Spring 1969

Artificial Limbs

A Review of Current Developments

COMMITTEE ON PROSTHETICS
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Prosthetics and Orthotics—U.S.A.—1969

Sidney Fishman, PHD'

A WEIGHT-BEARING walking cast for tibial fractures, a programmed "feeding arm" for the congenital-amputee child, the operating-room procedures for below-knee amputations, the analysis of prosthetic fit by x-rays and transparent sockets, and the use of externally induced intracavitary pressures to stabilize the spine are all current areas of concern in prosthetics and orthotics. Two or even one decade ago, hardly one of these matters would have been considered a legitimate concern of prosthetics and orthotics. These areas currently being investigated serve as clear and dramatic evidence of the growth in scope of the field. Prosthetics and orthotics has been characterized for some years by the frequent introduction of novel methods for treating an ever-increasing variety of orthopaedic conditions through the application of bioengineered devices and techniques.

Although progress in the directions mentioned above has been gratifying, recent efforts to specifically improve the more orthodox methods of fitting lower- and upper-extremity prostheses and braces have met with more limited success. It would appear that the singular developments of five to ten years ago *(e.g.,* patellar-tendon-bearing prostheses, quadrilateral total-contact sockets, and tibial torsion braces) have outstripped the basic anatomical, kinesiological, and biomechanical data and insights upon which all logical designs must be based. We therefore are, and have been for a period, on a plateau with regard to improving the more conventional methods of fitting artificial limbs and braces to any significant degree. This has stimulated a series of studies in some quarters in an effort to uncover new and more objective information regarding the factors which influence the interface between patient and device, such as the distribution of forces and pressures, and their effects upon the tissues and the functions of the wearer.

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Other researchers have been content to work within the framework of the existing data in the hope that advanced engineering technologies will permit a breakthrough towards improved function and comfort—as evidence, note the plethora of electric hands and elbows and related electronic control systems, the more sophisticated hydraulic and pneumatic knee mechanisms, and the variety of approaches to the casting of sockets for the lower-extremity amputee.

It is difficult to foretell which course will yield the most clinically useful improvements, although my predilection is that the more basic research approaches are more promising for the present.

What about prosthetic-orthotic education?—There are today three to six institutions of higher learning (depending on how you prefer to count) that offer systematic instruction in prosthetics and orthotics—not only to prosthetists but to physicians, surgeons, therapists, and other rehabilitation workers as well. Curricula and text materials have been developed for each significant category of orthopaedic impairment, such as upper-extremity prosthetics, lower-extremity orthotics, etc., and the information contained therein is becoming more broadly disseminated. Long-term training (2-4 years) as well as short-term training (1-6 weeks), designed for different purposes, is available, and a healthy interaction is being maintained between the clinician, researcher, and educator. These courses have continued long enough by now for "feedback" to have reached the teaching institutions so that instructional practices can be modified accordingly.

Lastly, there appears to be a healthy balance between the similarities and differences in the content of the courses being offered at the several institutions. This situation has been aided immeasurably by the activities of the University Council on Orthotic-Prosthetic Education and the Committee on Prosthetic-Orthotic Education, which have successfully performed for prosthetics and orthotics education the coordinating function that CPRD has performed in the related area of prosthetics and orthotics research.

As to clinical practice—the ultimate beneficiary of research and education—the picture is almost as happy. While by no means perfect, or even completely adequate, improved standards for prosthetic-orthotic service are now in effect. Through the efforts of the American Board for Certification, the American Academy of Orthopaedic Surgeons, the American Academy of Physical Medicine and Rehabilitation, several governmental agencies, UCOPE, CPOE, and CPRD, the means of evaluating and upgrading the varying qualities of prosthetics and orthotics practice are at hand. Unfortunately, the determination with which these standards are applied varies from one locale to the next; however, the general trend is in one direction—

upward. One cannot, however, avoid mentioning the troublesome effects that the serious shortages of trained personnel have had on the more rapid elevation of professional practice.

It would appear that two new subspecialties may be evolving because of changes in the relative rates of incidence of upper-extremity *versus* lower-extremity disabilities. Apparently, a prosthetics and orthotics facility can be fully productive working only with patients with lower-extremity impairments, leaving the far less numerous upper-extremity patients to specialized centers. Perhaps those responsible for certification may in time see fit to take official cognizance of this trend, since there is a considerably greater substantive relationship between lower-extremity *prosthetics* and lower-extremity *orthotics* (and upper-extremity prosthetics and upper-extremity orthotics) as specialty fields than is shown by the present division between prosthetics and orthotics.

Although there have been a few dramatic improvements, progress has been gradual, but nonetheless substantial and real. The changes could be seen clearly only if one had gone to sleep in 1945 when the Artificial Limb Program began and had reawakened in 1969; nothing would be recognizable in the practice of prosthetics—not the components, not the fitting techniques, not even the prosthetists. In the field of orthotics, some things would still appear to be the same, since this area has received systematic attention for a shorter time. Not that *optimum* orthotics practice, as applied by leaders in the field, today is not completely different than in 1945; however, we have been far less successful in our educational efforts in orthotics than in prosthetics. Also, less attention has been paid to the elevation of standards and the qualification of workers in the orthotics field. Awareness of this discrepancy in practice is now widespread, and one may hopefully await the redress of this imbalance in the next few years.

One thing is sure: in the field of prosthetics and orthotics, the proverb "Plus ca change, plus c'est la meme chose" simply does not apply—not in research, not in education, not in clinical practice, and not in the parameters of the prosthetics and orthotics field itself.

Amputations Below the Knee

Ernest M. Burgess. M.D.
Joseph H. Zettl, C.P.

AHE elective amputation must be considered plastic and reconstructive in nature. The need to create a dynamic and sensory motor end-organ should be foremost in the surgeon's mind in planning an amputation, and is emphasized here once more. The below-knee stump no longer hangs suspended in an open-end socket. The variable degrees of pressure and weight-bearing over the entire stump surface afforded by the total-contact patellar-tendon-bearing prosthesis enhance the surgeon's opportunity to fashion a functional terminal end-organ. Stump strength created by surgical muscle stabilization; pliable, sensitive, but nontender skin and scar; adequate soft tissue coverage of bone ends and other pressure-sensitive areas; high ligation and division of nerves to remove neuromata from pressure zones; meticulous rounding and tailoring of bone surfaces; all contribute to an ideal organ for substitute limb application. The atrophic, wasted, bony, below-knee stump so commonly encountered in years past is no longer acceptable.

Stump-muscle stabilization, *i.e.*, the attachment of sectioned muscles under appropriate tension to bone (myodesis) and to opposing muscles (myoplasty), is a prime requisite for dynamic stump activity. Muscle stabilization is especially needed in the through-knee and the above-knee amputee. Our experience also justifies its routine use in below-knee amputation. Muscle-to-bone suture does add operative handling of tissues and encircling sutures carry the potential of local muscle constriction. For these reasons myodesis is not recommended for use in the below-knee amputation for vascular disease. The new technique developed by the Prosthetics Research Study utilizes the long posterior myofascial flap sewn anteriorly to anterolateral deep fascia and tibial periosteum and provides a reasonable degree of muscle fixation without risk of strangulation. Muscle-to-bone suture is reserved for the nonischemic patient.

NONISCHEMIC PATIENTS

The optimum level for a below-knee amputation in the presence of adequate blood supply is at the junction of the middle and lower third of the leg. However, the level of amputation will often be determined by the causal pathology, including infection, the degree of scarring of the tissues, and related factors. The surgeon should save all effective length down to optimum level, consistent with providing a comfortable, nontender stump.

A cylindrical stump shape is desired. The surgeon should think in terms of producing a "foot-like" organ at the below-knee level. The total-contact socket is the "shoe on the foot." Just as plastic surgical techniques are required in operating on the hand and foot, the same techniques of gentleness in skin and other tissue handling are applicable to amputation surgery. When viewed in this light, the amputation becomes a surgical challenge instead of a distressing surgical exercise. Immediate postsurgical prosthetic fitting not only supports and augments the dynamic approach to rehabilitation, it offers certain

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physical advantages, *i.e.,* immobilization, appropriate continuous pressure relationships, and comfort. These benefits further justify its incorporation into the over-all management of the below-knee amputee.

AMPUTATION TECHNIQUE FOR THE NONISCHEMIC PATIENT

The patient is prepared for surgery in the usual manner. A pneumatic tourniquet is used. Short, broad fishmouth skin flaps are outlined to provide a mediolateral closure. In the nonischemic patient the flaps are fashioned approximately equal in length. It is advisable to cut the flaps long, then trim them at the time of closure to provide correct skin tension without puckering or undue tension. Skin and fascia are reflected together.

Scarring, infection, deformity, or other unusual circumstances may necessitate modification of the skin closure. Flaps can be outlined to permit closure in any plane or direction provided the resulting scar is nonadherent, nontender, and able to withstand properly and comfortably wearing of a total-contact socket. Anterior location of the scar, condemned in the past, actually is well tolerated even in elderly patients. The application of principles of plastic surgery in skin management must prevail.

In the average adult the tibia is transected 2 1/2 to 3 in. above the distal level of the skin incision. The fibula is divided 3/8 to 1/2 in. higher. A reciprocating power saw facilitates clean bone section. The tibial periosteum is elevated about 3/4 in. above the cut end of the tibia and the anteromedial angle beveled to provide a larger radius on the anteromedial aspect. Careful *rounding* of the edges with a sharp, fine-tooth file is now done. Bone surfaces must be smooth so as to eliminate the possibility of high unit pressures.

When the muscles are to be reattached to bone, a procedure recommended where it is physiologically feasible, 4 to 6 holes not more than 7/64 in. in diameter are drilled through the lateral and posterior periphery of the tibia about % in. proximal to the distal end. Muscles are sectioned long, the gastrocnemius-soleus is left as a myofascial flap sufficiently long to bring it around the end of the tibia to the anterior surface, and nerves and blood vessels are ligated and divided, the former well above amputation level, the latter at the level of tibial section. The nerves are ligated high, as indicated, but are not pulled down so forcibly that traction-avulsion injury results proximal to ligation.

Muscles are now sutured to the bone through the drill holes with medium braided polyester suture and tying the knots within the medullary cavity of the tibia. The loop sutures pass through the body of the major muscle groups and through deep fascia. They should be attached under moderate tension, slightly greater than rest length and therefore capable of providing maximum function. Muscle groups are now sectioned just beyond the end of the tibia except for the gastrocnemius-soleus flap which is left long, beveled, and brought over the end of the tibia as a thinned myofascial flap and sutured to anterior deep fascia and anterior periosteum. Good muscle stability and stump contour are provided by this technique. The moderately bulbous stump will rapidly contour to an ideal cylindrical shape in the rigid postsurgical dressing.

The skin flaps are trimmed and closed with interrupted fine polyester sutures in such a manner that no tension is present, yet a firm stump without redundant tissue is provided (Fig. 1). Drainage of the stump is optional. We prefer a through-and-through Penrose drain; however, suction drainage is convenient and some wounds will not require any drainage.

THE RIGID DRESSING

The wound is covered with a saline-dampened nonadherent silk or nylon dressing and a small amount of fluffed gauze (2 to 3) is placed over the distal stump end. A sterile three-ply Orion Lycra stump sock is rolled carefully over the stump to avoid damage to the suture lines. The superior portion of the stump sock is held firmly

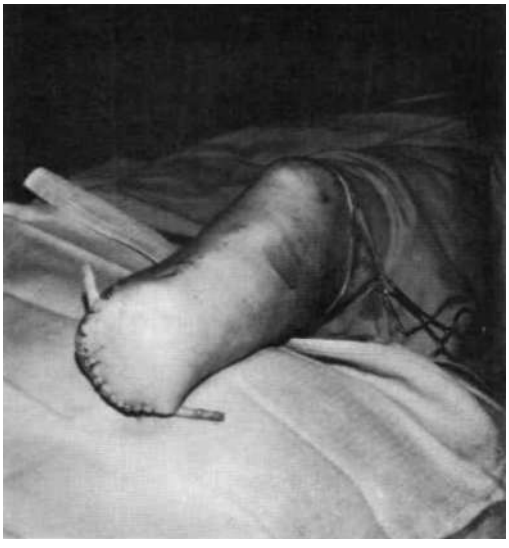

Fig. 1. Below-knee stump of nonischemic patient immediately after closure.

suspended anteriorly and in a proximal direction by an assistant. A simple adjustable shoulder-suspension harness which is interchangeable for right and left can be substituted to achieve the same result.

Relief pads of felt or polyurethane are glued to appropriate locations on the stump sock to provide relief for bony prominences. Prefabricated pads are available in a standard size, right and left, but must be trimmed, skived, and beveled in appropriate areas to suit individual requirements. The pads are designed and located to provide relief of pressures over the patella, the tibial tubercle including the tibial crest, and the distal-anterior (bevel) aspect of the tibia. Dow Corning medical adhesive is used to secure the felt relief pads in place while the polyurethane relief pads are provided with an adhesive backing. A sterile reticulated polyurethane distal pad of the proper size is selected and applied to the distal stump end over the tibial relief pads (Fig. 2).

For the initial part of the rigid dressing, elastic plaster bandage is used because, when pulled within limits of its elasticity, this bandage provides safe and beneficial compression to the stump while conforming well to its contours, providing a smooth, effective, rigid dressing.

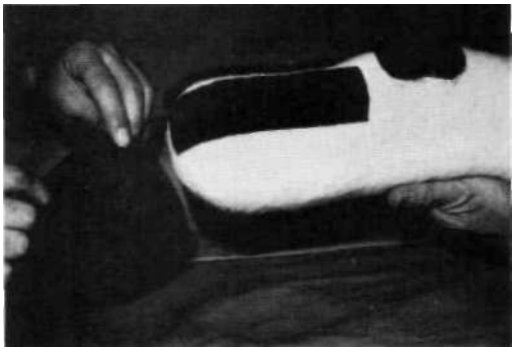

Fig. 2. Application of distal polyurethane pad. Other relief pads are already in place.

Before the wrap is started, the tibial relief and distal relief pads are secured in place with one-and-three-quarter turns of elastic plaster bandage (Fig. 3). Firm tension is applied to the distal portion of the stump from a posterior-to-anterior direction, while the plaster bandage is pulled almost to the limit of its elasticity. By supporting the posterior skin flap, tension on the suture line is reduced and the soft tissues are immobilized. The wrap is then started on the distal end and carried proximally to a level slightly past mid-thigh while tension is maintained in the bandage. A minimum of two layers is required. Circumferential wrapping is carried out from the lateral to the medial aspect, when viewed from the front, in order to avoid anterior displacement of the gastrocnemius

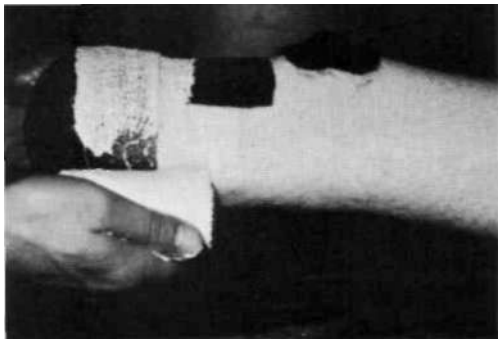

Fig. 3. Beginning the rigid dressing by securing the tibial relief and distal relief pads in place with elastic plaster bandage.

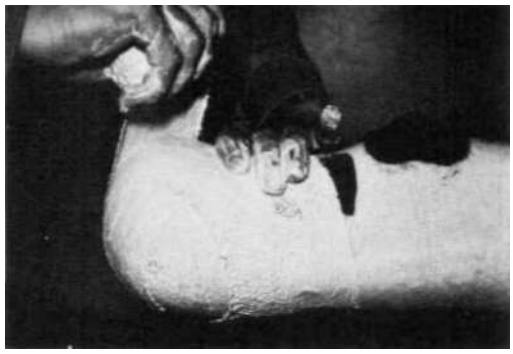

Fig. 4. Application of the first layers of the rigid dressing.

(Fig. 4). Tension in the wrap decreases progressively as the application proceeds proximally to the level of the knee joint where it is simply rolled on up to slightly past mid-thigh. It is important to apply the dressing with firm tension to the distal portion of the stump and to avoid proximal constriction to blood flow. The knee is held in 5 to 15 deg. of flexion controlled by longitudinal tension applied to the stump sock from the proximal end. Owing to the inherent structural weakness of elastic plaster bandage, the initial wrap must be reinforced with conventional plaster bandage and splints. Two splints are applied over the distal portion of the rigid dressing.

A minimum of two layers of conventional plaster bandage is applied starting at the distal third and wrapping proximally with even, overlapping circular wraps (Fig. 5). At the proximal border of the cast a suspension strap is incorporated anteriorly. For an obese patient with excessive soft tissue over the thigh, a second suspension strap is applied posterolaterally. With the plaster of Paris still wet, the cast is gently compressed with the base of each hand just proximal to the femoral condyles to provide an effective built-in suspension mechanism.

After the plaster has hardened sufficiently, the contoured waistbelt is applied to the patient and connected to the strap or straps of the rigid dressing. The prosthetic unit is located and attached to the cast with a roll of conventional plaster bandage (Fig. 6). The pylon is sized and cut to correspond to the length of the sound extremity. A window is cut out of the plaster over the patella to insure com-

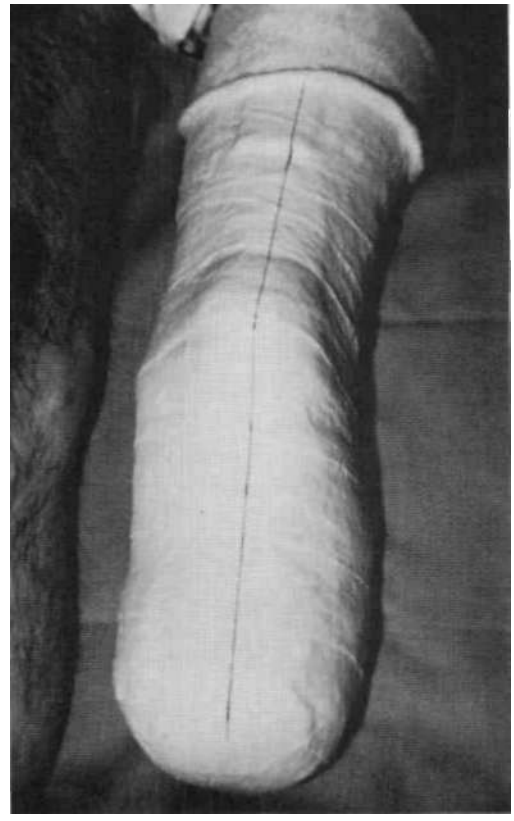

Fig. 5. Completed rigid dressing. Note alignment reference line.

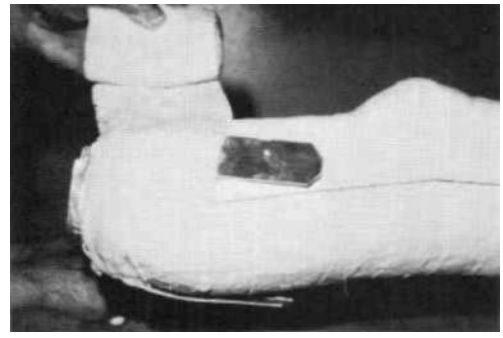

Fig. 6 Attachment of upper portion of prosthetic unit to the rigid dressing. Note alignment reference line.

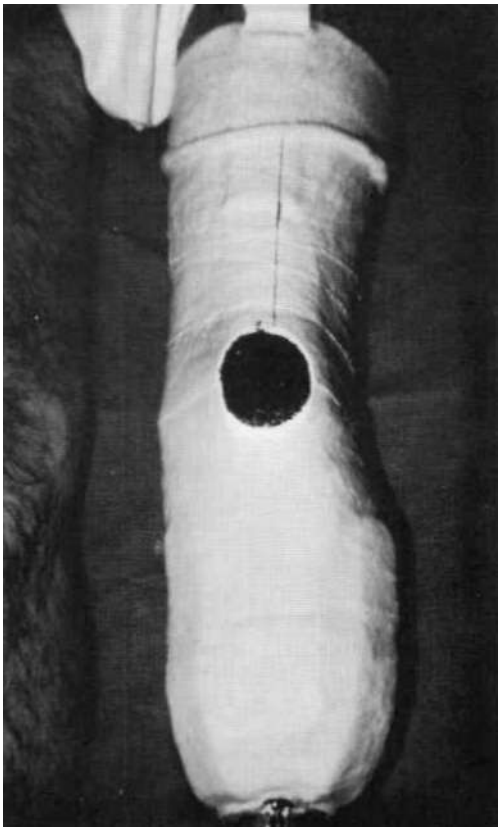

Fig. 7. Window in rigid dressing to provide complete relief over patella,

plete relief in this area (Fig. 7). The prosthetic unit is then disconnected from the cast socket before the patient is taken to the recovery room.

POSTSURGICAL CARE

As a rule, a minimum amount of pain is experienced by patients that have been provided with a rigid dressing. It is unusual for drugs stronger than mild opiates and sedatives to be required for relief. A slight degree of weight-bearing on the stump will usually tend to reduce any discomfort that might be present.

The patient should be encouraged to stand up and bear some weight on the prosthesis as soon after the first 24-hour period postoperatively as is practicable. The time and extent of ambulation must be determined by the responsible surgeon. Walking training should be carried out only under the direction of a physical therapist or other qualified personnel. Activity should be increased daily as the patient's condition permits. Parallel bars, walkerettes, crutches, and canes are used as aids in ambulation. Two bathroom scales may be used to determine the degree of weight-bearing that is taken on the amputated side. These measurements provide a good guide to the clinic team concerning the progress being made by the patient. The patient should never be allowed to ambulate without supervision. Furthermore, ambulation should not be permitted without the prosthesis because in this case the effect of gravity tends to pull the socket away from the stump, thereby reducing the pressure between stump and socket.

On the second postoperative day (48 hours after surgery) the drain is removed. If there does not appear to be any reason for removing the cast, such as elevated body temperature, extreme discomfort, or excessive looseness of fit, the cast is kept in place up to 14 days. If for any reason the cast is removed, whether intentionally or unintentionally, it is mandatory that, if a new cast is indicated, it be applied immediately. During the first two postoperative weeks edema will form rapidly upon removal of the cast and, unless a new cast is reapplied within a very short period, the patient will have to be treated in the conventional manner. The old cast should never be reapplied because of the trauma that is apt to result. When the socket is removed purposely, a cast cutter is used. Often the sutures can be taken out at the time of removal of the first cast, 10 to 14 days after surgery. Sometimes it is necessary to wait until removal of the second cast, 15 to 20 days postoperatively.

In many instances the stump will be sufficiently mature and stable for use of a definitive prosthesis at the time the second cast is removed. When this is so, a cast of the stump is taken and appropriate

measurements are recorded so that fabrication of a permanent prosthesis can proceed immediately. When the definitive prosthesis is delivered, a light plaster socket mobilizing the knee joint is provided for use when the definitive prosthesis is removed. Use of a plaster socket has proven to be superior to elastic bandages to prevent edema. If delays are anticipated in providing the patient with a definitive prosthesis, the prosthetic unit, pylon, and foot are applied to the short cast to continue ambulation activities.

THE ISCHEMIC PATIENT

Throughout the United States and Canada an estimated 80 per cent of all major, elective, civilian amputations result from ischemia. All but a relatively few involve the lower extremity. Significant advances in surgical and postsurgical management coupled with the use of improved prostheses now allow amputation below the knee in the great majority of these patients.

It is difficult to overestimate the importance of the knee in amputee rehabilitation, especially in the older, classical ischemic patient. Debility, impaired vision, poor balance, neuropathy, compromised circulation and joint function in the remaining lower limb, and chronic systemic illness, all emphasize the critical need to save the knee. The older bilateral leg amputee, especially, needs his knees to approach the rehabilitation goal that permits a reasonable degree of ambulation and self-sufficiency. In a consecutive series of 128 unselected major lower-extremity amputations for peripheral vascular disease (1964 through 1968), we have been able to obtain primary healing at below-knee level in 86 per cent. Once healed, the stumps remain healed. With adequate prosthetic care, secondary breakdown will seldom occur. These patients were among the approximately 300 cases requiring amputation of the lower extremity that were used in studying and developing the techniques of fitting prostheses immediately after surgery. As a result of these experiences, separate surgical techniques have been developed for the ischemic patient and for the nonischemic patient.

LEVEL OF AMPUTATION

The great achievements in surgical reconstruction of the peripheral vascular system represent a leading chapter in medical progress during the past two decades. Continuing basic and clinical research throughout the world supports the hope that an even higher percentage of limb salvage can be expected in the years ahead. However, despite the practical effectiveness of modern vascular reconstructive surgery, statistics indicate that amputations for ischemia are increasing both relatively and absolutely in relation to population throughout the western world.

When acute or chronic compromise of arterial blood supply reaches a level insufficient to support tissue viability and when reconstructive surgery and nonsurgical supportive measures fail, amputation will be required.

Patients requiring amputation are entitled to comparable medical and surgical consideration, comparable team effort, and the same high-level rehabilitation management attending similar patients whose ischemic limbs are treated by vascular reconstruction. Too often, ablative surgery does not command this high estate.

Decision to amputate may be simple and evident. Gross necrosis of tissue with demarcation, uncontrollable infection, pain, irreversible neuropathy, alone or in combination, and with results of specific tests to assay circulation, will establish the need to amputate. When all available information poses a serious question as to the possibility of limb salvage by reconstructive surgery rather than amputation, it has been common practice to attempt such surgery, even though extensive. Before questionable extensive reconstructive arterial surgery is carried out, the surgeon should consider critically the overriding probability of its failure with mandatory subsequent amputation. Will the proposed surgery compromise the level of amputa-

tion? Will amputee rehabilitation be additionally complicated by further deterioration of general health incident to the extensive surgical attempt at limb salvage? On a number of occasions, below-knee amputations have been performed in ischemic patients who were being considered for possible vascular surgical treatment but in whom, after review of all available information, such surgery might well have damaged the existing blood supply to a degree that an above-knee amputation would then have been required. It is important that the responsible surgeon understand the great rehabilitation value of the knee and weigh all facts relevant to the rehabilitation potential.

There is no single test or combination of tests now available that will demonstrate specifically the lowest effective amputation level. Successful below-knee amputations have been obtained repeatedly in patients whose arteriograms indicated complete occlusion of the superficial femoral artery.

A careful physical examination is the first requisite in determination of the level of amputation. Appearance of the soft tissues, temperature of the skin, the presence or absence of edema after elevation, growth of hair, level of sensation and acuity, together with palpation of pulses, are all important and cannot be supplanted by laboratory data. Arteriography, plethysmography, thermography, and a number of other objective techniques are useful. These include skin mapping with interarterial fluorescein, the use of radioactive Xenon #133, and transcutaneous ultrasonic Doppler recordings. Each adds to the available information and assists in level determination. Old established guidelines for determining amputation level are not valid when weighed against recent experience.

Unless it is *clearly evident* that a through-knee or above-knee amputation will be required, the surgeon should prepare the leg for both below-knee and above-knee amputation. Incisions through the skin and muscle preparatory to below-knee surgery can then be carried out quickly.

Bleeding and tissue viability can be observed directly and the final decision can now be made as to the level of amputation. Only a few minutes are added to the operative time should one elect the above-knee or through-knee level.

AMPUTATION TECHNIQUE FOR THE ISCHEMIC PATIENT

No tourniquet is used. The leg is draped free with the patient supine. Open and infected areas are walled off and shielded by sterile adherent plastic drapes prior to skin preparation. The level of amputation is 3-1/2 to 5 in. below the knee, *i.e.*, a short below-knee stump (Fig. 8). It has been recognized for many years that skin over the posterior leg has better blood supply than that anterior and anterolateral, and a long posterior and a short anterior skin flap are now used routinely. A long anterior flap, or even equal anterior and posterior flaps, should be avoided. The anterior scar resulting from use of a long posterior flap poses no problem in fitting the prosthesis. The modern total-contact below-knee prosthetic socket can accept a stump with scar placement in any position, provided it is nonadherent, well-healed, and nontender, and it is now standard policy in the Prosthetics Research Study to place the scar wherever it will heal most advantageously.

The anterior skin flap is fashioned approximately at the level of anticipated tibial section. The posterior flap must then be 5 to 6 in. longer to provide proper skin coverage without undue tension (Fig. 9).

