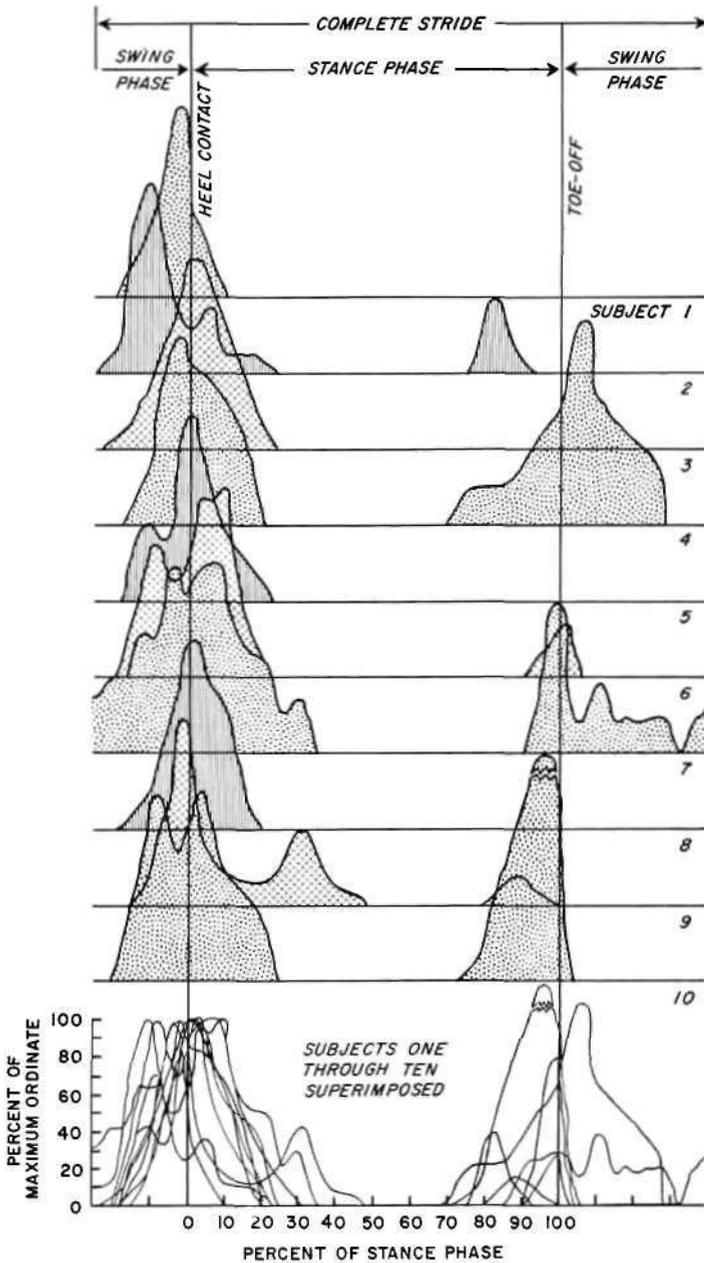
As a result of these investigations, in 1947 the group at Berkeley, noting the earlier work of Blix *{11)*,

Fig. 1. Typical electromyographic summary curves, in this case for the hamstring group. Ten subjects. Cadence: 95 steps per minute, level walking. Data from UC studies *{102)*.

Fig. 2. Relation between the maximum speed with which a muscle can contract and the weight with which it is loaded. When the length of the muscle is halved, its speed of contraction is also halved.

(17) of the University of Copenhagen as the best so far done on normal human muscle dynamics.

Energy Requirements

In another study, an investigation was made of the dissipation of energy (Fig. 3) in human locomotion *(13,14,15)*. Results showed that approximately 50 percent of the energy consumed in walking is used simply in bouncing up and down, that is, in vaulting over one leg and then the other. The other half is used in the oscillations of the legs. It is therefore apparent that, if the amputee is not to be subjected to unduly large energy demands, he must have a smooth pathway of displacement of the center of gravity of the body *(23,40,41)*. Any deviation from the smooth, natural locus of the center of gravity means excessive dissipation of energy and consequent degradation into heat *(53,97)*.

was first to call attention to the length-tension relationships existing in human muscles *(57,63, 64,83,84,85,86,87,90,91,93,94)* and thus laid the basis for the decision to use certain muscles for the cineplastic technique *(2,3,9,99)*. The characteristics of the length-tension diagram have since proved to be of fundamental importance in devising prosthetic aids for upper-extremity amputees *(10,101,107)*. The cineplastic muscle tunnel, comprising a skin-lined tube placed through the distal end of a muscle, permits an amputee to utilize effectively his own muscle forces for activating an artificial arm or hand. But in order to operate a cineplastic prosthesis efficiently, it is necessary that the muscle be near its rest length, so that it can generate a force sufficiently large and so that it can shorten enough to carry out necessary movements *(17)*. Appearing in publications as early as 1949, the work conducted at the University of California has been recognized by Buchthal


Fig 3. Typical moment-angle diagram for the leg of a normal subject during level walking. From Bressler *[sic]* and Berry *(14)*.

Contrary to much popular belief, man not only pushes his way through space. He also *pulls* his way *(15,102,105)*. Indeed, deceleration of the swinging leg, not push-off from the other toe, provides the greater part of the energy for locomotion, the proportion attributable to deceleration of the swinging leg being about 4, that attributable to push-off only 3. Energy is absorbed by the knee to decelerate the leg and foot during the swing phase, but not all of the energy so absorbed is lost *(4)*. A considerable portion is stored and returned to the system in the later part of the swing phase to impart continued forward acceleration at the time when most of the body's potential energy is lost *(48)*. Thus locomotion is due not only to the push of the member in support but also to the pull of the deceleration in the swinging knee.

Because the above-knee amputee has no calf group, and therefore cannot contribute the equivalent of this force at push-off, it was suggested that some conservation of energy might be effected in a prosthetic device without an ankle joint *(78)*. That this was a correct deduction has since been demonstrated (Fig. 4) in the Stewart-Vickers leg *(20,69,112),* in which the ankle is locked at toe-off until 20 deg. of knee flexion has occurred *(114)*. It has the

highest net output and the lowest total input of all legs tried to date (Fig. 5).

AMPUTEE PAIN

Intimate contact with amputees led to the early investigation of pain as related to the amputee patient *(65)*. In 1946 a team of interviewers set out to question amputees in various hospitals, particularly in the Veterans Administration Hospitals and in the Naval Hospital then at Mare Island. Over a period of a year and a half, detailed histories were obtained from 80 patients. As a result of this review, further funds were provided by ACAL to establish a Pain Clinic at the University of California, primarily to evaluate pain as found in the amputee. Established in August 1949, the clinic functioned until January 1953.

In June 1952, an analysis of 218 amputees was reported *(109)*. In this study, which constitutes one of the largest series on record, the type and frequency of pain in the amputee were explored. Because it was thought that perhaps deficiencies in stump circulation might contribute to the pain experienced by the amputee, circulatory studies were undertaken. Concurrently, innervation of the deeper tissues was studied *(54)*. Sections of tissue were taken from periosteum, muscle, and skin, and the
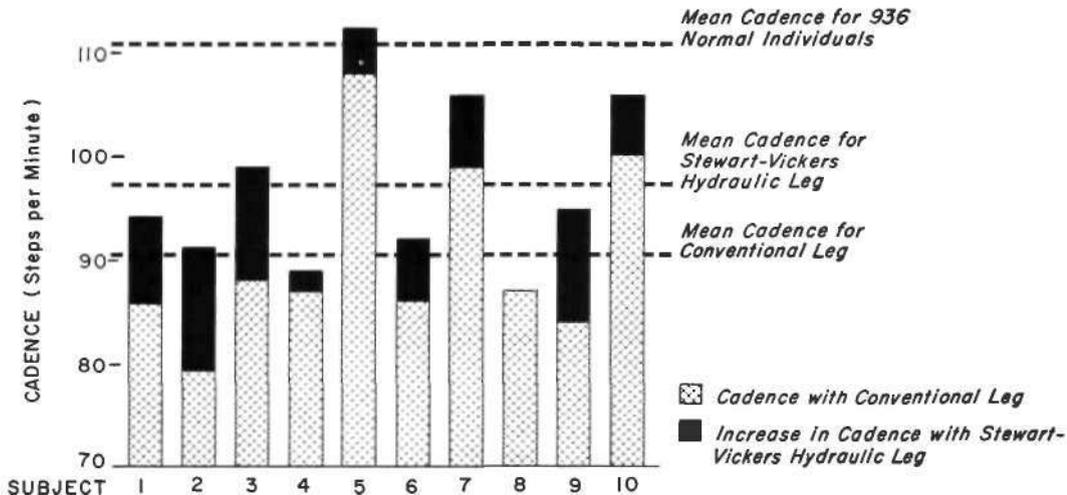

Fig 4. Cadence changes observed in above-knee amputees asked to walk at "normal" speed first with a conventional limb and then with the Stewart-Vickers (locked ankle) prosthesis *(114)*.

nerve supply to these tissues was demonstrated by a methylene blue technique.

One of the most intriguing aspects of this investigation was the work with normal individuals in whom irritative lesions purposely were produced in the deeper tissues *(45,46,109)*. With the authors, some 75 medical students, and three laboratory assistants serving as subjects, 0.5 to 1.0 cc. of 6-percent saline solution was injected systematically into the paravertebral muscles at each intervertebral level from the atlanto-occipital area to the lower sacrum. Five subjects were used in the testing of each injection site, a total of 140 individual observations being made. Although the distribution of pain approximated a segmental plan, it also overlapped considerably and differed in location from the conventional dermatomes. It was found that, in any irritation of deep somatic tissues, pain did not restrict itself to the area of injection but tended to radiate distally into the extremities. Injection of 6-percent saline into any given interspinous level produced in the normal a characteristic pain distribution that was remarkably constant from subject to subject.

The distribution of pain referral from deep structures in the normal suggested similar investigations in the amputee. To elicit the sensation of the phantom limb, it was necessary to inject the salt solution into the appropriate interspace. In the normal, radiation of pain into the lower limb was most marked when the interspinous tissue between L4 and L5 was affected, and in the above-knee amputee the L4-L5 interspace also gave the best response. The immediate reactions of amputees resembled those reported by normals—a rapid onset of pain close to the site of injection and then, in the case of L4-L5 injection, radiation into the buttocks and the posterolateral aspect of the thigh. In nearly all instances there occurred a rapid "filling" of the absent areas of the phantom limb, the subjects usually evidencing surprise at the sudden totality of a
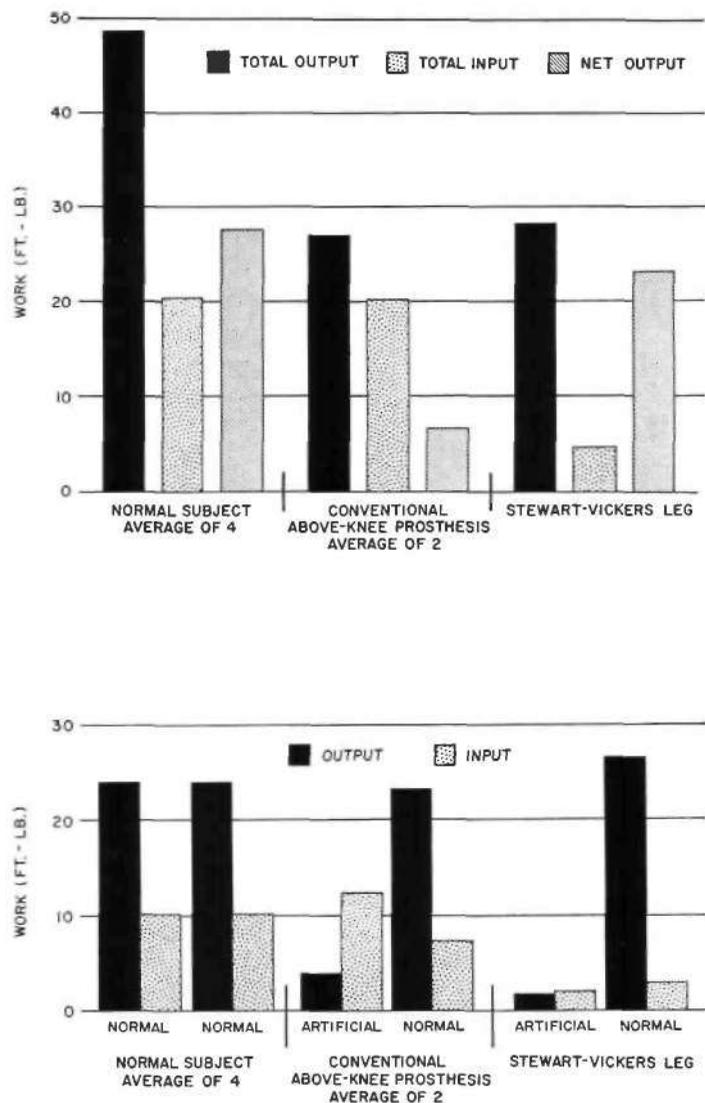

Fig. 5. Energy characteristics of the normal ankle compared with those of the conventional leg and the Stewart-Vickers leg. Top, total input, total output, and net output of both ankles per stride. Bottom, input and output of each ankle per step.

SITE OF INJECTION L4-5 →

ABSENCE OF PHANTOM

ABSENCE OF PHANTOM

ABSENCE OF PHANTOM

RADIATION OF PAIN

■ SEVERE ▨ MILD ▨ NUMBNESS

A *B* *C*

Fig. 6. Effect of interspinous injection of saline on the painful phantom limb of one subject. *A,* Phantom before injection. *B,* Radiation of sensation induced by injection of 6-percent sodium chloride solution. *C,* Residual sensation following injection.

phantom limb even though the new portions were seldom, if ever, immediately painful.

Severe pain was a frequent feature in the portion of the phantom present before injection. After injection the pain often spread into the newly "filled in" portion of the phantom limb. Transient pain following injection occurred in phantom limbs regardless of the existence of preinjection pain. But in many cases involving pre-existing phantom pain, a secondary decrease in the amount of pain followed the injection, in some but not in all instances the decrease being preceded by a transitory accentuation of the pre-existing pain. Occasionally, the decrease reached the point where no pain was felt, so that the amputee experienced the first complete relief in many months.

The decrease in pain is even more remarkable when one considers that it is brought about by the application of a noxious stimulus to a tissue remote from the phantom itself. For example, in an above-knee amputee who had undergone amputation two months before the investigation, there was a phantom sensation of the "foot" only, the phantom being very painful with the sensation of severe constriction of the great "toe" (Fig. 6). When saline was injected

into the L4-L5 interspace, much of the intervening phantom limb was filled in almost immediately, the anterior aspect of the "leg" becoming the most prominent part. Soon after the phantom was "completed," the pre-existing pain in the "foot" increased in intensity and area. This state continued for five or six minutes, whereupon the pain began to decrease and continued to do so until, in another five minutes, it had disappeared completely. Numbness, but not pain, remained in the "foot" only. In some instances even phantom awareness disappeared after saline injection.

In general, the saline injections had greater effects on phantom limbs than on real ones, a peculiar susceptibility best illustrated by the effects of mid-line injections. An accurately placed mid-line injection in a normal subject produces very little radiation, the severe pain being confined to a rather small area in the immediate vicinity of the injection. In the case of the amputee, however, such minimal radiation in the trunk is accompanied by profound effects on the phantom extremity. Every conceivable change in phantom form and phantom

pain can result from interspinous injection of an irritating hypertonic saline solution, the changes probably stemming from the sudden increase in the sensory inflow at the particular segmental level.

Out of these observations came, then, one method of treating phantom pain, for when a small amount of hypertonic saline was injected into the appropriate segmental interspinous ligament, the phantom experience was changed and pain occasionally was relieved. This finding led to the use of hypertonic saline for the treatment of various painful conditions. Although permanent cures resulting from such techniques are not numerous, the method may prove to be a valuable addition to the modern medicine chest, which is by no means rich in effective pain palliatives (98).

It deserves to be noted that, in seeking the origin of the phantom experience, one must look not only for direct involvement of the nerves of major nerve trunks. The entire segment of the extremity must be investigated for any irritative skeletal lesions arising from the joints, the muscles, or the connective
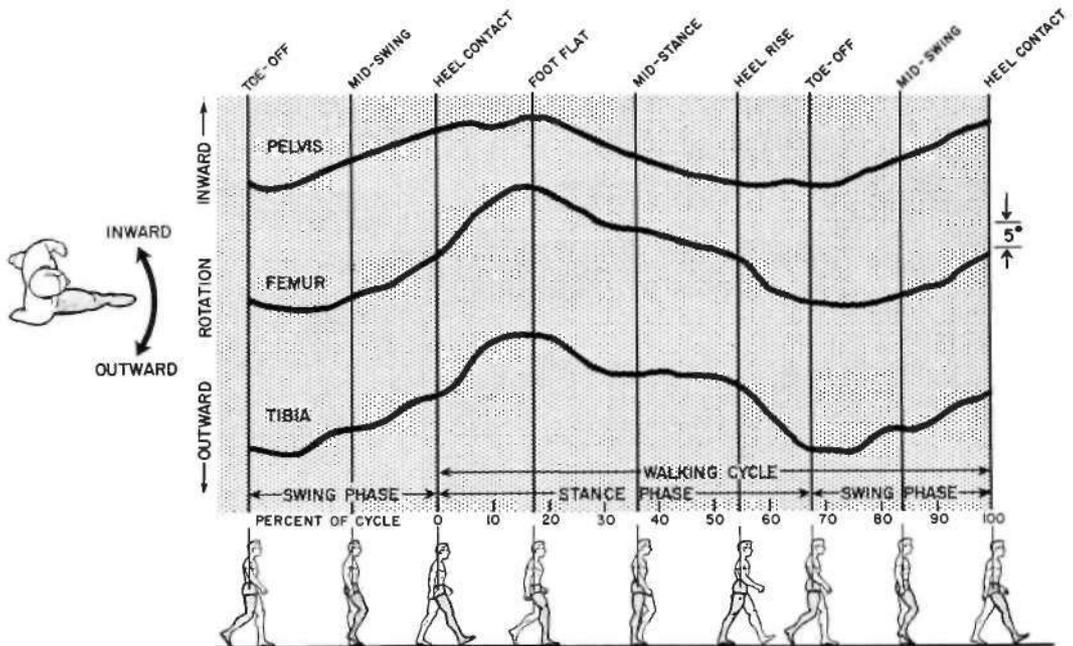

Fig 7. Typical relative rotations of the pelvis, femur, and tibia in normal, level walking. Data from UC studies (102).

tissues of the stump or from portions proximal to the stump.

EVOLUTION OF BASIC DATA

From the basic studies now has come much information of value in prosthetics. As early as 1947 it was determined *(59)* that in normal walking the leg rotates in space internally and externally about 15 deg. on the average (Fig. 7). That this horizontal rotation of the extremity might be of some importance in human locomotion has since been known as the "Berkeley fetish," and as far as is known no one has yet taken cognizance of the fact in any successful limb design. In 1950 it was suggested *(79)* that it would be of considerable value if deceleration at the end of the swing phase could be incorporated through some sort of variable-cadence knee joint. This has been done in at least one device, the U.S. Navy above-knee leg *(68,112),* now available commercially (see *Digest,* this issue, page 65). Several others currently are under development.

At the same time it was suggested that, inasmuch as the above-knee amputee can obtain no forward propulsion by contraction of the calf group, the ankle joint is of little use—that, indeed, if an ankle joint with rubber bumpers is used, energy is lost by hysteresis of the bumpers. As already mentioned, the improved performance of the Stewart-Vickers leg, in which the ankle is locked at toe-off up to 20 deg. of knee flexion, proves the validity of the original observation. Similarly, it was pointed out that, because of the interrelationship between the ankle-foot function and the knee-joint function, greater stability would be required of the knee joint were the articulated ankle to be abandoned.

In 1953, Saunders, Inman, and Eberhart *(97),* summing up the results of all the basic studies, pointed out that there is an interrelationship between all displacement patterns of all segments of the lower extremity, that there are six major determinants in locomotion, that modification of one results in modification of the others, and that any changes in the knee or ankle, either in normal or in amputee, are necessarily accompanied by compensatory changes in the remaining joints. Basically,

locomotion is the translation of the center of gravity through space along a pathway requiring the least expenditure of energy (Fig. 8). The six major determinants of the pathway are pelvic rotation, pelvic tilt, knee flexion, knee extension, knee and ankle interaction, and lateral displacement of the pelvis. Serial observations of irregularities in these determinants provide insight into individual variation and a dynamic assessment of pathological gait, which may be viewed as an attempt to preserve the lowest possible energy consumption by exaggerating motions at unaffected levels. Compensation is reasonably effective with the loss of one determinant, that at the knee being the most costly. Loss of two determinants makes effective compensation impossible, the cost of locomotion in terms of energy then being increased threefold, with an inevitable drain upon the body economy.

With regard to the surgery of amputation, the studies in muscle physiology suggested that considerable improvement might be effected in lower-extremity prosthetics were muscles fixed in the distal end of the stump so that they could not retract *(88).* As previously pointed out, retraction of these muscles means shortening, and shortening means an inability to develop natural tensions. More recently the studies have suggested that, in order to retain normal weight-bearing through the shaft of the femur, more attention should be paid to the possibility of end-bearing rather than to the more conventional method of weight transmission through the ischial seat. All of these ideas, derived from the results of the early studies on locomotion, were offered to the limb industry by the University group in the hope that designers or manufacturers would incorporate the recommended features into new prostheses.

THE CLINICAL STUDY

In the spring of 1953, after years of basic study, the question arose as to what might be done toward applying to the amputee problem some of the knowledge gained. After several months of discussion, the UC Prosthetic Devices Research Project accepted a proposal to institute the so-called "Clinical Study," the principal objective being to draw upon the

Fig. 8. The sum of the effects of the six determinants of gait. The pathway of the center of gravity is a smooth curve in both horizontal and vertical planes. From Saunders, Inman, and Eberhart *(97),* by permission of *The Journal of Bone and Joint Surgery.*

pool of fundamental knowledge, to attempt to apply it toward the solution of practical problems, and to see whether or not there would emerge certain definite devices or methods which could be passed on to the artificial-limb industry and to prosthetists. Last year, then, the clinical program was established, and currently it is the center of attention.

To organize such a clinical study obviously required a limbshop and examining rooms. Through the kindness of the Navy, space was afforded at the Navy Prosthetics Research Laboratory at the U.S. Naval Hospital at Oakland, California. There the setup includes a small limbshop where prosthetics work is done, a medical examination room, fitting and training rooms, an evaluation and photography room, and conference rooms, the entire operation being conducted in cooperation with the limb industry. Through the Industry Advisory Committee, amputees are selected on the basis of referral by limbshops, by physicians, by rehabilitation agencies, by the Veterans Administration, and by direct personal contact. After preliminary screening by the Clinical Study Group, an individual is selected only with the approval of the Industry Advisory Committee, and all of the work is done with the

knowledge, assistance, and cooperation of the artificial-limb industry.

Because it is concerned primarily with research, the Clinical Study is not a commercial operation, and consequently production is not high and is not supposed to be. Thus far only 16 subjects have entered the clinic. Of these, 10 are unilateral above-knee amputees ranging in age from the teens to the seventies, two are bilateral above-knee cases, one is a bilateral above-knee/below-knee case, two are hip-disarticulation cases, and one is a unilateral below-knee case. Five are in the follow-up stage, six in the postfitting adjustment stage, three in the fitting stage, and two in the pre-prescription stage. All save one have been complicated cases, presenting difficult problems that nobody else wished to tackle. From particular cases such as these have come practical answers for other difficult cases.

A thorough and complete study—from the medical, biomechanical, and prosthetic points of view—is made of each case, and individual problems are diagnosed and corrected. To find the best possible solution in any particular case requires a knowledge of what attempts have been unsuccessful and why they failed, for sometimes a great deal more is learned by

determining why one proposed solution failed than by determining why another was successful.

The clinic team consists of an orthopedic surgeon, a prosthetist, a physical therapist or amputee instructor, and sometimes an engineer (5). This group makes the initial evaluation and provides a prescription *(6)* based on complete data including a medical history, an analysis of existing condition of the stump and of the rest of the body, and an evaluation of the old prosthesis. The prescription is reviewed by the Clinic Study Panel, including several orthopedic surgeons, a psychiatrist, a prosthetist from industry, and an engineer familiar with prosthetic problems. Once the prescribed device is fitted, the results are viewed by the Panel, and the reasons for success or failure are documented fully so that the case may serve as an example for future reference. No experimental devices are used in the clinic program. Only those devices available commercially are fitted to the subjects.

INDUSTRY PARTICIPATION

Active participation by individual members of the artificial-limb industry has not yet started, but plans are now being made for such activity in the immediate future. That part of the program will involve working with prosthetists, screened by the industry, who will visit the clinic for a period of orientation. They will follow cases through the clinic study and then be assigned a shop case on a cooperative basis. The clinic team will act initially as a review committee in preparing the prescription, but the individual prosthetist will fill the prescription in his own shop. After fitting, the amputee and the prosthetist will return to the clinic for evaluation. This procedure provides a twofold check. It evaluates the prosthetist's degree of efficiency and tests the validity of the clinic's method of prescription.

PROSTHETIC PROBLEMS

Crotch Pressure

Because enough time has now elapsed to be sure that more than temporary success has been achieved, some general ideas can be discussed with a fair degree of confidence. The most common complaint heard by the group relates to crotch pressure. In every instance, however, the condition has been eliminated. Correcting for excessive crotch pressure involves two things—the right socket shape and correct alignment (page 35). Proper socket shape is ensured by providing for ischial-gluteal bearing (which prevents sinking into the socket), by controlling the anteroposterior dimension, and by raising the height of the socket brim.

Localized Socket Pressure

The next most common complaint relates to edema. Rarely has there been a case of the suction socket where edema could be traced to high negative pressure alone. Excessive crowding or tightness invariably were contributing factors. Edema may result principally from a high rate of pressure change at any point along the length of the stump. Because emphasis has been placed on socket shape near the top brim, not enough attention has been given to good fit throughout the length of the stump. Any constrictions or ridges, including those formed by muscle groups, cause pressure changes that interfere with venous return. The inside finish of the socket also may be a factor. In one instance, for example, a severe case of edema was alleviated by providing the socket with a smooth, high-gloss finish.

Socket Brim

Skin irritation around the socket brim also is a source of annoyance and discomfort. Accordingly, dermatologists are cooperating in the program. They examine amputees having skin problems and outline procedures for therapy, including the taking of biopsies of the skin. Pigmentation is evaluated to determine whether or not it is due to capillary hemorrhage caused by decreased suction or whether it is merely a pigmentation that often occurs in areas of friction. Out of this study should come a routine test and a new modality of skin care for the leg amputee.

Again, the condition can be eliminated by controlling the shape and height of the anterior and lateral brim above the ischial seat. Medial

width also is a controlling factor because it determines the total amount of pressure exerted by the front of the socket to maintain stability on the posterior weight-bearing surface. And, as in the case of edema, the inside finish is important in preventing skin damage. Sitting discomfort, a complaint often heard, usually is relieved by using a flat back, by not having the inside edge of the seat too sharp, and by ensuring that any channel for gluteal relief is not too large.

Alignment

Alignment is a continuing problem, and the development of guiding principles is most important. Although general principles are comparatively simple to state, to understand them fully and to apply them to individual cases is difficult. One of the objectives of the clinical program is to apply to typical problem cases the alignment principles developed through fundamental research and to develop examples showing how these principles can be applied, why they work, and the end-results that can be obtained. Naturally, the best results are obtained when the stump is so oriented as to take full advantage of the remaining hip musculature. There is a growing body of information relating to a number of common problems—problems associated with changing from a pelvic belt to a suction-socket leg; problems concerning the very muscular stump with prominent hamstrings or with some particularly firm muscle or muscle groups isolated in the stump; problems of the short and the long above-knee stump; problems caused by the flabby stump; and problems of inside finish.

MEDICAL PROBLEMS

Often the problems of the amputee, both in the lower extremity and in the upper, stem not from an ill-fitting prosthesis. More often the problems can more properly be termed medical. Accordingly, the Clinical Study includes investigation of those aspects of amputee rehabilitation related to physiological changes associated with loss of limb.

Pain—Phantom and Real

As pointed out long ago *(66.72),* loss of the normal limb so often is followed by the appearance of some form of phantom limb that, when a patient does not acknowledge one, it is suspected that he is withholding information or that the phantom has been repressed. Statistics show that the phantom is a normal phenomenon in the sense that most amputees have it. It is pathological, however, in the sense that the amputee perceives something that actually does not exist.

In general, awareness is a matter of degree and, to some extent, a matter of verbal conventions. Some amputees say that the phantom has the same unobtrusive quality as does the material counterpart, that it appears only when called upon. Sometimes the amputee has difficulty in remembering that the phantom is unreal and that it does not serve in the capacities of its living predecessor. The normal person is not particularly aware of his limbs unless his attention is drawn to them in some way. Except under the impact of a sudden stimulus, or when a special effort is made, preferably together with a movement, our awareness is potential and shadowy in nature. With the eyes closed, and with the limb at complete rest, awareness is, in fact, not too far removed from mere imagination. To make certain that the limb exists, we move it, look at it, or rub some part of it. The amputee cannot conduct such an empirical test.

Sometimes the patient can sense his lost limb as acutely as he can the remaining real one, and he often can imagine that he can "move" the phantom. More often, however, the phantom draws attention to itself by some "abnormal" sensation which makes the amputee more aware of it than he is of his real limb. Fortunately, only a small percentage of all phantoms habitually are painful. Some typical ones are shown in Figure 9.

Frequently the "foot" seems to shorten and approach the end of the stump. The patient illustrated in Figure *9C* experienced "telescoping" of the phantom, a phenomenon which, contrary to the observations of most other writers on the subject, was found infrequently in the Berkeley series. It is true that relatively undifferentiated parts like the calf and the forearm commonly are not felt. Some phantoms of distal parts are, from their onset, situated

at the normal distance from the trunk. Others always seem to be located closer to the stump than normal. A few patients experience a gradual shrinkage of intermediate phantom parts, as has occurred over a period of years in the subject illustrated in Figure 10. In this case, all that remains of the shrunken ghost are the "toes," and these have come to lie not in empty space, as is the rule, but inside the stump. Not infrequently a phantom which has shortened may, on application of a prosthesis, lengthen and actually become identified with
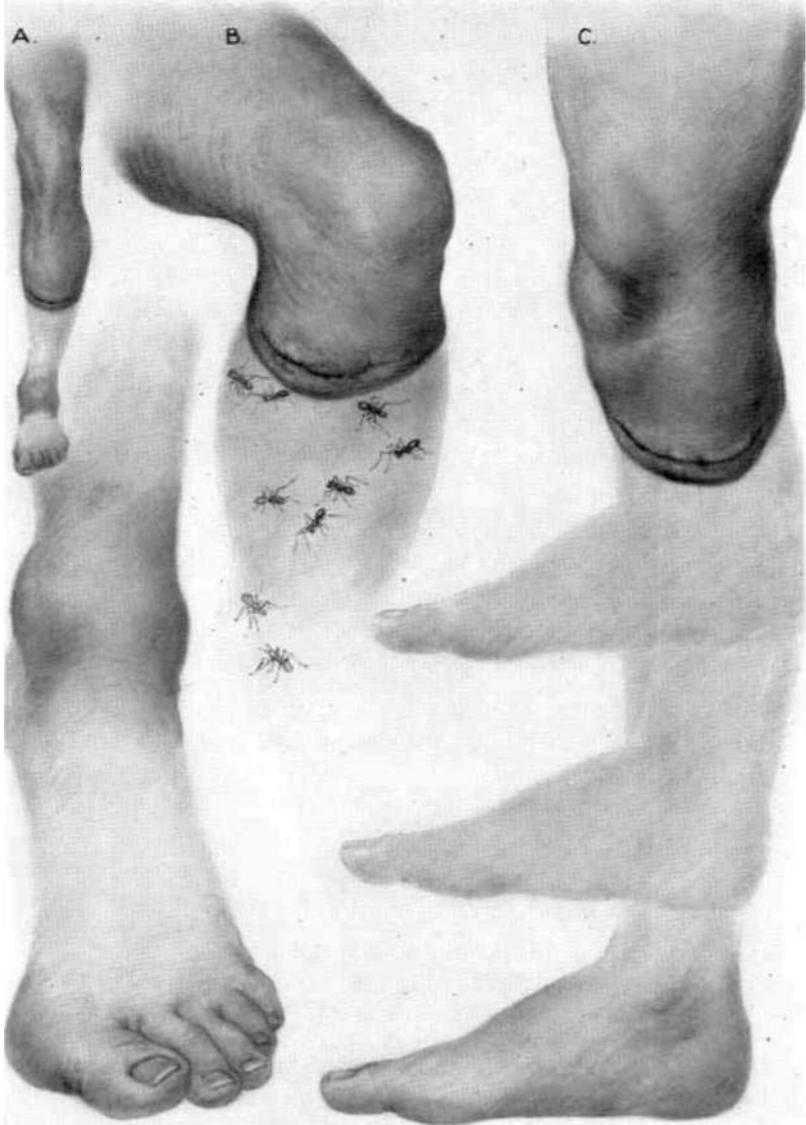
[Fig. 9. The phantom limb, a phenomenon of almost universal occurrence among amputees. *A*, Phantom toes and ankle, reported more frequently than are other phantom parts of the amputated lower extremity. *B,* Mild "tingling," characteristic of the painless phantom, is often described in terms of "crawling ants." C, The "telescoping" phantom, in which the foot, over a period of time, gradually approaches the stump and finally disappears within it.
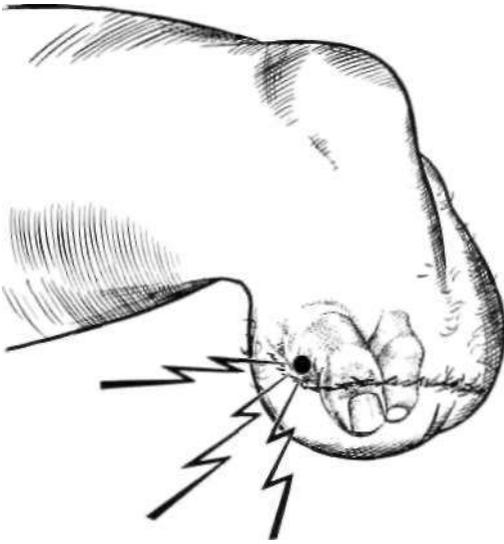
Fig. 10. A rare and peculiar form of phantom experience. Here the two "toes" seem to reside within the stump itself.

the artificial limb. Thus, in one instance, a young above-knee amputee felt as though the shortened "foot" were appended to the stump. When he wore his prosthesis, however, the phantom foot felt as though it were in the position corresponding to that of the artificial foot.

Awareness of the missing member may or may not be described as basically unpleasant, but it is subject to intermittent unpleasant sensations—itching, tingling, or pain (Fig. 11). As pointed out by Livingston *(65),* the pattern of the painless phantom bears no resemblance to the areas of distribution of the major peripheral nerves. Thus the partial nature of the phantom cannot be ascribed to the affection of certain nerve lesions in the stump. Rather, the pattern of the phantom seems to relate to the most mobile parts and to those serving the highest degree of sensory function. But a substantial number of amputees experience, at one time or another, some sort of painful phantom of varying duration (Table 1).

How many amputees have pain? Taking into consideration the inadequacies of follow-up information, the subjective character of the pain experience, and the semantic difficulties

beclouding the term "pain," a conservative estimate would be that 80 percent of all amputees are substantially free of pain and are either being trained for useful work or else are already actually so engaged. It is likely that, of the remaining group, possibly half are faced with severe intermittent or persistent pain. Because of persistent, incapacitating pain, approximately 10 percent of all amputees never get into a limbshop, never get out of the doctor's office. They become narcotic addicts and often commit suicide. Where pain enters the phantom syndrome, it may assume large clinical importance. If it is excruciating and persists for long periods, it may take a devastating toll of the whole personality and physical well-being.

In describing severe pain, we all use a vocabulary taken from common objects known to produce injury. Lesser pains are described in terms of cutaneous and deep sensations. Thus we speak of "pressure," of "pins and needles," of "sharp" pains and "dull" aches, of "stabbing" and "shooting" pains. It seems unlikely that man at his present stage of evolution ever will devise a specific terminology for pain because he has no special organ for observing his discomforts. No matter how introspective a person may be, his account of pain always is phrased in imagery taken from other fields of experience. Nothing could be more real than these sensations, but we say "as if" to give them intelligible expression. The vocabulary is metaphorical.

It is not surprising, therefore, to find am-

Table 1

INCIDENCE AND SEVERITY OF PHANTOM PAIN IN A BERKELEY SERIES OF 80 AMPUTEES

Pain Characteristics	No. of Patients	Percent of Total
No phantom pain at any time	24	30
Moderate pain at one time, now gone entirely	18	22
Moderate pain which still occurs	32	40
Severe pain at some time but now gone entirely	3	4
Severe pain which still occurs	3	4
Total	80	100

putees using language akin to that of the torture chamber when they try to do justice to their agonies. They hardly go further than anyone else in telling about physical suffering,

Nor do they hallucinate when they talk about "ropes" and "vises," for they remain aware of the imaginary character of these similies. It is possible, however, that, as the tearing and
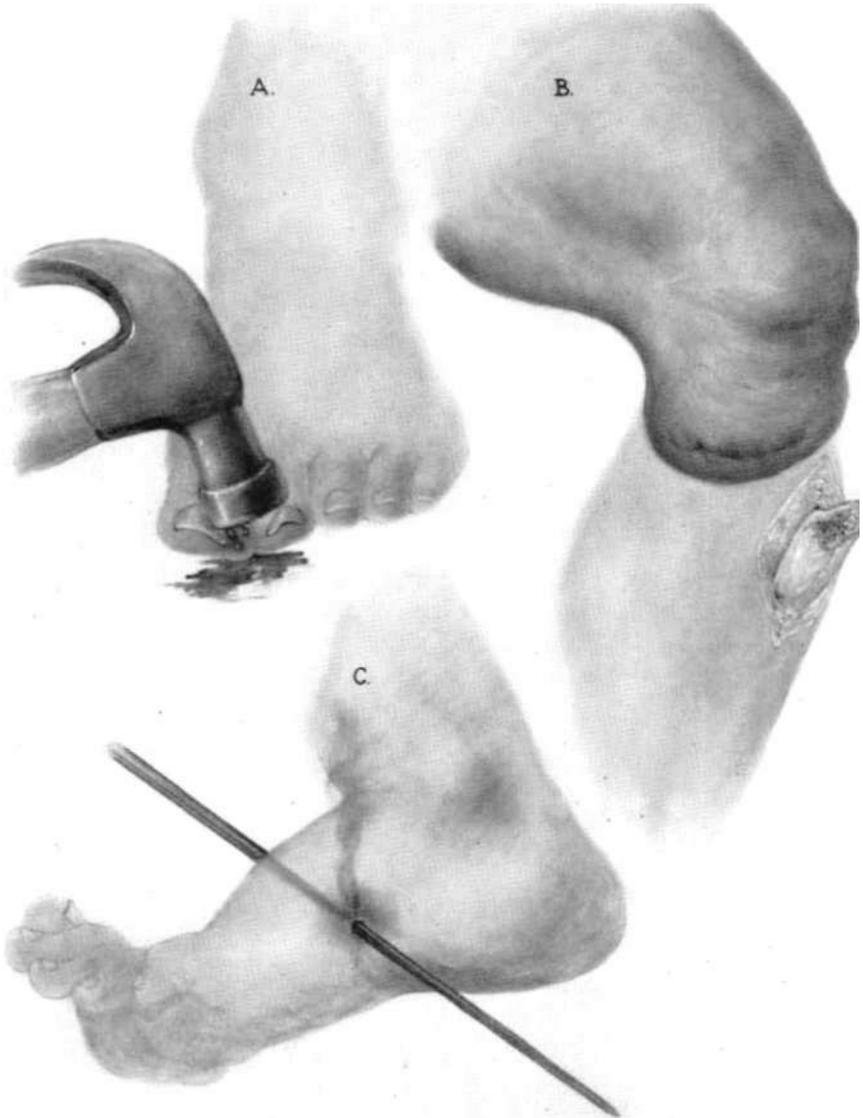
Fig. 11. The painful phantom, of fairly common occurrence among amputees at one time or another. Only some 30 percent experience no phantom pain at any time. Probably about 10 percent face persistent and sometimes incapacitating pain. *A,* Among the similes used to describe a phantom pain is "as if my toes are being crushed by a hammer." *B,* Pain experienced at the site of an injury leading to amputation, such as a fracture, often persists as a part of the phantom pattern. C, The "hot wire" sensation and involuntary cramping of phantom toes are among the other frequent manifestations.

squeezing sensations are felt in a part of the body known to be missing, the suffering is heightened and the imagery made more vivid by the ghostly character of the phantom.

It has been argued that phantom sensations are hallucinations because they entail a belief in the reality of an absent object, or that they are illusions because irritations of the stump are being misinterpreted, or that they are normal sensations because the cerebral representation of the once-present member still is intact. Some workers have correlated the type of sensation with the "level" of its origin in the nervous system, painful sensations being ascribed to pathological conditions of the cut nerve end in the stump or to mental aberrations. But classifications of either the amputee's descriptions or of the presumptive causes bringing about the sensations have thus far been unsatisfactory. The various frames of reference used in the statistical survey at Berkeley do, in fact, overlap. Duration and frequency of pain have some influence on the complaint of severity. Tingling and burning seem to be more superficial and, however annoying, more tolerable than do tearing, stabbing, cramping, squeezing, and crushing. It should be understood, however, that there are degrees of each of these and that, as such, intensities may, to a point, be compared with each other.

It is obvious that a patient's account of his painful feeling is colored by his personality. The way a person describes such experiences depends not only on the abnormal processes causing them but also on his imagination, his previous experience, his learning, his cultural inheritance, and his vocabulary. But any view which discounts the abnormal physiological processes and credits only their "mental" interpretation is probably in error. The complexity of the nervous system and its integration into one functioning whole does not favor the idea that there is one chief recipient and executive who sorts out the messages from the various parts of the body and, in the case of pain, edits them as writhings and groans or as sentences made up of more or less colorful language. It seems improbable that there is simply one stimulus arising somewhere in the organism and that the ego reacts to this stimu-

lus in a more or less stoic way. A so-called "neurotic" or "imaginative" disposition is likely to pervade the most "bodily" of processes, while a steadfast person is apt to have a stomach and blood vessels no more stable than his emotional display.

Regardless of individual personalities, however, there is a certain uniformity in the complaints of pain-stricken amputees. Although the matter has not been explored from the point of view of psychophysiological typing, it appears that pain phenomena cannot be predicted either from the age of the patient or from the age of his phantom. By the same token, racial or cultural background and physical or mental make-up cannot be used to predict pain phenomena. Nor have the local pathological factors before, during, and after amputation—the factors that might be held responsible for the appearance of pain—been elicited.

Aside from the problem of the painful phantom is that relating to painful stumps (Fig. 12). Amputees may have spontaneous stump pain. Or they may have so-called "trigger points," certain areas which, on slight pressure, tend to produce a flash of pain persisting for various intervals of time. Patients have complained of circumscribed areas of pain in the stump even though palpation revealed no corresponding point of tenderness. These two conditions usually are found together. Nodularities in the stump often are palpable, as indeed they are, on a minor scale, in other subcutaneous parts of the body. Some of these are tender, some are not; some are and some are not connected with phantom pain. In fact, separate places in the same stump may represent exclusive triggers—one for stump pain, the other for phantom pain.

But the conditions prevailing at the end of the stump, including such nodules as the famous "amputation neuroma," do not provide a basis for intelligent speculation. The mere fact that stimulation of a presumptive neuroma often produces pain in the phantom is no proof for the theory that the "cause" of this pain lies solely in the periphery. In order to be disabused of such a notion, one has only to look at certain cases of known diseases of the central nervous system or at. complete

transections of the spinal cord. In the latter, the brain receives no communications from the stump. In cases of painful diseases of the central nervous system, stimulation of the normal peripheral tissues having their nervous connections with the diseased part of the central nervous system often produces an abnormal sensation, including pain. This phenomena always is referred to the periphery. Nobody sounds convincing when he says that
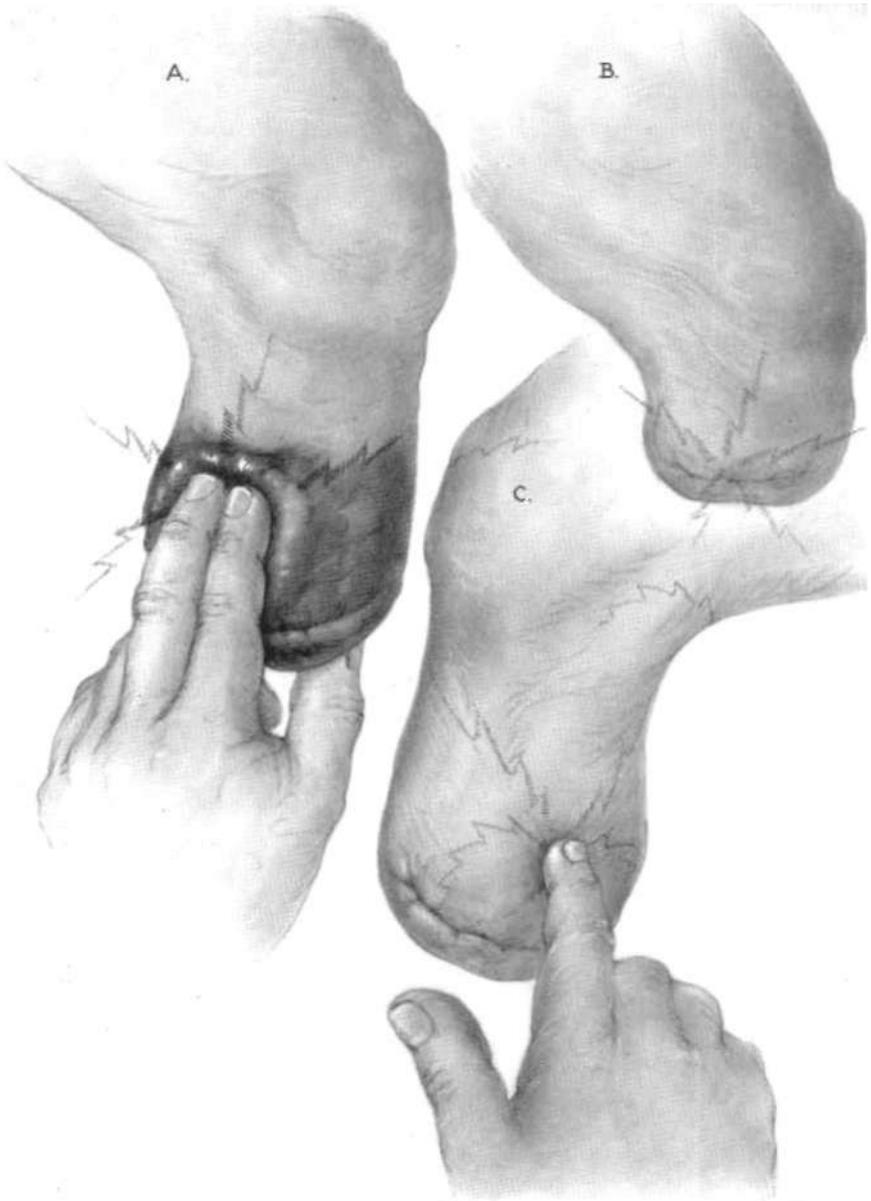

Fig. 12. Types of stump pain. About a third of the clinical reports of pain refer to discomfort in the stump rather than in a phantom part. Stumps may be painful to the touch *(A)* or spontaneously *(B)*. Frequently present are "trigger points," pressure upon which gives rise to pain over a larger area, either in the stump or in a phantom or both (C).
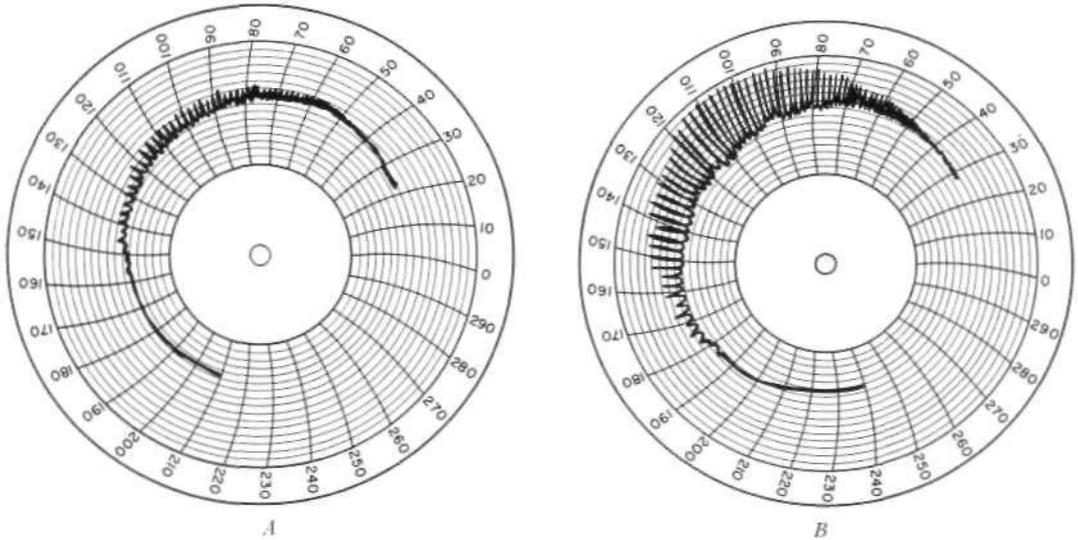
Fig. 13. Pulsations recorded during generalized vasodilatation in a below-knee amputee. Oscillometric records show a smaller amplitude of pulsation in the blood vessels supplying the stump *(A)* than in those supplying the sound limb *(B)*.

he feels pain in the brain or spinal cord. The central nervous system has no conscious sensory representation of itself. The mere description of a painful sensation does not permit detection of its origin. The origin has to be deduced from circumstantial evidence which, in the case of amputees, is lacking. Even where sensations are "triggered off" from the periphery, they can be completed only by participation of the central nervous system, and disturbances may occur anywhere along the line.