After outlining the skin flaps, dissection is carried down through the deep fascia to the tibia. The periosteum is incised and stripped proximally 1 in. The anterolateral muscles are divided down to the intermuscular septum; blood vessels and nerves are ligated appropriately and severed; and then the tibia and fibula are sectioned, preferably with a power saw. The fibula is cut no more than 3/8 to 1/2 in. above the level of the tibia. Soft tissues are dissected from the posterior aspect of

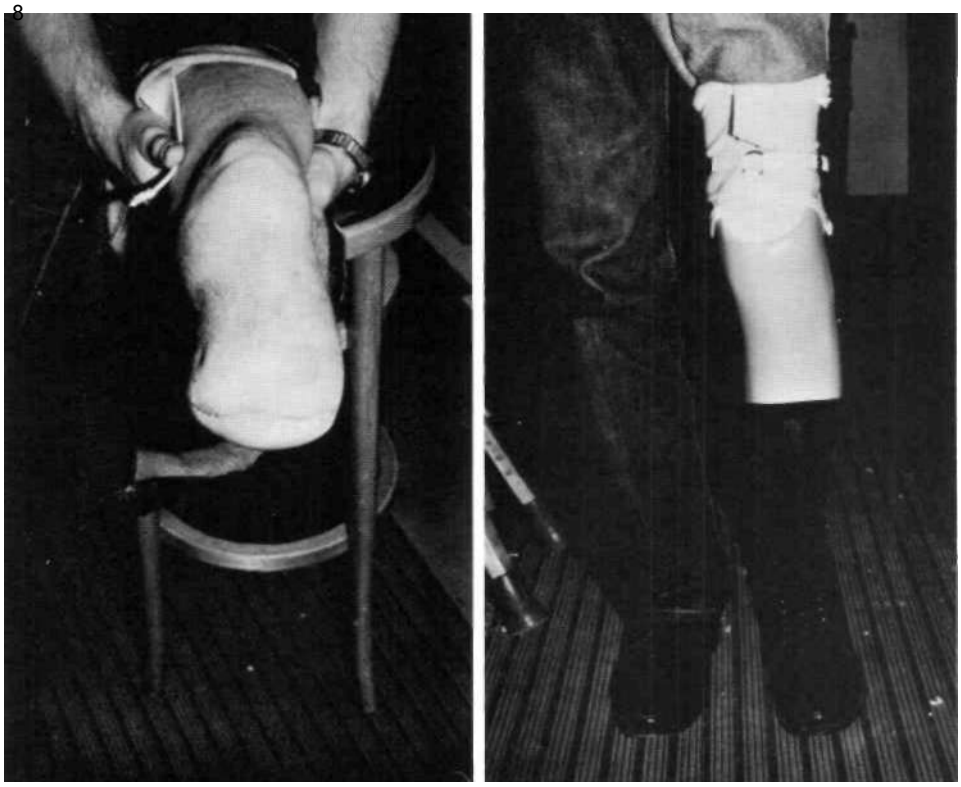

Fig. 8. *Left,* stump of 33-year-old patient on 26th day after amputation because of infection owing to nonunion of the tibia. *Right,* permanent prosthesis provided same patient on 26th day postoperative.

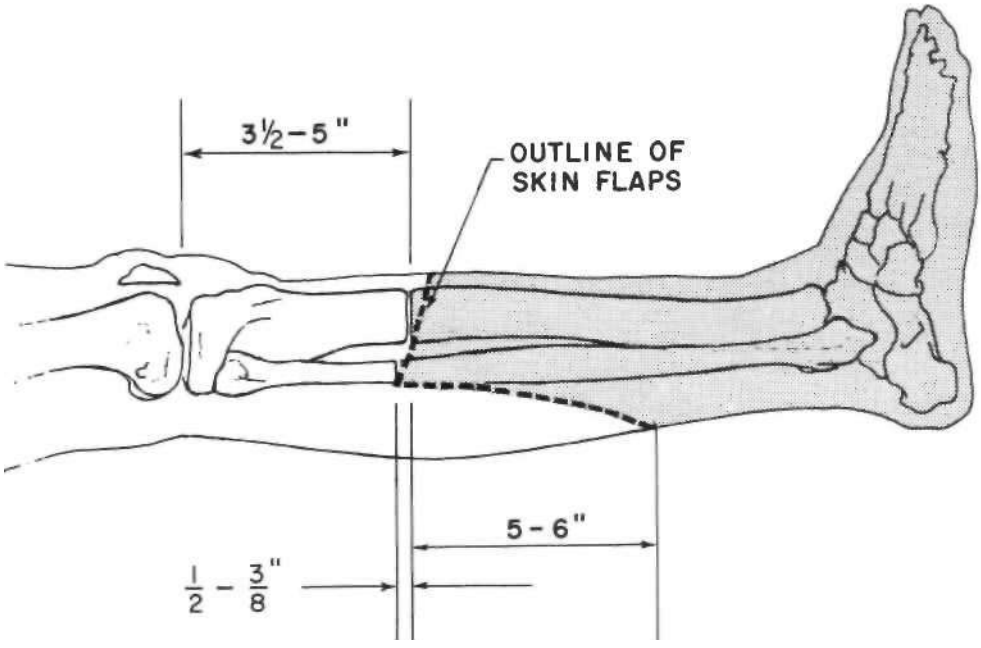

Fig. 9. Outline of skin flaps for below-knee amputation on typical ischemic patient,

the tibia and fibula down to the level of the posterior transverse division of skin. The leg is then separated and removed. The tibia is very carefully rounded with a short bevel over its anterior and medial aspects. It is important that no rough bone areas or ridges remain. A long bevel is specifically avoided. Nerves are pulled down and sectioned high with a sharp knife. They are not injected, crushed, or cauterized. The major nerves are ligated with a fine suture just above the site of division before the division is made. Encircling suture controls oozing from the blood supply that accompanies the nerve, and it also appears to localize neuroma formation and to lessen overgrowth and adherence to adjacent structures. The posterior muscle mass consisting of the gastrocnemius-soleus and deep flexor group is now beveled and tailored to permit the entire muscle flap to come forward and be sewn anteriorly to the deep fascia of the anterolateral muscle group and to the reflected periosteum over the anterior tibia. Contouring and trimming of the gastrocnemius medially and laterally gives a smooth musculofascial flap stabilized over the end of the bones. The skin is then brought up and closed without subcutaneous suture (Fig. 10). Medial and lateral "dog ears" are contoured moderately. They should not be taken back sufficiently to disturb skin circulation. The immediate postsurgical socket rapidly shapes the stump including moderate skin irregularity at the medial and lateral angles. The wound is drained deep to the muscle flap, *i.e.,* to bone. Through-and-through drain or suction drainage may be used. An immediate postsurgical rigid dressing and prosthesis are then applied.

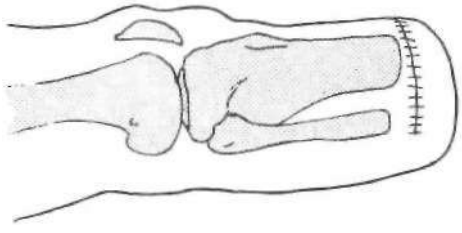

Fig. 10. Below-knee stump of typical ischemic patient showing position of suture line.

POSTSURGICAL CARE

Drains are removed 48 hours after surgery. If the patient's general condition permits, ambulation with guarded weight-bearing is begun 24 to 48 hours following surgery. The advantages of upright activity with limited stance and gait are obvious. However, only touch-down weight-bearing not exceeding 25 lb. is allowed until the initial cast is changed. Personnel in charge of the patient should be instructed carefully as to their responsibility in preventing the patient from bearing excessive weight or from falling.

The postsurgical management with an immediate prosthesis has resulted in much less pain than previously encountered. Postoperative pain is generally of a diffuse aching type. Complaint of localized pain almost always indicates abnormal pressure and requires inspection of the stump and change of the socket. Unless complications develop, *i.e.,* evidence of infection, excessive loosening of the socket, or severe pain, the initial rigid dressing should be left intact until the time of anticipated suture removal, usually two to two-and-one-half weeks following surgery. The cast is then removed, with the patient under sedation but not anesthesia, the wound is inspected, sutures are removed if indicated, and a new temporary prosthesis is applied. By this time the patient is usually ready for unsupported crutch ambulation and discharge from the hospital. A temporary prosthesis is worn continuously until a definitive limb is provided. Ordinarily the final limb can be fabricated, fitted, and worn four to five weeks following below-knee amputation. Typical ischemic patients are shown in Figures 11 and 12.

Necrosis of skin flaps can result from either inadequate blood supply or undue pressure. If the level of amputation is so low that the blood supply is insufficient to support a below-knee amputation, it will be evident at the initial cast change. The decision then to amputate at a higher

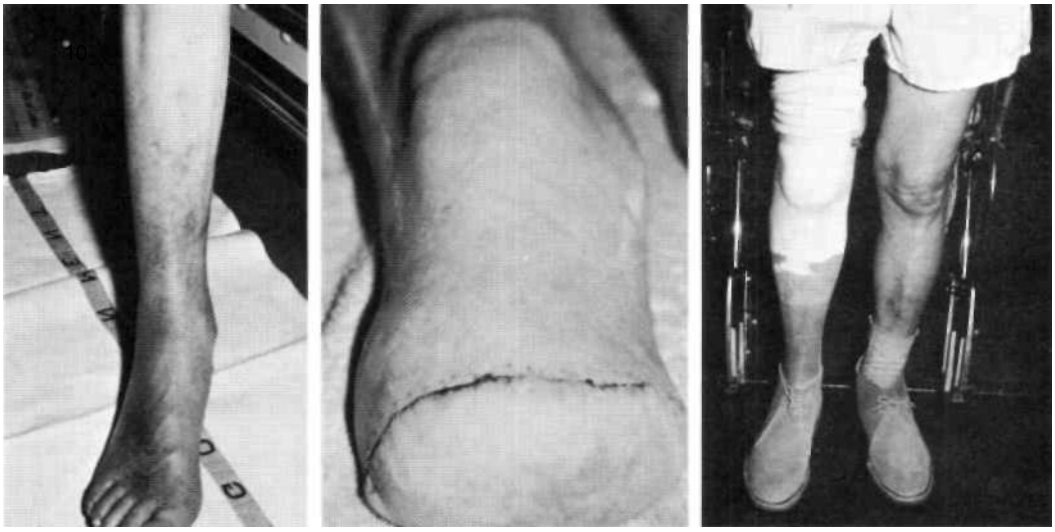

Fig. 11. A 69-year-old, white male had multiple difficulties consisting of arteriosclerosis obliterans with complete right superficial femoral occlusion, diabetes mellitus, arteriosclerotic heart disease with mitral insufficiency, and coronary occlusion. No reconstructive vascular surgery was considered to be feasible. The preoperative condition of his foot is indicated on the *left.* Good stump healing was achieved by the 25th postoperative day, *center.* The definitive prosthesis was applied on the 28th postoperative day, *right.*

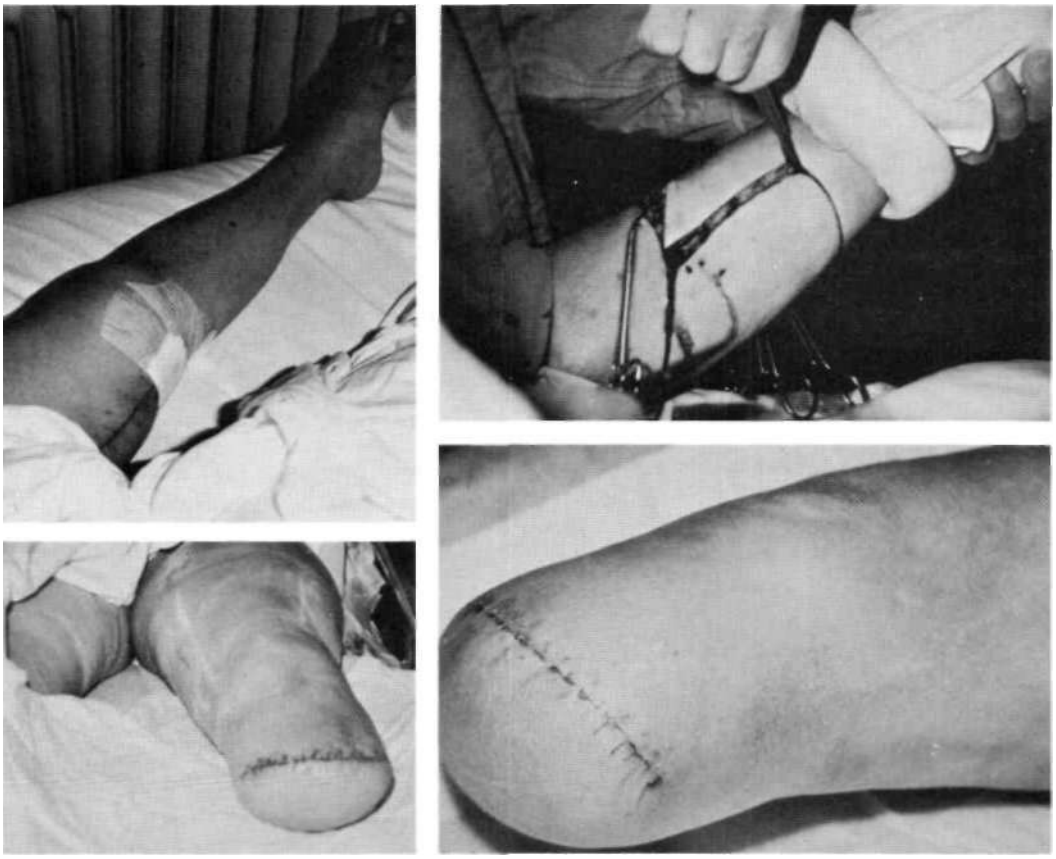

Fig. 12. A 73-year-old, white female with severe chronic peripheral vascular disease without diabetes. Two attempts at femoral popliteal bypass graft had been made in the three weeks prior to "breakdown" of the graft operative sites. Progressive gangrene of the foot had ensued with demarcation just above the ankle level. Figure in *upper left* shows the appearance of the leg prior to amputation. A short below-knee level of amputation was selected and a long posterior musculocutaneous flap developed, *upper right.* The appearance of the below-knee stump at 19 and 29 days following surgery is indicated in the *lower* figures. The definitive prosthesis was fitted on the 32nd postoperative day.

level should be made promptly. Re-amputation rate in the PRS series to through-knee or above-knee over the four-year period has been 9.4 per cent. As experience and techniques have improved, the re-amputation rate for below-knee cases with ischemia has continued to decrease. The surgeon, of course, likes to avoid all re-amputations. However, salvage of the knee is of such paramount importance that an occasional re-amputation may be required if we are to save all knee joints possible in view of our inadequate means for determining the best level for amputation.

SUMMARY AND CONCLUSIONS

Below-knee amputation is statistically by far the most important major amputation used today. The vast majority of major lower-extremity amputations performed for ischemia will heal primarily and remain healed at below-knee level. The below-knee amputation for ischemia is short in length, the posterior skin and myofascial flaps are fashioned long, and the technique is precise. The resulting stump is cylindrical in shape, well-padded, comfortable, and easily fitted with modern below-knee prostheses of the total-contact type. An immediate postsurgical prosthesis is an integral part of the over-all below-knee amputee management in both the ischemic and nonischemic patient. Restoration of function and rehabilitation of the below-knee amputee, both unilateral and bilateral, have improved in almost spectacular fashion when the guidelines and management which have been outlined are followed.

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Clinical Evaluation of the Engen Plastic Hand Orthosis

Hector W. Kay, M ED.[1]

A PRIMARY function of the hand is prehension, the ability to grasp an object. While the hand can perform numerous types of grasp, of major importance is the type involving flexion of the index and middle fingers towards or against the opposing thumb to provide what is sometimes referred to as "three-jaw-chuck" prehension.

Temporary or permanent paralysis can impair or completely inhibit the function of hand, wrist, or entire upper extremity, and the ability to oppose the thumb to the flexing fingers may be lost. In these instances, various types of orthotic systems have been designed to achieve the goals of prevention or correction of deformities, or restoration of function, or both. A key feature of these systems is the stabilization of the thumb in opposition to the fingers.

Pioneering efforts in the area of hand-splinting were undertaken at the Georgia Warm Springs Foundation where many types of assistive devices were developed to meet the needs of a large patient population having residuals of poliomyelitis. Although the number of polio patients has decreased in recent years, rehabilitative medicine has expanded to include patients with many other types of neuromuscular and skeletal disorders. A systematic method of hand splinting to meet the needs of these patients has continued to be of paramount importance. On-going efforts in this regard have been maintained not only at GWSF but also at Rancho Los Amigos Hospital and other institutions *(1,2).*

As part of Research Project VRA RD-1564, Thorkild J. Engen, Project Director, Baylor University College of Medicine, Houston, Texas, in 1959 initiated the development of a plastic hand orthosis having the basic configuration shown in Figure 1 (3). Based on the premise that preservation of hand posture is best maintained by support, rather than suspension, the device is designed to hold the thumb in the opposed position and simultaneously support the metacarpal arch. The aim has been to develop a standardized item shaped to conform to the natural contours of the hand which could then be adapted to meet individual needs. The Engen orthosis is made in four sizes: large, medium-large, medium, and small; and for both right and left hands. Because the orthosis is fabricated of polyester resins, it can be remolded upon application of heat.

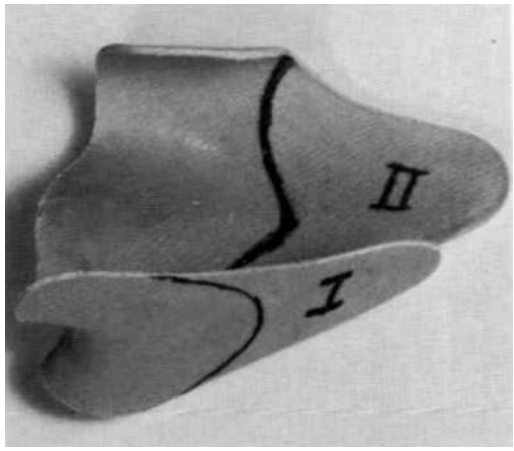

Fig. 1. Basic Engen plastic hand orthosis being prepared for individual application.

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In the early stages of redevelopment, the Engen orthoses were fabricated of epoxy resins with and without fiberglass reinforcement. Ultimately these models were discarded because of breakage problems. The plastic shells originally submitted to New York University for a laboratory evaluation program were made of fiberglass and polyester resins *(9).* The current shell is a polyester resin and nylon laminate prepared by means of a vacuum-molding technique. With the new materials, the fitting technique is essentially unchanged; the orthosis is molded and modified by the orthotist as necessary to provide a custom fit.

In the course of development, attachments were devised or adapted to provide wrist support and to provide prehension.

Three versions or adaptations of the Engen plastic hand orthosis were selected as the subject of the field evaluation: the short opponens orthosis, the long opponens orthosis, and the reciprocal wrist-extension, finger-flexion unit. Additional modifications of the basic concept involving the use of external power were specifically not included in the study.

SHORT OPPONENS ORTHOSIS

The so-called short opponens orthosis is the simplest application or adaptation of the Engen equipment *(3).* It consists essentially of the basic hand shell with a retaining strap (Fig. 2). The prime purpose of this device is to maintain the thumb in apposition to the index and long fingers and to support the metacarpal arch. The functional goal is the achievement of "three-jaw-chuck" prehension as distinct from "lateral" grasp. Patients said to benefit from this orthosis are those with neuromuscular disorders resulting in various degrees of muscle imbalance of the intrinsic and opponens muscle groups. Such patients would typically have spinal cord injuries at the C-7, C-8, and T-l levels, peripheral neuropathy (ulnar and median nerves), or hemiplegia.

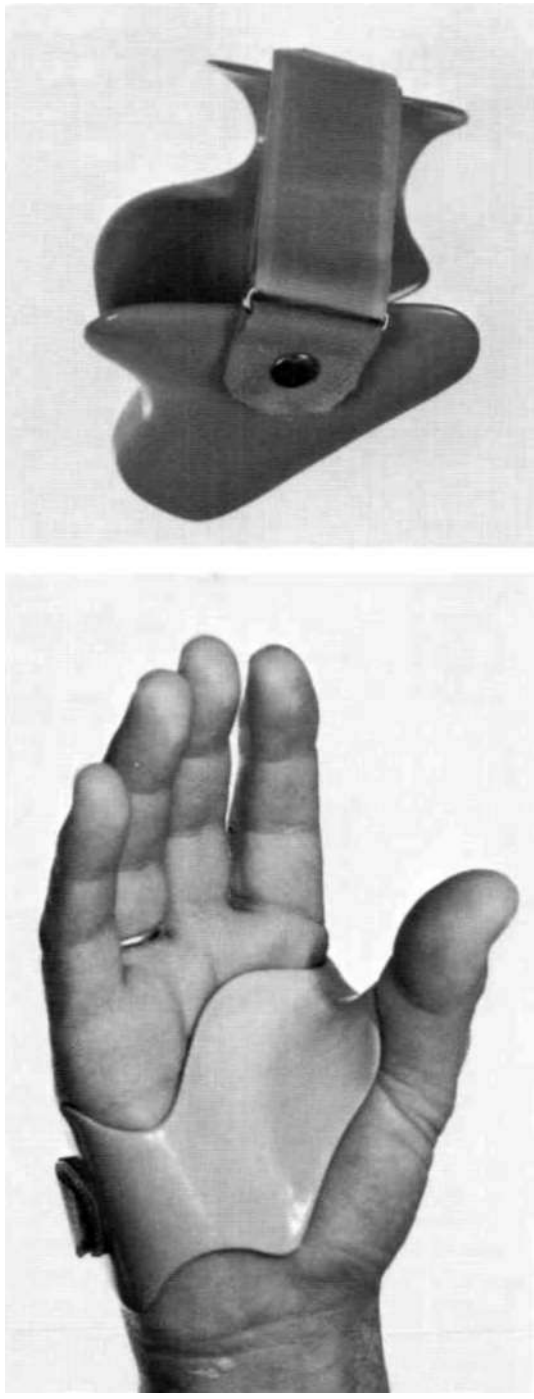


Fig. 2. Two views of the short opponens orthosis.

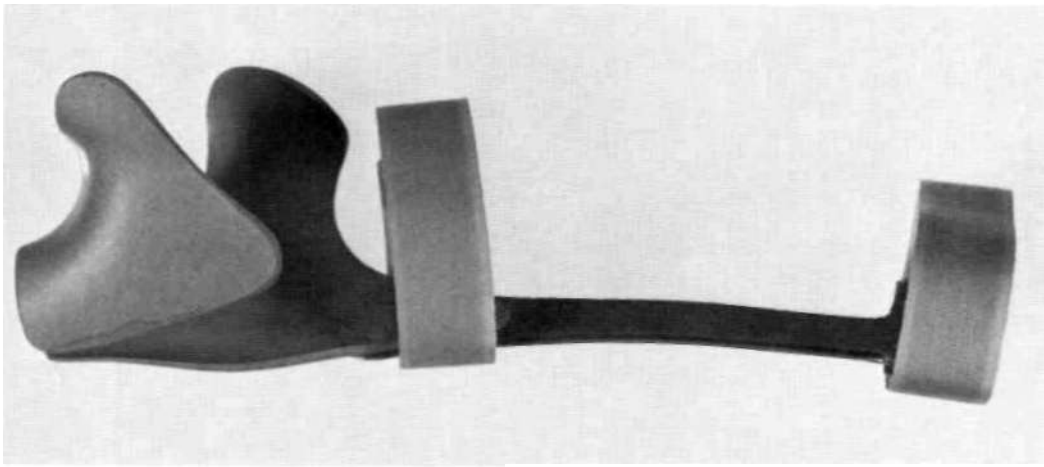
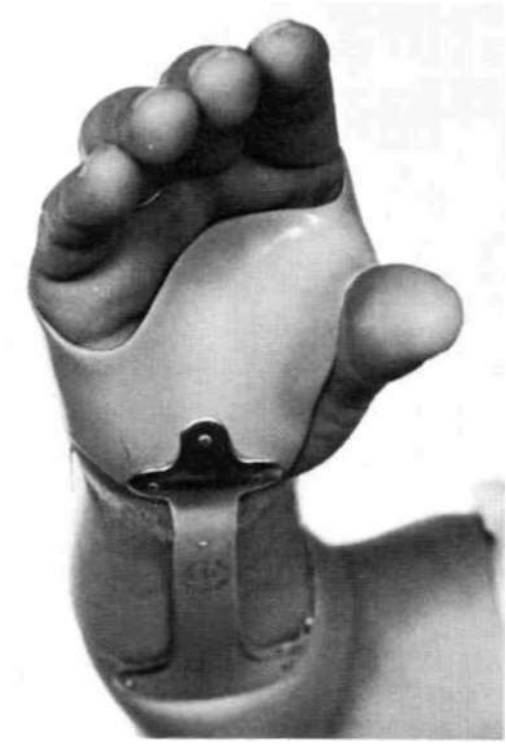


Fig. 3. Two views of the long opponens orthosis.



LONG OPPONENS ORTHOSIS

This adaptation (3) consists essentially of the basic plastic hand shell with an attached extension arm which is stabilized on the forearm by appropriate straps (Fig. 3). Like the short opponens orthosis, this device is designed to prevent deformity and achieve "three-jaw-chuck" prehension if the necessary residual muscle movements are present and can be controlled. Patients with spinal lesions at the C-5, C-6 levels, peripheral neuropathy involving the median or ulnar nerves, or both, and the radial nerve, or hemiplegia, are said to be suitable candidates for this device.

RECIPROCAL WRIST-EXTENSION FINGER-FLEXION ORTHOSIS

This adaptation, which is the most complex of those studied, is designed to provide prehension when voluntary wrist-extension power is available (Fig. 4). Quadriplegic patients who retained innervation to the wrist-extensor muscles are said to be appropriate subjects for this type of functional orthosis.

PROCEDURES

PARTICIPATING CLINICS AND PERSONNEL

As an initial step in the activation of the proposed field study, the Committee on Prosthetics Research and Development, through its staff and Subcommittee on Evaluation, selected five treatment centers known to be active and interested

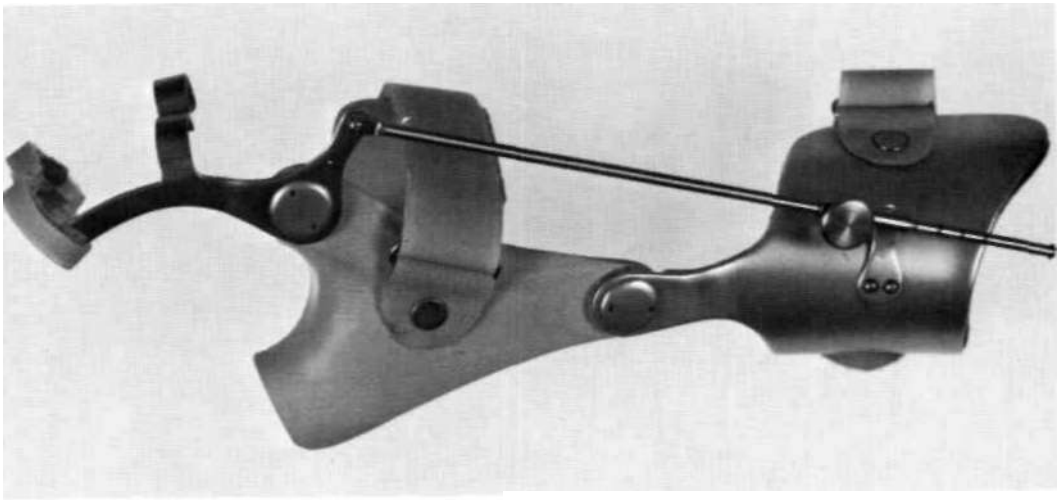
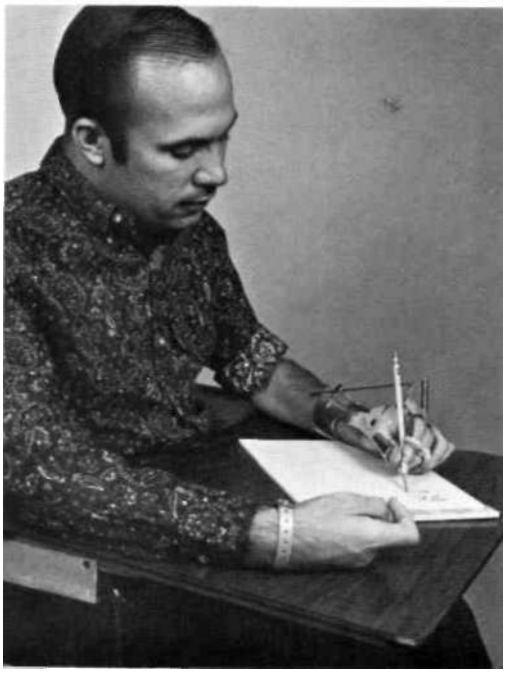


Fig. 4. Two views of the reciprocal orthosis.



in the application of hand splints. These clinics were approached and each agreed to participate in the study. The institutions and personnel involved were:

1. Duke University Medical Center, Durham, N. C. (Frank W. Clippinger, Jr., M.D.; Bert R. Titus; Felton Elliott).
2. Georgia Warm Springs Foundation, Warm Springs, Ga. (Edward Haak, M.D.; H. G. Bowden).
3. Highland View Hospital,[2] Cleveland, Ohio (Alvin A. Freehafer, M.D.; Arthur Guilford, Jr., G. A. Guilford and Sons).
4. Ohio State University, Columbus, Ohio (Marvin H. Spiegel, M.D.; Lawrence Czap; Charles W. Rosenquist, Columbus Orthopaedic Appliance Co.).
5. Veterans Administration Hospital, Hines, Ill. (James F. Kurtz, M.D.; Vladimir T. Liberson, M.D.; Walter J. Piotrowicz, CO.).