We are confronted with the anomaly that stimulation of a certain trigger point within the stump arouses not a distant, painful phantom but one incorporated in the flesh of its own trigger. The specificity of this trigger further is illustrated by the fact that, on the opposite side of the same stump, there may be another tender spot, stimulation of which sets up increased local stump pain.

Circulatory Problems

Investigation of circulation in the amputee reveals that the stump acts as though it were poikilothermic, that is, it has no ability to change its temperature. Rather, the temperature of the stump matches that of the surroundings, as occurs in a cold-blooded animal.

Studies concerning the relationship of the vascular system to pain in amputees have been conducted along three general lines. First has been evaluation of the status of the circulatory system in amputation stumps, both in patients suffering from phantom or stump pain and in amputees free of pain. The second has involved clinical and laboratory studies of selected nonamputee patients suffering from pain syndromes possibly related in pathophysiology to phantom pain. And finally tests have been conducted with various sympatholytic drugs and blocking procedures, first with respect to their effects on phantom-limb pain and related pain syndromes and second in regard to their effects on the circulation of blood in stumps and in painful limbs.

Studied in detail were 43 amputees, 31 without known vascular disease (Group A) and 12 suffering from vascular disease either as the underlying cause of amputation or as a concomitant to the amputation (Group B). Pain in the stump or phantom limb was an important problem for 15 of the patients in Group A and for 8 of those in Group B. The remainder described varying degrees of phantom awareness but denied that pain existed or, if it did exist, that it was disturbing.

One method of investigation was simple

clinical examination. In that survey, stumps appearing to have an adequate blood supply were found, when exposed to air at room temperature, to be almost uniformly cold to the touch as compared with the opposite extremities. In oscillometric tests, the pulse of arterial blood into the stump was found to be significantly smaller than that into the normal limb (Fig. 13). In skin tests with histamine, the appearance of normal flares and wheals indicated that local denervation could not account for the failure of the skin to warm during generalized body warming.

Figures 14 and 15 indicate graphically the results of surface-temperature measurements on the normal extremities and on the stumps of two amputees *(16)*. Skin temperature was measured after initial exposure of the body to cool air in a room with controlled atmosphere, the subject being exposed until finger and toe temperatures were stabilized. Recordings were made by means of thermocouples taped to the skin of the stump and to the contralateral extremities at multiple points along the length of the limb, the thermocouples being applied symmetrically so that points equidistant from the trunk could be compared. All such measurements were made with the subject in a basal state and exposed to room air between 17° and 21°C, conditions leading uniformly to constriction of the cutaneous vessels of the extremities in normal subjects. Under such circumstances, a temperature gradient exists between the proximal and distal portions of a normal arm or leg, so that the surface temperature of a finger or toe is several degrees lower than the temperature at points near the trunk.

Temperatures then were recorded during maximal vasodilatation induced by oral administration of whiskey and wrapping the trunk in an electric blanket. After vasodilatation, the gradient is abolished or reversed in the normal limb, finger and toe temperatures rising to 30°C or higher.

At the end of the initial cooling period, when subjects had been exposed to cool room air for periods of from 30 to 150 minutes, the surface temperature at the distal end of the stump almost invariably was cooler than was the skin at a symmetrical point on the cor-

responding intact limb. Analysis of the temperature gradients found after cooling showed further that, in at least a third of the Group A amputees and in half of the Group B amputees, the stumps were cooler than were the opposite extremities, not merely at the distal ends but for distances of from 20 to 55 cm. from the ends.

In one instance a patient was put in a room at 18°C with nothing across his body except a towel. Over a period of two hours the body temperature was lowered to a point just above that at which shivering occurred. The temperature of the toe in the normal extremity dropped to a low level. When the patient suddenly was given 2 ounces of whiskey and warm water and had an electric blanket placed across his chest, the temperature of the normal extremity rose rapidly. But the temperature of the stump remained constant during the entire procedure, a phenomenon characteristic of all amputation stumps.

A total of 40 amputees (28 Group A, 12 Group B) were subjected to one or more vasodilatation tests, and the responses of 45 stumps were observed. Of these, nearly two thirds failed to warm significantly at a time when the skin temperature of the normal extremities had risen to 30°C as a result of indirect or "reflex" vasodilatation. Only occasionally did stumps show evidence of significant vasodilatation. It occurred with higher frequency in those patients with underlying or concomitant vascular disease than in amputees of Group A. Thus, of 11 stumps in which the temperature rose to the same level as the corresponding point on the contralateral limb, or even to levels reflecting "ceiling" blood flow for skin, only six were among the 32 stumps of Group A patients, and five were among the 13 stumps of Group B patients. In brief, a smaller proportion of stumps showed vasodilatation in Group A patients (one fifth) than in Group B patients (two fifths).

In the majority of trials, experiments with other methods of inducing vascular relaxation were equally ineffective in causing a rise in stump temperature. In a total of eight intravenous injections of vasodilator drugs, the temperature of the stump increasedonlyslightly on two occasions (2.5°C or less). A rise in tern-

perature was effected once with Priscoline (2-benzylimidazoline hydrochloride) and once with tetraethylammonium chloride. Injections of prccaine in the region of the lumbar sympathetic ganglia produced a significant warming of the stump in one of two cases only. No correlation was found between the degree of phantom or stump pain experienced by these patients and the extent to which slump temperature fell during the initial period of exposure or the extent of stump warming during generalized vasodilatation. Amputees rarely complained of stump or phantom pain during these experiments, even though they were subjected to extremes of temperature requiring rapid vasomotor adjustments.

The ease with which stumps become cool on exposure to a cold environment can be attributed to two factors. First, surface-volume relationships in stumps favor cooling. Second, less blood passes through the stump than through comparable portions of the intact limb because, in the stump, distal tissues are absent. Apparently the shunts between the arterial and the venous side, which permit an increased volume of blood to flow through the extremity, are located distal to the wrist joint and to the ankle joint. In amputations at or above the wrist or ankle, therefore, flow of blood to the extremity is impaired. Normally, body heat is lost chiefly through radiation from hands, head, and feet. When the body is deprived of one of these radiating "fins," the remaining stump cannot be warmed. Neither can excess heat be radiated away, and for that reason an amputee often finds intolerable an environmental temperature that is quite acceptable to the normal. The amputee is distressed in a heated room, while the normal subject suffers no discomfort. Since the radiating mechanism is lost with amputation of an extremity, and since the only other means of cooling is through evaporation of sweat, the amputee is more likely to be troubled with problems of perspiration.

Skeletal Changes

In addition to problems of pain and changes in circulation, the amputee sometimes is troubled by decalcification of the stump and adjacent portions of the pelvis, a change that occurs when the body weight no longer is borne along the axis of the major articulations but along the prosthetic weight line (page 36). Because in an osteoporotic extremity the covering of the bone is more sensitive than is that in the normal, a decalcified bone often becomes exceedingly tender and develops spontaneous pain.

An interesting fact is that the joint itself, in Figure 16 the hip joint, begins to show early degenerative changes because it no longer transmits weight. In future studies it should be possible to evaluate more closely what changes are to be expected in the proximal articulations of an amputation stump, and more particularly in the joint cartilage covering the articulations, as a result of elimination of normal weighi-bearing through thesearticulations. Obviously, the only way la preveni osteoporosis and increased sensitivity is to resort to some type of end-bearing.

In the younger leg amputee, moreover, especially in growing children, other bony deformities develop (Fig. 17). Instead of the normal curvature of the neck of the femur, there develops a valgus deformity as is seen in polio and in dislocated hips. And finally, of course, because of loss of the mass of the limb, one must expect to find scoliosis and other abnormalities in the spine (Fig. 18).

SUMMARY

In summary, it may be said that, first, amputation produces changes in musculature, not only the familiar contractures and atrophy [50,88) but other changes as well. If a muscle is cut in half, its ability to shorten is decreased. A mid-thigh amputation decreases the effective normal range of motion of the hamstring group. If the hamstring group is cut in half, the velocity of contraction is halved, and an amputee thus afflicted cannot therefore perform certain functions with any degree of facility.

The mechanism of normal level walking requires the expenditure and distribution of considerable energy, for which the body depends largely upon the leg musculature. Thus, the handicap resulting from loss of any part of the leg is due not only to the loss of support but also to the loss of power available from
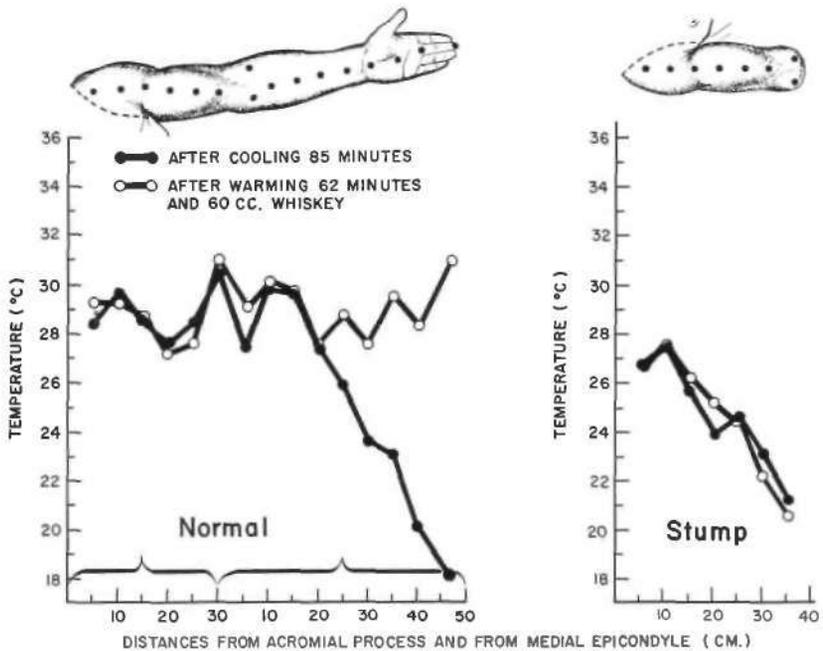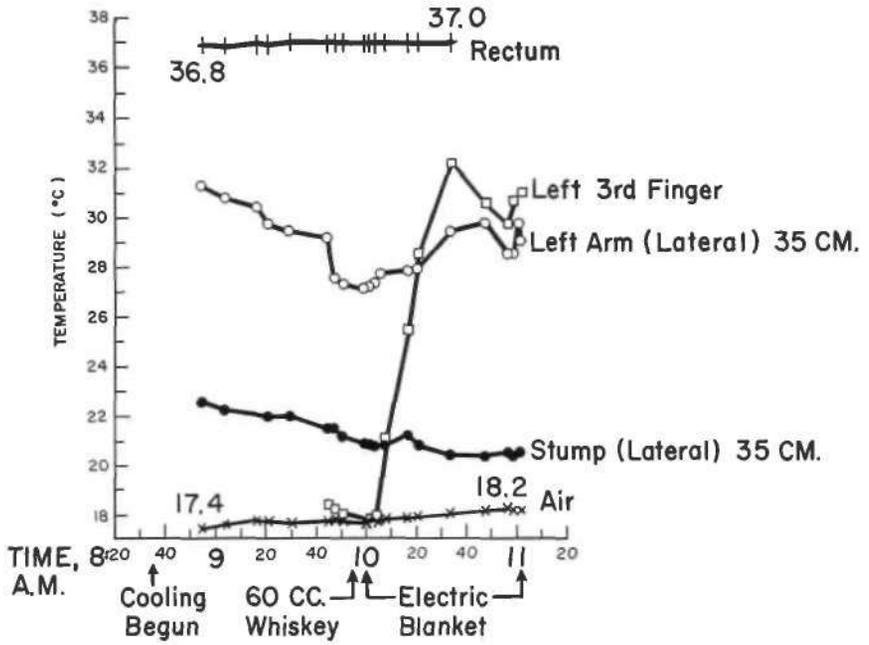
Fig. 14. Surface temperatures in the upper extremities of a below-elbow amputee during cooling and subsequent warming and vasodilatation. Above, time-temperature relations. Below, length-temperature relations. Points along the extremities indicate the locations of thermocouples. Relative humidity constant at 65 percent.
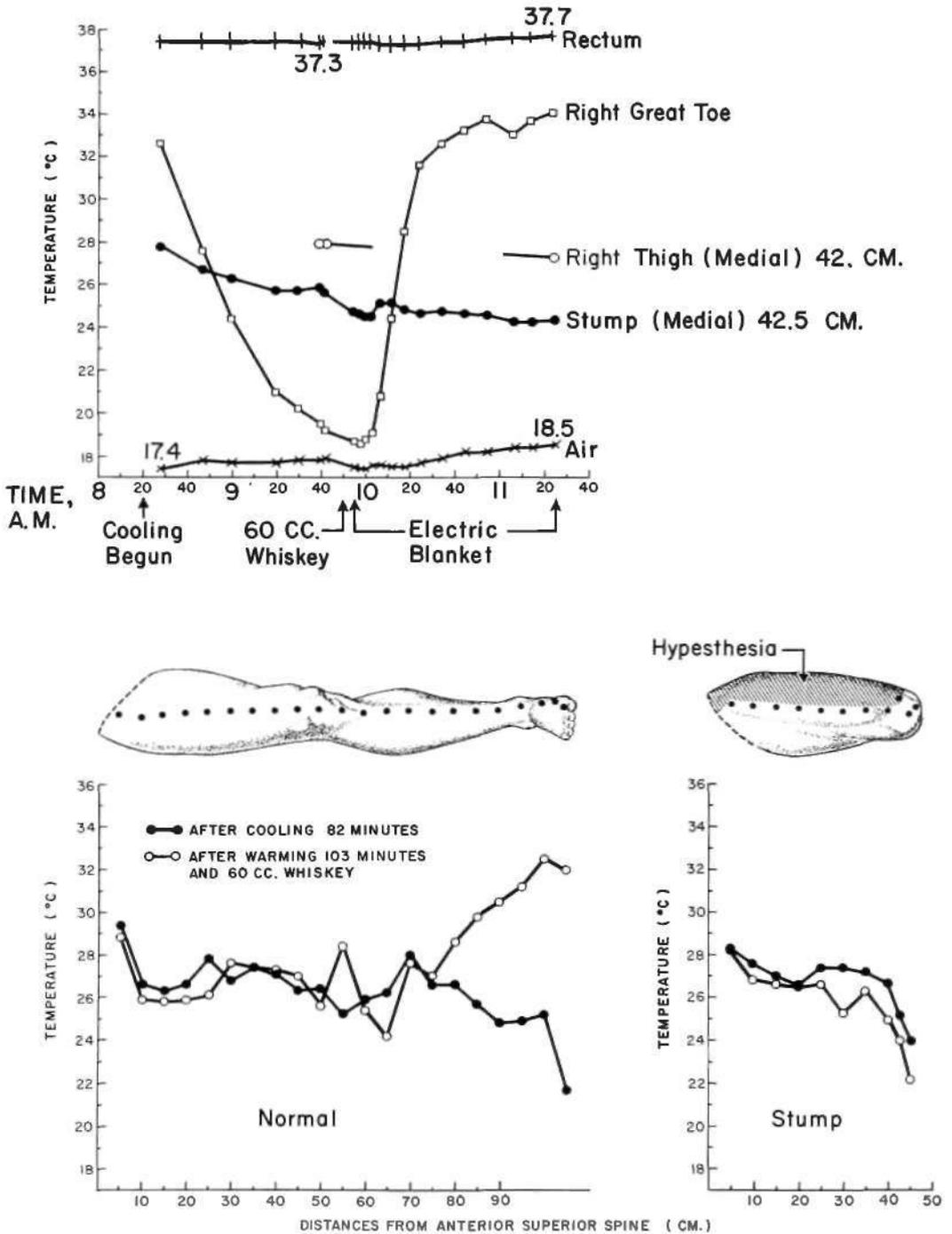
Fig. 15. Surface temperatures in the lower extremities of an above-knee amputee during cooling and subsequent warming and vasodilatation. Above, time-temperature relations. Below, length-temperature relations. Points along the extremities indicate the locations of thermocouples. Relative humidity constant at 74 percent.
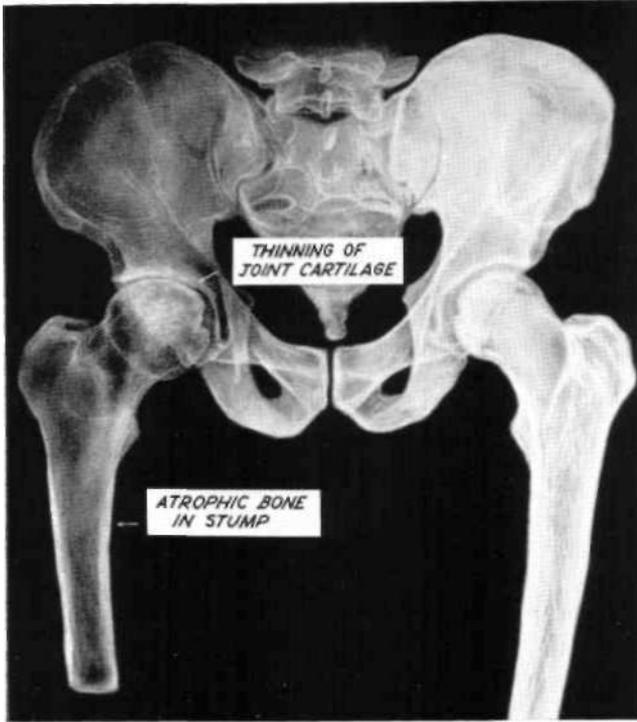
Fig. 16 Roentgenogram of an above-knee amputee, showing skeletal changes that occur when the hip and the remainder of the leg on the amputated side are deprived of the normal stimulation of weight-hearing

Fig. 17. Complicating deformities in juvenile amputees. When amputation is necessitated in childhood, defects often occur in the subsequent growth of related bony structures. Here, for example, the pelvis is smaller, and the pelvic-femoral angle larger, on the amputated side than on the sound side.
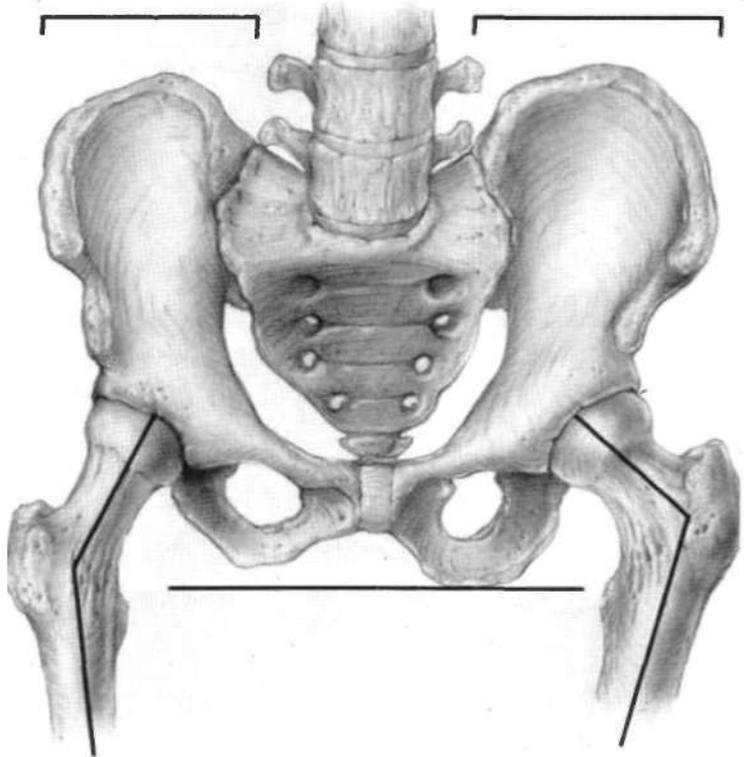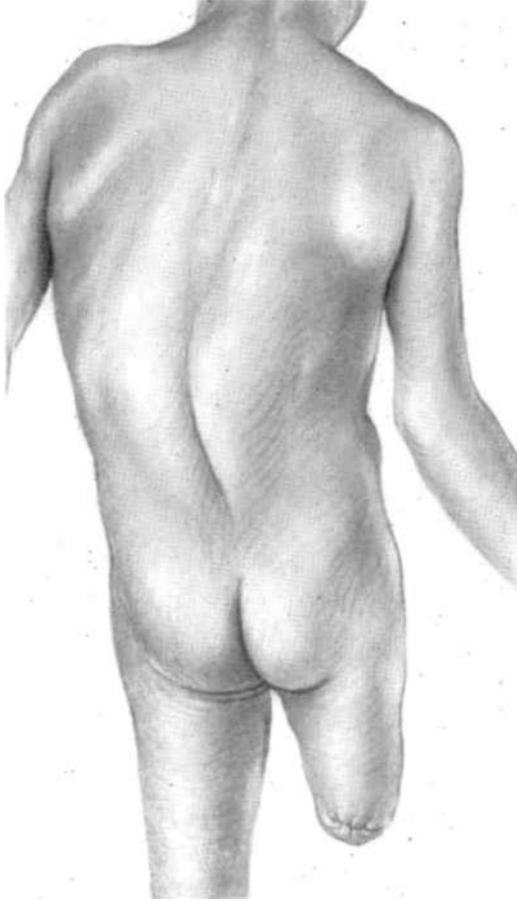
Fig. 18. Scoliosis, a postural defect often a sequel to amputation of the lower extremity. Loss of the weight of the amputated limb leads to habitual compensatory positioning of other body elements and thus complicates rehabilitation.

the muscles. The skeletal structure of a normal limb can more or less easily be simulated in a prosthesis, but such a device has little value without simultaneous provision for the necessary power. Accordingly, an understanding of the energy characteristics of normal level walking is important in considering the design criteria for artificial legs. Judging from the results of the energy studies at Berkeley, at a given pace an above-knee amputee uses two and a half to three times as much energy as does the normal. The adverse effect of this overexertion is only further complicated by the fact that heat production is increased at a time when the radiating mechanism has been impaired. In the manufacture of any lower-extremity prosthesis, then, an important consideration, is to design the substitute limb for maximum energy conservation.