INSTRUCTION IN FABRICATION PROCEDURES

The study of the Engen devices was initiated by an instructional course in the three applications to be evaluated. This course was conducted by the developer and his staff at the Texas Institute for Rehabilitation and Research, Houston, Tex., from Dec. 5 to 8, 1966 (orthotists, four days; physicians, one day). Instructional material and fitting check lists were prepared by the developer *(5,6,7,8),* and used as the basis for the course. A special training session for Mr. Sigars was conducted December 4-6, 1967, after he joined the Rancho Los Amigos Hospital team.

[2] Unfortunately, the Highland View Hospital team had to withdraw prior to the commencement of the study. It was replaced by a team from Rancho Los Amigos Hospital consisting of E. Shannon Stauffer, M.D., and Dale Fries, orthotist. In the course of the study, Mr. Fries transferred to another position and was replaced by Mr. Charles Sigars.

THE STUDY PLAN

Concurrent with the recruitment and training of participating clinic personnel, the CPRD staff, in collaboration with the developer, and under the guidance of its Subcommittee on Evaluation, prepared the schedule and data-recording forms for the study (8).

Essentially, each clinic was requested to seek patients appropriate for applications of the Engen devices. Data related to the fittings would be recorded on the forms developed by the Committee on Prosthetics Research and Development. Each patient fitted was to be followed for a period of 12 months unless treatment was terminated prior to that time. The CPRD staff was to provide liaison with the field clinics as necessary during the course of the study.

RESULTS

TECHNIQUE TRANSFERABILITY

With a new fabrication or fitting technique which is said to yield excellent results in the hands of the developer, an important consideration is whether or not the skill and "know-how" involved in the applications can be successfully transferred to others.

In the present study the means of achieving this transfer were: (1) Written instructional material prepared by the developer; (2) A course of instruction which included practice in the fabrication of devices; and (3) Follow-up visits made by the developer to each participating facility. Problems encountered locally were analyzed and supplementary instruction given.

It was the consensus of the evaluation team as well as that of the participants that the fabrication techniques for the three EPHO adaptations under study were successfully transmitted by these procedures. Moreover, while the orthotists participating in the evaluation were selected and highly skilled, indications were that less skilled technicians could be satisfactorily taught by the same methods.

PATIENT FITTINGS

The Sample

During the period of the evaluation program, 22 patients were fitted with the Engen Plastic Hand Orthosis. Distribution in terms of the three adaptations under study were: short opponens orthosis, 7; long opponens orthosis, 3; and reciprocal units, 12.

Moreover, data was available on an additional 48 patients distributed as follows: short opponens orthosis, 11; long opponens orthosis, 7; and wrist-driven reciprocal units, 30. These patients were fitted at Hines VA Hospital following the closure of the official phase of the study. Some findings of interest from these additional fittings are included.

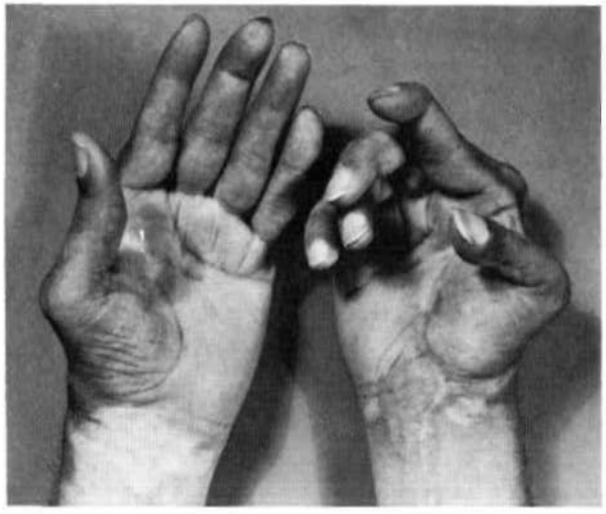
In the total of 70 fittings reported, 18 were with short opponens, 10 with long opponens, and 42 with reciprocal units, roughly a 2:1:4 ratio. Whether this ratio could be extrapolated to the general population is not known.

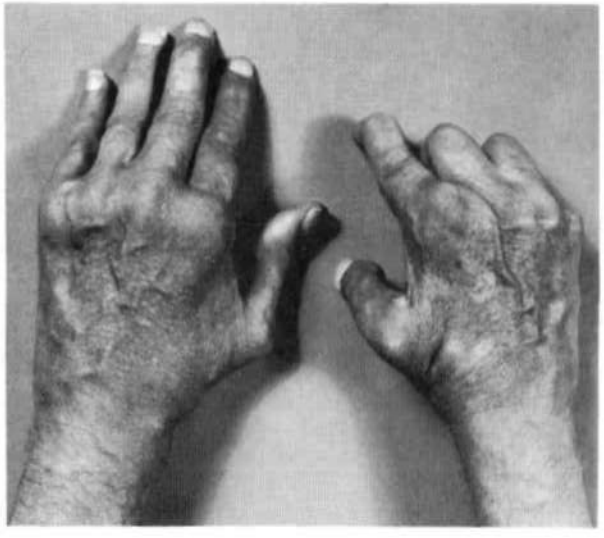
Typical conditions for which the three versions of the EPHO[3] were applied were: (1) short opponens orthosis: rheumatoid arthritis of the hands (Fig. 5); quadriplegia (to prevent deformities and support the hand in a position of function pending fitting of reciprocal units); contraction deformity of the wrist; (2) long opponens orthosis: quadriplegia (as a stabilizing device pending reduction of contractures and fitting with a reciprocal unit) (Fig. 6); or as a base for the addition of self-help devices (Fig. 7); reciprocal units: quadriplegia (Fig. 8).

Outcomes

Results of the fittings in the five participating clinics were variable, success or failure being related primarily to three factors:

[3] Utilizing the basic Engen items as modules to which accessory equipment was added if indicated by the patient's needs.





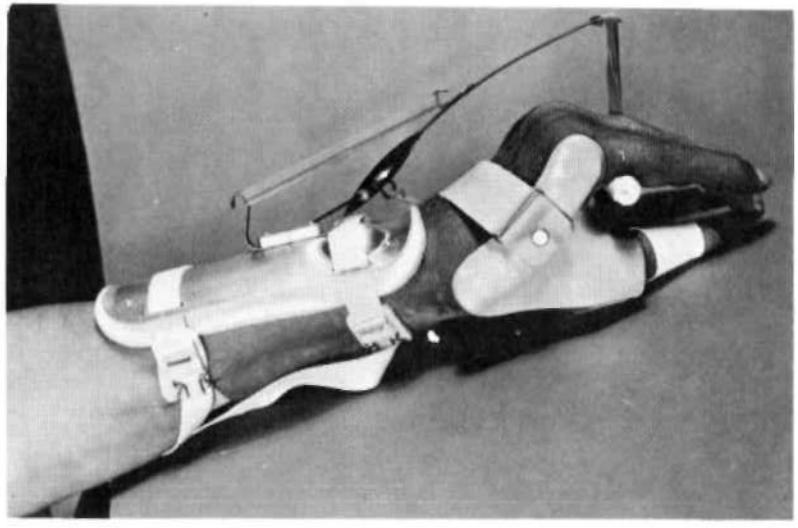


Fig. 5. *Left*, palmar and dorsal views of patient's arthritic hands. *Above*, left hand fitted with Engen short opponens orthosis and Thomas outrigger splint.

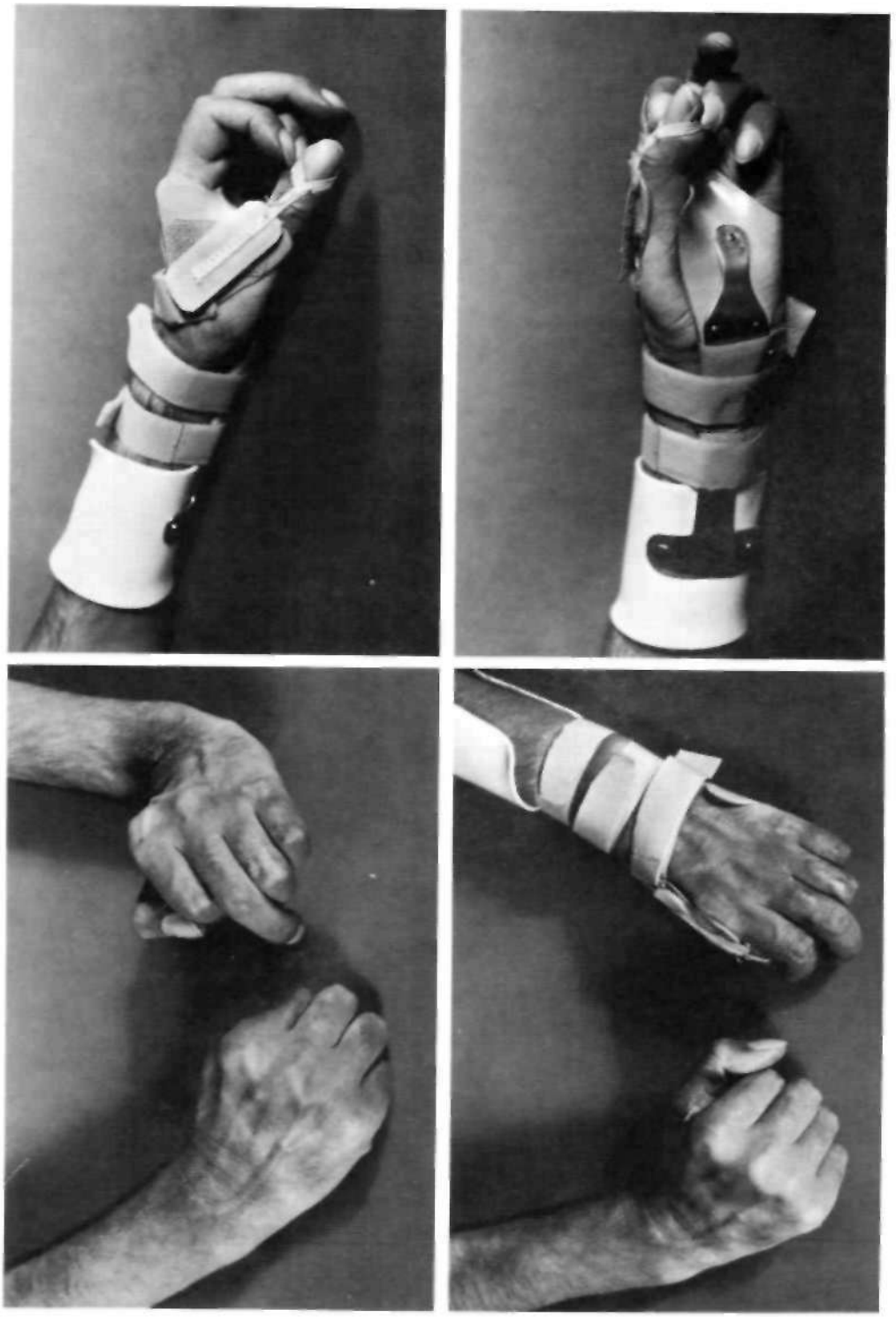


Fig. 6. Patient fitted with Engen long opponens orthosis as stabilizing device.

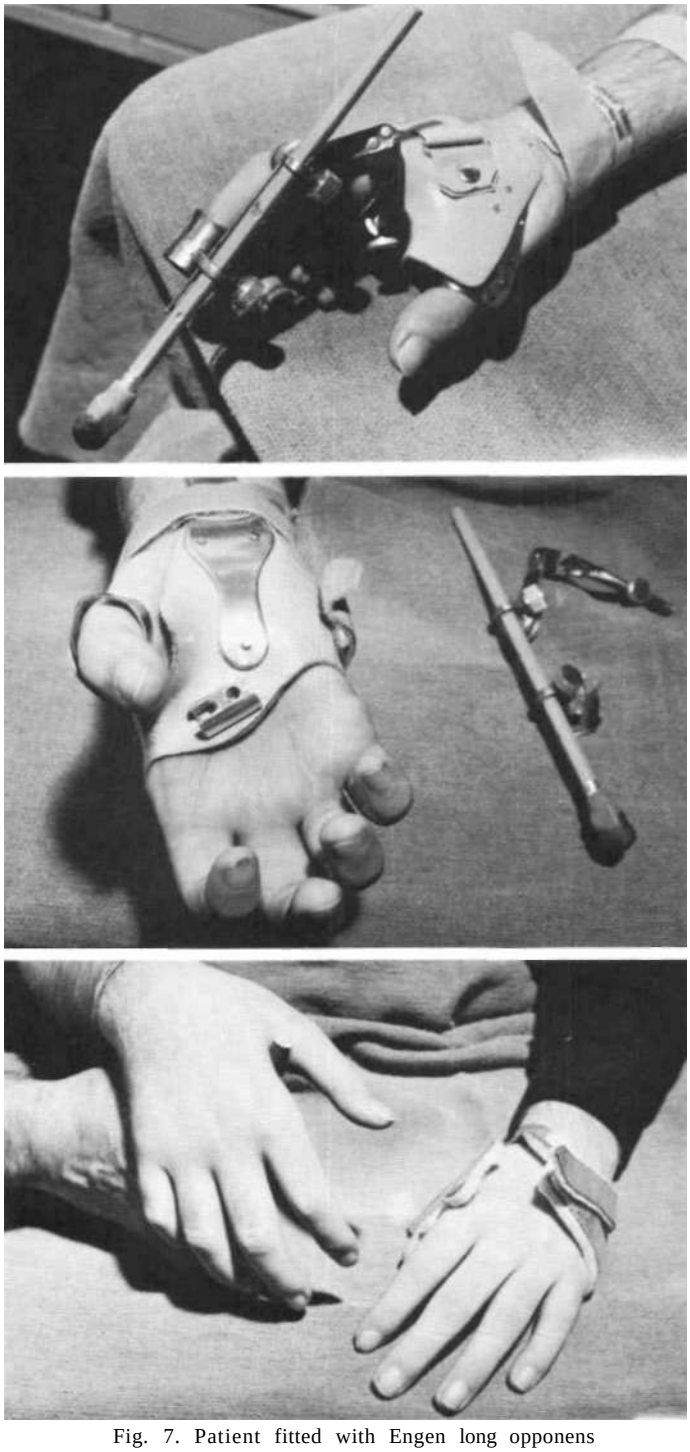


Fig. 7. Patient fitted with Engen long opponens orthosis with attachment for self-help devices. Note atrophy of thenar cleft.

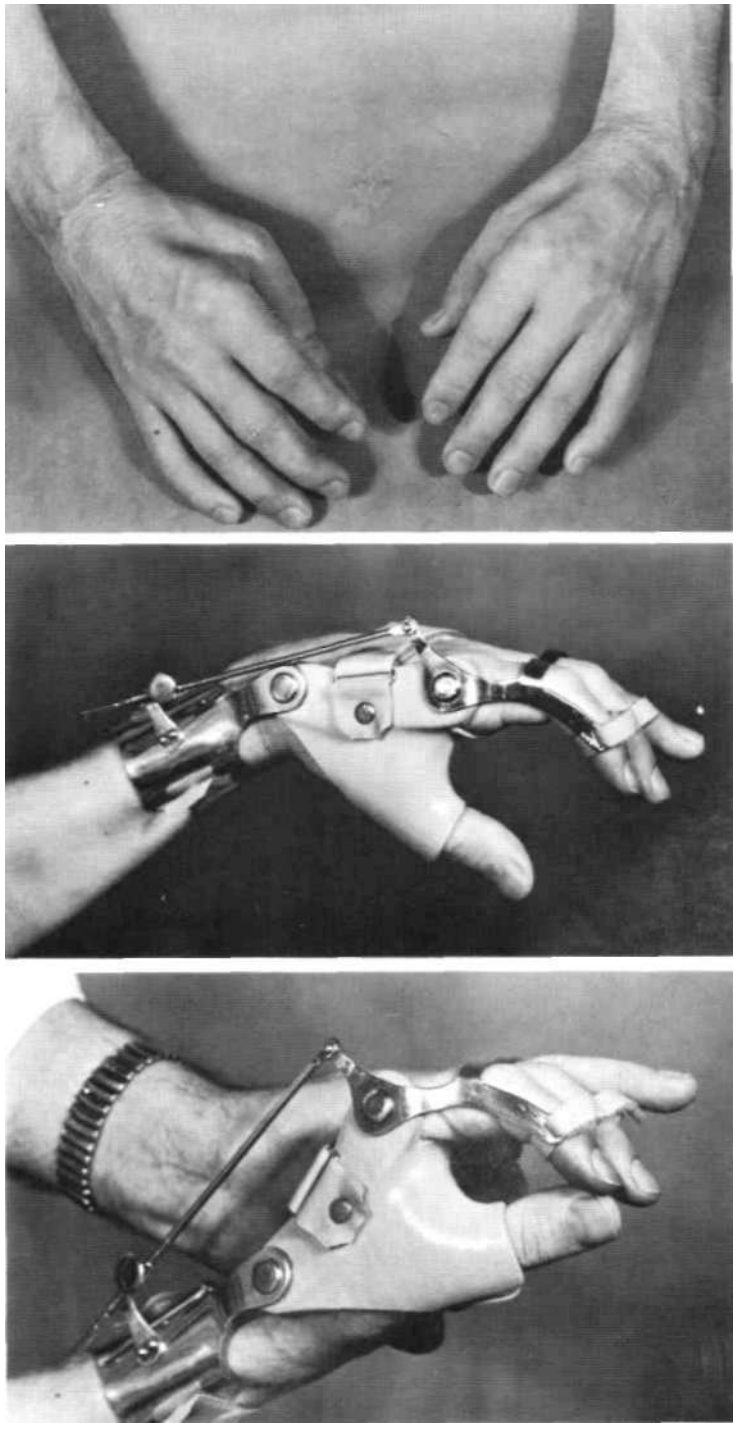


Fig. 8. Patient fitted with reciprocal unit.

1. Proper selection of patients. In several of the clinics patients were selected under somewhat experimental circumstances, that is, either the motivation of the patients was less than optimal or the anticipated benefit to be derived from the Engen device was marginal. In these instances, the fittings typically proved to be failures.

2. Objectivity in the evaluation of outcomes. Two of the clinics participating in the study had devices of their own design which were "competitive" with the Engen items. Personnel of these clinics were of the opinion that the Engen devices provided no features superior to their own devices other than perhaps the telescoping rod on the reciprocal unit application.

3. Meticulous care in application and follow-up. Although the Engen Plastic Hand Orthosis is essentially a prefabricated shelf item, it must be carefully tailored to the needs of the individual patient. This tailoring may involve: (a) some reshaping of the plastic shell to accommodate atrophy or size discrepancy in the patient's hand; (b) the addition of accessory finger pieces and other equipment to the basic Engen shell.

Moreover, since the condition of the patient's hand changes with time and with the use of the Engen splint, follow-up to maintain fit of the device is essential. This follow-up is obviously best accomplished when the patient is being treated on an in-patient basis, in-house orthotic facilities are available, and there is close cooperation between the disciplines involved in the care of the patient.

Where the foregoing conditions were satisfactorily met, excellent success was achieved in the fittings of the Engen devices. Selected cases which illustrate the applications and outcomes of the three EPHO modifications under study are presented below.

CASE PRESENTATIONS

SHORT OPPONENS ORTHOSIS

Case No. 1

A. M. was a 40-year-old male with a diagnosis of quadriplegia resulting from a physiologically incomplete lesion of the spinal cord at the C-5 level. A short opponens orthosis was prescribed for his right, dominant hand with a view to aiding in the restoration of function, and the prevention and correction of deformities. It was hoped that eventually Mr. M. would be a candidate for a right reciprocal unit. The patient was described as having a motivational level of fair and a tolerance to pain that was average.

Mr. M. was fitted with a medium-sized orthosis. The suitability of the preformed size and shape was rated as good and the ease of customizing and the clarity and completeness of the instructions for doing so were also rated as good. No special modifications of the shell were necessary for this patient.

A. M. was reevaluated at 1, 3, 6, 9, and 12 months following the initial fitting. The efficacy of the splint in achieving the objectives of the fitting was rated as good in all respects. The patient's performance in such activities as turning pages in a book and writing was rated as fair. The performance in feeding and using a toothbrush was cited as being poor. The patient's reactions to the orthosis were good with respect to fit, comfort, and cosmesis, and fair as regards function. During the course of his treatment the patient was given physical and occupational therapy and special instruction in the use of the Engen device. He was also given medication for spasticity which did not involve the hands.

The evaluation of the device with regard to this patient remained remarkably consistent throughout the entire 12 months of the test period except that the patient's own reactions to the functional assistance provided by the device declined from fair to poor from the third month on.

The outcome in this instance was considered to be excellent, but two other patients, D. R. and J. A., whose initial conditions were remarkably similar, withdrew from the study one and four months, respectively, after the initial fitting. In these two instances the restoration of function achieved with the orthosis was minimal and this factor, combined with low levels of motivation, resulted in the withdrawals.

Case No. 2

Patient N. E. was a 60-year-old male with a diagnosis of rheumatoid arthritis of some eight years' duration. He was prescribed an EPHO short opponens orthosis for his right, dominant hand, the

objectives being assistance in the restoration of function and the prevention and correction of deformities. His tolerance to pain was described as average, and his skin condition as thin, and his motivational level was said to be good.

N. E. was fitted with the large-sized EPHO shell. With regard to the fitting, the suitability of the preform size and shape was rated as good, as were the ease of customizing and the clarity and completeness of instructions. No special modification was necessary initially, but some five weeks later a Thomas outrigger suspension was applied to prevent further subluxation of the metacarpophalangeal (MCP) joints (Fig. 5). Mr. E. was reevaluated at 1, 3, 6, 9, and 12 months following fitting and then left the clinic area taking the provided splint with him.

Initially the achievement of objectives involving the prevention and correction of deformities was rated as good, but the restoration of function as poor. Mr. E.'s performance in typical activities of daily living were all rated as poor. The patient's reactions to the device were good with respect to fit, comfort, and cosmesis, but poor as regards function.

As Mr. E. continued to wear the experimental device his ratings in all performance activities were raised to fair, and finally to good in such activities as page-turning, writing, and feeding. The patient's rating of the functionality of the device gradually improved until finally it was reported as good.

In this fitting the outcomes appeared to be positive from the beginning with respect to the prevention and correction of deformities with gradually increasing benefit in the area of function.

LONG OPPONENS ORTHOSIS

Case No. 3

Patient J. K. was a 21-year-old male. His primary diagnosis was quadriplegia with a spinal-cord injury at the C-5, C-6 levels which was incurred some nine months prior to his inclusion in the evaluation program. He was fitted with an EPHO long opponens orthosis, medium-size, to the right hand which was less impaired than the left. His hands were atrophied, especially in the thenar-cleft area, and he had a slight lateral palmar drift on the (right) hand fitted. The patient's motivational level was said to be good and his pain tolerance average. The objectives of the fitting were restoration of function, and prevention and correction of deformities in the hope that he might eventually be fitted with a reciprocal orthosis.

The application of the device proceeded without difficulty except that the device was somewhat too large for the patient's atrophied thenar-cleft area. The splint tended to displace itself into this area. Three weeks after the initial fitting a reduction in the cock-up angulation was recommended by the developer, together with the addition of a T-bar to abduct the thumb and a dorsal strap for better retention.

The patient preferred the EPHO splint to his previously worn Royalite device and requested that the EPHO be modified to include the self-aid attachments worn on the earlier splint. The device was subsequently reinforced with a Monel metal piece and has held up well since that time. The patient's flexed lateral palmar drift was held in proper position by the orthosis.

At the one-month follow-up of this patient the ratings of outcomes were generally poor to fair with only the patient's reaction to the cosmesis of the device being designated as good. However, steady improvement occurred throughout the follow-up period, and by 9 months after initial fitting the device was rated as good in all characteristics specified in the evaluation program. Thus, in this instance, the outcomes of fitting the Engen plastic hand orthosis must be considered as excellent.

Case No. 4

On another patient, F. G., with a somewhat similar disability, the results of the

fitting were considerably less positive. This patient was a 40-year-old male with complete transverse severance of the spinal cord at the C-6, C-7 levels. The injury to this patient had occurred some six and a half years prior to the present study and he had had a surgical transfer of the brachioradialis tendon to the wrist extensors on his left hand several years previously. The hand tended to go into marked radial deviation on voluntary extension of the wrist. He could raise his elbows and shoulders bilaterally. He had muscle spasms.

F. G. was fitted with a medium-sized long opponens orthosis and it was immediately noticeable that the splint would not hold the patient's marked radial deviation. At the developer's suggestion the cock-up angle of the splint was reduced to prevent creeping and a plastic clip added on the proximal medial side. A lateral Velcro strap was added to pull the ulnar side of the wrist toward the radial side, and an elastic sling was added to correct the flexion of the interphalangeal (IP) joint of the thumb. The patient was to be considered for a reciprocal orthosis if his contractures could be reduced. The patient's motivational level was rated as poor with respect to any type of splinting.

The outcomes of this fitting initially were also mixed and failed to show appreciable improvement, particularly with regard to function, over a 6-month follow-up period. The patient was then taken off the program at his own request.

RECIPROCAL WRIST-EXTENSION FINGER-FLEXION ORTHOSIS

Case No. 5

Patient V. C. was a 42-year-old male who had sustained a spinal-cord injury at age 26. His primary diagnosis was "dislocation and compression of the spinal cord at the C-5, C-6 levels with complete paralysis." With no prior experience with orthotic devices, he was fitted with a reciprocal unit on his right, dominant hand. His motivational level was rated as good, but his pain tolerance was given as low. The objectives of the fitting were restoration of function and prevention and correction of deformities.

The fitting utilized a large reciprocal orthosis and finger pieces but a medium-sized forearm piece. The component sizes were considered to be good for this patient. However, the shape of the plastic shell did not provide good support for the arch of the hand or conform well to the thenar-cleft area. A thumb sling and a middle-finger IP stabilizer were added. A later review of this case indicated that the MCP and the wrist joints were incorrectly placed. With these conditions the patient had no desire to try and use the splint and did not wish to keep it. Replacement of the malpositioned joints effected a marked improvement in the function of the device and the patient's acceptance of it. This high level of performance and acceptance was maintained throughout the remainder of the patient's 12-month participation in the study. In this case, obviously the difference between success and failure hinged on the proper joint positioning, emphasizing the importance of this aspect of the fitting. This type of experience was repeated with a number of other patients in the evaluation.

Case No. 6

Patient W. M. was a 47-year-old male who sustained a spinal-cord injury approximately one year prior to being fitted with the Engen orthosis. His diagnosis was given as "compression of cord, level C-5, C-6 incomplete, C-7 complete." Mr. M.'s motivational level was said to be good, but his pain tolerance was given as low. He was fitted with a reciprocal orthosis on his right, dominant hand, the objectives being restoration of function, and prevention and correction of deformities.

The initial application of the device seemed to proceed satisfactorily, the component parts being a large plastic shell, a

larger finger unit, and a large forearm piece. The sizes and shapes of the various components seemed to be appropriate. Three days later a "knuckle bender" was added because of tightness of the MCP joints and a modified Oppenheimer splint was fitted to increase the limited range of wrist extension and thumb abduction.

A later review of this case indicated that the joint hinges had been incorrectly positioned and this deficiency was corrected. Again a dramatic improvement in the achievement of fitting objectives, functional level and patient acceptance, was evident, although this subject's function was not as good as that of the previous patient. This case again illustrates the importance of joint positioning and indicates the use of the Engen basic equipment as a module to which other accessories might be added.

SUMMARY AND RECOMMENDATIONS

In the present study it would appear evident that orthotists with prior experience and skill in the fabrication of hand splints can be taught to apply the EPHO variations successfully. In this connection the instructional manual and fitting checkout sheets developed in conjunction with the field study provided an excellent basis for the transfer of techniques from developer to field orthotists. However, this written material is not regarded as an adequate substitute for direct person-to-person instruction. Moreover, a follow-up visit to each of the clinics following initial fittings helps to insure that the techniques taught are being properly applied and assists in the solution of specific local problems.

The outcomes of the field fittings of the Engen equipment were mixed, positive results being related primarily to three factors: (1) proper selection of patients, including consideration of motivational factors; (2) meticulous care in application and follow-up of the devices; and (3) objectivity in evaluating outcomes. Where these considerations were observed, the successful outcomes achieved support the developer's claims for the device.

Fitting results for each subject in the study showed no significant changes after 6 months' wear of the Engen device. Hence, consideration might be given to reducing the follow-up period in similar future studies from 12 to 6 months.

THE DEVICES

Prescription Criteria

The criteria for prescription of the Engen adaptations as set forth above were re-affirmed by the results of the field study. The following additional comments also emerged:

1. Short Opponens Orthosis
   a. has been found useful as a stabilizing splint in several instances of postsurgical management;
   b. has been used in providing patients with various self-help devices as attachments to the basic shell;
   c. with special modifications has been used in rheumatoid arthritic cases to help prevent ulnar and radial finger drift and align the fingers in proper position for finger prehension;
   d. has been used as the stabilizing splint pending evaluation for application of a reciprocal unit.
2. Long Opponens Splint with Extension Arm Support
   a. has also been utilized for the same applications as the short opponens orthosis above.

SPECIFIC FINDINGS

Specific findings relating to the design and applications of the EPHO devices were:

1. Although the Engen Plastic Hand Orthosis is ostensibly a prefabricated shelf item, it must be carefully tailored to the needs of the individual patient. This tailoring may involve:
   a. some reshaping of the plastic hand shell to accommodate atrophy or size discrepancy in the patient's hand;
   b. the addition of accessory finger pieces and other equipment to the basic Engen shell.
2. In the installation of the EPHO reciprocal orthosis, great care must be exercised in the location of the joint axes.
3. Since the condition of the patient's hand changes with use of the Engen splint, follow-up to maintain fit of the device is essential. This follow-up is best accomplished when the patient is being treated on an in-patient basis, in-house orthotic facilities are available, and there is close cooperation between the disciplines involved in patient care.

4. The telescopic rod feature of the reciprocal unit was frequently cited as a most significant new characteristic of this type of orthosis.
5. Although definitely related to the level of experience gained in the application of the EPHO devices, saving of the orthotist's time was a significant feature of the system.
6. Some deficiencies in the design and materials of the EPHO were noted:
   a. The range of three sizes provided initially were considered inadequate but the addition of the fourth (medium-large) size virtually eliminated this problem.
   b. A very common problem was that of fitting the hand shell to atrophied thenar-cleft musculature. The likelihood that this problem would be encountered and measures for adapting the shell to meet it should be emphasized in the instructional material.
   c. Some problems were encountered with stripping and bending of the telescopic rods.
   d. Some tendency for the shells to revert to their original shape after heating and modification was reported. However, in general, the physical properties of the splints were considered adequate to last an indefinite period with proper care and maintenance.

In conclusion, the field evaluation of the EPHO adaptations clearly revealed that the devices are useful additions to the armamentarium of orthotic items available for the treatment of patients with disabilities of the hand. It is recommended that the outcomes of this study be forwarded to the prosthetics-orthotics schools with a view to the possible inclusion of instruction in this system as part of the orthotics curriculum.

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Causes of Death in a Series of 4738 Finnish War Amputees

Georg Bakalim[1]

THE loss of a limb and its replacement by a prosthesis create conditions deviating from the normal. Walking is always more difficult. Loon (2) found that the energy consumption of amputees increases with the level of amputation. In the case of an above-knee amputation the effort of walking is greater than in a below-knee amputation and, in cases of hemipelvectomy and disarticulation of the hip, energy requirements are still greater. In the same investigation, it was found that walking with crutches, without a prosthesis, requires more energy than walking with a prosthesis. In addition, it appeared that in the presence of disturbances in the stump that affect walking, the consumption of energy increases. A poorly fitted prosthesis has the same effect. During walking, the center of gravity should shift smoothly, not in a jerky way that makes it more difficult to maintain balance. Almost all amputees experience excessive sweating not only of the stump but in general. The tightly fitted socket and the thigh corset used in connection with the old, conventional type of below-knee prosthesis are contributory causes of sweating.

Owing to the loss of the weight and accompanying movements of the amputated limb, upper-extremity amputees find it more difficult to keep their balance in walking after amputation. Similarly, the strain on the remaining upper limb in lifting and carrying is greater than before. The increased consumption of energy taxes the circulation and the heart. In this connection, no further attention will be paid to the secondary changes in the weight-bearing structures, particularly the joints and spine, that result from the altered static conditions due to the loss of a limb (9).

The health of amputees has been the subject of many previous studies, *e.g.*, those of Rausche (6), Schneider (7), Schulze *(8),* and Bodechtel *(1).* Meyeringh, Stefani, and Cimbal (5) reported a higher rate of hypertension in obese amputees than in amputees of average weight. In an electrocardiographic investigation of 1033 amputees, performed by the same authors, no differences were observed as compared with a normal series. Likewise, in a series of 1128 amputees obesity was not more frequent than in a corresponding group of the general population (5). Loos *(3)* reported similar findings in a series of 647 cases. Solonen, Rinne, Viikeri, and Karvinen *(9)* observed no noteworthy increase in cardiac and vascular diseases in amputees.

The purpose of this study was to find out whether death from degenerative cardiac and vascular diseases is more common among amputees than in the general population. At the same time tuberculosis, cancer, accidents, suicide, and miscellaneous causes of death were surveyed from the same standpoint.

MATERIAL

The series consists of 4782 war amputees. Data was collected from the files of the State Insurance Department. Finger, hand, toe, and foot amputations have been omitted since these cause no major problems. Before the end of 1944, *i.e.*, during the war, 44 amputees died. These cases are also considered in this study. The age distribution in this group was the same as in the remaining 4738 cases which have been followed up from 1945 till the

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end of 1965. The causes of death were obtained from the death certificates. During the last 10 years a steadily increasing number of cases have been examined post-mortem. In case of a casualty, or when the cause of death is unknown, autopsy is invariably performed. As a rule, the autopsy records contain more than one diagnosis, but in this study only the main diagnoses have been utilized. Although many of the second diagnoses might have been of interest, taking them into account would have implied considerable technical problems and would have rendered the statistical treatment more difficult. Since 1945, 643 subjects have died. During the period 1940-1965 the total mortality was thus 687/4782 (14.4 per cent). The number of mortalities during each year is shown in Figure 1. A steady rise is seen from 1960 onward. This increased mortality is not surprising, considering that more than 20 years have elapsed since the war and the mean age of the war veterans is about 50. However, this curve alone permits no conclusions to be drawn. In order to form an opinion concerning the mortality of the war amputees, the figures have to be compared to the death rates for the corresponding age groups of the general population.

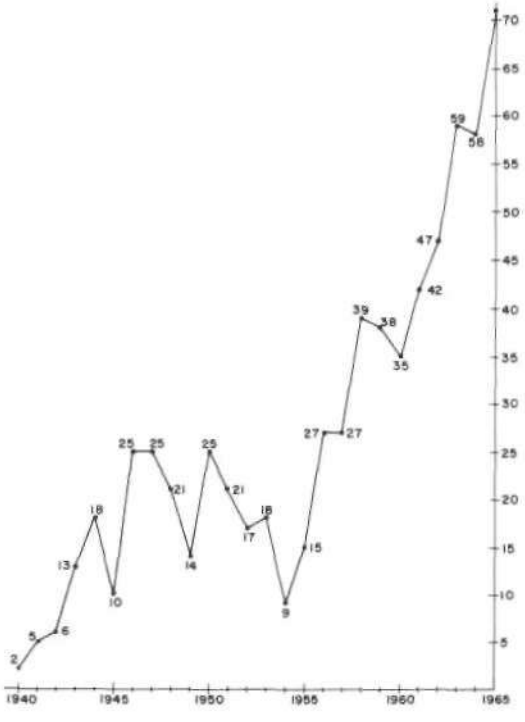

Fig. 1. Annual mortality of war amputees in 1940-1965.

AGE AND OCCUPATION

For the main causes of death the distribution of the dead war amputees by 5-year age groups is given in Table 1. Mostly, the age groups 40-50 years show the highest mortality. However, for conclusions to be drawn concerning the health of the group under review, comparable data for a "normal" group is required. The occupations of the dead, differentiated mainly on the basis of training, are given in Table 2. In this connection the main interest attaches to the proportion of heavy laborers.

Farmers (177) and unskilled workers (230) constitute the largest groups. Heavy labor is represented by 72.6 per cent, light occupations by 27.4 per cent. The handicraftsmen number 74 (10.8 per cent). There are as many as 31 shoemakers, which is accounted for by the fact that training for this occupation was offered after the war.

LEVEL OF AMPUTATION

The level of amputation appears in Table 3. Finger, hand, toe, and foot amputations were not included in this series because the trouble caused by them is considered to be so slight that it cannot lead to vascular disease. Two amputees in the present series had Chopart stumps, one had a Pirogoff stump, and in six cases disarticulation of the wrist had been performed. The ratio of above-knee to below-knee amputations is 1:2.

METHOD OF COMPARISON

The age distribution of the series followed up, exclusive of those who died before 1945, and the percentage figures for the corresponding age groups of the general Finnish male population are shown in Table 4. As may be seen in the table, the age distribution of the am-

TABLE 1. DISTRIBUTION OF THE CAUSE OF DEATH BY 5-YEAR AGE GROUPS IN A SERIES OF 687 DEAD WAR AMPUTEES

Age Group (years)	Degenerative Vascular Diseases of the Central Nervous System	Degenerative Cardiac and Vascular Diseases	Tuberculosis	Malignant Diseases	Injuries	Suicides	Other Diseases	Total
20–24			6		1	4		11
25–29		2	18	1	10	6	11	48
30–34	2	8	14	2	13	5	9	53
35–39	5	14	13	7	10	10	13	72
40–44	11	37	11	11	13	19	29	131
45–49	13	56	6	25	16	11	17	144
50–54	11	48	1	21	5	5	8	99
55–59	12	33		18		2	6	71
60–64	5	11	1	7	4	1	3	32
65–69	3	6		2			3	14
70–74		3		1			1	5
75–79	2			1				3
80–84		1					3	4
Total	64	219	70	96	72	63	103	687

TABLE 2. DISTRIBUTION ACCORDING TO OCCUPATION

Occupation	Number		Per Cent	
University-educated	20			
School teacher	3			
Musician	1			
Technician	16	81	11.8	
Businessman	17			
Office clerk	13			
Store keeper	11			27.4
Labor foreman	15			
Noncommissioned officer	3			
Craftsman	9			
Optician, watchmaker	10			
Baker	1	107	15.6	
Tailor	1			
Shoemaker	31			
Painter	3			
Carpenter	12			
Blacksmith	5			
Masseur	2			
Chauffeur	15			
Gardener	2	177	25.7	
Farmer	175			
Skilled worker	62	292	42.5	
Unskilled worker	230			72.6
Sailor	5	11	1.6	
Fisherman	6			
Doorkeeper	8	8	1.2	
Unknown occupation	11	11	1.6	
Total	687		100.0	

TABLE 3. CLASSIFICATION ACCORDING TO LEVEL OF AMPUTATION OF THOSE AMPUTEES WHO DIED DURING THE PERIOD 1945–1965

Shoulder disarticulation	3
Above-elbow	96
Below-elbow	63
Wrist disarticulation	6
Hip disarticulation	3
Above-knee	193
Knee disarticulation	1
Below-knee	319
Pirogoff	1
Chopart	2
Total	687

TABLE 4. AGE DISTRIBUTION OF 4738 AMPUTEES AT THE BEGINNING OF 1945 COMPARED TO THE GENERAL FINNISH MALE POPULATION

Age Group (years)	Number		Per Cent	General Male Population (per cent)
15–19	200	1475	31.2	16.0
20–24	1275			
25–29	1296			
30–34	1034	3177	67.0	27.0
35–39	572			
40–44	275			
45–49	55			
50–54	13	86	1.8	19.0
55–59	12			
60–64	6			
Total	4738		100.0	62.0

putees differs widely from the age distribution of the general Finnish male population as obtained from the Statistical Yearbook of Finland. For this reason, the death rates for the general Finnish male population could not be used as such for comparison with the mortality rate of amputees. It was necessary therefore to construct an equivalent, theoretical population with an age distribution corresponding to that of the amputees. The data required was obtained in part directly from the Statistical Yearbook, and in part by calculation based on the death rates for men and women and the sex ratio, or for the earlier years, on the total mortality and the age distribution of the dead, as indicated in the Statistical Yearbook. In the comparisons, it was deemed most appropriate to consider only the period from 1945 till the end of 1964. The amputees who died before 1945 numbered 44, and 71 died in 1965. When these 115 cases were subtracted from the total number of dead in the present series (687), 572 cases remained for the comparative analysis of mortality.

MORTALITY

As mentioned above, the total mortality for the period under review was 687/4782 (14.4 per cent). The causes of death are listed in detail in Table 5. The distribution according to the cause of death has been given in summary form in Table 1. Degenerative vascular diseases of the central nervous system and degenerative cardiac and vascular diseases have the same etiology but each forms a separate entity, and the Statistical Yearbook of Finland provides figures for comparison precisely on this basis. In addition, death rates were available for pulmonary tuberculosis, malignant diseases, accidents, and suicide, other causes falling into a miscellaneous group consisting of cases for which no comparative figures were found in the Statistical Yearbook. Many cases of poisoning and drowning were recorded under accidents. Alcohol abuse was a major etiological factor. It was sometimes difficult to decide whether the cause of death was an accident or suicide.

COMPARISON OF MORTALITY OF THE AMPUTEES AND THE GENERAL POPULATION

In what follows, the total mortality is analyzed first and then the mortality in the various groups listed above is analyzed, except for the miscellaneous group for which no comparable data was available.

TOTAL MORTALITY

On comparing the total number of deaths during the period January 1, 1945, to December 31, 1964, *i.e.*, 572, to the mortality of the general Finnish male population, the age distribution was taken into account in two different ways. In both methods, consideration was given to the fact that during the period under review the subjects passed into age groups with a lower expectation of life.

Method I

For each 5-year age group of amputees in Table 4 (age distribution at the beginning of 1945), the expected losses for the 5-year periods 1945-1949, etc., until the beginning of 1965, were calculated on the basis of the expectations of life indicated in the Statistical Yearbook of Finland, that figure being used which pertains to the mean age of the age group during the period in question. To exemplify, for those who were aged 20-24 years at the beginning of 1945, the expectation of life at 25 years was considered as the relevant figure for the period 1945-1949, since the youngest in the group had survived for 20-24 years and the oldest for 24-29 years. Correspondingly, the expectation of life at 30 years was applied to the period 1950-1955, etc. The 5-year losses were calculated on the basis of the total number of survivors. In Figure 2, the cumulative curve for the calculated losses from the level of 1945

TABLE 5. CAUSES OF DEATH OF 687 AMPUTEES IN A SERIES OF 4872

Degenerative vascular diseases of the central nervous system	
Cerebral hemorrhage	32
Subarachnoidal hemorrhage	10
Cerebral embolism and thrombosis	22
	64
Degenerative cardiac and vascular diseases	
Cardiac infarction	148
Pulmonary infarction	6
Arterial embolism	11
Myodegeneration of the heart	24
Cardiac insufficiency	25
Arteriosclerosis	5
	219
Rheumatic disease of the heart	7
Tuberculosis	
Pulmonary	70
Intestinal	2
Miliary	2
	74
Malignant diseases	
Mouth	2
Nose, pharynx	1
Jaw bones	2
Esophagus	4
Stomach	31
Small intestine	1
Rectum	4
Liver	2
Pancreas	4
Lungs	29
Mediastinum	2
Prostate	2
Testes	1
Kidneys	1
Bladder	2
Brain	3
Malignant lymphogranulomatosis	1
Myelomatosis	1
Leukemia	2
Undefined neoplasm	1
	96
Injuries	
Traffic accident	17
Other accident	10
Explosion	1
Work accident	4
Home accident	2
Drowning	11
Burns	1
Freezing	1
Alcohol intoxication	11
Barbiturate intoxication	7
Carbon monoxide poisoning	2
Homicide	5
	72
Suicides	
Hanging	24
Shooting	15
Poisoning	10
Drowning	5
Cutting or piercing weapons	3
Crushing	6
	63
Miscellaneous diseases	
Pneumonia	12
Bronchial asthma	1
Gastric and duodenal ulcer	6
Acute appendicitis	3
Intestinal obstruction	8
Peritonitis	1
Cirrhosis of the liver	7
Perforation of the gallbladder	1
Pancreatitis	2
Chronic nephritis	30
Meningitis, abscess of the brain	4
Paralysis agitans	2
Poliomyelitis	2
Septicemia	5
Paratyphus	1
	85
Senility	1
Unknown cause of death	6
Total	687

is compared to the cumulative curve for the actual losses.

The recorded death rates for the *5-year age groups* are slightly lower than the expected figures, but the difference is statistically insignificant. The same obtains to the death rates as expressed by *5-year periods* (Table 6). The differences between the recorded and the expected figures are of the order of 10 per cent. The greatest differences relate to the periods 1950-1954 and 1955-1959, while for the

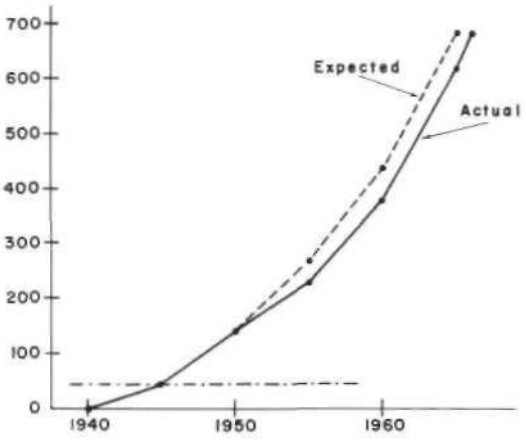

Fig. 2. Cumulative death rates-calculated for 5-year periods compared to cumulative expected death rates.

TABLE 6. MORTALITY AMONG 4738 WAR AMPUTEES DURING THE PERIOD 1945–1964 (5-YEAR PERIODS AND CUMULATIVE FIGURES) AND CORRESPONDING EXPECTED VALUES IN AN EQUIVALENT MALE POPULATION

Period	Deaths		Difference in Per Cent	Cumulative Deaths		Difference in Per Cent
	Recorded	Expected		Recorded	Expected	
1945–49	95	97.3	2.3	95	97.3	2.4
1950–54	90	124.6	27.8	185	221.9	16.6
1955–59	146	168.7	13.5	331	390.6	15.3
1960–64	241	246.4	2.6	572	637.0	10.4
1945–64	572	637.0	10.4			

periods 1945-1949 and 1960-1964, the recorded figures fall below the expected ones by about 2 per cent only.

Method II

In the Statistical Yearbook of Finland, the number of survivors among 100,000 men of the same age is indicated. On the basis of these figures, the numbers of expected survivors in all age classes represented in this series at the beginning of 1945 were calculated for the end of the age periods 20-24 years, 25-29 years, etc., and the expected death rates in the various age groups were expressed as percentages. The expected total mortality by the end of 1964, *i.e.*, 549, is in very good agreement with the actual figure of 572. All the 687 deaths considered, the percentile distribution between the age groups corresponds fairly well to the expected distribution (Table 7, Fig. 3).

If the causes of death are disregarded, it may be stated that the mortality in the present series corresponds very closely to the mortality in the corresponding general population. This obtains to the figures for the various 5-year periods and the total mortality as well as to the figures for the age groups. There seems to be a tendency toward a lower mortality for am-

TABLE 7. DISTRIBUTION ACCORDING TO AGE AT DEATH OF 687 WAR AMPUTEES COMPARED TO THE GENERAL MALE POPULATION

Age at Death	Deaths		Expected Mortality (per cent)
	Number	Per Cent	
20–24	11	1.6	1.2
25–29	48	7.0	3.8
30–34	53	7.7	7.6
35–39	72	10.5	12.4
40–44	131	19.1	16.7
45–49	144	21.0	18.7
50–54	99	14.4	17.3
55–59	71	10.3	11.9
60–64	32	4.7	6.5
65–69	14	2.0	2.2
70–74	5	0.7	1.0
75–79	3	0.4	0.6
80–84	4	0.6	0.1
Total	687	100.0	100.0

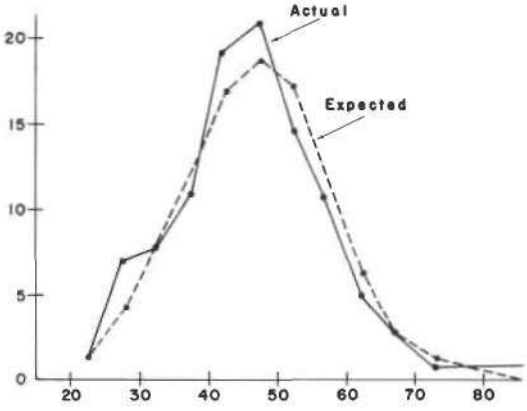

Fig. 3. Death rates for the different age groups compared to the expected death rates.

putees than in the general population, and, with regard to the age at death, it appears that among the amputees there may be a trend toward a lower age, though only by one or two years at the most.

DEGENERATIVE VASCULAR DISEASES OF THE CENTRAL NERVOUS SYSTEM

The mortality in degenerative vascular diseases of the central nervous system was 64/687 (9.3 per cent). Traumatic cerebral hemorrhages of course do not belong to this group. Comparable data relating to the general population was obtained from the Statistical Yearbook of Finland, and expected figures were calculated for the period 1945-1964 in the same way with respect to the total mortality. The expected number of deaths in this group of disease was 37.4. The actual number (64) was 71.2 per cent higher. In the age groups 25-44 years the actual number of deaths was 130.9 per cent higher than the expected number; in the age groups 45-64 years it was 49.6 per cent higher; and in the age groups 65-74 it was 42.6 per cent higher (Table 8). No consistent trend is discernible with regard to the age at death.

DEGENERATIVE CARDIAC AND VASCULAR DISEASES

This group includes cardiac infarction, pulmonary infarction, peripheral embolism, myodegeneration, cardiac insufficiency, and arteriosclerosis. The mortality in this group was 219/687 (31.9 per cent). The expected number of deaths in the general population was 134.3. The actual mortality was 63.1 per cent higher. As regards the different age groups, the actual mortality was 193.2 per cent higher than the expected in the group aged 25-44 years at death, 38.9 per cent higher in the group aged 45-64 years, and 28.6 per cent higher in the group aged 65-74 years (Table 8). One hundred and four amputees (47.5 per cent) died at an age of 45-54 years, 51 (23.3 per cent) at an age of 35-44 years, and 44 (20.1 per cent) at 55-

TABLE 8. DEATHS IN AMPUTEE POPULATION COMPARED WITH EXPECTED NUMBER IN GENERAL MALE POPULATION

Age at Death (years)	Disease of the Central Nervous System			Cardiac and Vascular Diseases			Pulmonary Tuberculosis			Malignancy			Accidents			Suicides			Miscellaneous		
	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference	Recorded	Expected	Per Cent Difference
24	(0)	(0.1)		6	2.2	+172.7	0	0.5		1	4.6	−78.3	4	1.0	+300.0						
25–44	18	7.8	+130.9	61	20.8	+193.2	56	62.6	−10.5	21	25.1	−16.4	46	72.1	−36.2	40	26.00	+53.8			
45–64	41	27.4	+49.6	148	106.5	+38.9	8	27.5	−70.9	71	90.0	−21.1	25	32.1	−24.1	19	18.7	+1.6			
65–74	3	2.1	+42.6	9	7.0	+28.6	(0)	(0.9)		3	3.8	−21.1	(0)	(0.6)		0	0.2				
75–	(2)			(1)						(1)											
Totals	64	37.4	+71.2	219	134.3	+63.1	70	93.2	−24.9	96	119.4	−19.6	72	109.4	−34.2	63	45.9	+37.3	103	147.4	−30.1

64 years. The remaining 20 deaths (9.1 per cent) were evenly distributed between the age groups 25-34 and 65-84 years (Table 1).

PULMONARY TUBERCULOSIS

The mortality in pulmonary tuberculosis was 70/687 (10.2 per cent). The actual mortality was found to be 24.9 per cent lower than the expected mortality (93.2 cases). In the group under 24 years of age the mortality was 172.7 per cent higher than the expected, while in the age groups 25-44 and 45-64 the actual mortality was 10.5 and 70.9 per cent lower, respectively, than the expected (Table 8).

MALIGNANT DISEASES

The mortality in malignant disease was 96/687 (14.0 per cent). The mortality was 19.6 per cent lower than the expected. In the age group 45-64 years the mortality was 21.1 per cent lower, and in the age group 65-74 it was also 21.1 per cent lower than the expected mortality. The frequency of malignant disease in different organs appears in Table 5. In none of the present cases was the disease a result of the amputation (Table 8).

ACCIDENTS

Accidents were the cause of death in 72/687 cases (10.5 per cent). The actual figures were in all age groups lower than the expected. In the age group under 24, the recorded number of deaths was 78.3 per cent lower than the expected mortality; in the group 25-44 years it was 36.2 per cent lower; in the group 45-64 years it was 24.1 per cent lower. The actual total mortality was 34.2 per cent lower than the expected. This group includes 17 (2.5 per cent) traffic accidents, but these could not be separately analyzed, because traffic accidents are not treated as a separate group in the Statistical Yearbook (Table 8).

It thus appears that the mortality from accidents was markedly lower among the amputees than in the general population. It might have been expected that amputees would be more accident-prone both at work and in the traffic, owing to their poorer mobility. The small proportion of traffic accidents among the total number of cases is also striking. Obviously, the amputees move about less than the general population, work at less dangerous places, and are, perhaps, employed to a lesser extent owing to their reduced working capacity.

SUICIDES

Since about 80 per cent of the suicides are committed by men, it seemed reasonable to use this age distribution as a basis when the expected mortality was calculated in the same way as for the other causes of death. The actual figures for the periods 1955-1959 and 1960-1964 are 68.1 and 36.0 per cent higher than the expected figures. The total number of suicides (63) for the period 1945-1964 is 37.3 per cent greater than the expected number. The greatest difference is noted for the period 1945-1949, the recorded frequency of suicides being 3.6 times higher (260.0 per cent) than the expected (Table 8). By contrast, the figure for 1950-1954 is 73.4 per cent lower than the expected mortality. If these two 5-year groups are added together the difference by which the actual frequency of suicides exceeds the expected has changed to a decrease (—13.8 per cent).

It appears that among amputees under 25 years of age, suicides were 300.0 per cent higher, and in the age group 25-44 years 53.8 per cent higher than was to be expected on the basis of the statistics for the general population. By contrast, the number of suicides committed by amputees aged 65-74 years was within 0.2 per cent of the expected figure. The total actual number of suicides exceeds the expected figure by a difference of 37.3 per cent (Table 8).

In addition, the rate of suicides among the dead amputees with the same occupation has been calculated. In this respect there is no major difference between heavy labor and other occupations. Tech-

TABLE 9. DISTRIBUTION OF 63 AMPUTEES, WHO COMMITTED SUICIDE, ACCORDING TO SITE OF AMPUTATION

Amputation	Number	Per Cent	
Above-elbow	5	7.9	30.2
Below-elbow	14	22.3	
Above-knee	13	20.7	61.8
Below-knee	26	41.1	
Others	5	8.0	8.0
Total	63	100.0	

nicians have the lowest rate of suicide, those with unknown occupations the highest. With regard to the former, it may be pointed out that their occupation is highly suitable for amputees, while the latter group includes subjects without regular employment, who lived in poor social conditions.

The possible relationship between the rate of suicides and the level and site of the amputation is analyzed in Table 9. Among lower-limb amputees the frequency of suicide was twice the frequency among upper-limb amputees. However, when the whole series is taken into account, the difference is not very great, the number of lower-limb amputees being double the number of upper-limb amputees.

The methods of suicide appear in Table 5. Alcohol abuse was known to have played a part in 11 cases, and 6 subjects had used barbiturates in addition. This group of 63 consists of only sure cases of suicide. In the group of accidents, at least a slight suspicion of suicide was present in many cases.

SUMMARY AND DISCUSSION

In a series of 4782 war amputees, the total mortality was 687 (14.4 per cent). The period covered by the present study is from 1945 till the end of 1965. In 1960, the mortality of the war amputees began to rise abruptly, and was one of the causes for undertaking this study. This mortality was compared to the mortality in the general Finnish male population. A theoretical, equivalent male population was constructed on the basis of data obtained from the Statistical Yearbook of Finland.