Medical problems are common to all amputees. Some of them, for example those related to circulation, cannot be solved, but proper surgical procedures help to preserve the musculature and skeletal structures of adjacent joints. Moreover, many things can be done to relieve pain, both spontaneous phantom pain and the tender trigger points occurring in stumps. All amputees suffer some discomfort at one time or another. They are bothered by skin changes occurring over the bony prominences, by edema at the distal end of the stump, and by attritional lesions occurring in the folds of the groin (Fig. 19). A minor skin lesion can disable a leg amputee completely, especially when it means staying off the leg or going on crutches. Increased perspiration and poor ventilation of the stump in the prosthesis may close the sweat glands and make the skin susceptible to fungal diseases, and contact dermatitis may result if the patient is allergic to certain materials used in the manufacture of the prosthesis. Such problems must be solved by socket fit, by alignment, or by other procedures.

From the Clinical Study have come valid recommendations concerning fit, alignment, and functional characteristics. As already noted, some horizontal rotation (between 9 and 15 deg.) is desirable in an artificial leg. Further, increased stability in the knee joint increases the leg amputee's sense of security. Some conservation of energy can be effected by eliminating the articulated ankle joint. And finally, the matter of appearance deserves consideration. In this regard, attention must be given to the color, contour, and texture of the artificial leg.

In the last analysis, the problem of the leg amputee is more than that of providing him with a prosthetic device. He has many medical problems, including pain, abnormalities in circulation, heat intolerance, and skeletal and muscular changes. The prosthetic device itself raises other problems—conservation of energy, proper alignment, comfort, and cosmetic ap-
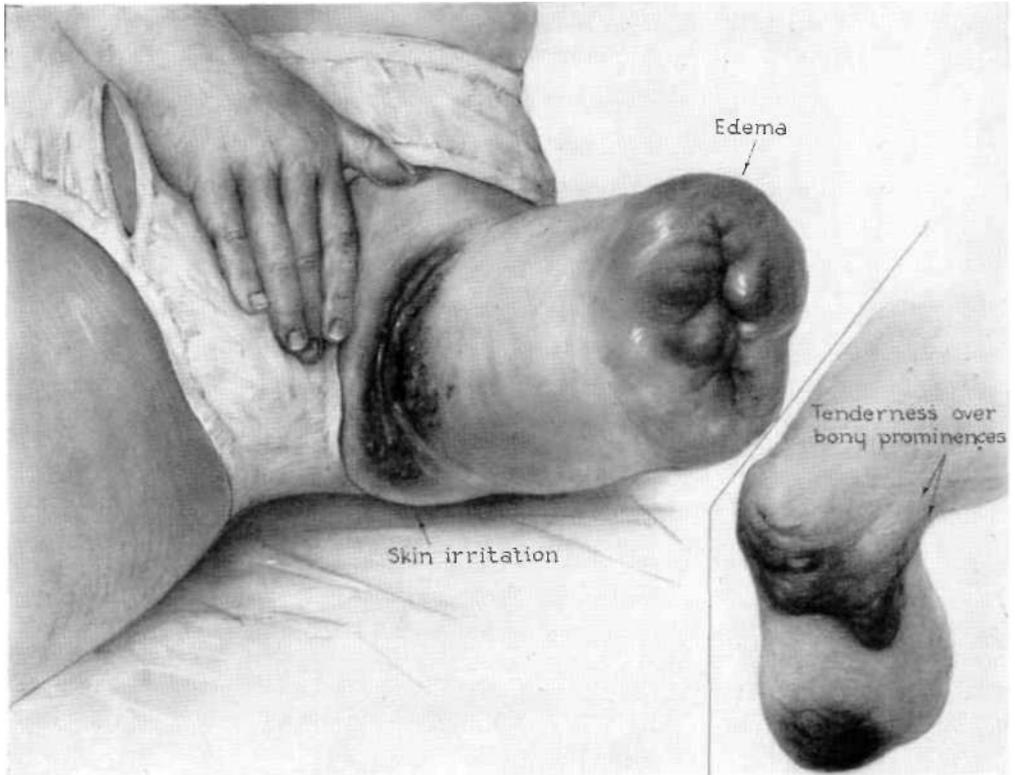
Fig. 19. Problems of fit. Among them are irritation and swelling in the crotch area, edema at the stump end, and tenderness at pressure points. Because such problems are more or less readily corrected by proper fit and alignment, they are less medical than prosthetic, although chronic skin irritation may need the attention of a dermatologist.

pearance. The Lower-Extremity Clinical Study is concerned with the solution of all these problems. The manner in which solutions are sought is shown in Figure e 20, where the central area represents the pool of fundamental knowledge accumulated over a period of nine years. As the amputee moves around the circle, each problem is studied and solved before he is allowed to move into the next phase of processing.

To date, pain and skin irritation have been the predominant problems, and study groups are being organized to investigate these areas in detail. Study groups also have been organized to investigate skeletal and muscular changes. At each step in the process, the panel itself often is faced with difficult problems. For example, the question of evaluation always is present, and it is not easy to determine whether or not the amputee actually has benefited from the time and effort devoted to his case. But as each difficulty is solved, the information derived is placed at the disposal of all those concerned, not only those within the Clinic Study Group but also all others whose interests lie in the field of amputee management. Seminars are held weekly to ensure that the information is brought to the attention of all interested persons. Eventually, all of the problem-solving data stemming from the investigations will appear in educational publications and will be available to members of the artificial-limb industry.

Finally, it may be said that the University group has no intentions of producing prosthetic devices and, indeed, makes excursions into that field only when it is necessary to develop experimental models pertinent to the study.
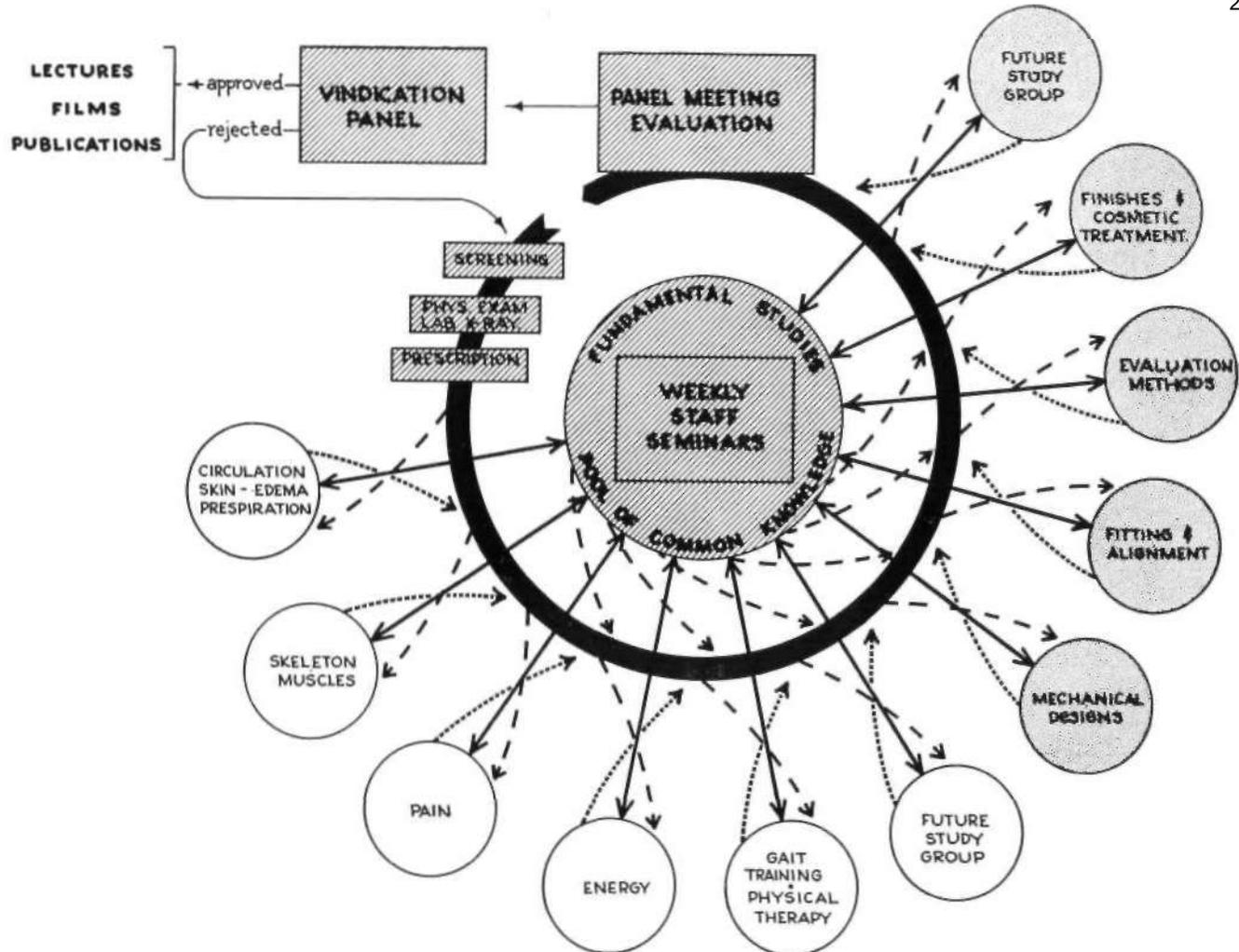
Fig. 20. Functional organization of theLower-Extremity Clinical Study:

The only function is to produce sound ideas that can be used by the artificial-limb industry in the manufactuie and fitting of improved prostheses. The study must, however, continue to be active until the basic scientific information can be translated into useful guides for the professions involved in the rehabilitation of the amputee.

LITERATURE CITED

1. Adel Precision Products Corp., Burbank, Calif., Subcontractor's Final Report [to the] Committee on Artificial Limbs, National Research Council, *The development of a hydraulically operated artificial leg for above knee amputations,* 1947.

2. Alldredge, Rufus H., *The cineplaslic method in upper-extremity amputations,* J. Bone & Joint Surg., 30A:359 (1948).

3. Alldredge, Rufus H., Verne T. Inman, Hyman Jampol, Eugene F. Murphy, and August W. Spittler, *The techniques of cineplasty,* Chapter 3 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

4.'Bartholomew, S. H., *Determination of knee moments during the swing phase of walking and physical constants of the human shank,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, January 1952.

5. Bechtol, Charles 0., *The prosthetics clinic team,* Artificial Limbs, January 1954. p. 9.

6. Bechtol, Charles O., *The principles of prosthetic prescription,* Chapter 6 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

7. Berry, F. R., Jr., *Angle variation patterns of normal hip, knee and ankle in different operations,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to] the Advisory Committee on Artificial Limbs, National Research Council, Series 11, Issue 21, February 1952.

8. Blaschke, A. C, *General energy considerations and determination of muscle forces in the mechanics of human bodies,* University of California (Los Angeles), Department of Engineering [Contractor's Memorandum Report No. 9 to the Advisory Committee on Artificial Limbs, National Research Council], September 1950.

9. Blaschke, A. C., and C. L.Taylor, *Biomechanical considerations in cineplasty,* University of California (Los Angeles), Department of Engineering, Special Technical Report 18, 1951.

10. Blaschke, Alfred C., and Craig L. Taylor, *The mechanical design of muscle-operated arm prostheses,* J. Franklin Inst., 266:435 (1953).

11. Blix, Magnus, *Die Lange und die Spannung des Muskels,* Scandinav. Arch. f. Physiol., 6:150 (1894).

12. Bradley, C. C, and Son, Inc., and Catranis, Inc., Syracuse, N. Y., Subcontractor's Final Report to the Committee on Artificial Limbs, National Research Council, *Artificial limb development for above-knee amputees including mechanical and hydraulic knee locks; suction socket and suction socket controls; knee lock controls operated by hip motion, stump muscles and foot position; toe pick up and foot providing lateral, plantar and dorsal flexion,* July 1947.

13. Bresler, B., *Use of energy methods for evaluation of prostheses,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to] the Advisory Committee on Artificial Limbs, National Research Council, September 1951.

14. Bressler *[sic],* B., and F. R. Berry, *Energy characteristics of normal and prosthetic ankle joints,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to] Advisory Committee on Artificial Limbs, National Research Council, April 1950.

15. Bresler, B,, and F. R. Berry, *Energy and power in the leg during normal level walking,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to] the Advisory Committee on Artificial Limbs, National Research Council, May 1951.

16. Brown, E., and N. Foreman, *Studies of skin temperature and of indirect vasodilatation in amputation stumps,* Am. J. Med., 10:112 (1951).

17. Buchthal, Fritz, and E. Kaiser, *Optimum mechanical conditions for work of skeletal muscle,* Acta Psychiat. et Neurol., 24:333 (1949).

18. Close, J. R., and V. T. Inman, *The action of the ankle joint,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, April 1952.

19. Close, J. R., and V T. Inman, *The action of the subtalar joint,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, Series 11, Issue 24, May 1953.

20. Contini, Renato, *Prosthetics research and the engineering profession,* Artificial Limbs, 1(3):47 (September 1954). p. 58.

21. Cunningham, D. M., *Components of floor reactions during walking,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, November 1950.

22. Eberhart, Howard D., *The objectives of the lower-extremity prosthetics program,* Artificial Limbs, May 1954. p. 4.

23. Eberhart, Howard D., Herbert Elftman, and Verne T. Inman, *The locomtor [sic] mechanism of the amputee,* Chapter 16 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

24. Eberhart, H. D., and V. T. Inman, *An evaluation of experimental procedures used in a fundamental study of human locomotion,* Ann. N. Y. Acad. Sci., 51:1213 (1951).

25. Eberhart, Howard D., Verne T. Inman, and Boris Bresler, *The principal elements in human locomotion,* Chapter 15 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

26. Elftman, H., *A cinematic study of the distribution of pressure in the human foot,* Anat. Rec, 69:481 (1934).

27. Elftman, H, *The measurement of the external force in walking,* Science, 88:152 (1938).

28. Elftman, H., *The rotation of the body in walking,* Arbeitsphysiol., 10:219 (1938).

29. Elftman, H., *The force exerted by the ground in walking,* Arbeitsphysiol., 10:485 (1938).

30. Elftman, H., *Forces and energy changes in the leg during walking,* Am. J. Physiol., 125:339 (1939).

31. Elftman, H., *The function of muscles in locomotion,* Am. J. Physiol., 125:357 (1939).

32. Elftman, H., *The function of the arms in walking,* Human Biol., 11:529 (1939).

33. Elftman, Herbert, *The work done by muscles in running,* Am. J. Physiol , 129:672 (1940).

34. Elftman, H., *The action of muscles in the body,* Biol. Symposia, 3:191 (1941).

35 Elftman, H., *Experimental studies on the dynamics of human walking,* Trans. N. Y. Acad. Sci., 11:1 (1943).

36. Elftman, H., *The bipedal walking of the chimpanzee,* J. Mammalogy, 25:67 (1944).

37. Elftman, H., *The carrying angle of the human arm as a secondary sex character,* Anat. Rec, 91:49 (1945).

38. Elftman, H., *The orientation of the joints of the lower extremity,* Bull. Hosp. Joint Diseases, VI-.139 (1945).

39. Elftman, H., *Torsion of the lower extremity,* Am. J. Phys. Anthropol., N.S. 3:255 (1945).

40. Elftman, H., *The basic pattern of human locomotion,* Ann. N. Y. Acad. Sci., 51:1207 (1951).

41. Elftman, Herbert, *The functional structure of the lower limb,* Chapter 14 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

42. Elftman, Herbert, and John T. Manter, *The axis of the human foot,* Science, 80:484 (1934).

43. Elftman, Herbert, and John Manter, *Chimpanzee and human feel in bipedal walking,* Am. J. Phys. Anthropol., 20:69 (1935).

44. Elftman, H., and J. T. Manter, *The evolution of the human fool, with especial reference to the joints,* J. Anat., 70:56 (1935).

45. Feinstein, Bertram, John N. K. Langton, R. M. Jameson, and Francis Schiller, *Experiments on pain referred from deep somatic tissues,* J. Bone & Joint Surg!, 36A:981 (1954)

46. Feinstein, Bertram, James C. Luce, and John N. K. Langton, *The influence of phantom limbs,* Chapter 4 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

47. Felkel, E. O., *Determination of acceleration from displacement-time data,* University of California (Berkeley), Prosthetic Devices Research Project, [Report to] the Advisory Committee on Artificial Limbs, National Research Council, Series 11, Issue 16, September 1951.

48. Felkel, E. O., *Determination of accelerations of the human leg during locomotion,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, Winter 1951.

49. Goodyear Tire & Rubber Company, Akron, Ohio, Subcontractor's Final Report [to the Committee on Artificial Limbs, National Research Council], *The development of a foot prosthesis incorporating a metal structure and a bonded rubber to metal ankle joint,* 1947.

50. Gordan, G. S., B. Feinstein, and H. J. Ralston, *Effect of testosterone upon atrophy of denervated skeletal muscle,* Exper. Med. & Surg., 7:327 (1949).

51. Hosmer Corp., A. J., Santa Monica, Calif., Subcontractor's Final Report to the Committee on Artificial Limbs, National Research Council, *Hydraulic weight bearing knee lock for knee disarticulation amputations; work arms for wrist disarticulations, below and above elbow amputations; work tools and devices for vocational rehabilitation; hydraulic control to actuate hooks and hands used on work arms; improved design hook,* 1947.

52. Inman, V. T., *Functional aspects of the abductor muscles of the hip,* J. Bone & Joint Surg., 29:607 (1947).

53. Inman, V. T., *Theoretical requirements of a lower-extremity prosthesis,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, December 22, 1950.

54. Inman, V. T., *Innervation of the extremities,* 3rd Biennial Western Conference on Anesthesiology, The California Society of Anesthesiologists and the Northwestern Society of Anesthesiologists, Los Angeles, 1953. p. 22.

55. Inman, V. T., B. Feinstein, and H. J. Ralston,

Some observations on electromyography, Am. J. Physiol., 155:445 (1948).

56. Inman, Verne T., H. J. Ralston, J. B. deC. M. Saunders, Bertram Feinstein, and Elwood W. Wright, Jr., *Relation of human electromyogram lo muscular tension,* University of California (Berkeley), Prosthetic Devices Research Project, and UC Medical School (San Francisco), Report to the Advisory Committee on Artificial Limbs, National Research Council, November 1951.

57. Inman, Verne T., and H. J. Ralston, *The mechanics of voluntary muscle,* Chapter 11 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

58. Inman, V. T., H. J. Ralston, J. B. deC. M. Saunders, B. Feinstein, and E. W. Wright, Jr., *Relation of human electromyogram lo muscular tension,* Electroencephalog. & Clin. Neurophysiol., 4:187 (1952).

59. Levens, A. S., V. T. Inman, and J. A. Blosser, *Transverse rotation of the segments of the lower extremity in locomotion,* J. Bone & Joint Surg., 30A:859 (1948).

60. Libet, B., *Neuromuscular facilitation by stretch, and the duration of muscular activation in locomotion,* Proc. 19th Internat. Physiol. Cong., Montreal, 1953. p. 563.

61. Libet, B., and B. Feinstein, *Analysis of changes in electromyogram (EMG) with changing muscle length,* Am. J. Physiol., 167:805 (1951).

62. Libet, Benjamin, and Bertram Feinstein, *Human electromyogram,* Surg. Forum, W. B. Saunders Co., Philadelphia, 1952. p. 415.

63. Libet, Benjamin, H. J. Ralston, and Bertram Feinstein, *The effect of stretch on action potential in muscle,* Biol. Bull., 101:194 (1951).

64. Libet, B., and E. W. Wright, Jr., *Facilitation at neuromuscular functions by stretch of muscle,* Fed. Proc, 11:94 (1952).

65. Livingston, Kenneth E., *The phantom limb syndrome: a discussion of the role of major peripheral nerve neuromas,* J. Neurosurg., 2:251 (1945).

66. Mitchell, S. Weir, *Phantom limbs,* Lippincott's Mag. Pop. Lit. So, 8:563 (1871).

67. National Research and Manufacturing Company, San Diego, Calif., Subcontractor's Final Report [to the Committee on Artificial Limbs, National Research Council], *An investigation of low pressure laminates for prosthetic devices; design and fabrication of above-knee and below-knee artificial legs; preparation of a production survey for manufacture of artificial plastic legs,* 1947.

68. New York University, Prosthetic Devices Study, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *Shakedown lest of the Navy above-knee prosthesis,* May 1951.

69. New York University, Prosthetic Devices Study, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *The functional and psychological suitability of an ex-perimental hydraulic prosthesis for above-the-knee amputees,* March 1953.

70. Northrop Aircraft, Inc., Hawthorne, Calif., Subcontractor's Final Report [to the] Committee on Artificial Limbs, National Research Council (Contract VAm-21223), *A report on prosthesis development,* 1947.

71. Northwestern Technological Institute, Evanston, III., Subcontractor's Final Report to the Committee on Artificial Limbs, National Research Council, *A review of the literature, patents, and manufactured items concerned with artificial legs, arms, arm harnesses, hands, and hooks; mechanical testing of artificial legs,* 1947.

72. Pare, A., from T. Johnson, *The works of that famous chirurgion, Ambrose Parey, translated out of the Latine and compared with the French,* Richard Cotes and Willi: Du-gard, London, 1649.

73. Polissar, M. J., *Concentration and potential pattern within the membrane and its relation lo penetration of ions and lo time constants of electrolonus and accommodation,* Fed. Proc, 11:124 (1952).

74. Polissar, M. J., *Physical chemistry of contractile process in muscle. I. A physiochemical model of contractile mechanism,* Am. J. Physiol., 168:766 (1952).

75. Polissar, M. J., *Physical chemistry of contractile process in muscle. II. Analysis of other mechanochemical properties of muscle,* Am. J. Physiol., 168:782 (1952).

76. Polissar, M. J., *Physical chemistry of contractile process in muscle. III. Interpretation of thermal behavior of stimulated muscle,* Am. J. Physiol.. 168:793 (1952).

77. Polissar, M. J., *Physical chemistry of contractile process in muscle. IV. Estimates of size of contractile unit,* Am. J. Physiol., 168:805 (1952).

78. Radcliffe, C. W., *Flexion stiffness of prosthetic ankle joints,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, May 1949.

79. Radcliffe, C. W., *Information useful in the design of damping mechanisms for artificial knee joints,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, April 1950.

80. Radcliffe, C. W., *Use of the adjustable knee and alignment jig for the alignment of above knee prostheses,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, August 1951.

81. Radcliffe, Charles W., *Mechanical aids for alignment of lower-extremity prostheses,* Artificial Limbs, May 1954. p. 23ff.

82. Radcliffe, Charles W., *Alignment of the above-knee artificial leg,* Chapter 21 in Klopsteg and Wilson's *Humanlimbs andtheirsubstitutes,* McGraw-Hill, New York, 1954. Especially pp. 686-688.

83. Ralston, H. J., *Muscle dynamics,* Surgical Forum

(1951), American College of Surgeons, Clinical Congress, W. B. Saunders, Philadelphia, 1952. p. 418.

84 Ralston, H J., *Isometric tension in the intact human quadriceps,* Proc. 19th Internat. Physiol. Cong., Montreal, 1953. p. 692.

85. Ralston, H. J., *Mechanics o] voluntary muscle,* Am. J Phys. Med., 32:166 (1953).

86. Ralston, H. J., J R Close, V T. Inman, and B. Feinstein, *Dynamical and electrical features of human isolated voluntary muscle in isometric and isotonic contraction,* Fed. Proc, 7:97 (1948).

87. Ralston, H. J., H. D. Eberhart, V. T. Inman, and M. D. Shaffrath, *Length-tension relationships in isolated human voluntary muscle,* Proc. 17th Internat. Physiol. Cong., Oxford, 1947. p. 110.

88. Ralston, H J., B. Feinstein, and V. T. Inman *Rate of atrophy in muscles immobilized at different lengths,* Fed. Proc, 11:127 (1952).

89. Ralston, H. J., V. T. Inman, B. Feinstein, and B. Libet, *Human electromyogram,* Am. J. Physiol., 163:743 (1950).

90. Ralston, H. J., V. T. Inman, L. A. Strait, and M. D. Shaffrath, *Mechanics of human isolated voluntary muscle,* Am. J. Physiol., 151:612 (1947).

91. Ralston, H. J., B. Libet, and E. W. Wright, Jr., *Effect of stretch on action potential of voluntary muscle,* Am. J. Physiol., 173:449 (1953).

92. Ralston, H. J., and B. Libet, *The question of tonus in skeletal muscle,* Am. J. Phys. Med., 32:85 (1953).

93. Ralston, H. J., M. J. Polissar, V. T. Inman, J. R. Close, and B. Feinstein, *Dynamic features of human isolated voluntary muscle in isometric and, free contractions,* J, Appl Physiol., 1:526 (1949).

94. Ralston, H. J., E. W. Wright, Jr., B. Feinstein, and V. T. Inman, *Effect of stretch upon action potential of voluntary muscle,* Am. J. Physiol., 159:586 (1949).

95. Ryker, N. J., Jr , *Glass walkway studies of normal subjects during normal level walking,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, Series 11, Issue 20, January 1952.

96. Ryker, N. J., and S. H. Bartholomew, *Determination of acceleration by use of accelerometers,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, September 1951.

97. Saunders, J. B. deC. M., Verne T. Inman, and Howard D. Eberhart, *The major determinants in normal and pathological gait,* J. Bone & Joint Surg., 35A(3):543 (1953)

98 Schiller, F., *Pain—controlled and uncontrolled,* Science, 118:755 (1953).

99. Spittler, A. W., and I. E. Rosen, *Cineplaslic muscle motors for prostheses of arm amputees,* J. Bone & Joint Surg , 33A:601 (1951).

100. Strait, L. A., V. T. Inman, and H. J. Ralston,

Sample illustrations of physical principles selected from physiology and medicine, Am. J. Physics, 15:375 (1947).

101. Taylor, Craig L., *Control design and prosthetic adaptations to biceps and pectoral cineplasty,* Chapter 12 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York. 1954.

102. University of California (Berkeley), Prosthetic Devices Research Project, Subcontractor's Final Report to the Committee on Artificial Limbs, National Research Council, *Fundamental studies of human locomotion and other information relating to design of artificial limbs,* 1947. Two volumes.

103. University of California (Berkeley), Prosthetic Devices Research Project, Preliminary Report [to the] Committee on Artificial Limbs, National Research Council, *The suction socket above-knee artificial leg,* September 1947.

104. University of California (Berkeley), Prosthetic Devices Research Project, |Report to the] Committee on Artificial Limbs, National Research Council, *The suction socket above-knee artificial leg,* revised edition, April 1948.

105. University of California (Berkeley), Prosthetic Devices Research Project, Supplementary Report 2, *The forces and moments in the leg during level walking,* revised August 10, 1948.

106. University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *The suction socket above-knee artificial leg,* 3rd edition, April 1949.

107. University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *Biceps cineplasty and prosthesis for below-elbow amputations,* April 1950.

108. University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *Functional considerations in fitting and alignment of the suction socket prosthesis,* March 1952.

109. University of California (Berkeley), Prosthetic Devices Research Project, and UC Medical School (San Francisco), Progress Report [to the] Advisory Committee on Artificial Limbs, National Research Council, *Studies relating to pain in the amputee,* June 1952.

110. University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, *Functional considerations in fitting and alignment of the suction socket prosthesis,* 2nd ed., August 1953.

111. University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, *The pattern of muscular activity in the lower extremity during wilking,* September 1953.

112. Wagner, Edmpnd M., *Contributions of the lower-extremity prosthetics program,* Artificial Limbs, May 1954. p. 8.

113. Wagner, Edmond M., and John G. Catranis, *New developments in lower-exlremity prostheses,* Chapter 17 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954. p. 482.

114. Wagner and Catranis, *op. cit.,* p. 511.

115. Weddell, G., B. Feinstein, and R. E. Pattle, *Electrical activity of voluntary muscle in man under normal and pathological conditions,* Brain, 67:178 (1944).

116. Wohlfart, G., B. Feinstein, and J. Fex, *Uber die Bieziehung zwischen electromyographischen und anatomischen Befunden in normalen Muskeln und bei neuromuskularen Erkrankungen,* Arch. f. Psychiat. u. Ztschr. Neurol., 191:478 (1954).

Functional Considerations in the Fitting of Above-Knee Prostheses

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IN THE fitting of any artificial limb, the goal of the prosthetist is simply to restore to the amputee the ability to perform everyday activities in an easy, natural, and comfortable manner. The basic requirements are therefore three in number—comfort, function, and appearance, the latter embracing both cosmetic appearance and appearance in use. Unless a prosthesis is reasonably comfortable, the amputee will be unable to wear it. Unless it performs the necessary functions with reasonable ease and dexterity, the amputee is not apt to find the device very useful. Unless it is reasonably acceptable cosmetically, and unless it can be operated in a natural manner, the limb is likely to be disagreeable both to the wearer and to his friends and associates.

But this seemingly simple set of requirements is vastly complicated by the fact that the three are all mutually interrelated. That is to say, the degree of satisfaction attained in one condition is influenced greatly by the situation prevailing with respect to the other two. Cosmetic appearance, for example, is necessarily limited by details of mechanism, and vice versa. No matter how elaborate a prosthetic device may be, it cannot be made to function properly unless it can be manipulated with ease and without discomfort. And conversely, no device can be comfortable in use unless its functional characteristics are properly integrated with the residual biomechanics of the wearer. Any change aimed at improvement in one condition unavoidably affects the other two—sometimes favorably, sometimes unfavorably.

In the lower extremity, cosmesis presents no serious problem. Since it is comparatively easy to fashion an artificial leg to an external shape and appearance more or less like that of its normal counterpart, and since in both sexes the lower extremity may be concealed beneath some sort of clothing, the actual cosmetic properties of a lower-extremity prosthesis amount to refinements to be added after all other requirements have been met. More critical in the lower extremity are comfort, function, and appearance in use. The leg prosthesis is in almost constant service, and it must provide both adequate support and a natural-appearing gait with as modest consumption of energy as possible. In fitting an above-knee limb, therefore, correct practices based on established biomechanical principles are mandatory if success is to be had.

Because during all activities the suction-socket above-knee leg *(3,4,19)* is controlled by the amputee through the use of remaining hip musculature, every effort must be made to ensure that these muscles are used to the fullest possible extent without causing discomfort. The intent here[2] is to present the basic concepts that apply to the fitting of all

[2] It should be understood that no new theory of alignment is intended, that the aim is simply to explain logically some of the problems facing prosthetists in the construction of above-knee legs and to provide rational solutions for those problems. The views presented are the combined result of experience gained at the University of California Prosthetic Devices Research Project during limbshop trials of the adjustable leg and alignment duplication jig *(8,9,10),* of a study of methods presently in use by the artificial-limb industry, and of a survey of information presented in the German literature *(15,16,17).*

above-knee prostheses, regardless of type of suspension, but which have particular application to the suction-socket above-knee leg. Although the details of fitting must necessarily be modified as dictated by the individual case, the basic features apply to all cases.

THE PRINCIPLES OF ABOVE-KNEE ALIGNMENT MEDIOLATERAL STABILITY

When one watches the walk of a typical above-knee amputee, two characteristics of gait often are particularly apparent. First, sidesway, *i.e.,* lateral movement of the torso from side to side, is exaggerated. Second, the amputee usually walks with his feet farther apart than does a normal individual of similar build. The average individual walks in such a manner that the lateral distance between successive points of heel contact is from 2 to 4 in. In order for the gait of an amputee to appear as normal as possible, therefore, he must walk with a base equally narrow. The amputee with a walking base of from 6 to 12 in. never can achieve a normal gait appearance. If such an amputee is asked why he walks with a wide base, he usually gives as the reason that it is more comfortable or that he feels more secure with his feet farther apart.

This circumstance is accounted for by the fact that, as an amputee attempts to walk with his feet closer together, certain functional requirements are placed upon the fit of the socket and upon orientation of the socket in space. In general, these requirements are not fulfilled in a prosthesis aligned for a wide-base gait. If an attempt is made to use such a prosthesis with a gait of narrow base, difficulties arise because certain forces come into play that cannot be accommodated by the stump in a comfortable manner. Although a poorly fitted prosthesis may be reasonably comfortable for many months provided the amputee walks so as to compensate for errors in fit and alignment, the same prosthesis may be very uncomfortable if the wearer attempts to change to a more normal-appearing gait. It is, however, possible to construct for the average above-knee amputee a prosthesis that allows a reasonably normal gait, that is comfortable in all normal activities, and that

eliminates common points of stump irritation such as those in the crotch area and near the end of the femoral stump.

The Weight-Bearing Line

One of the most common terms used by the prosthetist in the fitting and alignment of an above-knee prosthesis is the "weight-bearing line." It serves as the guide for many phases of setting up the prosthesis, but its exact position is subject to considerable difference of opinion. One prosthetist may use a weight line drawn from the ischial tuberosity through the center of the ankle joint; a second may select a line falling along the medial side of the foot; and a third may advocate use of a line drawn from the geometric center of the socket at the ischial level to the center of the heel. It is possible to get many other definitions of the weight-bearing line. As a matter of fact, they probably are all equally helpful in the alignment of prostheses. In considering the manner in which the weight-bearing line is used, it becomes apparent immediately that such a line actually serves as a "reference line" or "construction line."

In the discussion that follows, the term "weight line" is used to establish a mental picture of a theoretical line in space along which the force of the body weight acts. This concept differs from "weight-bearing line" in that "weight" is due to the gravitational attraction of the earth, whereas "weight-bearing" refers to the transmission of a force through the structural elements of the anatomy and the prosthesis. Although it would appear difficult to establish any one line which accounts for the net effect of the weight of the various and widely separated parts of the anatomy, that can be done in a theoretical, idealized way by defining a point within the body at which the effect of all body weight can be assumed to be concentrated. This point is usually designated as the "center of gravity" of the body as a whole. With all the weight assumed to be concentrated at the center of gravity, the body weight must then always be considered as acting directly downward from this point, as though it were a plumb bob suspended on a string hanging from the center of gravity. The string would represent
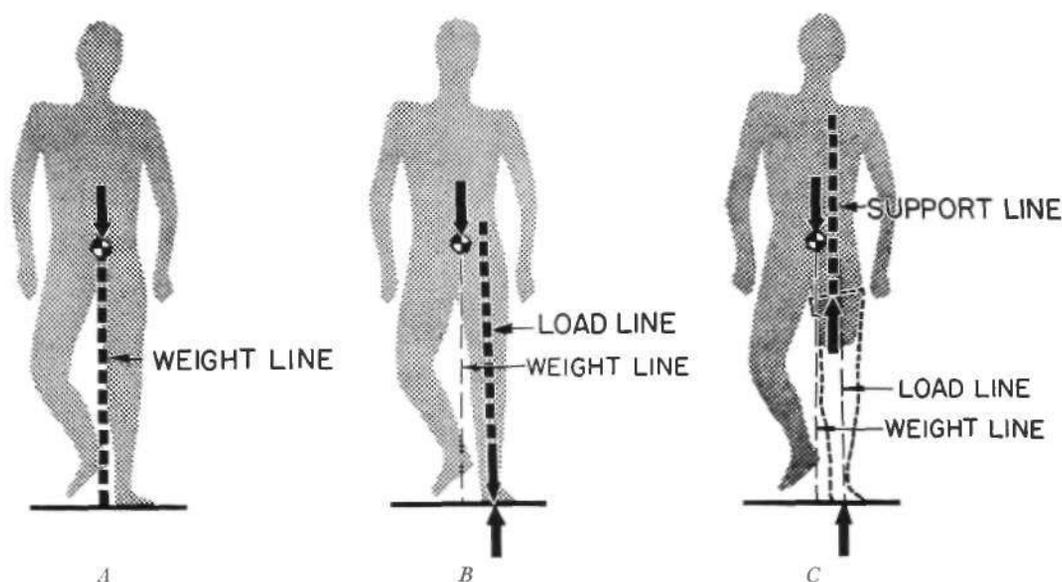
Fig. 1. Definitions in alignment of the lower-extremity prosthesis. *A,* The "center of gravity" of the body is a point at which all body weight can be assumed to be concentrated. The effective body weight passes through the center of gravity and acts vertically downward along the "weight line." *B,* The "load line" is a line along which the force between the foot and the floor acts. In general, it is not perpendicular to the floor surface, since this force has two effects. First, it supports the body weight in a vertical direction, and second, it provides the horizontal forces necessary to cause motion of the body in the forward and medial directions. *C,* The "support line" is a vertical line along which the effective supporting force exerted between the rim of the socket and the stump of the amputee is assumed to act. In general, the support line does not pass through the center of gravity or through the center of foot pressure.

the body weight line. A short definition of the weight line as shown in Figure \A might read as follows: *The weight line of the body is a line through the center of gravity along which the body weight can be assumed to act vertically downward at all times.*

Variations in Vertical Force

Thus far we have considered only the effect of the body weight acting downward. For either an amputee or a person with two good legs, the body weight must be supported by the contact between foot and floor. For many reasons, the force of contact between foot and floor is very difficult to measure accurately because, for either foot, the contact force is extremely variable over the short time the foot is supporting weight. Shortly after the heel strikes the floor, the leg receives an initial load which, because of the slight reduction in the rate of progression of the body as a whole, quickly increases to a value greater than body weight. During the mid-portion of the stance phase, as the center of gravity of the body is reaching the lowest point in its path of motion, the load on the leg decreases to a value somewhat less than that of body weight. As the body is being elevated and propelled forward into the next step, the load builds up again to a value greater than that of body weight.

Forces in Shear

While all this is occurring, the person also is swaying from side to side and varying in speed slightly as he walks. This condition requires that the contact force must also provide some horizontal frictional forces along the floor, as everyone has realized after slipping on ice or when making a sharp turn. The forces acting on the foot during walking are, then, of two kinds—those acting perpendicular to the floor, which support the body weight, and

those acting parallel to the floor, which are necessary to provide resistance to the impetus of the body moving forward, backward, or sideways.

Floor Reaction and Load Line

The total force exerted on the sole of the foot—the combination of all these effects—is known as the "floor reaction." It acts along the same line as does the total force exerted by the amputee on the socket of the prosthesis. The floor-reaction force is the load which the leg, whether normal or prosthetic, must transmit upward from the floor. In general, the line of these forces, known as the "load line" (Fig. 1*B),* is not perpendicular to the floor but is directed upward, inward, and forward or backward with an inclination that varies continually during the time either foot is supporting the body. It is very definitely not a line drawn from the center of the hip joint through the knee and ankle joints. A line so drawn should, instead, be designated as the "mechanical axis of the lower extremity."

The Support Line

An additional necessary concept is that of the "support line" (Fig. 1C). In order to define the support line, it is necessary first to identify a "support point," which may be defined as the center of action of all the vertical supporting forces at the top rim of the socket, including the ischial-bearing force, support in the gluteal region, and support in other weight-bearing areas around the socket rim. Where such a point lies is very difficult to establish, its actual location depending largely upon the individual prosthetist's methods of fitting. In a typical ischial-bearing socket, the support point is probably somewhere anterior and lateral to the point of contact of the socket with the ischial tuberosity. The support line is defined as a vertical or plumb line, passing through the support point, along which the effective supporting force between the socket rim and the stump can be assumed to act. In general, the support line coincides neither with the weight line nor with the load line.

Use of the Hip Abductors

Figure 2 presents a rear view of an above-knee amputee, walking with a narrow base, at an instant during the walking cycle when the full weight is carried on the prosthesis. During the stance phase, the amputee, like the normal individual (5), keeps his pelvis


Fig. 2. Use of the hi]) abductors for lateral stabilization of the pelvis.

Fig. 3. Lever action of the pelvis in stabilization of the torso,

The Lever Principle:

$$T(y) = W(x)$$
$$T = W\left(\frac{x}{y}\right)$$

wall of the socket and .an therefore be extremely uncomfortable. Anticipating this action, the amputee makes appropriate compensation. He maintains his balance either by leaning over the prosthesis, which results in the familiar amputee list, or by walking with a wide base and swaying from side to side. In the alignment of an above-knee prosthesis, then, one of the most important objectives is to construct the prosthesis in such a way that the hip abductors may be used in a normal and comfortable manner to prevent this tendency toward pelvic drop, torso list, or sidesway, and to allow a reasonably normal and comfortable gait.

The Pelvic Lever

As indicated in Figure 1*A,* the center of gravity of the body is defined as the point at which the entire weight would have to be concentrated were it to have the same effect on the body as a whole as does the actual weight distribution. On the strength of this concept, the pelvis can be assumed to act as a lever in the stance phase while the amputee supports his weight on the prosthesis (Fig. 3). Using the ischium as a supporting pivot or fulcrum, the pelvic lever supports the body weight (which acts vertically downward through the center of gravity and along the weight line) by the balancing action of the hip abductors, the process being similar to normal hip action in which vertical support is through the hip joint. If this lever action is to prevent dropping of the pelvis toward the unsupported side, the tension in the hip abductors must be sufficient to balance the body weight. The abductor muscle force can perform this function only if abduction of the stump is prevented by firm contact against the lateral wall of the socket. Otherwise the muscle action would simply cause abduction of the femoral stump inside the socket.

horizontal primarily by action of the hip abductors on the supporting side, as shown by abductor tension in Figure 2. If, for one reason or another, the hip abductors are unable to exert the necessary force, the pelvis has a tendency to drop toward the unsupported side. When, therefore, the above-knee amputee stands upon his prosthesis, his pelvis may tend to drop toward the normal side owing either to inadequate hip abductors or to inadequate support on the lateral side of the stump—support which is necessary to stabilize the femur and to form a firm base for action of the hip-abductor musculature.

Dropping of the pelvis toward the normal side generally results in an increase in pressure in the crotch area. It often allows the pubic ramus to come into contact with the medial

Distribution of Lateral Pressure

The necessary stabilization of the stump against the lateral wall of the socket can be accomplished comfortably if the stabilizing pressure is distributed widely over the lateral side. For a stump of average length, stabilization is achieved by fitting the lateral wall snugly over its entire length. A slight flattening of the lateral wall, with relief near the distal end of the femur, usually ensures that the stabilizing forces are not only comfortable but that they are directed medially as required (Fig. 2). If, with the stump improperly supported against the lateral wall, an attempt is made to use the hip abductors for pelvic stabilization, the result may be a gap around the lateral brim and a painful concentration of pressure near the end of the stump.

Considerations of Mechanical Advantage

Two other factors enter into the lateral stabilization of the pelvis by the hip abductors. First, in balancing the body weight on the ischial fulcrum, the tension in the hip abductors has greatest mechanical advantage when the lever arm between the abductor tension and the support point is as long as possible. Support of a substantial portion of the body weight by the ischial seat and of a smaller amount by the gluteal musculature gives the abductor tension sufficient mechanical advantage to balance the body weight with little or no conscious effort on the part of the amputee. The characteristics of this lever system are shown in the schematic diagram of Figure 3, where the required tension T is reduced by decreasing the distance x and increasing the distance y.

Adduction of the Stump

A second factor in making allowance for normal use of the hip abductors is the degree of stump adduction in the socket. The "rest-length" theory of muscle action (1,6,7,11,12, 13,14) has shown that the muscles of the body act most efficiently when they are at approximately their normal rest length. To make the action of the hip abductors efficient, the stump, when fitted in the socket, must be adducted in such a manner that the outward movement of the femur within the muscle mass of the stump is anticipated and that the normal pelvic-femoral angle is maintained as closely as possible while the body weight is being supported on the prosthesis. For the average amputee, this requirement can be met in a practical way by aligning the medial wall of the socket perpendicular to the floor, the lateral wall being sloped definitely inward. Although exceptions are necessitated on the basis of stump length, the short stump being aligned with less adduction, every effort should be made to adduct the stump as much as conditions permit.

An additional advantage of alignment in adduction becomes apparent immediately. As a result of the accompanying decrease in tension of the adductor musculature, pressure in the crotch area is decreased. As a result of this relaxation, the pressure in the crotch or medial area (Fig. 2) is then predominantly lateral rather than vertical and no longer causes painful pressure on stretched adductor tendons or in the region of the ramus. It should be emphasized here that a socket properly fitted and aligned carries little or no weight on the medial wall.

Foot Position

Alignment of the foot in a medial position, a fundamental consideration if the amputee is to walk without excessive sidesway or torso list, helps to ensure that the body weight will be borne chiefly on the ischial seat. The average amputee walks well with the centerline of the foot located directly below the ischium during the time the prosthesis is supporting the entire body weight. But this rule-of-thumb, illustrated by the reference line shown in Figure 2, must vary depending upon the capacity of the amputee to use his hip abductors. If an amputee with a very short stump attempts to use it for lateral stabilization, he cannot tolerate the increased and usually localized pressure resulting from the short stump length and the concentration of force in a small area. He must, therefore, walk with more limited use of his hip abductors, and compensation is effected by leaning over the prosthesis to shift the weight line closer to the support line and by walking with a wider

base, an expedient which increases lateral stability but leads to excessive sidesway. Because of these factors, and because of the probability in such cases of some degree of abduction contracture, the amputee with a very short stump should have his prosthesis aligned to accommodate a gait of wider base.

Recapitulation

In summary, mediolateral stabilization of the pelvis accompanied by a decrease in the amount of sidesway and list can be achieved by alignment of the foot in a medial position relative to the socket, by fitting the stump in an adducted position where possible, and by providing firm support for the stump against the lateral wall of the socket to allow efficient use of the remaining abductor musculature of the hip.

KNEE CONTROL

Involuntary Control

Generally, the tendency of the articulated knee joint of the above-knee prosthesis to collapse under load is controlled involuntarily through alignment or by mechanical devices which lock or restrain flexion while the body weight is being transferred through the prosthesis (20). Although involuntary control is desirable as an aid in achieving a smooth and natural-appearing gait, a proper balance must be obtained between the amount of involuntary and voluntary control of knee stability, taking into account the amputee's coordination and age and the condition of his stump.

Involuntary control of knee stability during weight-bearing is made possible by so placing the knee axis that it is at all times posterior to the load line of the prosthesis (10). A prosthesis with the socket placed well forward on the knee block or aligned in hyperextension and with the knee joint located posterior to the ankle joint is said to have a high degree of "alignment stability." That is to say, under load the knee joint is forced to extend until the extension stop makes contact and prevents further motion. This expedient often is necessary for amputees who have a fear of falling or when it is required because of age, insuffi-

cient stump power, excessive weight, or the prevailing terrain. But it has the disadvantage of making the prosthetic knee hard to flex under even a light load and thus results in poor gait and difficulty in negotiating stairs and slopes.

Voluntary Control

An attempt should therefore always be made to minimize the amount of involuntary alignment stability and to provide for a maximum of voluntary knee control by stump action because this type of functioning results in the smoothest and most effortless gait possible. The average above-knee amputee has a reasonable amount of strength remaining in his hip flexors and extensors and is able to extend and flex his stump throughout an appreciable range of motion, and it is important that the fullest use be made of this musculature in voluntary control of knee stability. That this control may be exercised in the most efficient manner possible, the stump should never approach the limits of its motion as the amputee performs normal activities. If, for example, the stump is able to extend a maximum of 20 deg. to the rear, then at push-off any forced extension in excess of the 20 deg. results in a forward rotation of the pelvis. To compensate for such a forward pelvic rotation, the amputee must arch his back, an expedient which leads to the development of lordosis. Alignment of the socket in a position of initial flexion, as shown in Figure 4, eliminates much of this difficulty.