When the causes of death were not differentiated, the mortality of the amputees was found to be in good agreement with the mortality of the general population. This obtains to both the whole series and the different 5-year periods. There was even a tendency towards slightly lower figures for the amputees.

On the other hand, when the causes of death were differentiated, certain features of interest emerged. The recorded death rates were higher than the expected figures with regard to degenerative diseases of the central nervous system (+71.2 per cent), degenerative cardiac and vascular diseases (+63.1 per cent), and suicide (+37.3 per cent). These were the causes of death in half the cases. One-fourth of the deaths were due to pulmonary tuberculosis or malignant disease. In both these groups the actual death rate was lower than the expected (—24.9 per cent and —19.6 per cent). In the age group under 25, the mortality in pulmonary tuberculosis was 2.7 times higher than in the corresponding group of the general population, but in all other age groups it was lower than the expected death rate. The number of deaths due to accidents (72) fell below the expected mortality by 34.2 per cent. Obviously, amputees move about considerably less than the general population, and they are less exposed to accidents owing to their limited working capacity.

In order to give a general survey of the findings, the main causes of death are listed in Table 8. In addition to the number of deaths, the mortality in each group is expressed as a percentage. Likewise, the expected mortality is given both in absolute figures and as percentages, and the differences between the actual and expected figures are indicated in percentages. In this connection, it has been assumed that the total expected mortality is the same as the actual mortality, as was also suggested by the analysis of the

total mortality carried out at the beginning of this study. The amputees seem to be more afflicted with fatal degenerative diseases of the central nervous system and fatal degenerative cardiac and vascular diseases, and suicides seem to be more common among them, as compared with the general population. On the other hand, the mortality from pulmonary tuberculosis, accidents, and a large group of miscellaneous diseases *(e.g.,* various diseases of the lungs and abdominal disorders), was lower among the amputees than in the general population.

It may be assumed that the higher frequency of suicides among the amputees is due in part to psychological causes connected with the loss of a limb. Also, a postwar depression may have become more pronounced with the lapse of time. Economic problems and poor social conditions may be regarded as contributory causes.

In the care of amputees, the factors of importance are: a satisfactory prosthesis, good condition of the stump, rehabilitation, suitable employment, and judiciously administered subvention. The question arises as to whether all that could have been done for the war amputees was done. Perhaps something had been neglected that could have prolonged the lives in some cases.

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The Mechanical Properties of Bone[1]

F. Gaynor Evans. PH.D.[2]

BONE is the material with which the orthopaedic surgeon deals. Consequently, some knowledge of its mechanical properties is of importance for an understanding of the mechanism and management of fractures, as well as the design of prosthetic or orthotic appliances and protective gear, *e.g.,* crash helmets. The behavior of a body under a load or force is a function not only of the form and structure of the body, but also of the mechanical properties of the material composing the body. For example, a steel beam will support a higher load before breaking and will behave differently under loading than will an oak beam of exactly the same shape and dimensions because of differences in the mechanical properties and structure of steel and of wood.

The mechanical properties of bone are determined by the same methods used in studying similar properties of metals, woods, and other structural materials. These methods are based on certain fundamental principles of mechanics, a knowledge of which is essential for understanding the terminology employed.

Mechanics, the science dealing with the effect of forces upon the form or the motion of bodies, has two subdivisions—statics and dynamics. *Statics* is the study of bodies at rest or in equilibrium as a result of the forces acting upon them. *Dynamics* is the study of moving bodies. The mechanical properties of materials are usually studied under static conditions, *i.e.,* under a slowly applied force or load, because the behavior of the test specimen can be more easily analyzed when the load is slowly applied.

A *force* is anything which tends to change the state of a body with respect to its motion or the relative position of the molecules composing the body. More simply stated, a force is a push or a pull. There are three primary kinds of forces: (1) *compressive* or pushing together forces, (2) *tensile* or pulling apart forces, and (3) *shearing,* or forces which make one part of the body slide with respect to an adjacent part (Fig. 1).

When a force is applied to a body, it produces stress and strain within the body. *Stress* (Fig. 1) is the ratio between the force and the area upon which it acts, *i.e.,* force per unit area. Stress is generally computed in terms of pounds per square inch (psi) or kilograms per square millimeter (ksm). Recently, some investigators of the strength characteristics of bone and other biological materials have been recording stress values in terms of kiloponds, dynes, or newtons per unit area, instead of pounds or kilograms because pounds and kilograms are units of mass as well as units of force. There will be no misunderstanding, however, if one specifies that stress values are in terms of *pounds force or kilograms force per unit area.* Stress is often used synonomously with strength, but the term has little value unless the kind of strength, *i.e.,* tensile, compressive, etc., is indicated. All strength values in the following discussion are in terms of *pounds force per square inch.*

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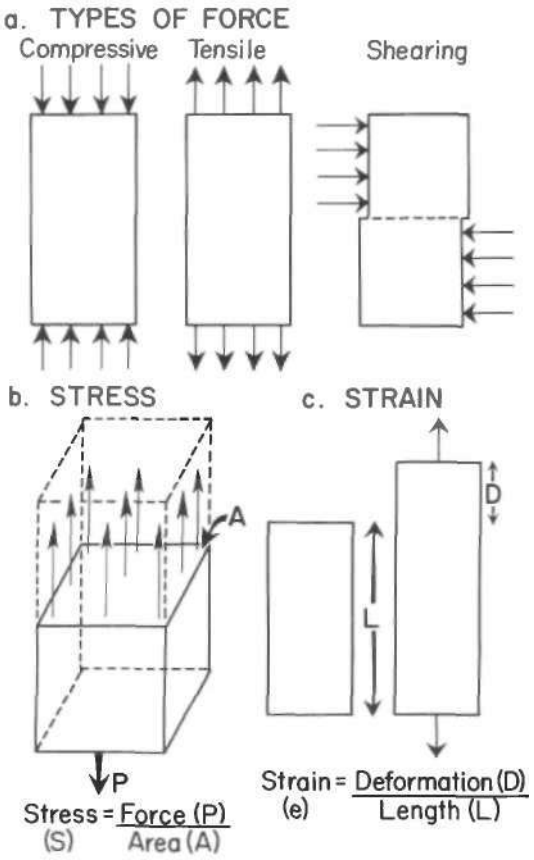

Fig. 1. Types of pure force—stress and strain.

Strain is a change in the linear dimensions of a body as the result of the application of a force (Fig. 1). Since there are no standard units of measurement for strain, it can be recorded as percentage, inches/inch, centimeters/centimeter, etc. Strain can be seen if it is sufficiently large, *e.g.,* as in stretching of a rubber band, but stress, which is only the ratio between force and area, is always invisible. The kind of stress and strain in a body is the same as the kind of force producing it.

When stress is plotted against strain, a *stress-strain curve* is obtained (Fig. 2). From a tangent drawn to the straightest part of the stress-strain curve the *modulus of elasticity* of the material, or the ratio between unit stress and unit strain, can be computed. The modulus of elasticity is a measure of the *stiffness* of a material, not its elasticity as one might assume from the name. *Elasticity* is the property of a material that allows it to return to its original dimensions after the removal of a force or load. The *energy* the specimen absorbs to failure can be determined by measuring the area below the stress-strain curve.

The method of choice in determining the tensile or compressive strength of a material is to make a test specimen of a standardized size and shape and test it under a pure tensile or a pure compressive force. Under these conditions the cross-sectional area of the specimen is known, or can be easily computed, and only one force—tension or compression—

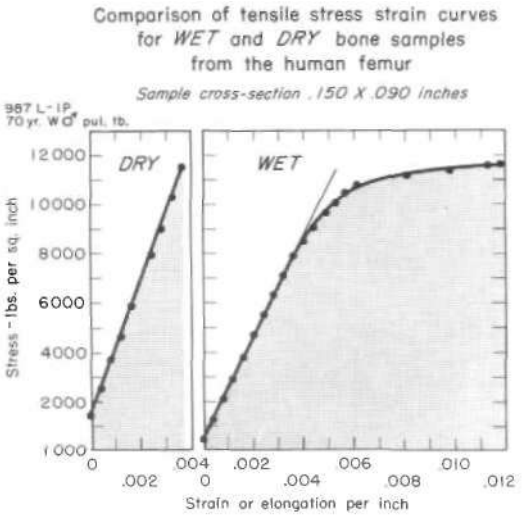

Fig. 2. Stress-strain curves for a dry- and a wet-tested specimen of compact bone from the posterior quadrant of the proximal third of the femoral shaft of a 70-year-old white man who died from pulmonary tuberculosis. The stress values are in *pounds force per square inch (7).*

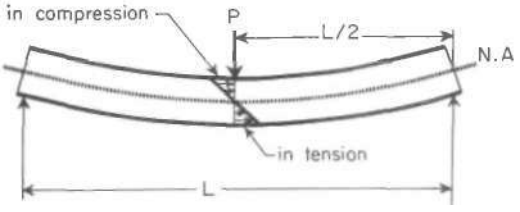

Fig. 3. Distribution of tensile and compressive forces in a body tested like a simple beam *(6). L* = length or span between supports; *N. A.* = neutral axis or plane; *P* = force or load.

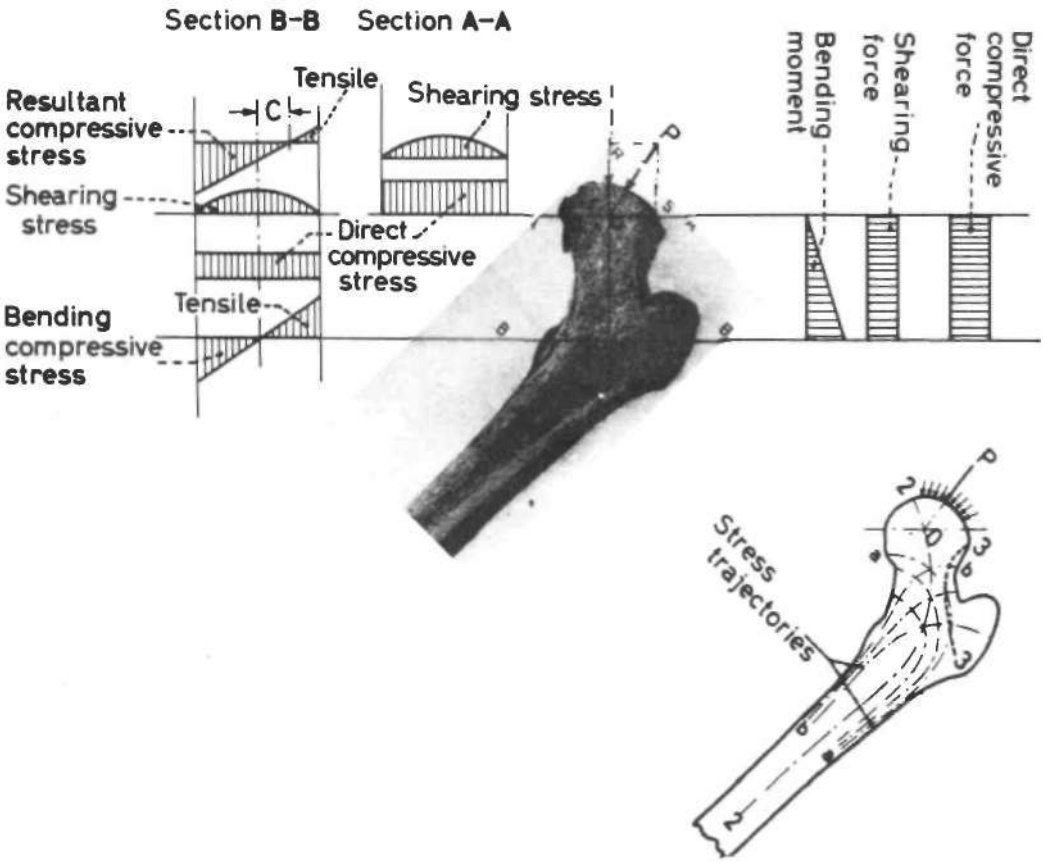

Fig. 4. Stress distribution in the neck of the femur *(20).*

is involved. Furthermore, the force is uniformly distributed over the cross-sectional area of the specimen. Consequently, the ultimate tensile or compressive strength of the material can be easily calculated from the formula $S = P/A$, in which S is stress, P is force or load, and A is the cross-sectional area of the specimen (Fig. 1).

If the specimen is tested like a simple beam (i.e., supported at the ends and loaded midway between the supports) and bending occurs, tensile, compressive, and shearing forces are all involved. Tensile forces develop on the convex side of the bent specimen while compressive forces occur on the opposite (concave) side (Fig. 3). Both types of forces are maximum at the surface and decrease inwardly to zero at the neutral plane or axis. There are also shearing forces which, like the tensile and compressive forces, are not uniformly distributed over the cross section of the specimen. Under bending conditions, the force responsible for failure as well as its magnitude is more difficult to determine. The bending forces in the neck of the femur, as a result of the load applied to the head of the bone (Fig. 4), have been determined by Zarek *(20),* an engineer who is currently working in biomechanics. For further discussion of forces in bending, see Harris' *Strength of Materials (11).*

The speed at which a force is applied to a specimen influences the values obtained for some of its mechanical properties. McElhaney and Byars *(17)* found that the ultimate compressive strength and the modulus of elasticity of fresh and embalmed femoral cortical bone from cattle and man increased with higher strain rates of loading while the energy-absorbing capacity and the strain at failure decreased. The

effect of high strain rates of loading on specimens of beef bone, cut and tested in different directions, has recently been investigated by Bird *et al. (1).*

Embalming also affects the mechanical properties of bone, at least those of compact bone. Thus, the mean ultimate tensile strength (in the long axis of the specimen and of the intact bone) is greater at the 0.01 significance level in embalmed wet- and dry-tested tibial specimens than in similarly tested unembalmed specimens *(4).* Furthermore, embalmed, wet-tested tibial specimens have a higher mean tensile strain, a greater mean single shearing strength (perpendicular to the long axis of the specimen) and are harder (Rockwell No.) than similarly tested embalmed specimens (5). However, the latter type of specimens has a higher mean modulus of elasticity. An analysis of variance showed that the increase in the hardness of the embalmed specimens was significant at the 0.01 level. As far as I am aware, there are no similar studies concerning the effect of embalming on the mechanical properties of spongy bone.

Two types or forms of bones are found in the foot—irregularly shaped bones (the tarsals) and miniature long bones (the metatarsals and the phalanges). The tarsal bones are essentially shells of compact bones filled with spongy bone, fat, marrow substance, blood, etc. The actual amount of osseous material in bones, such as the tarsals and the bodies of vertebrae, is not very great. According to Policard and Roche *(18)* the talus and the calcaneus are about 80 per cent nonosseous tissue. The percentage of bone in the bodies of 92 human lumbar vertebrae studied by Bromley *et al.* (2) varied from a maximum of approximately 24 per cent to a minimum of 15.5 per cent in males and from 21 per cent to 12 per cent in females at 5 and 70 years of age, respectively. As far as I am aware, there are no studies on the mechanical properties of spongy bone from the foot. Therefore, examination of such properties will be based on data obtained from the human femur.

Two types of specimens were used—a rectangular bar (the standard specimen) 0.79 cm. x 0.79 cm. x 2.5 cm. and a cube 0.79 cm. on a side. The specimens were obtained from the head, neck, greater trochanter, and condyles of the femur with the long axis of the standard specimens oriented in different directions.

The specimens were tested under direct compression in a Riehle 5000-lb. capacity testing machine, equipped with an automatic stress-strain recorder and calibrated to an accuracy of ±0.5 per cent. The low range scale of the machine (0-200 lbs.) was used with the load registered on the dial of the machine in units of 0.5 lbs. The specimens were loaded at a speed of 0.45 in. per min.

All specimens were tested wet to more nearly approximate the condition in the living foot. Drying of compact bone increases its ultimate tensile strength (in the long axis of the specimen), its modulus of elasticity, and its hardness (Rockwell No.) but decreases its single shearing strength (perpendicular to the long axis of the specimen) and its tensile strain (7, 8). Similar studies have not, to my knowledge, been made on spongy bone.

The ultimate compressive stress (strength) and strain, the modulus of elasticity, and the energy absorbed to failure were computed from stress-strain curves for wet-tested specimens. The density of air-dried specimens was determined with a strontium 90 densitometer developed by Evans, Coolbaugh, and Lebow *(9).* Dry specimens were used to avoid the effects of moisture that might be trapped within the interstices of the specimens. A total of 69 rectangular (standard) specimens and of 15 cubic specimens from 1 adult, white female, 3 adult, Negro males, and 6 adult, white males were tested. All specimens were kept in saline solution until tested. A minimum of 20 load-deformation readings were taken for each specimen during the test period.

The results of the study showed that the mean compressive stress (strength) of the cubic specimens was greater than that of

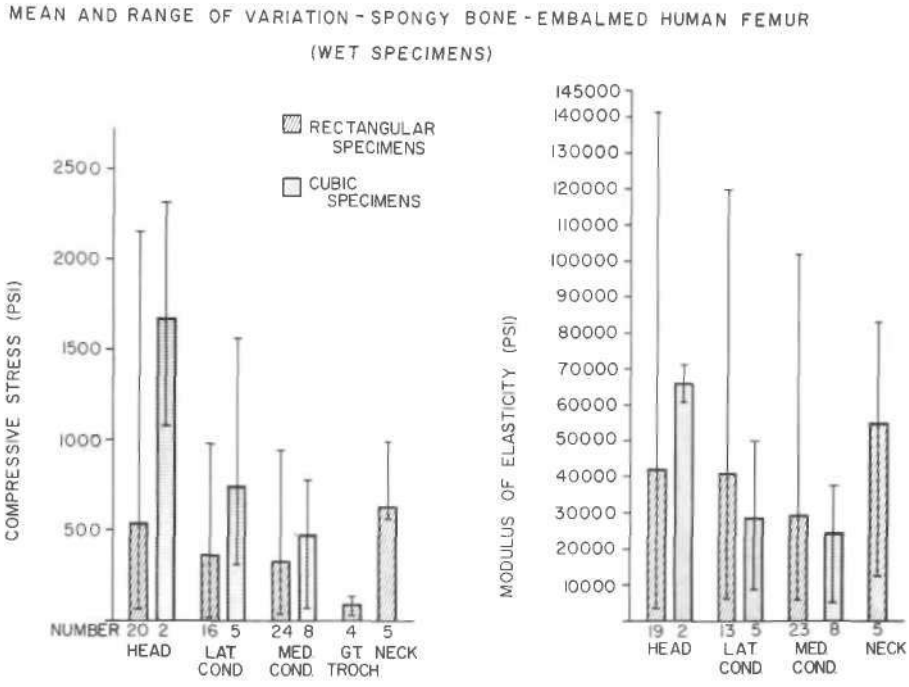

Fig. 5. Mean and range of variation in some mechanical properties of spongy bone from different regions of the femur. Compressive stress values in *pounds force per square inch. Gt. troch.* = greater trochanter; *Lat.* = lateral; *Med.* = medial; *Cond.* = condyle.

the rectangular (standard) specimens from the same region (Fig. 5). This phenomenon is characteristic of practically all materials. In cubic specimens high frictional forces developed between the ends of the specimen and the testing machine to resist the tendency of the specimen to be squeezed out of the machine. Furthermore, the upper part of the cube tends to be impacted into the lower part. Both of these factors contribute to higher values for compressive stress and modulus of elasticity in cubic than in specimens which are longer than wide. Because of these factors, it is felt that the values obtained from the rectangular (standard) specimens more accurately represent the true mechanical properties of spongy bone.

In the living body, most of the bones are subjected to bending action as a result of gravity, muscular activity during movement, and blows. Consequently, the bones are subjected to a combination of tension, compression, and shearing rather than to a single pure force. The question then arises as to why the strength of bone is usually determined by testing the specimens under a pure force. The answer to this question, on mechanical grounds, has already been given. There are, however, other valid reasons for testing the strength of bone under pure tension or compression.

Experimental studies with strain sensitive lacquers on bones within the living body as well as outside of it demonstrate that certain types of linear fractures of the skull, the pelvis, and the long bones all arise from failure of the bone from tensile stresses and strains produced in it by bending (3). The determination of the tensile strength of bone under pure tension thus has direct application to the mechanics of fractures of those types. Clinical experience also indicates that tensile forces are important in the production of many types of fractures.

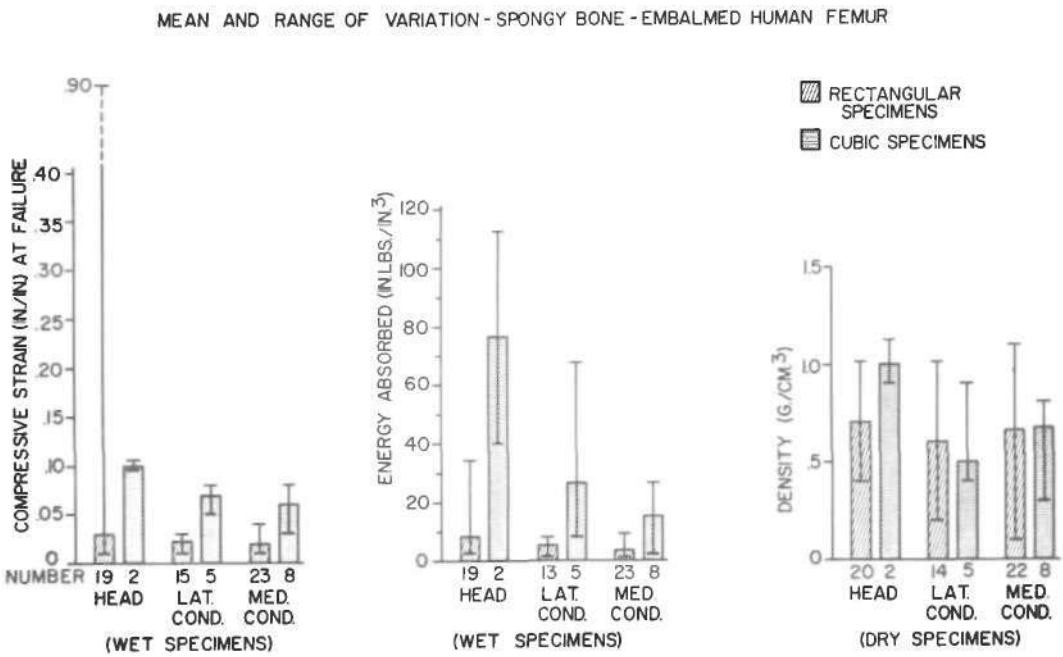

Fig. 6. Mean and range of variation of some mechanical properties of spongy bone from various regions of the femur.

Compression fractures are quite common in the bodies of the vertebrae, especially those in the lumbar region, and in the calcaneus, the most frequently fractured of the tarsal bones *(12).* Compression fractures of the talus also occur. There is, consequently, a sound practical reason for investigating the compressive strength of the tarsal bones, especially the calcaneus and the talus although, to my knowledge, it has not been done. The rationale for determining the strength of spongy bone from the femoral head and condyles under direct compression is that these regions of the bone are normally subjected to compression forces in the erect posture *(13).* Specimens from other regions were similarly tested for comparative purposes.

When the results of the tests were compared according to the region of the bone from which the specimens were obtained, without regard to the direction of loading, several differences were found. The rectangular (standard) specimens from the neck had the highest and those from the greater trochanter the lowest mean compressive stress. Among the cubic specimens the highest and the lowest mean compressive stresses were found in specimens from the head and the medial condyle, respectively.

Regional variation was also found in the modulus of elasticity (stiffness) of the specimens (Fig. 5). The mean stiffness of the rectangular specimens exceeded that of the cubic specimens from the same region except for the specimens from the head. The rectangular specimens from the neck and the medial condyle, respectively, had the highest and the lowest mean modulus. The maximum and the minimum stiffness means of the cubic specimens were found in those from the head and the medial condyle, respectively.

Comparison of the mean compressive strain, mean energy absorbed to failure, and mean density of the rectangular and cubic specimens from different parts of the femur also reveals interesting differences (Fig. 6). The cubic specimens showed somewhat more variation in the mean compressive strain than did the rectangular ones, the strain being greatest in the specimens from the head and least in those

from the medial condyle. Little difference was found in the mean compressive strain of the rectangular specimens, those from the head having a slightly greater strain than those from the condyles. The cubic and the rectangular specimens from the head had the highest while those from the medial condyle had the lowest mean energy absorbed to failure. However, the former specimens showed more regional difference than did the latter. The mean density for both types of specimens was greatest in those from the head and least in the ones from the lateral condyle.

A statistical analysis of the above data from the rectangular (standard) specimens revealed the following significant differences between the means. The mean compressive stress of the strongest specimens (from the neck) was greater, at the 0.02 significance level, than that of the weakest specimens (from the greater trochanter). The difference between the mean compressive strain of the specimens from the head, which had the highest, and that of specimens from the medial condyle, which had the lowest, was significant at the 0.01 level.

The mean energy absorbed by the specimens from the head was significantly greater, at the 0.02 level, than that absorbed by specimens from the medial condyle. The differences between the means for the other mechanical properties of the rectangular specimens were not statistically significant. The number of cubic specimens tested was not sufficiently large for statistical analysis.

Comparison of the maximum compressive stress and modulus of elasticity (Fig. 7) of the rectangular and cubic specimens

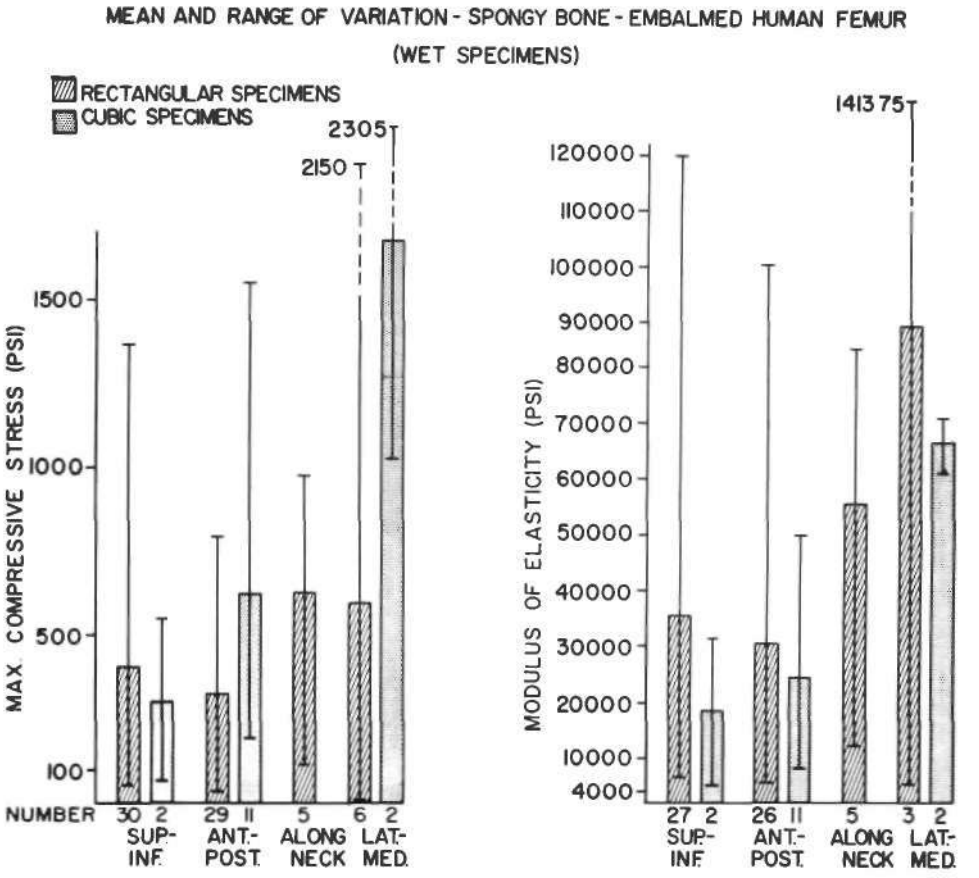

Fig. 7. Mean and range of variation in some mechanical properties of femoral spongy bone according to the direction of loading. Stress values in *pounds force per square inch.*

according to the direction of loading showed that spongy bone is an anisotropic material, *i.e.*, a material that is not equally strong in all directions. The rectangular specimens loaded in the direction of the long axis of the neck of the femur showed the highest, while those loaded in the anterior-posterior direction showed the lowest mean compressive stress. Among the cubic specimens, the highest mean compressive stress was found in specimens loaded in a lateral-medial direction and the lowest in specimens loaded in a superior-inferior direction.

The rectangular specimens loaded in a lateral-medial direction had the highest mean modulus of elasticity and those loaded in the anterior-posterior direction the lowest. The cubic specimens loaded in a lateral-medial direction had the highest mean modulus of elasticity while the lowest was found in the specimens loaded in a superior-inferior direction.

Considerable variation was also found in the energy absorbed to failure, the compressive strain at failure, and the density of the specimens when evaluated with respect to different directions of loading (Fig. 8). The rectangular specimens loaded in a lateral-medial direction had the highest mean energy-absorbing capacity whereas those located in an anterior-posterior direction had the lowest. The highest mean energy-absorbing capacity among the cubic specimens was found in those loaded in a lateral-medial direction and the lowest in the specimens loaded in a superior-inferior direction.

The rectangular specimens loaded in a lateral-medial direction had the highest average compressive strain and those loaded in the direction of the long axis of the neck had the least. The compressive strain of the cubic specimens loaded in a lateral-medial direction far exceeded that of all other specimens. The lowest com-

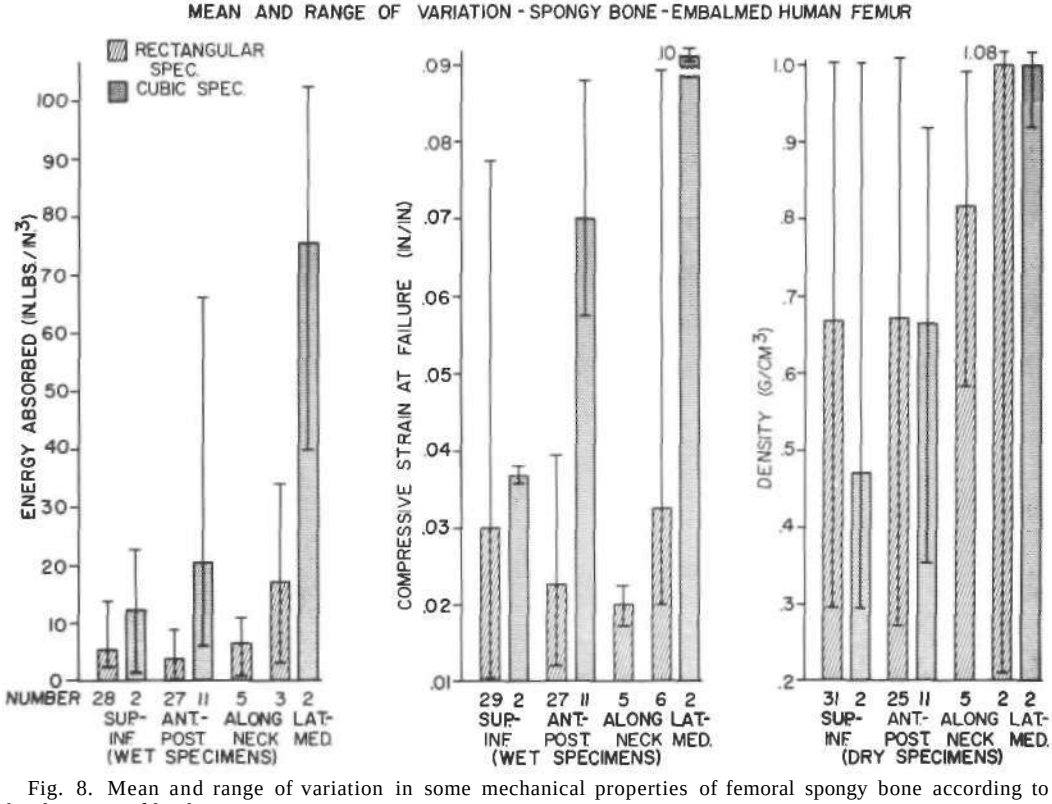

Fig. 8. Mean and range of variation in some mechanical properties of femoral spongy bone according to the direction of loading.

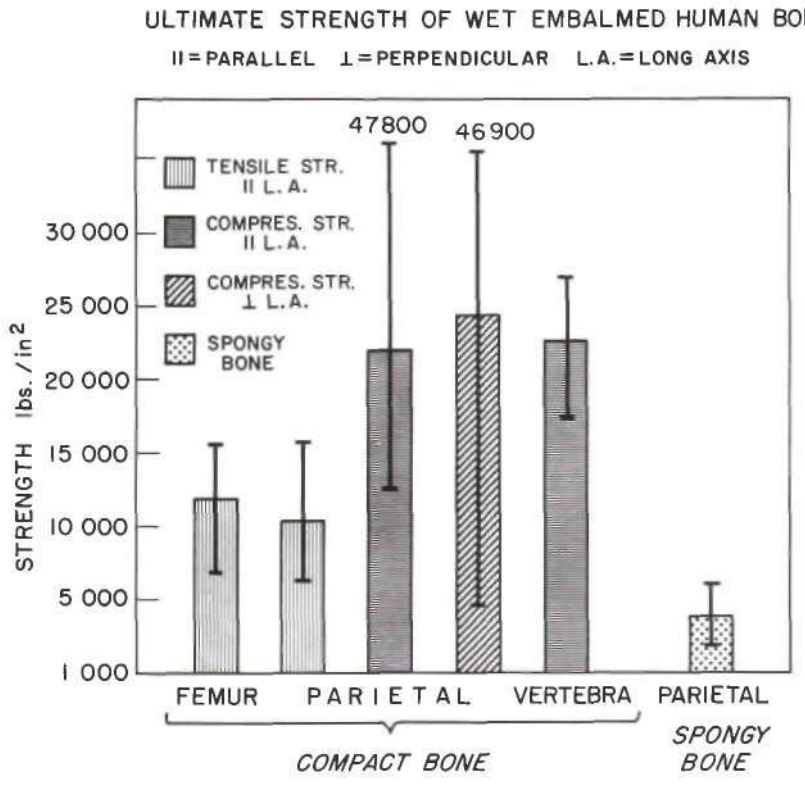

Fig. 9. Mean and range of variation in strength of various bones according to type (compact or spongy) and direction of loading (6).

pressive strain among cubic specimens was found in those loaded in the superior-inferior direction.

Surprising differences were found in the density of specimens cut in different directions. The density of rectangular and cubic specimens cut in the lateral-medial direction was the same but greater than that of any other specimens. The rectangular specimens cut in the superior-inferior and in the anterior-posterior direction were the least dense. Cubic specimens were the least dense when cut in the superior-inferior direction. These differences in density of the specimens suggest directional variation in the orientation and abundance of trabeculae in various parts of the femur.

A statistical analysis of the means for the various mechanical properties with respect to the direction of loading revealed the following significant differences. The variation between the energy absorbed by rectangular specimens, loaded in the lateral-medial direction, was significantly greater at the 0.01 level than that of the specimens subjected to anterior-posterior and to superior-inferior loading. The difference between the maximum compressive strain (found in lateral-medial loading) and the minimum strain (found in specimens loaded in the direction of the long axis of the neck) was significant at approximately the 0.04 level. No other significant differences were found between the means for the other mechanical properties when analyzed with respect to the direction in which the specimens were cut and loaded.

Although spongy bone is much weaker than compact bone (Fig. 9), its foam-like structure makes it a good energy-absorbing material, as demonstrated experimentally more than a century ago by Dr. Physick *(19)* and more recently suggested by Evans, Pedersen, and Lissner *(10).* The presence of fat, marrow substance, and blood in the

TABLE 1. RANGE OF VARIATION IN THE FATIGUE LIFE OF HUMAN METATARSALS TESTED WITH A LOAD OF 15 LB. (*15*)

Metatarsal	Dry-Tested		Wet-Tested	
	No. of Specimens	Repetitions to Failure	No. of Specimens	Repetitions to Failure
II	7	2,000- 1,343,000	3	151,000-11,117,000
III	9	1,000- 870,000	2	6,860,000-13,908,000
IV	6	14,000- 2,273,000	3	150,000- 1,177,000
V	8	10,000-10,297,000	2	195,000- 521,000

interstices of spongy bone in the living condition enhances its energy-absorbing capacity by making it act like a quasi-hydrostatic system. The capacity of bone to absorb energy is one of its important mechanical properties as far as fracture mechanics is concerned because, as pointed out by Lissner and Evans *(16),* all physical injuries arise from the absorption of energy. Most fractures are produced by impacts or blows and thus involve energy absorption.

Another mechanical property of bone to be considered is its fatigue life. This is especially important in relation to march, stress, or fatigue fractures which are most common in the metatarsal bones although they have also been reported in other bones. These fractures are thought to be the result of repetitive loading such as occurs during marching, hence the name "march" fracture.

The only investigation known to me on the fatigue life of intact bones is one we made several years ago *(15).* In this study the strength of intact human metatarsal bones was determined by loading them to failure in a Sonntag Flexure Fatigue machine equipped with an automatic counter (which recorded the number of cycles to failure) and shutoff. The chief advantage in using this type of fatigue machine is that it has an inertia force-compensator spring which absorbs or eliminates all unknown inertia forces. Consequently, the force in the specimen being tested, regardless of its rigidity, is equal to the known force produced by the oscillator assembly.

Forty-one bones were tested with a force of 15 lbs. (the maximum that could be applied with our machine), 3 bones with 12 lbs., and 8 bones with 10 lbs. Only the second through fifth metatarsals were tested because the first one was too large for the fatigue machine. The influence of moisture upon the fatigue life of the specimens was investigated in 10 bones by allowing water to drip on them during a test. The bones were not degreased and all were tested at room temperature. None of the bones exhibited any known pathologic condition. In order to hold the bone in the fatigue machine during a test, the ends were embedded in Selectron 5026 plastic. The number of repetitions to failure was automatically recorded and the machine shut off as soon as the specimen broke. A cycle means the bone is bent once up and once down.

Comparison of the results obtained for the wet- and the dry-tested specimens showed that drying tended to decrease the fatigue life of the bones (Table 1). The probable explanation is that drying increased the modulus of elasticity of the bone and hence the specimens were stiffer. The number of repetitions to failure, with a 15-lb. force, varied from 1,000 to 10,297,000 for the dry specimens and from 150,000 to 13,908,000 for the wet specimens. Metatarsals 2 and 3 showed the greatest fatigue life when tested wet. No consistent relations were found between the fatigue life of the bones and their size or age of the individuals from whom they were obtained. The type of fracture produced experimentally (Fig. 10) was similar to some reported *(14)* in the clinic literature (Fig. 11).

It is interesting to speculate how long an individual must walk before the metatarsals would be subjected to the same num-

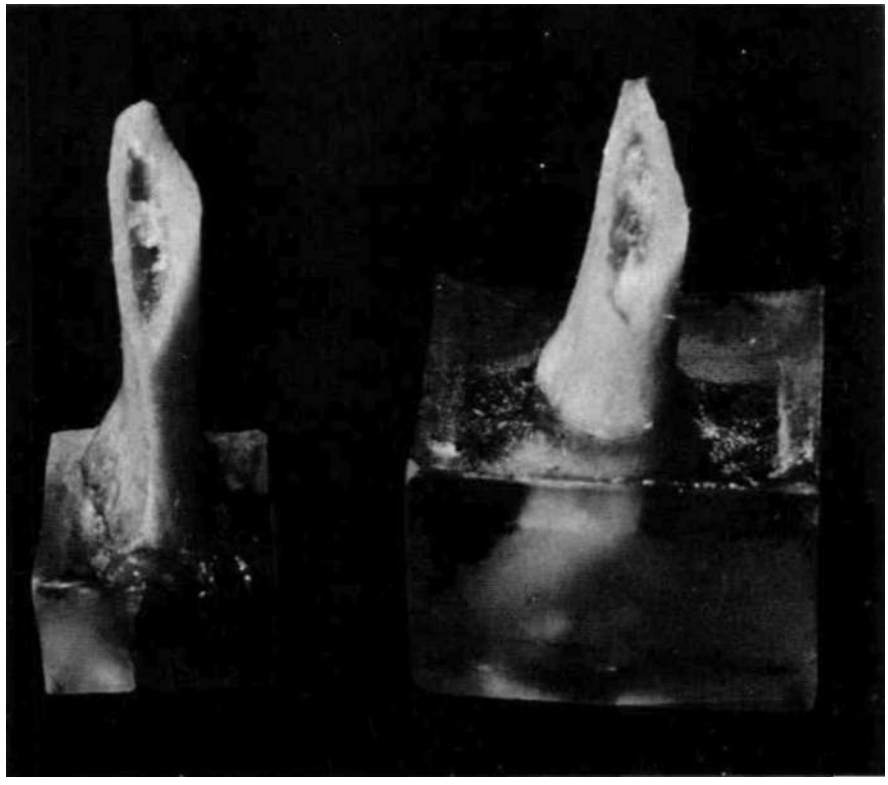

Fig. 10. Experimentally produced fatigue fracture of an intact human metatarsal bone.

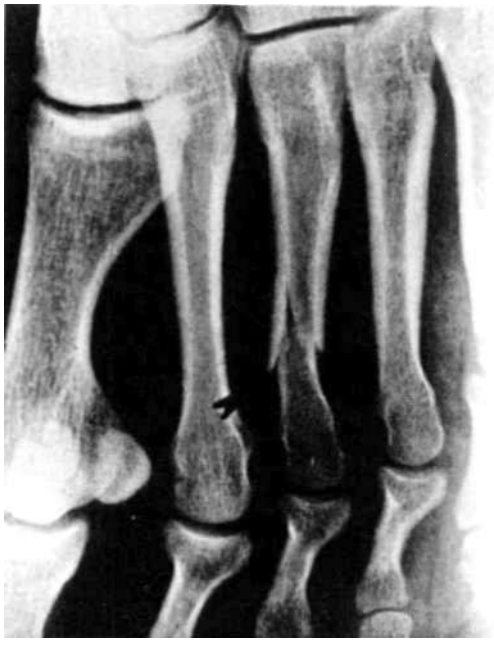

Fig. 11. A clinical fatigue fracture of a metatarsal bone *(14).*

ber of repetitions at which failure occurred in our experiments. If it were assumed that an individual walked at the army pace of 120 steps per min., walking 50 min., resting 10 min., one would have to walk continuously for almost a month before the second metatarsal would be subjected to the number of repetitions at which the failure occurred in the present study. During each cycle of loading, the bone was bent up and down in a vertical plane. The fracture was probably a tensile failure initiated on the side which, at the instance of failure, was the convex or tensile side.

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Dynamic Structure of the Human Foot

Herbert Elftman[1]

THE FOOT is one of the most dynamic structures in the human body. The lively interplay of forces which makes its function possible is easily forgotten and it is too often treated like the graven image of a static structure. The success of modern therapeutic measures in solving other problems has owed much to close cooperation between Nature working from within and assistive devices from without. The forces within the foot can be powerful allies in such a partnership.

The human foot acts in concert with the rest of the body during standing and movement. It provides man with his most effective physical contact with the environment and is especially responsible for successful regulation of initial and final contact of the body with the ground. The foot must also provide adjustable support during the characteristic human occupations of manipulating the environment or of simply standing in line.

Human bipedality was made possible by the redesign of an ancestral foot with five long toes used for the grasping of the limbs of trees. We still testify to our heritage by having a big toe larger than the rest but no longer opposable. The heel bone was brought down into contact with the ground to provide additional area of support. Each of these changes traded an old advantage for a new one and the barter is still going on.

THE FOOT IN MOTION

Walking is more characteristic of human movement than running, since man has substituted cunning in the management of external devices for fast movement of body parts when speed is desired. The foot must constantly adjust to the varying loads imposed upon it. Particularly important are the stresses it must withstand at the initiation of contact with the ground and again at its termination.

INITIATION OF CONTACT

The heel is the first part of the foot to touch the ground in walking. It is consequently entrusted with the delicate mission of gradually bringing the foot to rest on the ground. In running this can be done without the help of the heel since the limb is already in the midst of its backward swing with respect to the body and the ball of the foot can touch the ground at zero velocity.

In walking, the advanced leg has barely started its backward swing with respect to the body when the heel touches the ground. The initial velocity of the ankle after contact is only slightly less than that of the hip joint, making heel-roll imperative. As the ankle approaches zero velocity at ball contact, the forward velocity of the hip joint is preserved by ankle and knee flexion (Fig. 1). Failure to do this properly is one of the most common deficiencies of assistive devices.

The normal human heel is specialized for the part it plays in walking. Resilience is supplied by the construction of the connective tissue under the heel. The collagenous fibers are arranged so as to produce cylindrical compartments filled by more fluid tissues. Since the fluid changes volume only slightly in compression, pressure is accommodated by elastic deformation of the surrounding connective tissue. While

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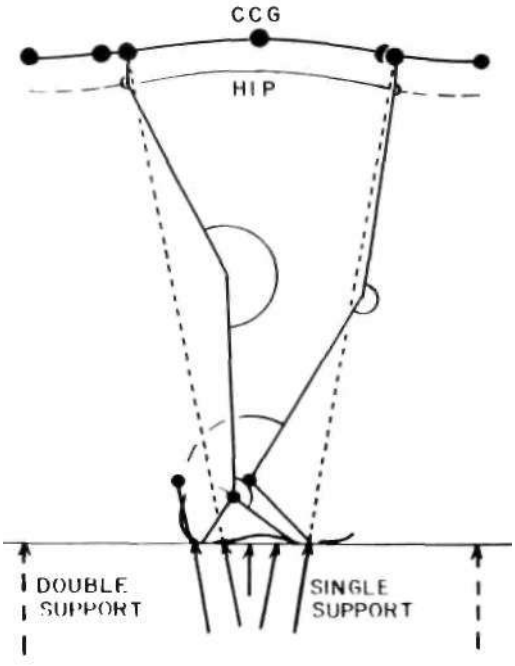

Fig. 1. Forces acting on the foot during two important phases of its activity: (1) completion of heel roll; (2) initiation of rolling off on the hall. *From Elftman, 1967.*

this elastic deformation is taking place the foot rolls forward on the heel. The character of this movement is determined by the contour of the calcaneous combined with the shape which the heel-pad assumes under pressure. Artificial heels can be of assistance if properly shaped, usually achieved when wear erases original design.

TERMINATION OF CONTACT

Although the foot moves only slightly in the interval between ball contact and heel rise, it is subjected to constantly changing stresses. As the body moves forward over the ankle until the knee becomes almost straight, tension is built up in the calf muscles in preparation for the critical events which terminate ball and toe contact. In this phase of walking the transformation of the ape foot into a human foot shows its functional worth. With grasping no longer the chief function of the toes, they have been shortened and the connective tissue pad beneath the ball has become stronger. The great toe has lost its opposability and is permanently aligned parallel to the others. This relieves the peroneus longus muscle of its ancestral responsibility of adducting the hallux and enhances the aid which it gives to the tibialis posterior in resisting splaying of the foot. The first metatarsal and its attendant phalanges retain the size which they had attained in the ape. This led to the accentuated use of this toe during push-off and the important role which the flexor hallucis longus plays in terminal contact with the ground.

Rolling over the ball of the foot has a function similar to that of the heel but acting in reverse. It must control the gradual acceleration of the ankle so that the lower limb as a whole is moving forward with body speed close to the time at which the advanced heel makes contact and double support begins. Here again knee flexion adjusts the relative velocities of the limb segments and allows the calf muscles to push off the limb as it begins its forward swing.

CONTROL OF FOOT POSITION BY HIP AND KNEE

Primary control of foot position is exercised at the hip joint with assistance from the knee when it is flexed. After the primary position of the foot is determined by these distant factors, fine control is added by joints of ankle and foot. The forces and moments which act on the foot are largely determined by the disposition and accelerations of other parts of the body. The importance of knee and hip joints in controlling the spatial relationships of the foot is emphasized frequently by unwelcome responses in these joints to abnormal stresses in the foot.

FUNDAMENTAL ARCHITECTURE OF THE FOOT

The foot consists of 26 bones controlled by 42 muscles and is held together by an almost unbelievable number of ligaments. Fortunately, in the normal performance of its major functions, many of these parts co-operate so closely that an initial work-

able concept of the foot can be based on very few units. The talus is the uppermost of these. When it is removed, the subtalar part of the foot reveals two major divisions: the calcaneus and, articulating with it by the calcaneocuboid joint, a semirigid constellation of bones terminating in the ball of the foot. This leaves the toes jutting out, to become of importance in activities which require forward extension of the base of support beyond the ball.

THE ANKLE-JOINT COMPLEX

The talus is a bony meniscus which allows the movements of the foot with respect to the shank to be divided between a pair of articulations: the subtalar below and the ankle joint above. Since the same external forces act on both joints, the normal body is interested in their combined movement but the clinician is frequently faced with the results of differential insult.

In the ankle joint, normal pressure is transmitted from the tibia to the trochlear surface of the talus and lateral bending moments are resisted, within limits, by the malleoli and ligaments. When the joint is compressed, as in weight bearing, the instant axis is determined by the curvatures of the surfaces in contact at the moment. The classical concept of an invariant axis passing horizontally through the lateral malleolus to emerge just below the medial malleolus has been revised in recent years. Barnett and Napier (1952) have described the difference in curvature between the parts of the talus used as movement progresses. Close and Inman (1952) have emphasized a component of vertical rotation conforming to the curved lateral surface of the talus. Both of these factors are sufficiently variable to require assessment in each individual.

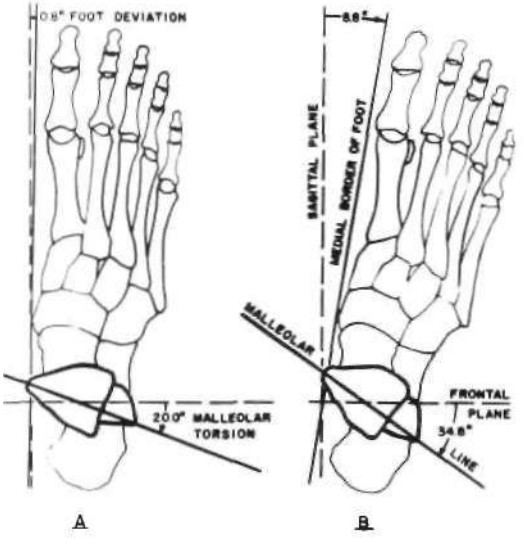

Fig. 2. Range of variation in the orientation of the axis of the ankle joint. Two-thirds of the individuals measured were within the limits shown here. *From Elftman, 1945.*

Even more variable is the orientation of the axis of the ankle joint with respect to the foot and to the transverse axis of the knee. The situation in any individual can be estimated by observing the position of the malleoli; the results of such measurements recorded by Elftman (1945) are shown in Figure 2. It is obvious that the orientation of the ankle joint determines the plane in which dorsi- and plantar flexion occur and this influences the amount of movement required in the subtalar joint.

The subtalar joint is guided in its movement, when it is under compression, by the areas of contact between the calcaneus and the lower surface of the talus. These surfaces are beautifully sculptured to form parts of a helical or screw-shaped surface. The helix is right-handed in the right foot; the resulting advance of the talus during eversion is important for the control of the transverse tarsal joint, but may be neglected during consideration of the ankle. For this purpose the major axis of the helix, also called the compromise axis, suffices. Its position in one foot is shown in Figure 3. This axis emerges from the talus so as to pierce the tendon of the tibialis anterior; its other end is variably located on the lateral surface of the calcaneus. The movements about this axis are called inversion and eversion.

The obliquity of this axis confers on the subtalar joint its most significant proper-

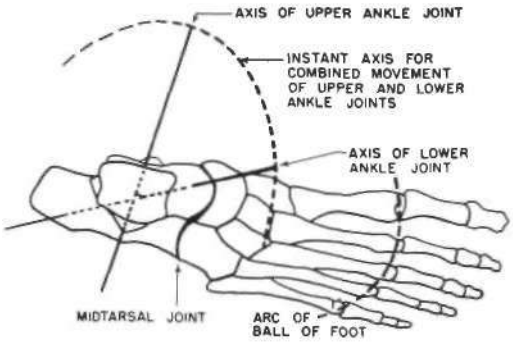

Fig. 3. The instant axis for the combined movement in the upper ankle joint and the subtalar joint lies in the thin disc represented by the dashed circle. Attention is also called to another variable functional feature, the arc of the ball of the foot. *From Elftman, 1954.*

ties. Indispensable for our ancestors in tree climbing, it is still our chief accommodation to rough terrain. Its large component of vertical rotation gives us the possibility of transverse rotation at the ankle under gravitational control.

Since the ankle joint and the subtalar joint are not subject to independent regulation, the resultant movement when the two are combined is of greater practical value than the separate components. The location of this resultant axis is indicated in Figure 3. If the two joint axes actually intersected, the resultant would lie in the plane determined by the two axes. Since they almost intersect, but not quite, the resultant is confined within a thin disc which may be treated as a plane for practical purposes. Once this plane is determined, the problem of substituting new artificial axes for the old ones is simplified.

Movement in the ankle-joint complex is controlled by: (1) moments due to the ground reaction; (2) constraints due to joint surfaces and ligaments; and (3) moments produced by the leg muscles which pass over the ankle. The part played by the ankle muscles can be studied quantitatively from the data shown in Figure 4. This is essentially an oblique section through the ankle oriented so as to include the axes of the ankle joint and the subtalar joint. The lever arms of the muscles with respect to these axes can be read from the diagram; the relative maximum strengths of the muscles are proportional to the areas of the circles which represent them. The resultant moment of various muscle combinations can then be found. Important points to note are: (1) the tibialis anterior is a dorsiflexor and not an invertor in this position; (2) the gastrocnemius and soleus are strong invertors as well as plantar flexors; (3) the peroneal muscles are stronger for eversion than for plantar flexion.

TRANSVERSE TARSAL JOINT

The part of the foot which lies immediately in front of the talus and calcaneus forms a semirigid unit articulating with the rear part of the foot by means of two joints, the calcaneocuboid and the talonavicular. Since they act together much of the time, it is convenient to call the combination the transverse tarsal joint.

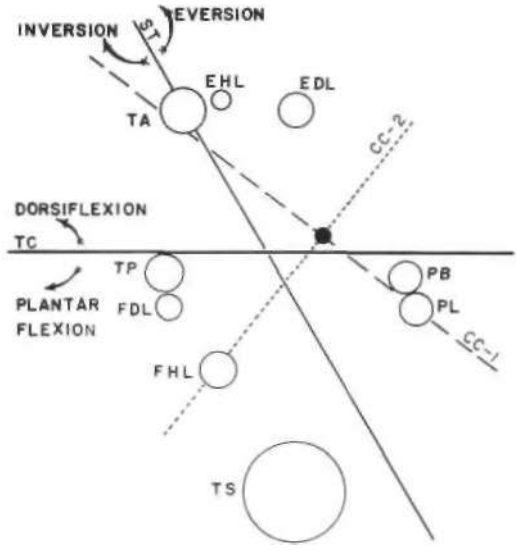

Fig. 4. Muscular control of the ankle. The figure is essentially a section through the right ankle in the plane of the disc shown in Figure 3 and includes the ankle-joint axis *(TC)* and subtalar axis *(ST).* The circles representing the muscles are proportional in area to the physiologic cross sections. The muscles may be identified by their initials, *e.g.,* triceps surae *(TS). From Elftman, 1960.*

The calcaneocuboid joint was described as a saddle-shaped joint by Adolf Fick in 1854; only one other joint of this type is present in man, at the base of the first metacarpal. More than a century elapsed before an adequate description of this joint was provided by Elftman in 1960. For practical purposes a simplified description will suffice. The principal axis (labeled CC in Fig. 5) passes obliquely through the calcaneus in such a fashion that an extension of it would almost intersect the subtalar axis in the neck of the talus. Associated with the major movement of rotation about this axis is a slight translation parallel to the axis. The total movement is known as supination and pronation. The man in the street calls these raising and lowering of the arch.

The talonavicular joint is the controlling element in the transverse tarsal joint complex. The head of the talus is a cam of ellipsoidal shape which is not concentric about the subtalar axis but makes a considerable angle with respect to it. As a consequence of this, rotation of the head of the talus during rotation about the subtalar axis changes its orientation, and movement in the transverse tarsal joint ensues to bring the navicular concavity to a conformable position. The important thing to remember is that inversion produces supination and eversion causes pronation. At the extremes of this range of association, the transverse tarsal joint becomes independent of the subtalar in extremely pronated (flat) feet and the subtalar motion can occur alone at extreme supination.