Initial Flexion

When the socket is aligned with initial flexion, several other advantages become apparent. Since the length of the hip extensors is increased by the additional degree of hip flexion, the amputee has greater control of knee stability during the entire stance phase of the walking cycle. Since the extensor muscles are thus elongated slightly, they are able to develop the required tension easily. With much less conscious effort on the part of the amputee, therefore, the stump is able to exert the force necessary to keep the prosthetic knee back against its extension stop.
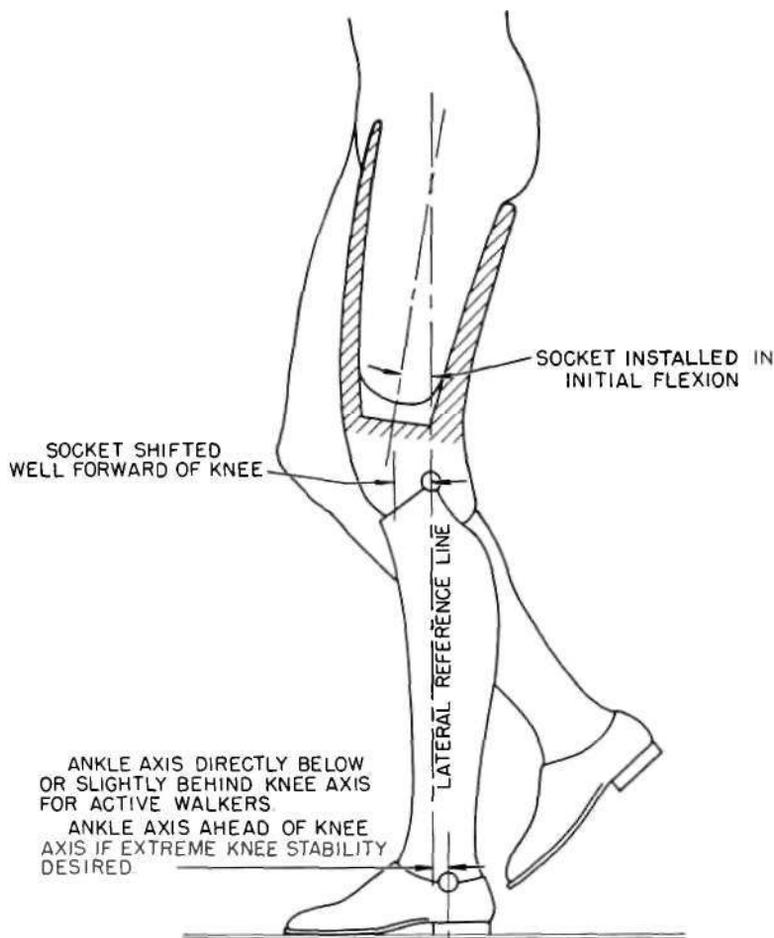
Fig. 4. Influence of alignment on control of knee stability, socket aligned in initial flexion to avoid exces. sive pelvic rotation.

Again, in an amputee with overdeveloped hamstring musculature there often is a tendency, as the stump extends at push-off, for the muscles to force the tuberosity of the ischium off the ischial seat, thereby causing pressure on the hamstring muscle and attachments and against the anterior brim of the socket. Initial flexion of the socket reduces this tendency and allows a portion of the body weight to be borne comfortably upon the hamstring attachments.[3]

If the same degree of alignment stability

is to be maintained, initial flexion of the socket must be accompanied by a shifting of the socket anterior to the knee axis. Merely changing the extension stop to decrease knee extension never can achieve the desired end-results. But less alignment stability is necessary under these conditions because of the increased voluntary control of the knee. Anterior positioning of the socket relative to the knee axis allows the prosthetic knee to be flexed a great deal more easily as weight is transferred from the prosthesis to the normal leg at the end of the stance phase. The result is a smoother gait. Although increased use of the hip extensors owing to their greater

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[3] Too much initial flexion results in a decrease in stride length, which may be undesirable in some cases.

working length produces some decrease in the power available in the hip flexors, the loss is not serious since during ordinary activities the hip flexors never approach the limit of their range of flexion and since the force requirements are small as compared with those of the hip extensors.

Ankle Position and Toe Break

Another important factor in achieving the proper amount of knee stability is the fore-aft position of the ankle joint relative to the knee joint. For the active above-knee ampu-tee, it usually is desirable to have the ankle joint directly below or slightly posterior to the knee joint, as shown in Figure 4. Such an arrangement has several effects. First, as the foot is moved to the rear, the distance out to the toe break decreases to give the foot more of a "rocker" action and to allow the knee to flex easily at the end of the stance phase. Second, the major portion of the weight can be carried on the ball of the foot while stand-ing. And third, the amount of toe clearance during walking is greater for a given angle of knee flexion. To move the ankle joint too far to the rear, however, results in instability at heel contact and excessive shortening of the stride.

Many of these advantages can be achieved by use of a double toe break *(i.e.,* a flexible forefoot), which also gives the foot more of a rocker action and decreases the amount of vaulting over the prosthetic foot. But too much flexibility or too short a distance from ankle to toe break causes the leg to feel too short at the time of push-off.

DYNAMIC ALIGNMENT

For the major part of the time that the amputee is supporting himself on the pros-thesis during the stance phase, the motions are relatively smooth, and the forces act on the prosthesis in essentially the same way as if the amputee were standing still with all weight carried on the artificial leg. During the swing phase, however, and during the times of transition from stance to swing and from swing to stance, the behavior of the prosthesis is influenced largely by dynamic forces varying rapidly with time. It is often relatively easy

to fit an amputee so that he is comfortable in the stance phase, but in many cases it is more difficult to construct the prosthesis so that the amputee is able to walk with a smooth, natural-appearing, effortless swing-through. The first requirement for a smooth swing phase is a smooth transition from stance to swing, since, if the prosthesis is to swing properly, it must be given a good start.

Knee Stability and Toe Break

Of particular importance during these transi-tion periods are knee stability, as affected by alignment and by the stiffness of dorsi-flexion and plantar flexion at the ankle, and the combined effect of toe-out and orientation of the toe break in the foot. For security, the knee axis should be positioned far enough behind the hip-ankle line so that the amputee is conscious of a stable knee while standing. The amount of security desired depends upon the particular amputee. If, as the amputee attempts to walk, the knee feels insecure, the dorsiflexion position and stiffness in the ankle should be investigated as a possible additional cause of knee instability.

In general, placing a stiff dorsiflexion bumper in the ankle and having the foot plantar-flexed in the neutral position, close to the point where the amputee has the sensation of "walking over a hill," produces the most desirable knee stability and allows smooth flexion of the knee at the start of the swing phase. The amount of toe-out usually is adjusted to the individual amputee. In all cases, however, the toe break should be at right angles to the line of progression to prevent insecurity resulting from the rapid shifting of the center of pressure during push-off.

Whip in the Swing Phase

One of the more obvious indications of poor dynamic alignment is the so-called "whip" of the prosthesis during the swing-through (Fig. 5). This lateral movement of the knee accompanied by medial movement of the foot, or vice versa, usually is caused by an incorrect amount of adduction for the particular socket being fitted, an improper angle of the knee axis with respect to the
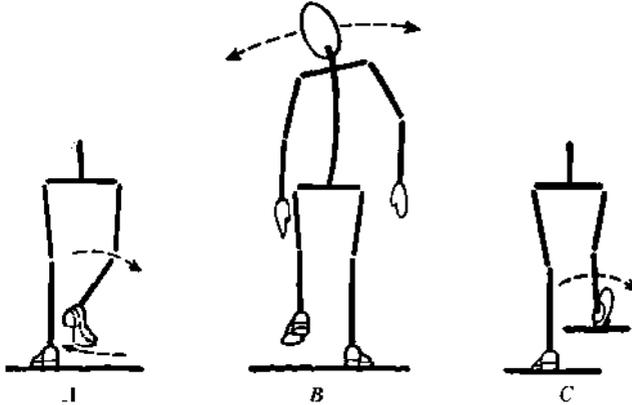
Fig. 5. Common indications of incorrect alignment. *A,* Whip of the prosthesis during the swing phase. *B,* Mediolateral instability. *C,* Rotation at heel contact. For specific causes of these difficulties, see Radcliffe *(10).*

frontal plane, the natural tendency of the femoral stump to twist inward as it is brought forward, or a combination of these factors.

An above-knee prosthesis often is "knocked" at the knee to position the foot laterally for greater stability while standing. Sufficient two-leg standing stability thus can be attained, but a stable, narrow-base gait is not then possible. The tendency of the prosthesis to whip also is aggravated because, as it swings like a pendulum, the leg has a natural tendency to swerve medially after toe-off and then to swerve out again just before heel contact. A prosthesis having the foot aligned medially for a narrow base during the stance phase need only move forward in a straight line from toe-off to heel contact.

Rotation of Knee Axis

Studies of normal human locomotion *(2,18)* show that the femur rotates an average of 3 to 4 deg. medially as the hip is flexed to bring the knee forward. Medial rotation of the femur causes a lateral displacement of the foot, as can be verified easily by observation of a person standing and flexing the hip while the shank hangs vertically. Accordingly, the knee axis in an above-knee prosthesis usually is rotated laterally to compensate for the tendency of the femur to rotate medially as the hip is flexed.[4] When the prosthetic

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[4] The amount of medial rotation in the stump de-

knee axis is aligned in a position laterally rotated with respect to the socket, the foot moves somewhat medially with knee flexion, thus compensating for lateral movement of the foot caused by the medial rotation of the socket during the swing phase and allowing the foot to travel in a straight path.

Ankle Stiffness

The stiffness of plantar flexion at the ankle determines, to a large degree, the stability of the knee at heel contact. A stiff ankle does not allow the foot to rotate forward into the stable flat position and thus tends to cause the knee to buckle forward as the weight is transferred to the prosthesis. An ankle joint with insufficient plantar-flexion stiffness, however, allows the foot to slap at heel contact. A proper balance between these two effects must therefore be attained for the individual amputee. Proper swing-through is achieved by proper dynamic alignment, which, in turn, is effected by a comfortable, stable, and functional prosthesis in the stance phase; a smooth transition from stance to swing phase; proper ankle stiffness; and adjustment of the knee axis in lateral rotation to compensate for medial rotation of the stump during hip flexion.

SOCKET SHAPE AND ORIENTATION

Considered thus far are the means by which the amputee can make most efficient use of the remaining hip musculature to control body movements and to control the prosthetic knee during the stance and swing phases. There are, however, many functional details of socket shape and fit which make it possible for the amputee to derive these benefits comfortably.

pends upon the inherent physiological characteristics of the hip joint and upon the loss of muscular function after amputation. Some amputees have even been observed to have lateral rotation of the stump upon hip flexion.

The Lateral Wall

As already indicated, for the amputee having sufficient stump length and power, sidesway and leaning over the prosthesis during the stance phase can be eliminated almost entirely by making provision in the socket for full use of the remaining abductor muscles of the hip, primarily the gluteus medius. This can be achieved in two ways. First, the stump is adducted in the socket so that the lateral wall is sloped downward and inward, the medial wall remaining essentially vertical. Second, a slight flattening of the lateral wall, and undercutting for relief of pressure points where necessary, ensures a comfortable distribution of the pressure directed medially against the stump. The hip abductors then can develop tension as needed because the excursion of the femur is blocked comfortably against the lateral wall of the socket. If, after the fit of the lateral wall is considered satisfactory, the socket is too tight, relief should be provided along the medial wall of the socket to avoid disturbing the fit required to block excursion of the femur.

The Anterior Wall

The lateral pressures, acting with the horizontal counterpressures in the upper portion of the medial wall, tend to maintain the ischium on its seat medially. To hold the ischium in place still more firmly, it is necessary to provide stabilization at the front of the socket. Accordingly, the anterior wall of the socket should fit the stump firmly in the area of Scarpa's triangle, and a very accurate measurement should be made of the distance from the ischial tuberosity to the tendon of the adductor longus so that the anteromedial apex may be fitted snugly around the adductor tendons. The socket brim should be rounded and fitted high on the anterior side. If fitted properly, the anterior brim usually can be brought up to the level of the inguinal crease without producing discomfort when the wearer is seated. The actual height of the anterior brim varies with the individual and is limited by contact with bony prominences. It usually extends from 2 to 2-1/2 in. higher than the ischial seat, but it should extend at least high

enough so that the brim will press into the abdominal muscles rather than pinch a roll of flesh near the top of the stump. Distributed over the upper portion of the entire anterior wall of the socket, such anterior counter-pressure easily can prevent the ischium from sliding into the socket and can prevent the discomfort that would result in the crotch area.

The Adductor Region

Incorporation of the proper distance from the adductor tendons to the ischial tuberosity, combined with a well-fitted, high, anterior brim, usually eliminates entirely any unwanted pressure in the crotch area. Some lateral counterstabilization by pressure in the crotch area is unavoidable, but it should be predominantly by lateral rather than by vertical pressure, and it can be tolerated comfortably if distributed over the widest possible area. Flattening the medial wall of the socket is one means of ensuring a comfortable distribution of pressure in the adductor region.

The Anteroposterior Dimension

Weight-bearing in the gluteal region makes it possible to reduce the size of the ischial seat. If the anteroposterior dimension is shortened, the socket may be widened in the mediolateral dimension, a feature having several advantages. First, it allows a greater area for gluteal weight-bearing on the posterior rim of the socket. Second, the ischium is moved laterally, allowing the ramus to be carried within the brim of the socket and thus easing a major source of irritation. Finally, because the ischium bears no weight in the posteromedial apex, there is less tendency for crowding of the adductor and hamstring musculature. Relaxation in this area owing to stump adduction also helps to relieve uncomfortable vertical pressures.

Shape at Ischial Level

As a result of these functional requirements, the socket shape shown in Figure 6 has evolved. When coupled with the proper alignment, it has proved to be extremely beneficial to the average amputee. As with any method of fitting, variations in shape must be made in
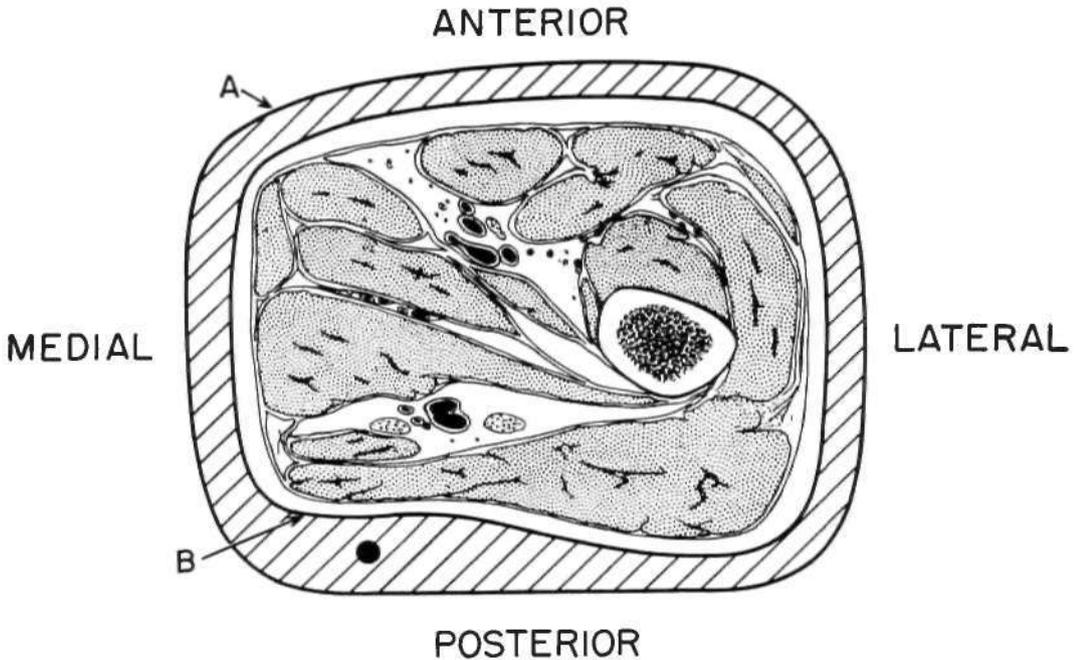
Fig. 6. Anatomical features of an above-knee stump in weight-bearing, shown in cross section 1/2 in. below schial level.

accordance with the muscular development and condition of the individual stump. The influence of muscular development at the ischial level is shown in Figure 7.

Entrances of the adductor tendons in the anteromedial apex, shown as *A* in Figure 6, can be made more comfortable by a slight flaring of the socket brim in this region. Flaring of the socket brim in the hamstring area *B* has no function while the amputee is walking, but it contributes remarkably to his comfort while sitting. Many amputees experience a burning sensation while sitting because the hamstring attachments attempt to stretch over an ischial seat located high or medially, especially when the ischial seat has been placed diagonally across the posteromedial apex. The socket shape shown in Figure 6, however, allows the ischial seat to be placed laterally to provide relief in the hamstring region and does not disturb the functioning of the limb during walking.

CONSTRUCTION OF THE SOCKET

STUMP EXAMINATION AND MEASUREMENTS

Before construction of an above-knee prosthesis is started, it is essential that a very careful evaluation be made of the amputee and his stump. A prosthesis may thus be planned and constructed to take full advantage of the individual patient's capabilities. Of particular importance is a thorough examination of the stump with regard to its functional characteristics. Answers to the following questions are helpful in planning the prosthesis, and they should be included in the examination data:

1. What degree of stump flexion contracture is present?
2. What degree of stump abduction contracture is present?
3. Is the stump musculature soft, average, or hard?
4. Is the hamstring group soft, average, hard, or prominent under tension?

SOFT STUMP
(considerable subcutaneous tissue)

A slight decrease in the anterior-posterior dimension (m) is necessary to maintain the ischium on its proper position.

AVERAGE MUSCULATURE
Notice the decrease in the dimension (n) due to the hamstring relief which results in more sitting comfort.

VERY MUSCULAR STUMP
Note the deeper gluteal channel and rounding of the socket walls to accommodate the larger muscular development.

PROMINENT RECTUS FEMORIS

PROMINENT HAMSTRING TENDONS
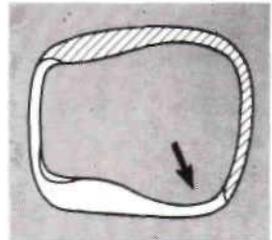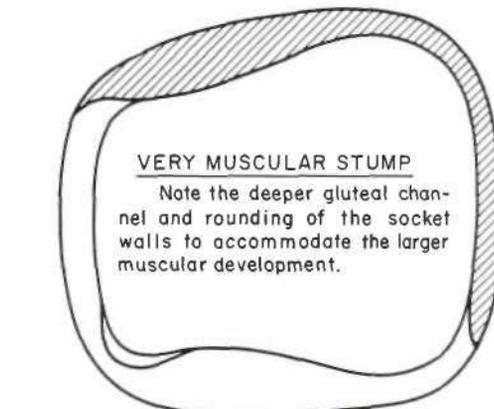
PROMINENT GLUTEAL GROUP
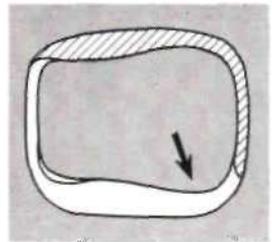
UNDERDEVELOPED GLUTEUS MAXIMUS

Fig. 7. Influence of stump muscular development on socket shape at ischial level.

5. Is the gluteal group soft, average, hard, or prominent with stump extension?

6. Is the stump contour along the lateral side convex, concave, or essentially flat?

7. Is the rectus femoris muscle prominent with stump flexion?

8. Is the adductor longus soft, average, or hard?

9. Is the ischium toughened, pressure sensitive, padded with muscle, or prominent?

10. Has the amputee been accustomed to ischial-bearing?

11. What is the amount and location of redundant tissue?

12. What is the extent, location, and adherence of scars?

13. Are there areas of prior irritation as shown by blisters, boils, pimples, scars, darkened skin areas, and so forth?

14. Are there areas which are sensitive because of bone spurs or other prominences?

15. Is there any prior history of edema?

In addition to this general information about the condition of the stump, which can be recorded on a form such as Figure 8A, the series of measurements indicated in Figure 8B should be recorded carefully.

PLANNING THE SOCKET SHAPE

After the information gathered during the examination is recorded, the limbfitter is ready to begin planning the prosthesis, a phase essential to proper fit. The socket contours and the over-all alignment to be incorporated into any lower-extremity prosthesis depend upon the interrelation of many factors. First, the amputee's general physical condition must be determined. Will the amputee be an active walker? Will ease of walking be more important than knee security, or vice versa? Has the amputee developed gait habits that require corrective training? Second, the stump must be evaluated on a functional basis. In terms of its potential usefulness in control of the prosthesis and of body movements, is it classed as short, medium, or long? Is there a normal range of motion in all directions? Are there any sensitive areas that restrict stump function? The answers to these questions affect the alignment of the prosthesis as well as the fit of the socket.

It is important to plan for alignment before the socket contours are considered because the orientation of the socket on the stump and the alignment of the socket on the pros-

thesis may affect considerably the method of fitting the socket. Shown in Figure 9 are some general features of alignment based upon the functional capacity of the stump—short, medium, and long. There are exceptions, of course, and these illustrations should serve only as a guide.

After the general type of alignment has been decided upon, the necessary features can be incorporated into the orientation of the socket on the stump, a matter requiring a decision regarding the approximate amount of initial flexion and adduction to be anticipated in the final alignment. The socket contours are determined by reference to the information on stump muscle development recorded during the examination. Figure 7 shows a typical socket shape for an amputee of average musculature and indicates the variations possible with different types of stump muscle development. Undersize patterns for use in roughing out the socket contours are shown actual size in Figures 10 and 11. The dimensions shown along the medial side of the patterns are typical measurements of the distance from the ischial tuberosity to the anterior aspect of the adductor longus tendon. The perimeter measurements shown correspond to actual stump dimensions. But these patterns may require modification to provide for individual stump characteristics, an example of such a pattern modification being shown in Figure 12.

MATERIALS

The primary features required of a material to be used in making a suction socket are ease in forming to the proper shape, adaptability to a surface finish which is nonirritating and easy to keep clean, and ease in making alterations as required by changes in the stump. Wood and plastic laminates have, so far, proved to be the most satisfactory. But major changes in the size of the stump often take place during the first several months of wear. Hence, wood is recommended for the first socket because it is relatively simple to shape and allows alterations to be made as required. After the stump size is stabilized, a socket can be made of plastic laminates, which seem better than wood

because of their flexibility, their ability to stand cleansing with soap and hot water, and their greater resistance to the action of perspiration.

SHAPING THE WOODEN SOCKET

The three stages in shaping a typical socket are shown in Figure 13. In the first, the postero-medial shelf is cut after laying out the socket pattern on the top of the socket block. The ischiogluteal shelf is cut in such a way as to be horizontal when the socket is oriented vertically in space. For the average socket, the medial wall is parallel to the vertical reference line (Fig. 2), and therefore the horizontal ischiogluteal shelf is cut at right angles to the medial wall of the socket. After the ischiogluteal shelf is cut, the missing portion of the socket pattern line is transferred down to the ischial level.

The second construction stage shows the roughed-out socket, where considerable extra wood has been left above the ischial level to allow for the protrusion and flaring of the anterior brim in this area. The finished socket is shown in the third stage with all areas of the socket brim flared and rounded to prevent irritation of the stump, especially important in the anteromedial apex where the adductor longus tendon enters the socket.

Figure 6 indicates the principle muscle groups and other anatomical features considered in preparing the patterns used as a guide in the preliminary layout of the socket outline. Because of the atrophy of certain muscle groups in the above-knee stump, and because the cross section shows the stump in the weight-bearing condition, the shape differs slightly from that of the normal. When the stump is bearing weight, it is necessarily compressed slightly in areas of relatively soft tissue which support load, such as the gluteal channel.

The Lateral Wall

The lateral side is always higher than the level of the ischial seat. In most cases, it is possible to extend it over the trochanter. To do so is especially important when the slump is short and when the height of the socket in this region may be required to maintain suc-

tion. If the muscular development requires it, the lateral side of the socket is, in some cases, undercut above the ischial level. Examination of the amputee determines the amount of undercut required, and, if it is necessary, it should be done with caution. The lateral wall should taper in acutely below the ischial level to provide adduction and lateral support for the femur upon weight-bearing above the distal end. Because the femur has been established as the body stabilizer during the stance phase, an undercut below the ischial level may distribute the pressure unevenly and thus allow most of the pressure to be taken at the top of the socket and near the distal end of the stump. The lateral wall should be shaped to fit the stump accurately and should, if necessary, be flattened to distribute the lateral-support pressure over a large area so that it can be tolerated comfortably.

The Medial Wall

The length of the crotch-line area that receives the adductor longus, gracilis, and adductor magnus muscles should be determined accurately by skeletal measurements. As indicated in Figure 12, the measurement from the anterior aspect of the adductor longus tendon to the weight-bearing portion of the ischial tuberosity, less about half an inch, gives the approximate length of the medial side of the socket. In general, the upper third of the medial wall is flattened, and the superior brim is flared to prevent skin irritation.