BALL OF THE FOOT

The structures which allow the heads of the metatarsals to transmit pressure to the ground consist of connective tissue and skin which have been modified in the human foot to spread the pressure in the hope of preventing painful concentrations. When weight is not borne by this region, a transverse metatarsal arch is visible. Even slight pressure is sufficient to bring the heads of the metatarsals in alignment with the ground and the arch disappears. The extreme variability in the lengths of the metatarsals has important consequences for foot action. The distribution of pressure as the heel is raised is very closely dependent on the contour of a line connecting the metatarsal heads, as shown in Figure 3. Morton (1935) has stressed the difficulties resulting from first metatarsals

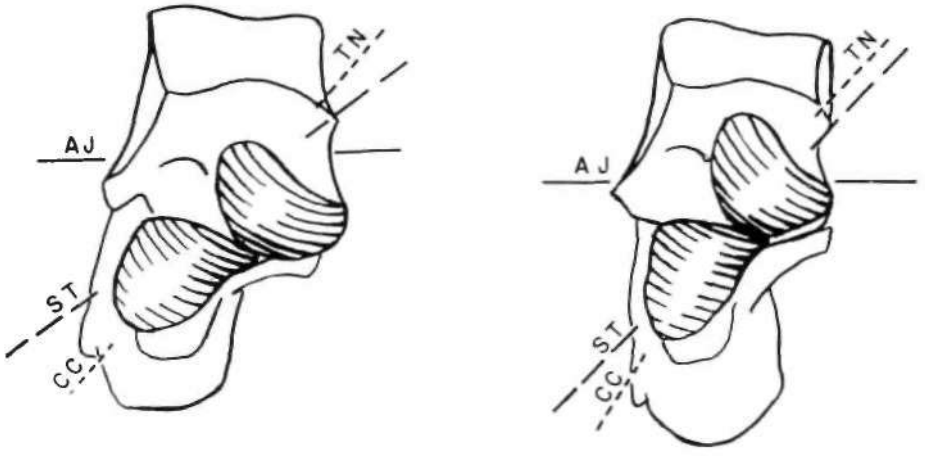

Fig. 5. Transverse tarsal joint, pronated at left, supinated at right. The joint axes are labeled as follows: *AJ,* ankle joint; *ST,* subtalar; *CC,* calcaneocuboid; *TN,* talonavicular. When the heel is placed on the ground in the supinated position, inversion in the subtalar joint restores the vertical orientation of the shank and rotates the head of the talus so as to lock the transverse tarsal joint.

which are short or have posteriorly located sesamoids. Equally disastrous effects can come from contours which are sharply curved or hairpin in shape.

Among a number of variable features in this part of the foot is the extent to which the base of the fifth metatarsal transmits weight to the ground. Another condition, splaying of the foot, can result when the cooperative efforts of the tibialis posterior and the peroneus longus are insufficient to give transverse stability.

TOES

Although human toes can be used for grasping when occasion demands, their customary use is accessory to the ball of the foot which lies behind them. The toes are the anchors for the long flexors which play an important part in managing the ankle-joint complex. By differential contraction of the flexors of the toes it is possible to adjust the distribution of pressure between parts of the ball of the foot. Because of the strength of the big toe and the long flexor attached to it, this part of the foot is usually the last to leave the ground and contributes the final touch to the control of movement.

CONTROL OF THE FOOT BY THE HEEL

When the body rolls forward on the heel until the foot rests on the ground, the position which the foot assumes is determined by the manner in which the calcaneous rolls forward. Proper contouring of the sole of the shoe where the heel nests in it will not only provide assistive forces but will also originate sensory feedback to stimulate better foot alignment.

If the heel cup is so constructed that its anteromedial quadrant is elevated, the calcaneus will come to rest with a predetermined amount of inversion about the subtalar axis. This places the contact area of the calcaneus more nearly under the vertical thrust of the body, decreasing its rotational moment. Since the ankle-joint axis strives for a horizontal position, the talus is forced into inversion and this drives the transverse tarsal joint into supination. Sensory feedback, in the course of a few steps, will encourage the hip joint to bring the foot down in a slightly toed-in position, thus restoring the knee joint to its usual orientation.

The details of the sculpturing of the heel cup need not be left to chance since the desired conformation of the internal

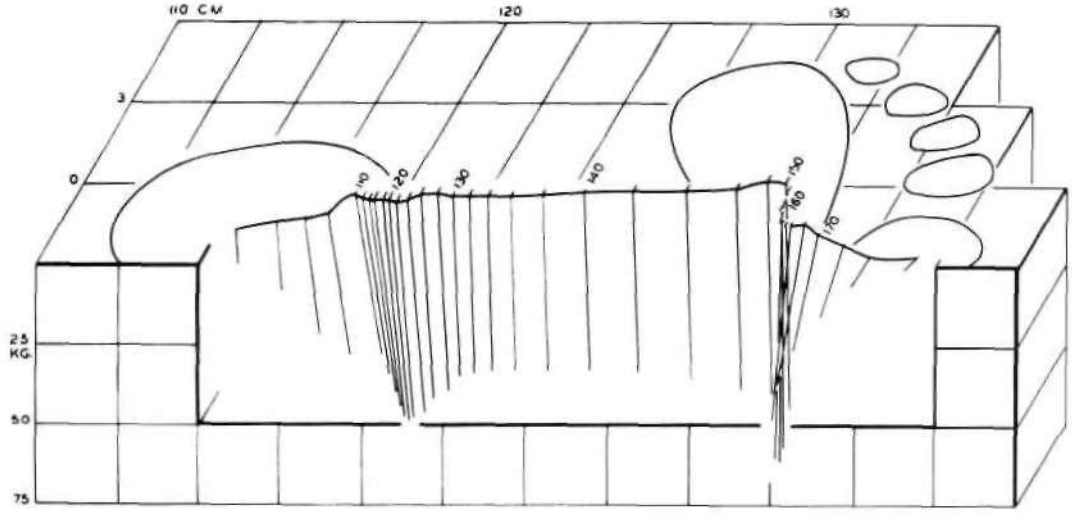

Fig. 6. Force plate record of the ground reaction acting on the foot of J. T. Manter during a step described by Elftman, 1939.

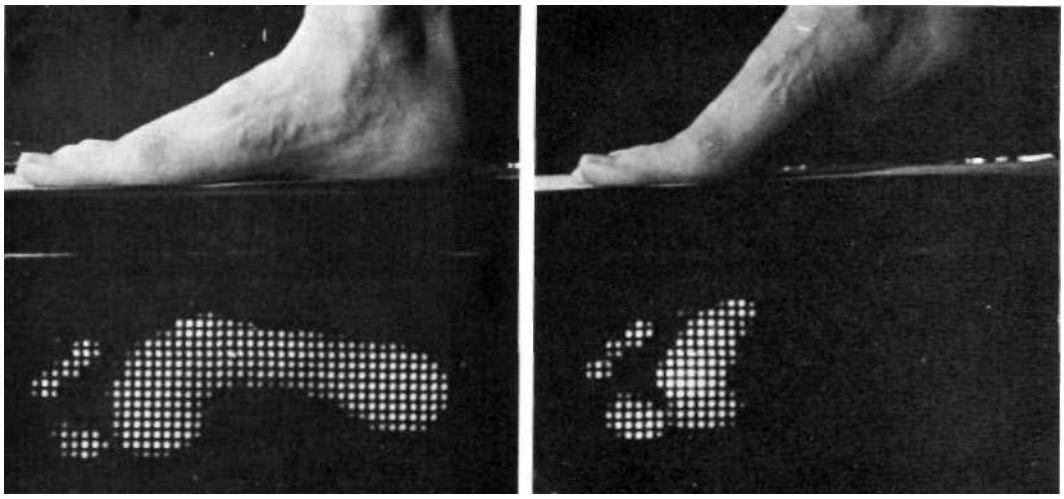

Fig. 7. Barograph record of the distribution of pressure at two phases of the step.

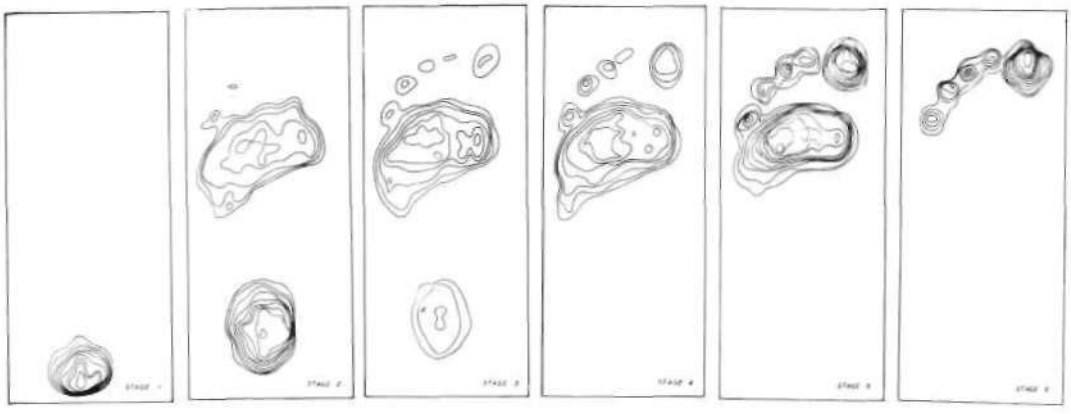

Fig. 8. Load distribution on the human foot during one step of J. T. Manter. (Isobars at 4 lb. per sq. in.) The records made on the original barograph and published in Elftman, 1934, were measured after calibration of the pressure transducer.

architecture of the foot is almost identical with that which it assumes when the subject stands on an inclined plane. Instant orthotics can be achieved by placing the proper compound in the shoes and having the subject stand in them, with heels supported at a proper elevation, to impress the functional shape.

MEASUREMENT OF FOOT FUNCTION

The foot is sandwiched between the pressure of the ground below and the weight and inertia forces of the body above. Since these are the forces to which the foot must accommodate, their measurement assumes primary importance.

The total pressure of the ground on the foot and the point at which its resultant is applied can be measured easily when the individual is standing. The only equipment needed consists of three reasonably accurate scales and a ruler. The usefulness of the information which can be obtained should not be underestimated; it is sufficient to tell whether many therapeutic devices achieve their objectives.

When the body is in motion, measurement of the ground reaction is more important and becomes more difficult. This can be accomplished by means of force plates, the earliest results of which are shown in Figure 6 from Elftman (1939). From data such as this and photographic determination of the location of joint axes, muscle moments and joint forces can be obtained.

In foot problems the distribution of the ground reaction over the foot is frequently of greater interest than its total value. Many interesting methods of making such measurements have been recorded and some are still useful; they have been reviewed by Elftman (1934). Since the distribution of pressure changes in the course of movement, instantaneous recording is of value. This can be accomplished by means of the barograph, introduced by Elftman in 1934. The changes in area of a pressure transducer placed under the foot are recorded photographically. Figure 7 shows two phases of a step; when the pressure is on the ball of the foot the structural characteristics of this region reveal themselves. Calibration of the pressure transducer allows the derivation of quantitative data from the photographic record. In Figure 8 it is even possible to recognize the concentration of pressure under the sesamoid bones beneath the head of the first metatarsal.

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The Influence of the Foot-Ankle Complex on the Proximal Skeletal Structures[1]

Verne T. Inman, M.D[2]

Since the human foot seems to have evolved concomitantly with bipedal locomotion, it seems unreasonable to study its functions isolated from those of the remaining segments of the lower extremity *(3).* It is my belief that a complete understanding of the foot can only be obtained by relating it to the other segments of the lower extremity.

Walking is more than merely placing one foot before the other. There occurs a series of axial, rotatory movements which can be measured in a horizontal plane; many of these are familiar to any observant person. I call your attention to the obvious rotations of the pelvis and the shoulder girdle. What is not so apparent is that similar horizontal rotations occur in the femoral and tibial segments of the extremities. The tibiae rotate about their long axes internally during swing phase and into the first part of stance phase and rotate externally during the latter part of stance; this motion continues until the toes leave the ground. The amount of these rotations is subject to marked individual variations. The minimal amount of horizontal rotation of the tibia in space was recorded in our series of twelve male subjects to be 13 deg. and the maximal to be 25 deg., with an average of 19 deg.

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These rotations are particularly interesting when one reflects that a great portion of this rotation occurs when the foot is firmly placed on the floor; normally, the shoe does not slip but remains fixed. The rotations, however, generate a torque which can be recorded from force plates. This torque is of considerable magnitude, being measured at 60 to 75 in.-lb.

We are now confronted with several observations which require an explanation. First is the measured horizontal rotation of the tibia of 19 deg. (plus or minus) about its long axis while weight is being borne on the foot. Second is the fact that the foot does not slip on the floor, although it may on a slippery surface. Finally, there is a measurable torque that may be recorded. A mechanism which would explain all three of these observations would have to have the following characteristics: It would permit the abovementioned rotations to occur, but it would offer resistance to them of a magnitude such that they would be transmitted through the foot to the floor and would be recorded on the force plate as torques.

The question immediately arises whether all or a portion of this mechanism is located in the ankle joint. Without presenting all the data concerning the mechanics of the ankle joint, one can state that the plantar flexion and dorsiflexion of the ankle joint which occur during the stance phase of walking do produce relative rotations of the talus in relation to the tibia. These rotations, however, are in the opposite direction to absorb the horizontal rotations of the leg.

The structure that permits the leg to rotate upon a fixed foot is the subtalar joint. From the viewpoint of descriptive anatomy, the subtalar joint appears to be a complicated structure. Functionally, it is simple. It is a single-axis joint and its major movement is essentially that of a mitered hinge. In normal standing, the axis of the subtalar joint is inclined at a little less than 45 deg. to the horizontal (studies of small numbers of cadaver specimens by both Manter (9) and Isman and Inman *(4)* have shown their averages to be about 42 deg.).

A simple mechanical model may be constructed consisting of two pieces of wood attached to each other at a right angle by means of a hinge whose axis is at 45 deg. (Fig. 1). Rotation of one board causes rotation of the other. In a similar way, rotation of the leg about the longitudinal axis is transmitted to the foot, imposing rotation on it about its long axis. The opposite is also true. Rotation of the foot from a pronated to a supinated position results in a horizontal rotation of the leg through its longitudinal axis.

We have now identified a mechanism located within the foot which is nicely designed to permit the horizontal rotations of the leg as measured. The next step is to determine the behavior of this mechanism during locomotion and its effect upon the remaining structures in the foot.

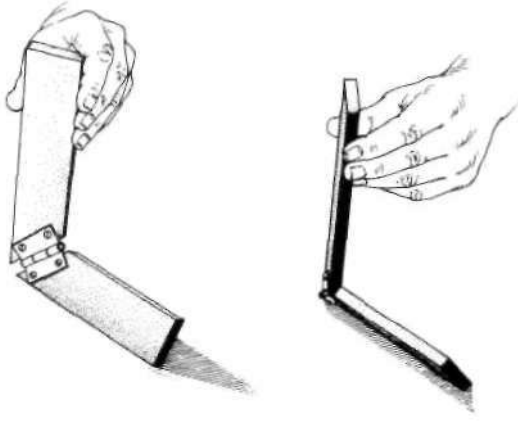

Fig. 1. Simple mechanical model constructed to show motion about the subtalar joint.

However, many events are occurring in the leg, ankle, and foot during stance phase. These events are sequential and interdependent. The behavior of each anatomical part is dependent upon some immediate prior event. This makes the presentation extremely complex and difficult. In an attempt to simplify it, the entire stance phase, from heel-strike to toe-off, will be divided into three periods. The first and third periods are when both feet are in contact with the floor (double weight bearing). The second period is when the foot is flat on the floor and weight is borne solely on that foot (single weight bearing). The first and third periods are naturally equal in length if the gait is symmetrical, and the third period equals the time of the swing phase of the opposite leg. The time of the *second* period is twice that of the first or the third and is equal to the sum of the first and third.

The first period extends from the moment of heel-strike to full weight bearing, which occurs when the toe of the opposite foot leaves the ground. The impact of the heel as it contacts the floor and the rapid loading of the foot result in a vertical floor reaction, which exceeds the total body weight by over 20 per cent. The suddenness of this impact is partially absorbed by lowering of the body through plantar flexion of the ankle. The entire leg continues to rotate internally during the loading of the foot, or through this entire first period. The internal rotation of the leg, through the action of the subtalar mechanism, causes the foot to pronate.

During this first period no muscular activity in the leg or foot is recorded by the EMG except in the anteriorly placed extensor group which is acting to prevent "foot slap" (Fig. 2) *(8).* The degree of pronation that occurs in the foot is dependent solely on the ranges of motion in the intrinsic articulations of the foot and on the laxity of the ligamentous structures.

Before turning our attention to the second period of stance phase, some further

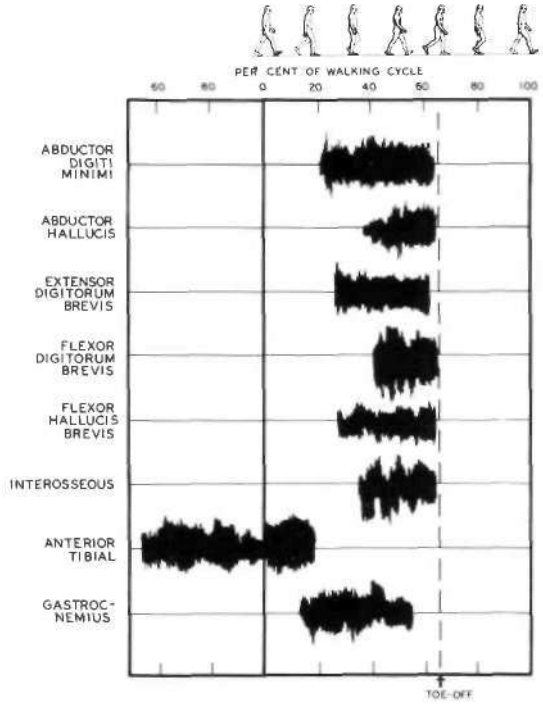

Fig. 2. Recordings of myoelectric signals of muscles of the leg during the stance phase of walking. *Courtesy J. Bone JointSurg.*

remarks are necessary concerning the subtalar joint. If the axis of the subtalar joint is inclined at 45 deg. to the horizontal, the angular rotation of the leg will be the same as the angular rotation of the foot about its long axis (pronation and supination). If, however, the axis of the subtalar joint is closer to the horizontal, the same amount of angular rotation of the leg will impose a greater motion upon the foot. Conversely, if the axis of the subtalar joint is closer to the vertical, then the same amount of rotation of the leg will produce less pronation and supination of the foot. This may be the explanation why in flat-footed individuals, in whom the axis may be more horizontal, the same amount of rotatory movement of the leg requires a greater amount of pronation and supination. Conversely, in persons with a cavus foot, in which the axis is more vertically placed, the same amount of rotation of the leg requires less pronation and supination of the foot.

Pronation of the foot seems to reduce all the inherent skeletal stability of the foot. Maximal motion of the midtarsal

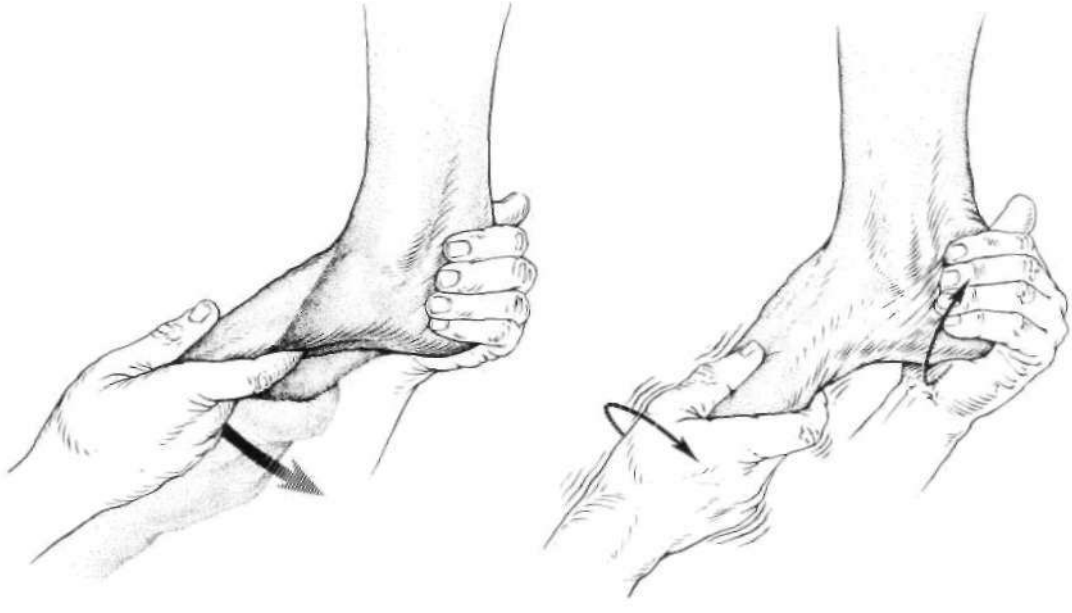

Fig. 3. The foot becomes a pliant structure when it is placed in pronation.

joint is easily demonstrated to occur with the foot pronated. In this position the greatest dimension of the articular surface of the head of the talus is parallel to the transverse axis of the calcaneocuboid joint. Furthermore, since no muscular activity can be detected to add support, the foot becomes a pliant structure able to mold itself to irregularities of the walking surface (Fig. 3).

The second period of stance phase is characterized by vigorous muscular activity. Almost simultaneously, all the intrinsic muscles of the foot, together with the calf musculature, contract (see Fig. 2). The leg reverses the direction of rotation and begins to rotate externally (Fig. 4). The arch of the foot is elevated and the foot supinates with concomitant motion in the subtalar joint. Whether the supinating

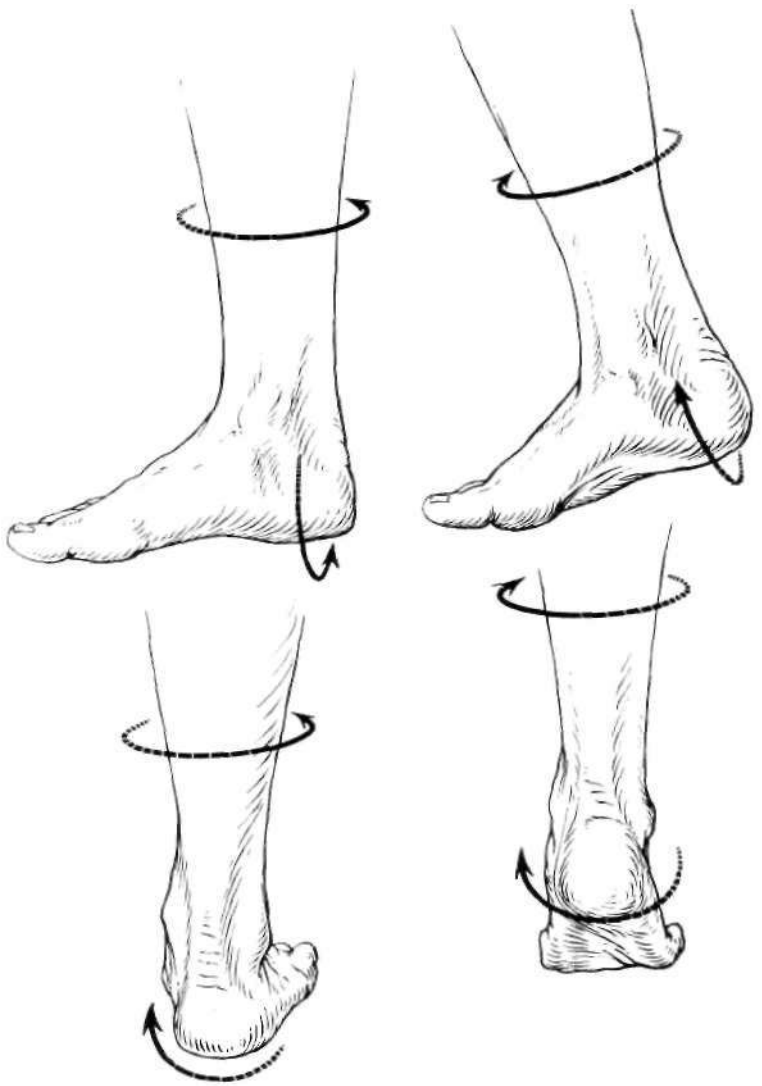

Fig. 4. Sketches showing the reversal of motions that occur between the first and second phases of stance phase.

foot, through the subtalar linkage, initiates the external rotation of the leg or whether it is independently controlled by the external rotator muscles of the hip is not definitely determined. The probability is that both contribute. Certainly some rotating force is imposed from above, since in amputees the same horizontal rotations are discernible in the prosthesis and measurable by a suitable device in the prosthetic ankle.

The amount of motion occurring in the subtalar joint has been measured by Wright *et al. (12).* Although this rotation is not as apparent as ankle-joint motion to the casual observer, it should be noted that the angular movement of the subtalar joint, as measured in degrees, approaches that of the ankle (Fig. 5).

The orthopaedic literature has recently furnished us with experimental evidence of the relative importance of the subtalar joint.

Surgical arthrodesis of this joint has been a frequent and favorite orthopaedic procedure. This procedure has been carried out for stabilization of the flail foot in poliomyelitis, as well as for the relief of flaccid flatfoot and of traumatic arthritis of the subtalar joint after fractures of the calcaneus. Granted that this procedure does give stability and better appearance to the foot, but the effects of the loss of subtalar motion upon the ankle joint have only recently been evaluated.

In 1959 Robins *(11)* reported on a series of 60 patients with flail feet resulting from poliomyelitis who had all had subtalar fusion. They had been followed for a minimum of 10 and a maximum of 25 years. It is interesting to note that when this procedure was carried out on younger individuals they tended to develop a "ball-and-socket" ankle joint (Fig. 6) which had increased rotation to replace the function of the lost subtalar mechanism. Many of these feet were asymptomatic; a few became so unstable as to require braces, and several others showed early but asymptomatic arthritic changes in the ankle joint.

The orthopaedic literature also contains isolated observations on congenital synostoses, between the tarsal bones, which suppress motion of the subtalar joint. Lusby (7), del Sel and Grand *(2),* Jacobs

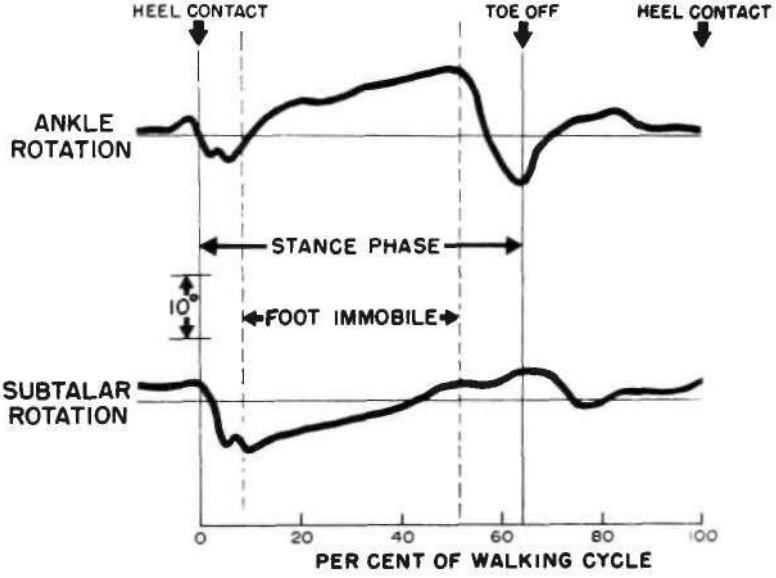

Fig. 5. Rotation about ankle and subtalar joints during the stance phase of walking. *Courtesy J. Bone Joint Surg.*

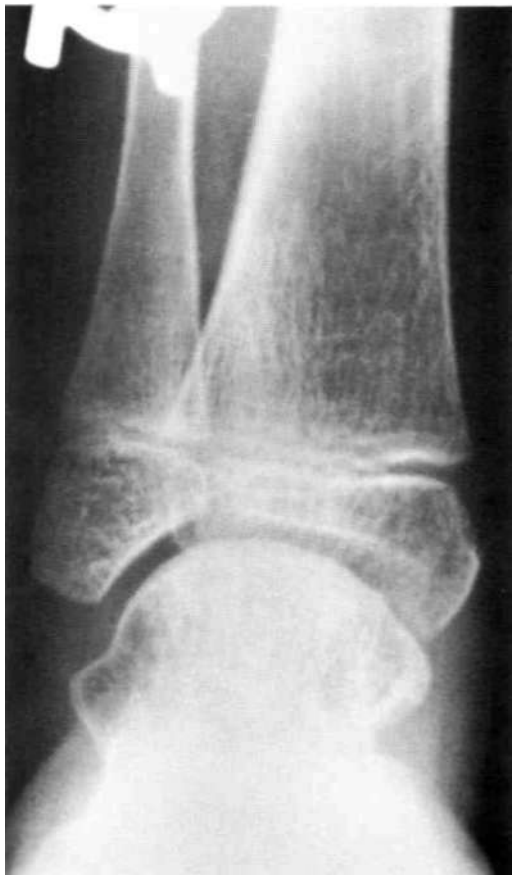

Fig. 6. The "ball-and-socket" ankle joint produced as a result of earlier fusion of the subtalar joint.