In almost every case, the crotch-line height varies with respect to the level of the ischial seat, but it should always be as high as is tolerable. In the typical socket, the crotch area is from 1/8 to 1/4 in. lower than the ischial seat. A pelvic tilt lowers the ramus of the ischium and may require a lowering of the medial side of the socket. In a properly fitted ischiogluteal weight-bearing socket, little or no weight should be borne on the medial side. From the ramus to the antero-medial apex, the medial brim can be raised as governed by comfort. If a medial adductor roll is present, the socket is enlarged slightly (never lowered) on the medial side to accommodate the excess tissue, which then is pulled into the socket and eventually diminishes.

PROSTHETIC INFORMATION — ABOVE-KNEE PROSTHESIS
A. STUMP DESCRIPTION

Amputee_____ Date_____

Height_____ Prosthetist_____

Weight_____

Age_____

Sex_____

Show location of stump details, identifying with letter code:

Lateral   Anterior   Posterior   Medial
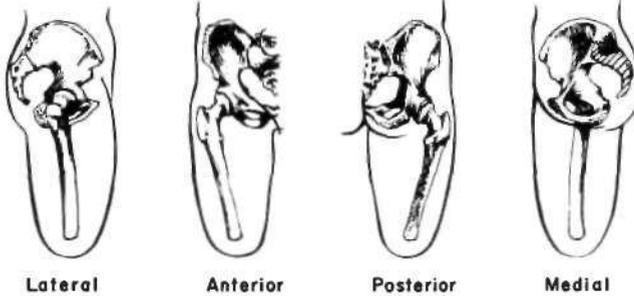
A. Abrasion
B. Boil
BS. Bone Spur
D. Discoloration
E. Edema
I. Irritation
M. Muscle Bunching
P. Pressure Point
R. Redundant Tissue
S. Scar
T. Trigger Point

Ischium: Toughened ( )   Pressure sensitive ( )
Muscle padding ( )   Prominent   ( )
Previous ischial bearing? yes( ) no( )

Subcutaneous tissue: Heavy ( )   Skin condition: Tough ( )
Light ( )   Thin ( )

Stump lateral contour: Convex outward ( ) Flat( ) Concave inward ( )

Stump Musculature	Soft	Average	Hard	
General				
Hamstring Group				Prominent with stump extension?
Gluteal Group				Prominent with stump extension?
Rectus Femoris				Prominent with stump flexion?
Adductor Longus				

Stump flexion contracture: _____degrees; abduction contracture:_____ degrees

Show modification of basic socket shape, if required:

Muscle code:
a. Sartorius
b. Rectus femoris
c. Pectineus
d. Vastus medialis
e. Vastus intermedius
f. Vastus lateralis
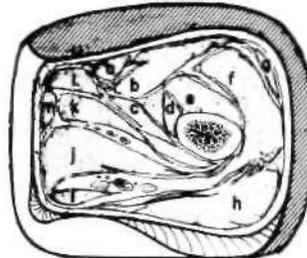
g. Tensor fasciae latae
h. Gluteus maximus
i. Hamstrings
j. Adductor magnus
k. Adductor brevis
l. Adductor longus
m. Gracilis

Fig. 8/1. Form used at the University of California for recording stump characteristics and measurements in above-knee fitting.

PROSTHETIC INFORMATION — ABOVE-KNEE PROSTHESIS
B. PROSTHETIC MEASUREMENTS

Amputee_____ Date_____

Right or Left Amputation_____ Prosthetist_____

Distance from Ischial Tuberosity to Tendon of Adductor Longus

Distance below Perineum	Stump Circumference
O	
	O

Ischial Tuberosity (standing)

Forefoot-Heel Circumference

Knee Width (sitting)

Top of Knee (sitting)

Tibial Plateau

Calf Circumference

Ankle Circumference

Shoe Size
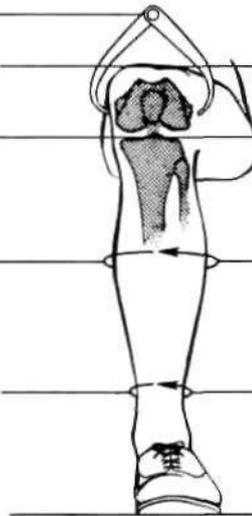
Distance Up from the Floor (Shoes On)

FORM 7B

Fig. 8B.

SHORT FUNCTIONAL LENGTH

- SOCKET SLIGHTLY ABDUCTED FOR SHORT STUMP
- POSTERIOR VERTICAL REFERENCE LINE THROUGH POINT OF CONTACT OF ISCHIUM
- STRAIGHT SHANK
- HEEL CENTER OUTSET WITH RESPECT TO VERTICAL REFERENCE LINE
- AS MUCH INITIAL FLEXION IN SOCKET AS PRACTICAL
- SOCKET ALIGNED FORWARD ON KNEE BLOCK
- LATERAL VERTICAL REFERENCE LINE THROUGH KNEE AXIS
- ANKLE ALIGNED AHEAD OF VERTICAL REFERENCE LINE

MEDIUM FUNCTIONAL LENGTH

- MEDIAL WALL VERTICAL
- LATERAL WALL SLOPED INWARD TO PROVIDE STUMP ADDUCTION IN SOCKET
- POSTERIOR VERTICAL REFERENCE LINE THROUGH POINT OF CONTACT OF ISCHIUM
- STRAIGHT SHANK
- HEEL CENTER DIRECTLY UNDER POINT OF CONTACT OF ISCHIUM
- TROCHANTER CONTACTS LATERAL WALL IN LINE WITH REFERENCE LINE
- LATERAL VERTICAL REFERENCE LINE THROUGH KNEE AXIS
- INITIAL FLEXION IN SOCKET IS SATISFACTORY WHEN LINE OF SIGHT INTERSECTS TOE BREAK
- ANKLE JOINT DIRECTLY UNDER KNEE JOINT

LONG FUNCTIONAL LENGTH

- POSTERIOR VERTICAL REFERENCE LINE THROUGH POINT OF CONTACT OF ISCHIUM
- INSET SHANK
- HEEL CENTER INSET WITH RESPECT TO VERTICAL REFERENCE LINE
- TROCHANTER CONTACTS SOCKET BEHIND REFERENCE LINE
- INITIAL FLEXION LIMITED BY LENGTH OF STUMP
- LATERAL VERTICAL REFERENCE LINE THROUGH KNEE AXIS
- ANKLE JOINT ALIGNED BEHIND KNEE JOINT

Fig. 9. Variations in alignment to accommodate stumps of different functional lengths. With the short stump, the slow or hesitant walker, having limited use of the hip abductors and extensors, needs considerable alignment stability. The moderate walker, with stump of medium functional length, has average use of the hip abductors and extensors. Alignment for the long stump is for an active walker having good use of the hip abductors and extensors.

The Anteromedial Apex

The socket shape at the anteromedial apex (Fig. 6) should conform to the contour of the adductor longus and gracilis muscles. The shape varies in each case, however, because these muscles form a cordlike tendon which must be fitted accurately. Tightness in this region, a common source of irritation in suction sockets, usually is caused by excessive length of the medial side of the socket. This condition allows the ischium to slide forward into the socket and to wedge the stump into the anteromedial apex. If tightness in the anteromedial apex persists, it is apt to be due to inadequate support of the stump across the anterior brim and down the anterior aspect of the adductor group.

The Anterior Wall

The primary function of the anterior brim of the socket is to maintain the ischium in place on the ischial seat so that ischial weight-bearing causes no discomfort. In many cases of amputees who are unable to tolerate ischial weight-bearing, the trouble can be traced to improper contact between ischium and socket. Ischial bearing on the edge of a flat ischial seat is especially uncomfortable. To maintain the ischium in place properly, considerable counterpressure from the front of the socket is required. Since, by and large, the portion of the stump in contact with the region of the anterior brim is soft tissue, some compression of the stump is necessary. This is accomplished by a flattening and inward protrusion of the anterior brim in the area of Scarpa's triangle.

The upper portion of the anterior brim is fitted 2 to 2-1/2 in. higher than the ischial seat and with a generous flare along the superior brim. When the socket is fitted with such a "high front," the anterior brim can hold the ischium in place comfortably. The high front does not interfere with sitting or with the amputee's ability to bend over far enough to tie his shoes. As the stump is flexed, the higher brim of the socket is accommodated by the abdominal musculature and does not pinch a roll of flesh on the upper portion of the thigh. The brim should be lowered only as necessary to prevent contact with bony prominences such as the anterosuperior spine. A channel should be provided below the brim for the rectus femoris muscle, which usually becomes prominent with stump flexion.

The Posterior Wall

The back of an ischial-bearing socket deserves particular attention. Channelization for the gluteus maximus muscle depends on the individual, but, in most cases where there has been little atrophy or distortion, this region of the socket should be kept on the same level as the ischial seat with a gradual enlargement in the posterolateral apex. The gluteus muscle should carry a considerable amount of body weight on a flared socket brim.

Relief for the adductor muscles or the crotch line often can be made by relieving the gluteus maximus. Too tight a fit over the gluteus maximus can cause crowding of the adductor muscles in the crotch section. If the space for the gluteus muscle is lowered and widened, the ischial tuberosity can be moved posteriorly and laterally on the ischial seat of the socket. Lowering this section, however, increases pressure on the ischial tuberosity and should, therefore, be avoided. Should additional room be needed within the socket, the lateral side of the gluteal region can be made wider. The gluteal area should be widened instead of cut deeper posteriorly because a deeper section forms a hump or radius on which the leg rotates during sitting and thus causes a burning sensation of the skin over the ischial tuberosity.

The outside shape of the socket in the posterior region is important to sitting comfort, but no attempt should be made to complete its shaping until the inside has been made comfortable and until the leg has been aligned properly and tested by walking. After these things are done, the back then is flattened for comfort and alignment while sitting.

The Ischial Seat

The ischial seat cannot be overemphasized. It should be located accurately under the ischial tuberosity, and, in the determination of its location, individual variations in anatomy must be taken into account. The seat should be adequate but not so wide as to cause discomfort while sitting. Slipping of the ischial tuberosity either to the inside or to the outside

ANTERIOR

MEDIAL

LATERAL

RADCLIFFE

MEDIAL WIDTH OF PATTERN
EQUALS DISTANCE FROM ISCHIAL
TUBEROSITY TO ANTERIOR ASPECT
OF THE ADDUCTOR LONGUS TENDON
MINUS ONE HALF INCH

$3\frac{3}{4}$"

$3\frac{1}{2}$" SOFT or AVERAGE MUSCULATURE

$3\frac{1}{4}$"

SIZE OF PATTERN
DETERMINED BY MEASUREMENT
OF STUMP CIRCUMFERENCE
AT THE ISCHIAL-SEAT LEVEL

18" 20" 22"

POSTERIOR

APPROXIMATE POINT OF
CONTACT OF ISCHIUM

Fig. 10. Undersize socket patterns (shown actual size) for stump with soft or average musculature,

ANTERIOR

MEDIAL WIDTH OF PATTERN
EQUALS DISTANCE FROM ISCHIAL
TUBEROSITY TO ANTERIOR ASPECT
OF THE ADDUCTOR LONGUS TENDON
MINUS ONE HALF INCH

FIRM MUSCULATURE

SIZE OF PATTERN
DETERMINED BY MEASUREMENT
OF STUMP CIRCUMFERENCE
AT THE ISCHIAL-SEAT LEVEL

$3\frac{3}{4}$"

$3\frac{1}{2}$"

$3\frac{1}{4}$"

MEDIAL

LATERAL

ABOVE-KNEE FITTING

18"

20"

22"

APPROXIMATE POINT OF
CONTACT OF ISCHIUM

POSTERIOR
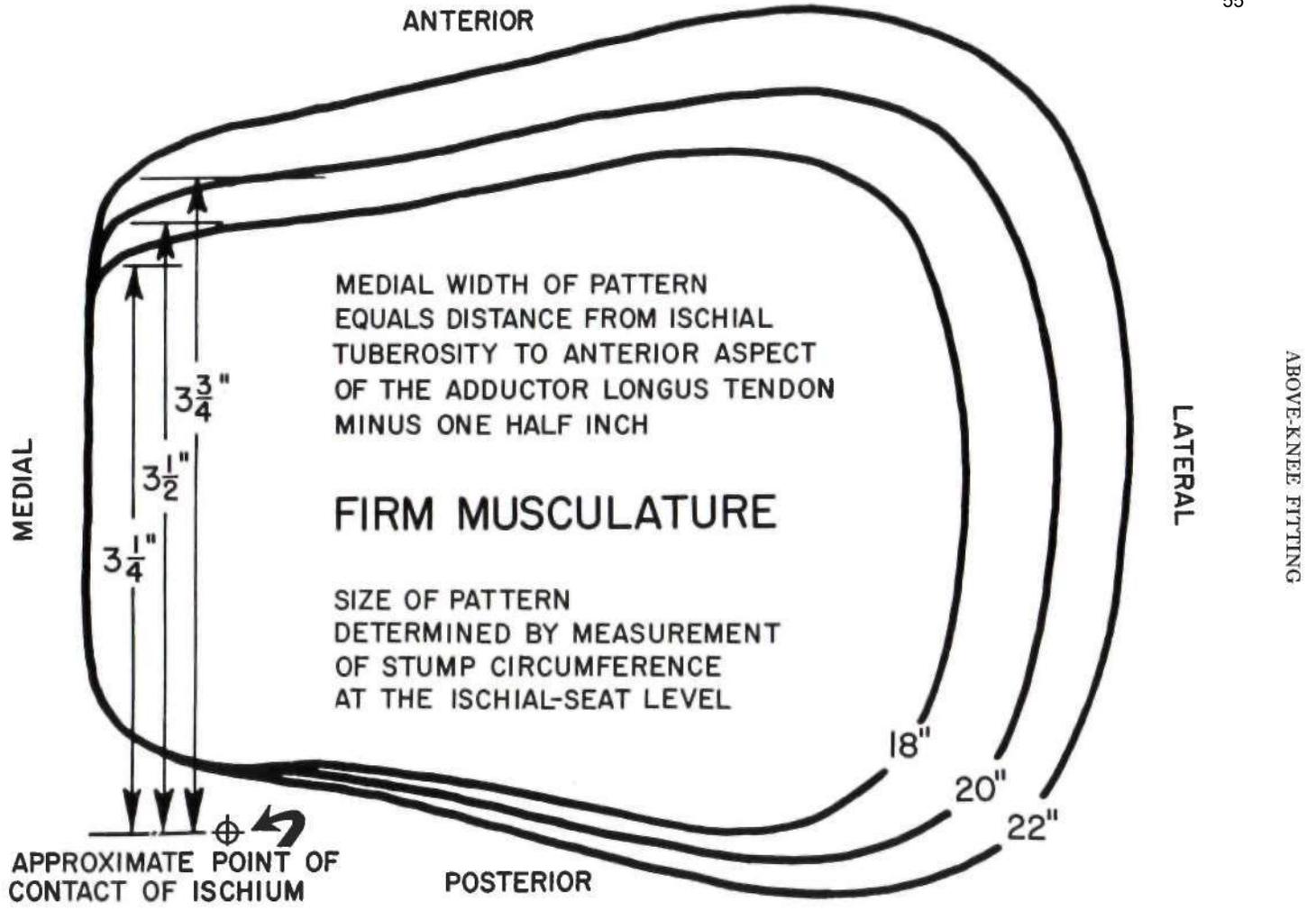
Fig. 11. Undersize socket patterns (shown actual size) for stump with firm musculature.
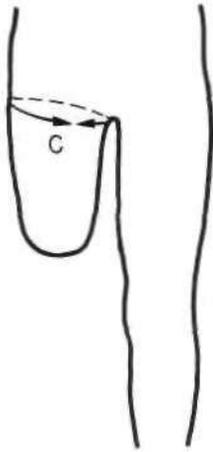
BASIS OF LAYOUT OF SUCTION-SOCKET PATTERN:
1. Muscular development of stump
2. Circumference of stump at ischial level (C)
3. Medial width of stump, measured from adductor longus tendon to tuberosity of the ischium (W)

A.
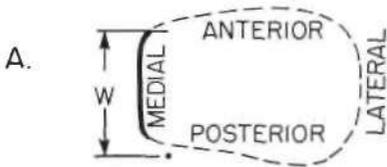
Choose from Figure 7 the pattern which most nearly conforms to muscular development and ischial circumference of the stump. Alter medial width of pattern if necessary. Medial width of pattern should be one half inch less than distance from adductor longus tendon to the tuberosity of the ischium.

B.
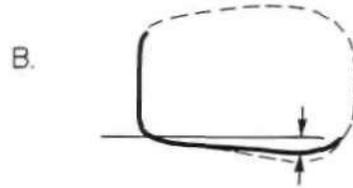
Estimate depth of gluteal channel required. This figure illustrates modification of basic pattern for stump with soft or flabby gluteal musculature.

C.
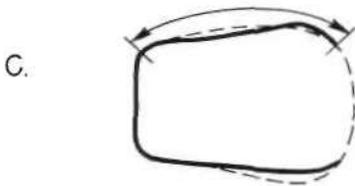
Modify anterior contour as necessary. This figure illustrates modification for stump with soft adductor musculature and prominent rectus femoris muscle.

D.

Cut pattern and vary over-all width to give a pattern circumference of from 2" to 3" less than measurement of stump ischial circumference. The socket contours will be enlarged as necessary during the fitting.

Fig. 12. Modification of socket shape to accommodate individual stump characteristics.

of the seat, conditions which create a great deal of discomfort, can be prevented by shaping the bearing surface in such a way that the seat slopes slightly toward the inside of the socket to render it more comfortable. Sloping increases the radius of the edge of the ischial seat and lessens the burning sensation of the skin in this region.

If the ischial seat is too prominent, or if the ischium rides on the edge of the seat, a jabbing sensation or a marked increase in pressure is felt near the end of the stance phase. Lowering the ischial seat allows more weight to be distributed to the gluteal region and, if the ischial tuberosity is located properly on the seat, results in less discomfort and a shorter break-in period.

Amputees with highly developed stump muscles may not require a well-defined ischial seat. In some cases, the muscles may push the ischial seat away from the tuberosity of the ischium and cause the weight to be carried by the muscles around the top of the socket. Such a condition is not objectionable, provided that the socket is designed with proper modification of the ischial seat. Indeed, such a design

may be necessary in unusual cases, as for example those with end-bearing stumps.

SPECIAL CONSIDERATIONS IN THE SUCTION SOCKET

Tightness of Fit

In the case of the suction socket, better results are obtained by having proper contours than by having a tight fit *(3)*. If, in the course of donning the leg, much difficulty is encountered in removing the sock, the fit is too tight. The superior brim of the socket should fit the contour of the stump while the muscles are tensed, and the fit should be so accurate that the socket can be suspended for short periods by skin friction without the aid of negative pressure *(i.e.,* without a valve).

Free Space Below the Stump End

The volume of unoccupied space at the lower end of the suction socket is not critical in obtaining sufficient suction. In most cases, it is convenient to have approximately 2 in. of space below the end of the stump to provide room for installation of the valve and for elongation of the soft tissue. In general, the
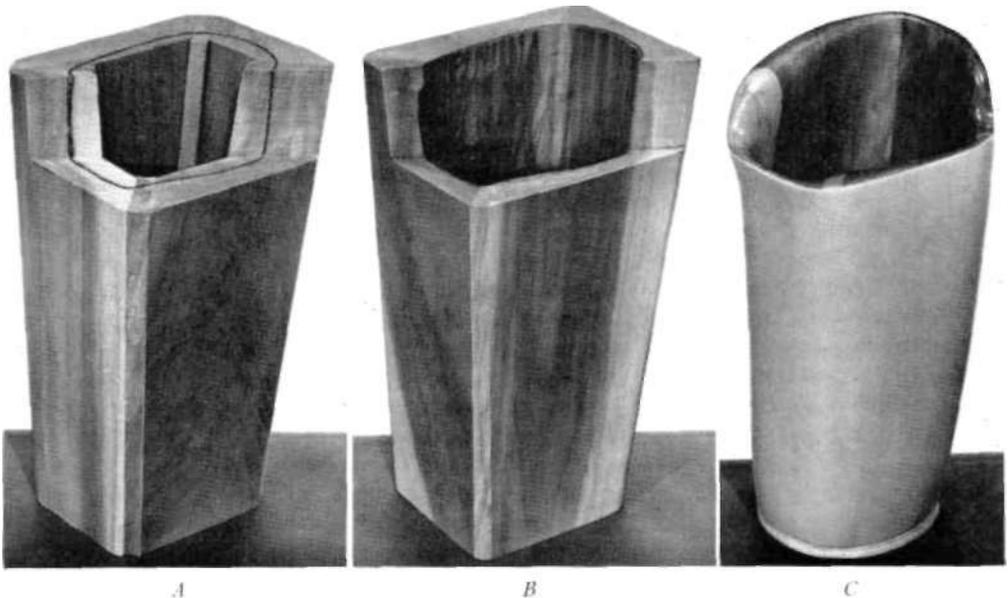

Fig. 13. Three stages in the construction of a wooden socket. *A,* Block cut to form posteromedial shelf. *B,* Roughed-out socket. *C,* Completed socket with inside finished and rawhide covering on outside.

smaller the volume in the end of the socket the less the excursion, but in itself the amount of free volume has no significant effect on the magnitude of the negative pressure.

End-Bearing

If it can be tolerated, end-bearing is recommended because it relieves the load on the ischium. Felt or foam-rubber padding placed in the bottom of the socket permits comfortable end-bearing, the thickness of the padding governing the amount of weight carried on the end of the stump. Although little free space remains in the socket, adequate suction and control are not affected. For example, Gritti-Stokes amputations, which are principally end-bearing, have been fitted successfully.

Inside Finish

No single recommendation is made regarding adequate nonirritating finishes. Industrial and perspiration-resistant lacquers common to the limb industry are being used routinely. Some subjects have reported slipping of the socket because of perspiration. In some cases, perspiration also has caused the lacquer finish to deteriorate and to produce a roughness resulting in skin irritation. In general, however, these industrial lacquers have proved satisfactory when applied according to manufacturers' specifications. In cases of excessive perspiration, the socket may have to be refinished every few months. Whenever perspiration creates a severe problem, the amputee should be referred to a dermatologist for possible treatment.

Bottom Seal

The bottom of the socket should be sealed with a piece of hard wood 1/8 in. thick or more, cut so that the surface goes along the grain, and sealed with a waterproof glue. The bottom may be given additional protection by applying a thin coating of one of the thermosetting plastics common to the limb industry.

Control of Negative Pressure

Several different types of valves have been used in suction sockets with good results. A simple type of plug valve with a manual suction release is satisfactory. Automatic expulsion valves permit some change of air in the socket, a beneficial feature during hot weather and at times when the amputee perspires. They have proved successful in all cases and are now in general use.

The valve opening should be positioned for ease in removing the fitting sock when the leg is donned and for convenience in operating the manual control, and it should be placed where the distal end of the stump is least likely to touch the inner face of the valve. The optimum location is toward the front on the medial side below the stump end.

The magnitude of the negative pressure or suction required to hold a suction socket in place is only slightly greater than the value given by dividing the weight of the prosthesis by the cross-sectional area of the stump near the distal end—in most cases about 1-1/2 lb. per sq. in. With the additional support given by contracting the stump muscles during each step, a negative pressure of 1-1/2 lb- Per sq. in. is sufficient. Some amputees prefer somewhat greater suction, with its accompanying feeling of security, but excessive suction may cause edema. A negative pressure greater than 1-1/2 lb. per sq. in. indicates the presence of forces tending to pull or push the leg off the stump. This action may occur when the stump muscles are contracted, or it may be caused by an improper fit resulting in constriction of the muscles. Use of a gauge for measuring the maximum negative pressure at the time of the rough and the final fittings serves as a check on the quality of fit and is essential to good and consistent results.

Accurate records should be made of the variations in pressure inside the suction socket during normal walking. With the automatic expulsion valve now in general use, these records should show a small positive pressure during weight-bearing and a negative pressure when the leg is in the swing phase. Figure 14 is a record of the pressure variations in a suction socket during two complete walking steps, the valve used during this test permitting automatic exhaust starting at a positive pressure of 1/2 lb. per sq. in.

The stiffness of the spring in the valve has, in itself, no direct effect on the magnitude of the maximum negative pressure. It does,
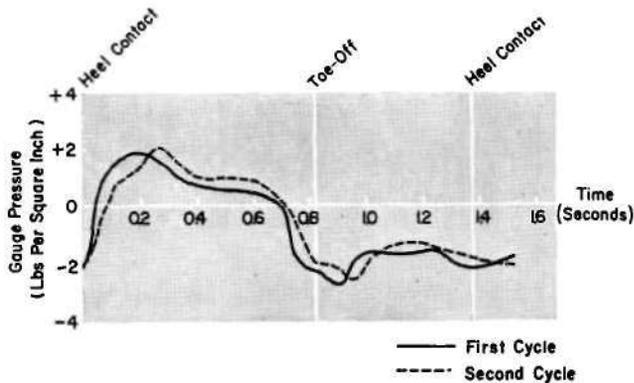
Fig. 14. Typical pressure variation in an above-knee suction socket during level walking. Body weight: 145 lbs.

however, allow a greater or lesser amount of air to be expelled with each step and thereby affects the amount of positive pressure developed during weight-bearing. Fairly high positive pressure within the socket during the stance phase generally is found desirable because it increases the pavex action of the socket on the stump, with consequent benefit to the circulation. High positive pressures help to control edema and to give the amputee a sense of "walking on air." But, as already mentioned, too great a positive pressure in the stance phase may tend to push the leg off or to increase the piston action of the stump in the socket. Springs permitting expulsion at a positive pressure of 1/2, 1-1/2 or 2 lb. per sq. in. now are commercially available. The choice should be based upon individual circumstances.

Some leakage generally occurs either in the valve or between the socket wall and the stump. A regulated amount of leakage is, however, desirable because it relieves the suction during periods of inactivity. If the leak rate is too great, the leg may fall off or the piston action may be excessive and cause discomfort. If the leak rate is too small, however, edema may result. A good test for leak rate is to measure the time required for the negative pressure to drop to half its initial value while the prosthesis is suspended on the relaxed stump. If the time is 50 to 80 sec, the leak rate is satisfactory, but if it is greater than 100 sec, the manual release should be used during periods of inactivity.

CONCLUSION

In summary, then, it may be restated that,

in the construction of an above-knee artificial leg, the objective of the prosthetist is to provide the wearer with optimum security in standing and walking, the best possible walking pattern, a minimum requirement for expenditure of energy in usual activities, and a generally comfortable leg that can be used more or less continuously without injuring the stump and without causing undesirable postural deformities. The above-knee prosthesis is called upon to replace as nearly as possible the functions of the normal leg, but it must do so under the influence of a residual motor mechanism deficient in power and sensory control. The necessary features are therefore to be obtained only by observance of certain functional rules established on the basis of anatomical, physiological, and mechanical considerations.

Of first importance is that the prosthetist well understand the mutual interdependence of the details of alignment of the various components and of the fit and orientation of the socket. Since, unlike the normal limb, support in the above-knee prosthesis is not through the shaft of the femur but through some other axis, due cognizance needs to be taken of the new set of musculomechanical relationships and of the influence of these relationships on the static and dynamic characteristics of the artificial replacement. When proper compensation for these factors is made by the limbfitter, undesirable compensation by the amputee is avoided, while the requirements of comfort, function, and acceptable gait are satisfied. In no other way can so much satisfaction be afforded the above-knee amputee.

LITERATURE CITED

1. Buchthal, Fritz, and E. Kaiser, *Optimum mechanical conditions for work of skeletal muscle,* Acta Psychiat. et Neurol., 24:333 (1949).
2. Eberhart, Howard D., Verne T. Inman, and Boris Bresler, *The principal elements in human locomotion,* Chapter 15 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.
3. Eberhart, Howard D., and Jim C. McKennon, *Suction-socket suspension of the above-knee prosthesis,* Chapter 20 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.
4. Haddan, Chester C, and Atha Thomas, *Status of the above-knee suction socket in the United Stales,* Artificial Limbs, May 1954. p. 29.
5. Inman, V. T., *Functional aspects of the abductor muscles of the hip,* J. Bone & Joint Surg., 29:607 (1947).
6. Inman, Verne T., and H. J. Ralston, *The mechanics of voluntary muscle,* Chapter 11 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.
7. Libet, B., H. J. Ralston, and B. Feinstein, *Effect of stretch on action potentials in muscle,* Biol. Bull., 101:194 (1951).
8. Radcliffe, C. W., *Use of the adjustable knee and alignment jig for the alignment of above knee prostheses,* University of California (Berkeley), Prosthetic Devices Research Project, Report to the Advisory Committee on Artificial Limbs, National Research Council, August 1951.
9. Radcliffe, Charles W., *Mechanical aids for alignment of lower-extremity prostheses,* Artificial Limbs, May 1954. p. 23ff.
10. Radcliffe, Charles W., *Alignment of the above-knee artificial leg,* Chapter 21 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.
11. Ralston, H. J., *Mechanics of voluntary muscle,* Am. J. Phys. Med., 32:166 (1953).
12. Ralston, H. J., H. D. Eberhart, V. T. Inman, and M. D. Shaffrath, *Length-tension relationships in isolated human voluntary muscle,* Proc. 17th Internat. Physiol. Cong., Oxford, 1947. p. 110.
13. Ralston, H. J., V. T. Inman, L. A. Strait, and M. D. Shaffrath, *Mechanics of human isolated voluntary muscle,* Am. J. Physiol., 151:612 (1947).
14. Ramsey, R. W., and S. F. Street, *Isometric length-tension diagram of isolated skeletal muscle fibers of frog,* J. Cell. & Comp. Physiol., 15:11 (1940).
15. Schede, Franz, *Theorelische Grundlagen für den Bau von Kunstbeinen; insbesondere für den Oberschenkelamputierten,* Ztschr. f. orthopad. Chir., Supplement 39, Enke, Stuttgart, 1919.
16. Schnur, Julius, *Beinbelastungslinie und Schwerlinie,* Medizinische-Technik, 5(3):54 (March 1951).
17. Schnur, Julius, *Die Aquilibral-Kontakt Prothese,* Orthopadie-Technik, 4(2):36 (February 1952).
18. University of California (Berkeley), Prosthetic Devices Research Project, Subcontractor's Final Report to the Committee on Artificial Limbs, National Research Council, *Fundamental studies of human locomotion and other information relating to design of artificial limbs,* 1947. Two volumes.
19. University of California (Berkeley), Prosthetic Devices Research Project, [Report to the] Advisory Committee on Artificial Limbs, National Research Council, *The suction socket above-knee artificial leg,* 3rd edition, April 1949.
20. Wagner, Edmond M., and John G. Catranis, *New developments in lower-extremity prostheses,* Chapter 17 in Klopsteg and Wilson's *Human limbs and their substitutes,* McGraw-Hill, New York, 1954.

Digest of Major Activities of the Artificial Limb Program

Technical Committee on Prosthetics

Since 1948 responsibility for the technical aspects of the Artificial Limb Program has been in the hands of two capable groups, the Lower- and Upper-Extremity Technical Committees (ARTIFICIAL LIMBS, January 1954, pp. 31-33). Composed of engineers, prosthetists, surgeons, and others, these groups have met from time to time to review current projects and to recommend general courses of action. As progress has taken place, there have arisen more and more problems common to both committees.

For this reason, and because a large proportion of committee members have been participating in both groups, it was decided at a joint meeting on October 2 to combine the Lower- and Upper-Extremity Technical Committees into a single committee to be known as the Technical Committee on Prosthetics (TCP). By popular vote, Dr. Charles O. Bechtol, long associated with both the lower- and upper-extremity programs, was selected as chairman.

Until his appointment last July as Associate Professor of Surgery and Chief of the Division of Orthopedic Surgery, Yale University, Dr. Bechtol was affiliated with the School of Medicine, University of California, San Francisco, and also conducted a private practice in Oakland. His participation in the establishment of the Veterans Administration Orthopedic Clinic Teams and his research work in cineplasty are well known in orthopedic circles. Dr. Bechtol also has been responsible

-Alburtus, YaleNews Bureau
DR. BECHTOL

for the physicians and surgeons section of the UCLA Upper-Extremity Training Courses, has served as a member of the American Board for Certification of the artificial-limb industry, and is active in the American Academy of Orthopaedic Surgeons.

Membership of TCP, with affiliations, is as follows:

CHARLES O. BECHTOL, M.D. *(Chairman).*

TONNES DENNISON, Field Engineer, Advisory Committee on Artificial Limbs *(Secretary).*

SAMUEL W. ALDERSON, President, Alderson Research Laboratories, Inc., New York City.

MILES H. ANDERSON, Ed.D., Educational Director, Artificial Limbs Project, University of California, Los Angeles, Calif.

ERNEST A. BRAV, Col., MC, USA, Chief, Orthopedic Section, Walter Reed Army Medical Center, Washington, D. C.

THOMAS J. CANTY, Capt., MC, USN, Director, Navy Prosthetics Research Laboratory, U.S. Naval Hospital, Oakland, Calif.

JOHN G. CATRANIS, President, Catranis, Inc., Syracuse, N. Y.

CLINTON L. COMPERE, M.D., Orthopedic Consultant, Veterans Administration Regional Office, Chicago, Ill.

RENATO CONTINI, Project Director, Prosthetic Devices Study, College of Engineering, New York University, New York City.

HOWARD D. EBERHART, Professor of Civil Engineering, University of California, Berkeley, Calif.

HERBERT ELFTMAN, Ph.D., Associate Professor of Anatomy, College of Physicians and Surgeons, Columbia University, New York City.

SIDNEY FISHMAN, Ph.D., Assistant Project Director, Prosthetic Devices Study, College of Engineering, New York University, New York City.

MAURICE J. FLETCHER, Lt. Col., MSC, USA, Director, Army Prosthetics Research Laboratory, Walter Reed Army Medical Center, Washington, D. C.

CHESTER C. HADDAN, President, Gaines Orthopedic Appliances, Inc., Denver, Colo.

VERNE T. INMAN, Ph.D., M.D., Professor of Orthopedic Surgery, School of Medicine, University of California, San Francisco, Calif.

FRED LEONARD, Ph.D., Chief, Plastics Development Branch, Army Prosthetics Research Laboratory, Walter Reed Army Medical Center, Washington, D. C

HANS MAUCH, Aero-Medical Laboratory, Wright-Patterson Air Force Base, Ohio.

EUGENE F. MURPHY, Ph.D., Chief, Research and Development Division, Prosthetic and Sensory Aids Service, Veterans Administration, New York City.

CHARLES W. RADCLIFFE, Acting Assistant Professor of Engineering Design, University of California, Berkeley, Calif.

AUGUST W. SPITTLER, Col., MC, USA, Assistant Chief Surgical Consultant, Office of the Surgeon General, Department of the Army, Washington, D. C.

CRAIG L. TAYLOR, Ph.D , Professor of Engineering

and Biophysics, University of California, Los Angeles, Calif.

ATHA THOMAS, M.D., Professor of Orthopedic Surgery, School of Medicine, University of Colorado, Denver, Colo.

HOWARD R. THRANHARDT, J. E. Hanger Company, Atlanta, Ga.

Lucius TRAUTMAN, Minneapolis Artificial Limb Company, Minneapolis, Minn.

EDMOND M. WAGNER, Consulting Engineer, San Marino, Calif.

The next meeting is scheduled to be held at the University of California at Los Angeles on February 4.

Human Limbs and Their Substitutes

Klopsteg and Wilson's *Human Limbs and Their Substitutes,* the 840-page treatise prepared by more than 30 collaborators working in cooperation with ACAL, appeared in mid-November under the imprint of the McGraw-Hill Book Company, New York. Price is $12. Originally scheduled to appear in October (ARTIFICIAL LIMBS, September 1954, p. 77), this comprehensive volume on prosthetics inadvertently ran into conflict with the publisher's usual autumn rush of textbook production, so that final printing and binding had to be postponed for a period of several weeks. That *Human Limbs* is complete at last attests the perseverance of all who had a part in its making. The Maple Press Company of York, Pennsylvania, was the printer.

R and D Panel Meetings

Meetings of the Panels on Lower- and Upper-Extremity Research and Development held in Berkeley, California, on December 1 and 2, respectively, were preceded by working sessions of the various Phase Subcommittees held on November 29 and 30. Reports on the status of the more active projects were prepared by the Phase Subcommittees for use by the Panels in making recommendations concerning the course of these projects through the transition procedures.

Meetings of both Panels are scheduled to be held at the University of California at Los Angeles the week of February 7.

Upper-Extremity Prosthetics Training Center

The eleventh session of the Upper-Extremity Prosthetics Training Course at the University of California at Los Angeles was completed on November 19. That particular session, offered to those prosthetists, therapists, and physicians who were from regions covered by the first ten sessions but who were unable to participate previously, was attended by a capacity enrollment. Graduates and their affiliations are as follows:

PROSTHETISTS

RUBEN HERRARA
Walter Reed Army Medical Center
Washington, D. C.

HERBERT HENNING
11138 S. Indiana Ave.
Chicago, Ill.

RICHARD JONES
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Fabens, Texas

JOHN KOLMAN
3410 Whittier Blvd.
Los Angeles, Calif.

JERRY LABATE
Veterans Administration
252 Seventh Ave.
New York City

CHARLES LE MOYNE
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Van Nuys, Calif.

RICHARD NIMS
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Long Beach, Calif.

ROBERT PLUMB
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JOHN SNYDER
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Tulsa, Okla.

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J. J. YOPEK
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Detroit, Mich.

FRANK ZEMAN
USN Reservation
San Patricio, P. R.

THERAPISTS

Louis BENECK, P.T.
Veterans Administration Hospital
Long Beach, Calif.

ELEVENTH tTPER-EXTREMITY SCHOOL—« full hull*'.

MARGARET BRYCE. P.T.
University of Wisconsin
Madison, Wis.

SlXTA DOMINGUEZ, P.T.
Veterans Administration
San Juan, P. R.

JANE L. FRITSON, O.T.
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.ROSEMARY KELLY, P.T.
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ODETTA MF.STEK, P.T.
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CAPT. ELIZABETH XACBOD, O.T.
Walter Reed Arm\ Medical Center
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JOHN XYGAARD. P.T.
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University of Illinois Hospital
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PHYSICIANS AND SURGEONS

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WILLIAM L. BETTISON, M.D.
Michigan Crippled Children's Commission
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Philadelphia, Pa.

EMILY R. HESS, M.D.
College of Medicine
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Cincinnati, Ohio

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THOMAS P. REEDER, M.D.
Los Angeles County General Hospital
Los Angeles, Calif.

FRANCIS L. WENGER, M.D.
George Washington University Hospital
Washington, D. C.

The twelfth session—to begin January 3 for prosthetists, January 24 for therapists, and February 7 for physicians and surgeons—is planned for residents of cities in the southeastern and southwestern United States and certain additional cities not otherwise covered.

Activation of Regional Training Courses

Now that the Upper-Extremity Prosthetics Training Courses at UCLA are about to come to a close, it is possible to begin directing attention to making new techniques and knowledge in both lower- and upper-extremity prosthetics available to clinic teams with the least possible delay after evaluation studies have been completed. In order to develop a plan to accomplish this goal in a way most beneficial to all concerned, a steering committee has been appointed. It consists of Miles H. Anderson *(Chairman),* William M. Bernstock, Sidney Fishman, Herman Hittenberger, Verne T. Inman, Glenn E. Jackson, Charles W. Radcliffe, Howard R. Thranhardt, and A. Bennett Wilson, Jr.

Although it is too early to outline in detail the preliminary plans considered by this group, thought now is being given to conducting courses of about two weeks' duration in regional centers, probably with the cooperation of a local university. These courses will be given as new techniques emerge from the Artificial Limb Program. In all probability the first series of sessions will be devoted to fitting and alignment of the above-knee prosthesis. To this end, a manual covering the subject matter is in preparation.

Because of the uncertainty of the completion date of the present series of courses in upper-extremity prosthetics, no date or place has been established for the pilot school.

Presentation at OALMA Assembly

On "Prosthetics Day," Wednesday, September 29, several persons closely associated with the Artificial Limb Program delivered talks before the Scientific Assembly of the Orthopedic Appliance and Limb Manufacturers Association at Chalfonte-Haddon Hall, Atlantic City (ARTIFICIAL LIMBS, September 1954, p. 79). Although Dr. Craig L. Taylor of UCLA was unable to be present as planned, his report on European progress was read in a very able manner by Dr. Miles H. Anderson. That presentation was followed by additional reports by Mr. Carlton E. Fillauer of Fillauer Surgical Supplies, Inc., Chattanooga, and Dr. Eugene F. Murphy of the VA's Prosthetic and Sensory Aids Service, New York City. Both of the latter speakers also had just returned from the Continent.

Using slides, Dr. Verne T. Inman of the University of California Medical School then presented a discussion of the philosophy underlying the Lower-Extremity Clinical Study.

He was followed by Professor Howard D. Eberhart, who described the lower-extremity fundamental research studies and device development at the University of California at Berkeley. These two lectures form the basis for the article entitled *The Lower-Extremity Clinical Study—Its Background and Objectives*, which appears in this issue of ARTIFICIAL LIMBS (page 4).

In the afternoon Dr. Sidney Fishman of NYU conducted a panel session covering various aspects of the Upper-Extremity Field Studies. Participating in the discussion were Mr. Howard R. Thranhardt, Air. Chester Nelson, Mr. Alvin Muilenberg, and Dr. Harriett Gillette.

Mr. Chester C. Haddan, originally scheduled to give his views on the effect of upper-extremity developments on the artificial-limb industry, was unable to attend the meeting. At Mr. Haddan's request, Mr. Thranhardt discussed the subject.

A display depicting the objectives and methods of the UC Prosthetic Devices Research Project attracted much interest.

The Navy Above-Knee Leg

Having undergone successfully a series of comprehensive tests, the U.S. Navy above-knee leg, developed at the Navy Prosthetics Research Laboratory, U. S. Naval Hospital, Oakland, California, was recommended last year by ACAL for supply to eligible Veterans Administration beneficiaries. This device (ARTIFICIAL LIMBS, May 1954, p. 16) consists of the so-called "Navy functional ankle" with two-durometer rubber block, plastic shank, and Navy variable-friction knee mechanism. Of special interest is the knee unit which, by effecting an increase in mechanical friction at the beginning and at the end of the swing phase, limits heel rise and provides deceleration of the shank prior to heel contact. The resulting gait thus differs less from the normal than does that obtained when the conventional constant-friction knee bolt is used.

To arrange for supply of the necessary units, bids were invited for the manufacture of the leg according to drawings and specifications furnished by the Navy. As a result of the competitive bidding, Western Wholesale Parts Company of Los Angeles has been awarded a

contract by the National Academy of Sciences —National Research Council to furnish a number of Navy above-knee legs for use in the lower-extremity field research program of the Advisory Committee on Artificial Limbs. With the permission of the Surgeon General of the Navy, the company also is offering these units to the artificial-limb industry as setups.

Honorary Award to OALMA

The 1954 Grand Award of the American Trade Association Executives was presented on November 8 to the Orthopedic Appliance and Limb Manufacturers Association "for having rendered outstanding service to the industry which it represents as well as to the American Public." Using as its material the story of the Suction-Socket Training Schools for prosthetists and orthopedic surgeons, OALMA won out over the entries of almost 60 other trade associations. Presentation was made at the national convention of ATAE at the Hotel Statler in Los Angeles. The Honorable Sinclair Weeks, United States

—Rothschild, Los Angeles

OALMA WINS HONOR—George W. Robbins of the University of California faculty presents award to Glenn E. Jackson, Executive Director of the Orthopedic Appliance and Limb Manufacturers Association. Harvey Lanham, Regional Director of OALMA for Southern California, looks on,

Secretary of Commerce, served as Chairman of the Jury of Awards. Other members of the jury included Mr. Clem D. Johnston, President of the Chamber of Commerce of the United States; Mr. Harold C. McClellan, President of the National Association of Manufacturers; Mr. Neil H. Borden, President of the American Marketing Association and Professor of Marketing at Harvard University; and Dr. George W. Robbins, Acting Dean, School of Business Administration, University of California.

The Suction-Socket Schools constitute a unique example of cooperation between an industry association and the Federal Government. Initiated in 1947 by OALMA and the U. S. Veterans Administration, with the active help and encouragement of the National Research Council working through the Advisory Committee on Artificial Limbs, the courses are intended to provide better technical service to the leg amputee. To date schools have been held in more than 20 cities. Graduates receive a certificate bearing the signatures of an executive of OALMA and of the Director of the Prosthetic and Sensory Aids Service of the Veterans Administration. Although actual operating expenses are met by OALMA, the VA provides part of the faculty and furnishes a considerable number of its orthopedic consultants and technicians for guidance.

Commenting on the results so far, Glenn E. Jackson, Executive Director of OALMA, pointed to several major benefits aside from the improved fitting of individual amputees. He said that, through the medium of the schools, prosthetist, therapist, and orthopedic surgeon have been brought into closer cooperative relationship in the form of the clinic team and that education of the limb technician has been a major influence in the development of limbmaking from a craft to the profession it should rightly be. Jackson said also that the administration of the schools has already built up a pattern of cooperation between agencies of the Federal Government and trade associations—a pattern which has already shown considerable "carry-over value" for other industries whose associations are developing educational programs.

The ATAE Award Program, established in 1929 by Margaret Hayden Rorke, long-time Managing Director of the Textile Color Card Association of the United States, has among its chief purposes "to foster and promote true service to American industry through the medium of the trade association" and "to arouse the appreciation of industry and the public in the trade association movement in America." According to ATAE, in 1954, as in previous years, the winning entry portrays "the trade association as the channel through which business has chosen not only to operate, but more important, to cooperate."

ABC EXHIBIT AT AMA MEETING—Left to right, Mary S. McLain, Manager of the J. E. Hanger Company, Miami, Florida, and Lester A. Smith, Assistant Director of the American Board for Certification, show research reports to Dr. Walter L. Bierring of Des Moines, Iowa, Past-President of the American Medical Association, and Dr. Thomas G. Hull, Secretary of the Council for the Scientific Assembly of AMA.

ABC Exhibit

The certification program for orthotists and prosthetists, in which effective use is

made of latest research findings, was the subject of an exhibit at the National Clinical Meeting of the American Medical Association held at Miami, Florida, November 29 through December 2. Prepared and sponsored by the American Board for Certification of the Prosthetic and Orthopedic Appliance Industry, Inc., the display was presented for the benefit of the 3000 physicians and 1200 medical technicians in attendance.

Among the features of the exhibit were various below-elbow prostheses developed in the Artificial Limb Program and now available to amputees generally. Literature distributed to visitors included the *Official Registry of Certified Facilities* and the September 1954 issue of ARTIFICIAL LIMBS.

Lester A. Smith, Assistant Director of the Certification Board, arranged the presentation and was in charge during the meeting. He was assisted by Mary S. McLain, Manager of the J. E. Hanger Company at Miami, and prosthetics technicians Jack L. Caldwell of the Hanger Company of Tampa and Nicholas M. Treuhaft of the Wheatly Limb and Brace Company, Miami.

The program of the Clinical meeting of the AMA Scientific Assembly described the exhibit as showing the results of the Certification Board's program during the last six years, including the development of apprenticeship standards, use of advanced training at the Prosthetics Training Center, UCLA, and the in-service training program now under development.

NATIONAL ACADEMY OF SCIENCES-NATIONAL RESEARCH COUNCIL

The National Academy of Sciences—National Research Council is a private, nonprofit organization of scientists, dedicated to the furtherance of science and to its use for the general welfare.

The Academy itself was established in 1863 under a Congressional charter signed by President Lincoln. Empowered to provide for all activities appropriate to academies of science, it was also required by its charter to act as an adviser to the Federal Government in scientific matters. This provision accounts for the close ties that have always existed between the Academy and the Government, although the Academy is not a governmental agency.

The National Research Council was established by the Academy in 1916, at the request of President Wilson, to enable scientists generally to associate their efforts with those of the limited membership of the Academy in service to the nation, to society, and to science at home and abroad. Members of the National Research Council receive their appointments from the President of the Academy. They bclude representatives nominated by the major scientific and technical societies, representatives of the Federal Government designated by the President of the United States, and a number of members-at-large. In addition, several thousand scientists and engineers take part in the activities of the Research Council through membership on its various boards and committees.

Receiving funds from both public and private sources, by contribution, grant, or contract, the Academy and its Research Council thus work to stimulate research and its applications, to survey the broad possibilities of science, to promote effective utilization of the scientific and technical resources of the country, to serve the Government, and to further the general interests of science.