(5), Brahme *(1)*, and Lamb (6) all report late effects on the ankle joint caused by the loss of subtalar motion. It seems quite characteristic that such congenital abnormalities also result in ball-and-socket ankle joints. It appears that the failure to provide subtalar motion in the growing child is reflected in the maldevelopment of the ankle joint which must accommodate for the lost subtalar motion.

When a subtalar or a triple arthrodesis is carried out in adults there is a frequent complaint that pain is referred to the lateral side of the ankle. Povacz *(10)* in 1965 reported on a series of 27 patients who had had subtalar arthrodesis; more than half complained of discomfort on the lateral side of their ankle joints.

Because subtalar motion is not as obvious as ankle-joint motion during walking, the importance of this joint has been overlooked. However, all evidence, both experimental and clinical, emphasizes its importance in normal locomotion. This evidence indicates that loss of subtalar motion in congenital abnormalities of the foot or operative procedures designed to obliterate motion in this joint causes, in the child, the development of a ball-and-socket ankle joint with ankle instability. In the adult, surgical arthrodesis of the subtalar joint may result in pain and arthritic changes in the ankle joint.

Because even a ball-and-socket ankle joint cannot absorb all the rotatory movements transmitted through the femur and the tibia, and because a fused subtalar joint may change the pattern of transmission of forces to and from the foot, there may well be effects on the proximal musculoskeletal structures; it should therefore behoove us to watch for them.

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Technical Notes from the Artificial Limb Program

This section of ARTIFICIAL LIMBS *is intended as an outlet for new developments in limb prosthetics and orthotics which, though not deserving of a long feature article, nevertheless ought to be brought to the attention of the readers of this journal. Notes may vary in length from a single paragraph to several pages of manuscript, as appropriate. Elustrations also are acceptable.*

A New Concept in the Swivel Walker

A number of Canadian-type swivel walkers *(1)* (Fig. 1) have been manufactured and used successfully by the Repatriation Department's Central Development Unit (C.D.U.). While satisfactory results were obtained, it was felt that the full potential of the swivel-ankle concept was not being achieved. The two knife edges of the foot-plates resulted in some lateral instability and an abrasive action on the floor surface. During some stages of the walking cycle the knife edges of the foot-plates did not land flat on the floor. This lack of full plantigrade contact caused undesirable rotation of the foot-plate, and to some degree up-

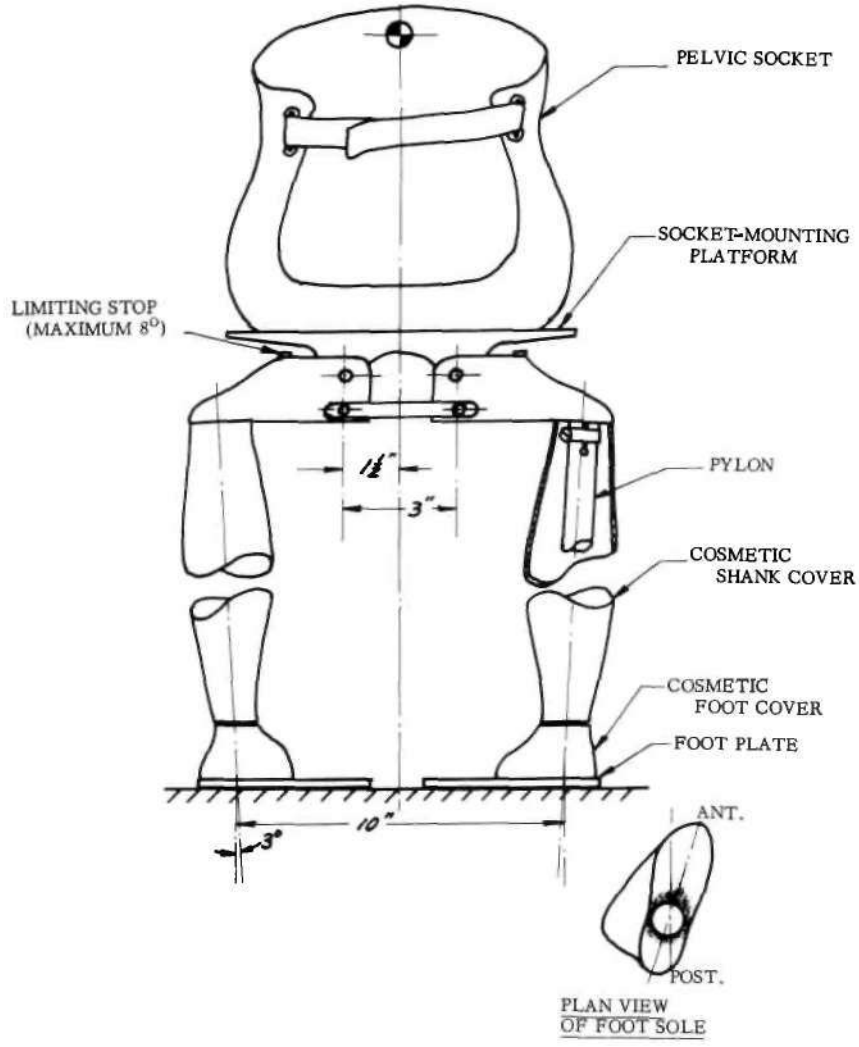

Fig. 1. Canadian-type walker.

set the harmonious relationship between lateral displacement and forward rotation.

Further, it was noticed that, when manufactured and aligned as set out in the Manual (1), the rotation actually commenced before the center of gravity crossed the line of support owing to the inertia produced by lateral thrust of the body in the dynamic condition. This factor seemed to limit forward progression. Moreover, a considerable amount of lateral displacement was apparently necessary in order to maintain a reasonable rate of progression.

The actual tilt of the ankle bearings with respect to the vertical in the Canadian-type swivel walker (a maximum of six deg. with the foot flat) depends upon the position of the patient's center of gravity and thus varies continually during walking. It was felt that if the angle of tilt of the bearings could be maintained in a constant position with respect to the vertical a smoother walking pattern would result.

In a new design (Figs. 2 and 3) suggested by one of the authors (Duncan) to overcome these deficiencies, the torsion-type "ankle" joints originally designed for the ambulator (2) were used. The ankle units are "neutral" in that they can be set for use on either the left or right.

A four-bar linkage is used between the socket and pylons in an attempt to reduce the amount of lateral displacement re-

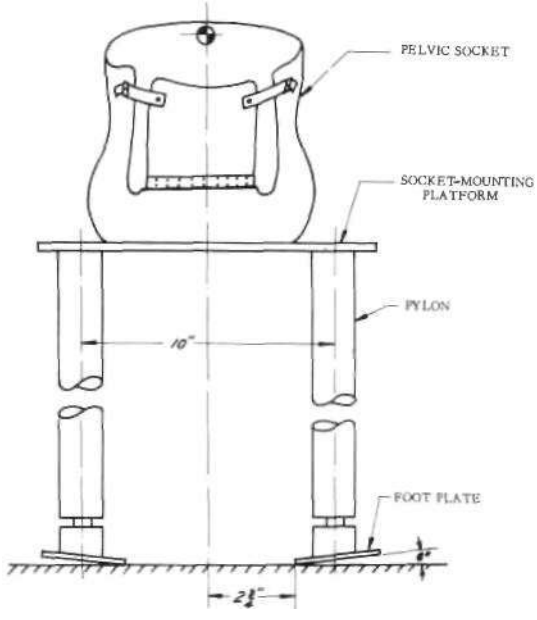

Fig. 2. C.D.U. modified walker.

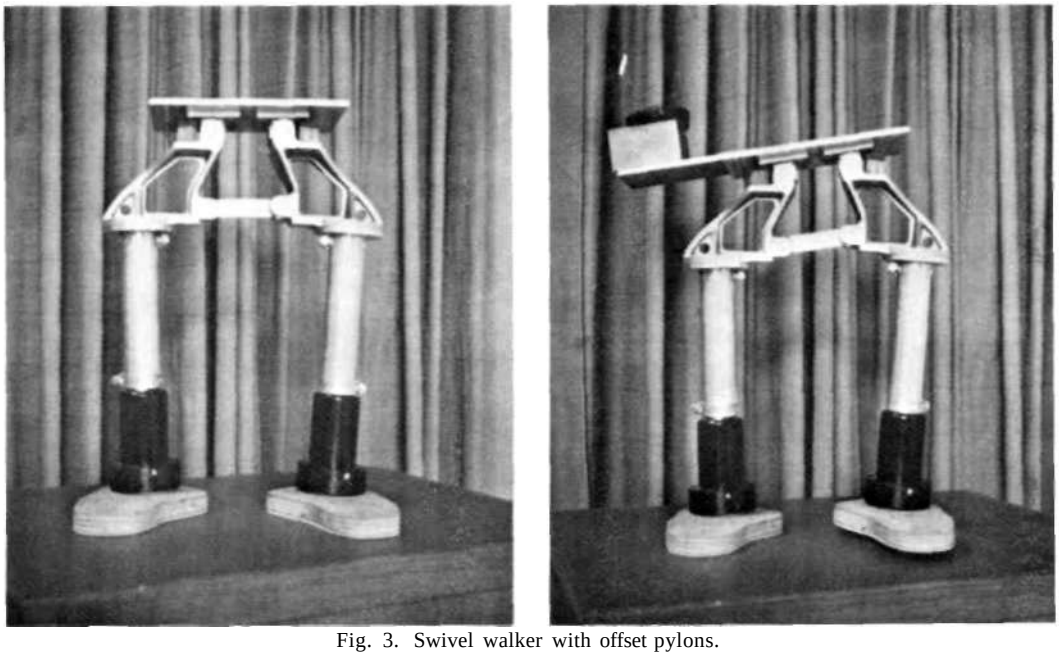

Fig. 3. Swivel walker with offset pylons.

quired during ambulation. With lateral tilt of the pelvic socket, vertical lift of the opposite pylon occurs, thus providing floor clearance for each step taken. Also, the foot-plates are maintained in a horizontal position and thus make total contact with the floor. The stops on the four-bar linkage may be adjusted to vary the amount of lateral tilt and hence floor clearance. Because the axes of the four-bar linkage are more closely spaced than are the lines of support of the Canadian-type walker, the inclusion of small medial extensions on the foot-plates is necessary (Fig. 2). Experience indicates that the significant increase in stability thus achieved offsets the cosmetically undesirable feature of the extension. It is hoped that a suitable transparent material can be found for the foot-plates.

The pylons are adducted at an angle of three deg. to the vertical to accommodate the long rubber torsion bars of the local "ankle" mechanism. If units as originally shown by the Toronto group were used, the pylons could be verticals. The foot flat characteristic offers greater stability both in standing and in walking. The combination of these features (foot flat and a small, fixed degree of socket tilt) contributes to a straighter line of progression than is the case with the Canadian design.

The modified design allows the patient to stand on one foot. Because lateral displacement and rotation can be adjusted independently, the fine positional maneuvering required for duties such as writing at a desk and working at a bench can be achieved more easily.

The pelvic socket can be constructed so as to permit self-entry and exit at walker height by patients with some residual upper-extremity function. Flesh-colored cosmetic shank and foot covers can be placed over the tubular pylons to afford a better appearance.

Experience with Canadian swivel walkers in Melbourne has indicated that the lateral shift or displacement of the body mass is about 3-1/4 in. in each direction in order to produce an average step length of 2-1/2 in. The corresponding amount of lateral shift of the body mass on the modified (C.D.U.) walker is about 2 in. As the time taken for each lateral body sway would, for a given rate of progression, be equal for both designs, the velocity of lateral mass transfer will be proportionally less for the C.D.U. walker. It is suggested, therefore, that the energy consumption would also be reduced similarly with the use of the new design. Limited trials of the one prototype fitted to date appear to support this hypothesis (3).

ACKNOWLEDGMENTS

The authors wish to acknowledge the general assistance and guidance of Dr. R. W. Klein, M. B. E., Director, and Mr. R. E. Spielrein, Senior Engineer, of the Central Development Unit. Their thanks are also due to the Chairman of the Repatriation Commission for his permission to publish this material.

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While this issue of the journal was in preparation, word was received of the death of Mr. Barry in mid-December 1968 at the age of 43. On behalf of the Committee on Prosthetics Research and Development, *Artificial Limbs* expresses deep regret at his passing.

A Suction Socket for the Geriatric Above-Knee Amputee

Although nearly every geriatric above-knee amputee could benefit by the several advantages offered by the total-contact suction socket, it is seldom prescribed for the older patient because of the difficulty encountered in donning sockets of the original design. Instead, relatively loose-fitting quadrilateral sockets, generally without provisions for total contact and suspended by a pelvic band and metal hip joint, are usually provided. Thus, many of the advantages of the more advanced fitting concepts are lost, such as improved control, better proprioception, decreased weight, and greater range of motion.

To eliminate the difficulties encountered in donning the suction socket, a design utilizing an inner socket and an outer shell (Fig. 1) has been devised and used successfully.

Construction of the Socket

A plaster-of-Paris model of the stump is obtained and modified in accordance with standard practices. Over the modified stump model is laminated a flexible inner socket consisting of 90 per cent flexible and 10 per cent rigid polyester resin and four layers of nylon stockinette (Fig. 2). Inner and outer polyvinylalcohol (PVA) bags are used along with a vacuum of approximately 20 in. of Mercury. A total-contact-type suction-socket valve is included as the laminate is laid up. It is important that the resin be distributed evenly (usually by stringing) to avoid the formation of weak areas, especially in the ischial seat and femoral triangle areas.

The outer shell is fabricated over the inner socket after it has been cured. The laminate consists of one layer of Dacron felt, six layers of nylon stockinette, and a mixture of 90 per cent rigid and 10 per cent flexible polyester resin. The proximal borders of the socket and shell are trimmed in the conventional manner except that a cutout must be provided in the proximal lateral aspect of the outer shell to accommodate the Velcro suspension strap. The Velcro strap is attached to the outer shell and passes through a "D" ring that is attached to the inner socket (Fig. 1). An aperture may be cut out of

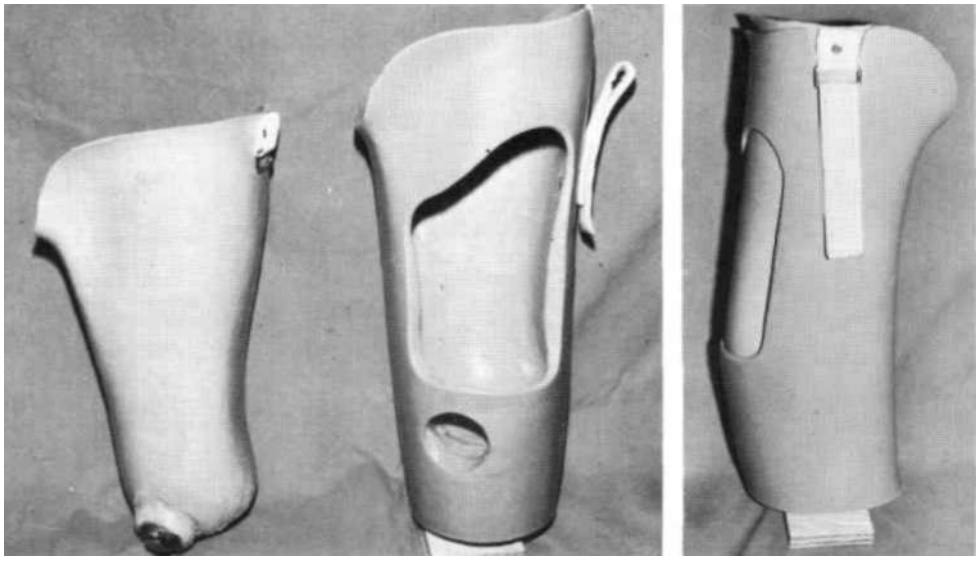

Fig. 1. Two-component type of total-contact above-knee suction socket designed especially for geriatric and generally weakened above-knee amputees. *Left,* flexible inner socket; *center,* rigid outer shell with anterior window; *right,* complete assembly of socket and shell held together by Velcro strap on lateral side.

the anterior wall to permit muscle bulging.

Donning the Prosthesis

To don the prosthesis the patient assumes a sitting position, applies two or three wraps of elastic bandage to the stump (Fig. 3), slides the flexible inner socket over the bandaged stump, and removes the bandage through the aperture for the suction-socket valve (Fig. 4). The valve is then installed, the inner socket is slipped into the outer socket, and the Velcro strap is inserted through the "D" ring, pulled back, and attached to itself. Thus this two-stage procedure for donning the socket eliminates the need for the patient to bend over appreciably, especially when trying to support the body weight on one foot (Fig. 5).

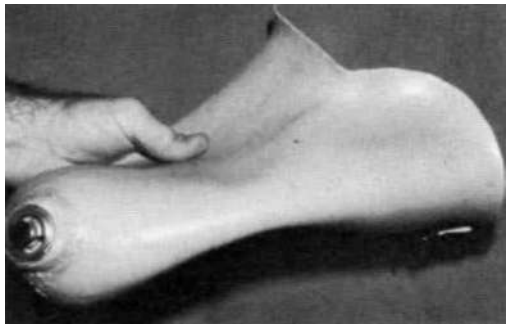

Fig. 2. Photograph of inner socket showing the degree of flexibility afforded by use of a mixture containing 90 per cent flexible polyester resin.

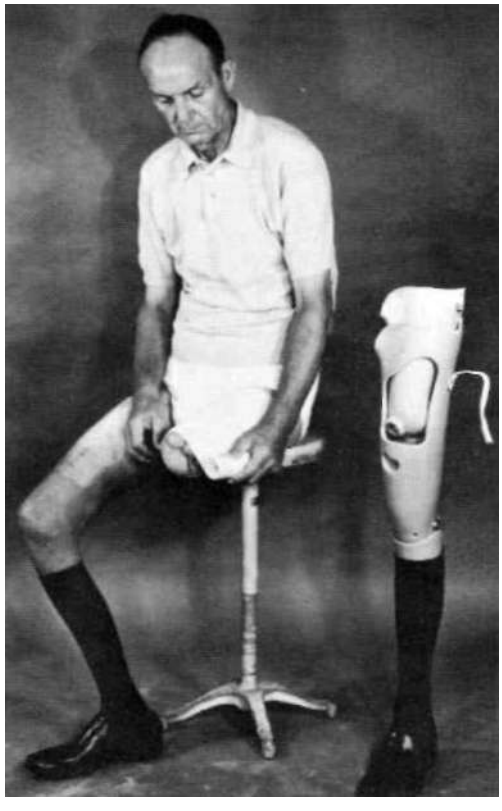

Fig. 3. Application of two or three layers of elastic bandage to the stump is the first step in donning the two-component socket.

Fig. 4. Removal of the elastic bandage from between the stump and inner socket to facilitate donning the two-component socket.

Results

Sockets of the two-component design have been fitted to four patients with excellent results. No problems with loss of suction or donning of the prosthesis from the sitting position were encountered. Ad-

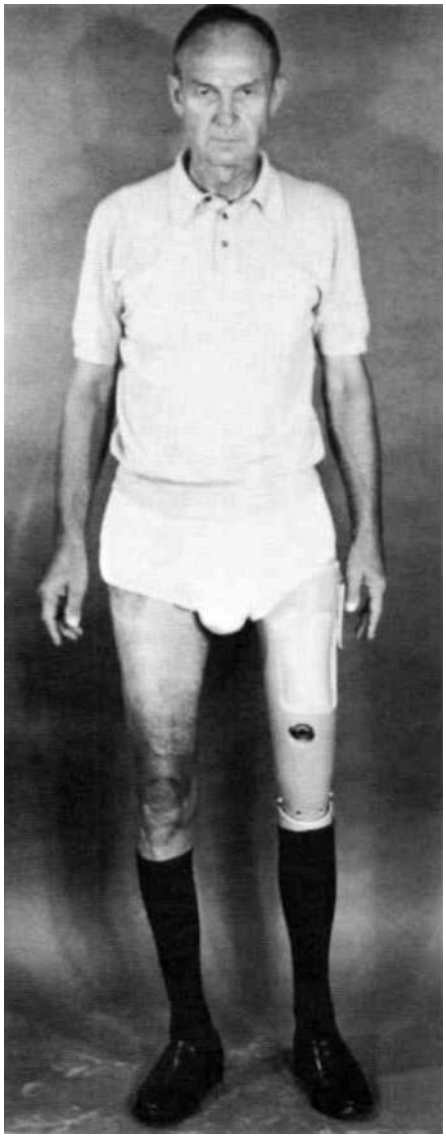

Fig. 5. The two-component total-contact suction-socket prosthesis in place.

justment for atrophy can be made effectively by providing buildups of leather or plastic between the inner socket and outer shell.

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News and Notes

Evaluation of Direct Forming of Below-Knee Sockets

Even before the development of laminated plastic sockets, prosthetists have recognized the desirability of having a material which would be formed over the stump at a temperature low enough not to be painful and with sufficient strength to withstand the loads generally imposed on sockets. The availability of such a material should make it possible to eliminate the need for a male plaster-of-Paris model and the messy conditions inevitable in its preparation. At the same time research groups have been continuously trying to develop techniques for taking impressions of the stump that yield uniform, useful results and require a minimum amount of skill.

A group at the Veterans Administration Prosthetics Center, led by Henry Gardner, have developed a process which eliminates the need for the male model and also reduces the amount of skill required to achieve uniform results, by forming a synthetic balata tube directly over the stump by means of an air-filled pressure sleeve. Total contact is achieved by introduction of a foam-in-place Silastic compound. In addition to the reduction in time and skill required to fabricate the socket, other advantages claimed by the developers are the ease with which modifications can be made to the finished socket and the "feel" of the synthetic balata to the skin.

The CPRD, under the terms of Contract No. SRS 69-9 with the Social and Rehabilitation Service, is conducting a clinical evaluation of the new technique. Clinics at the University of Miami, Duke University, Rancho Los Amigos Hospital, Veterans Administration Sawtelle Hospital, and the VA Regional Office, Syracuse, N.Y., were invited to participate. Prosthetists from these centers were trained in the technique November 6-8 at the VAPC and asked to fit 10 patients, five new amputees that have not been fitted previously and five satisfied PTB wearers, and to keep accurate records of the results.

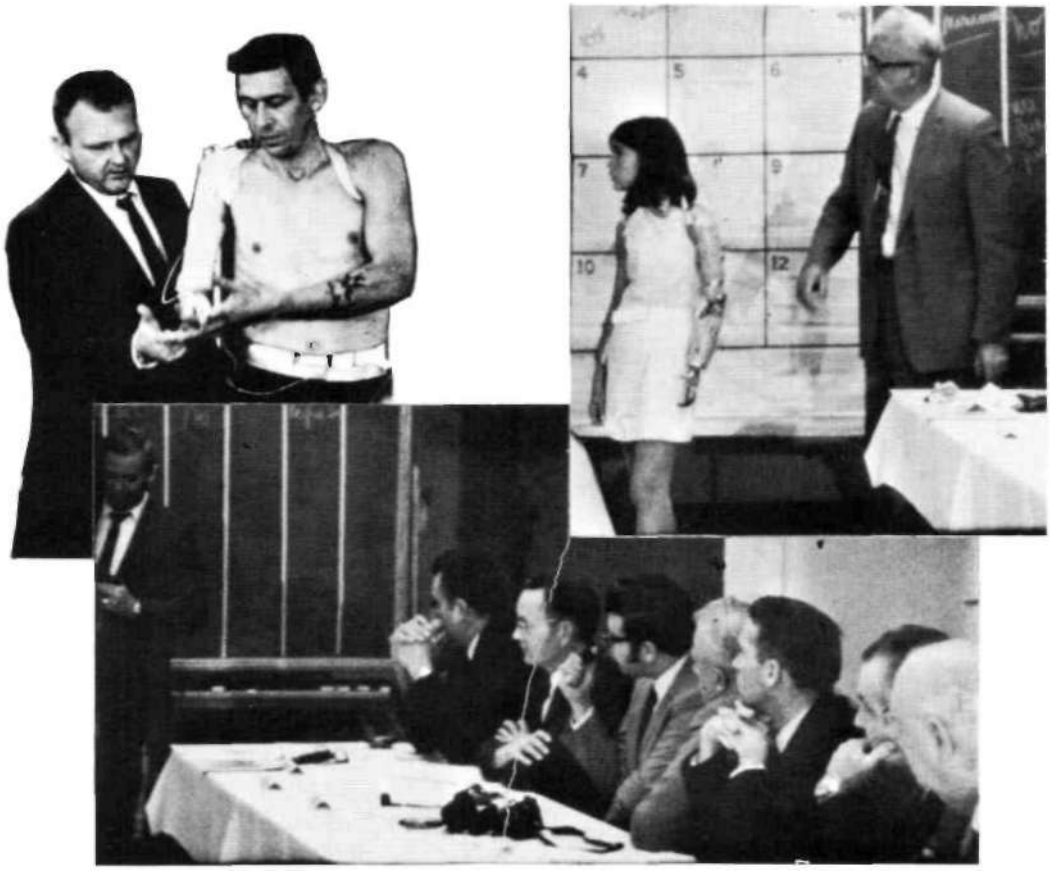
The synthetic balata sockets will be tested for usefulness both as an interim prosthesis and as a definitive, or "permanent," prosthesis. Time studies of the new fabrication method and of conventional lamination methods will be made. Results concerning use as an interim prosthesis will be published as soon as conclusions can be drawn and will not be delayed until results can be obtained on use for long periods as a definitive prosthesis.

Panel on Upper-Extremity Prosthetics

Because of the tendency of the Panel on Upper-Extremity Prosthetics Fitting and the Panel on Upper-Extremity Prosthetics Components to overlap in interest, the two groups have been merged into a single entity known simply as the Panel on Upper-Extremity Prosthetics Components. The chairman is Dr. Edward Peizer, Chief, Bioengineering Research Service of the Veterans Administration Prosthetics Center.

The new group held its first Workshop October 21-23, 1968, in Santa Monica, Calif. The first day was devoted to the study of externally powered elbow systems and in the subsequent two days virtually all other matters of interest in the development of improved upper-extremity prosthetics were considered.

To gather together as much useful information as possible arrangements were made to have at the meeting patients who had been fitted with elbow systems. The seven systems known to be either available or well advanced in development were obtained and fitted to seven patients. The seven systems were those developed by the Army Medical Biomechanical Research Laboratory, Gilmatic, Rancho Los Amigos, Ontario Crippled Children's Centre, VAPC, American Institute for Prosthetic Research, and Liberty Mutual Insurance Company (Boston Arm). The units developed at Rancho Los Amigos and Gilmatic were fitted by those groups,



respectively. The other five were fitted on October 24 by Messrs. Rodney Chupurdia. Maurice LeBlanc, and Thomas Pirrello.

Out of the conference came the recommendation that models of the AMBRL and Gilmatic designs be procured for clinical evaluation studies. It was noted that all of the designs offered had merit and that plans for further evaluation were under way for most of the other designs.

It was also recommended that a similar workshop be conducted on externally powered terminal devices during the Spring of 1969.

Subcommittee on Child Prosthetics Problems

On October 28, 1968, the Subcommittee on Child Prosthetics Problems met in Washington, D.C. In the absence of Dr. George T. Aitken, Mr. Colin A. McLaurin was chairman. The various development and evaluation projects concerned with child prosthetics were reviewed and recommendations for future action were made.

The Subcommittee approved the Child Amputee Clinic, Shriners Hospital, Portland, Oregon, as a participating clinic, raising the total to 28. The Subcommittee is in contact with 16 additional clinics that are interested in becoming affiliated with the participating clinical research program.

Listed below are the clinics and the clinic chiefs now participating in the Child Prosthetics Research Program.

Area Child Amputee Center, Michigan Division of Services to Crippled Children, 920 Cherry St. S.E., Grand Rapids, Mich. 49506. *George T. Aitken, M.D.*

Amputee Clinic, Children's Division, Institute of Rehabilitation Medicine, 400 E. 34th St., New York, N. Y. 10016. *Leon Greenspan, M.D.*

Amputee Clinic, Newington Hospital for Crippled Children, Newington, Conn. 06111. *Earl E. VanDerwerker, Jr., M.D.*

University of Dlinois Amputee Clinic, 840 S. Wood St., Chicago, Ill. 60612. *Claude N. Lambert, M.D.*

Birmingham Child Amputee Clinic, State Crippled Children's Service Bldg., 621 S. 18th St., Birmingham, Ala. 35205. *Chestley L. Yelton, M.D.*

Duke Orthopedic Amputee Clinic, Duke Medical Center, Durham, N. C. 27706. *J. Leonard Goldner, M.D.*

Georgia Juvenile Amputee Clinic, Crippled Children's Service, Box 15089, Emory University Branch, Atlanta, Ga. 30333. *Richard E. King, M.D.*

Amputee Clinic, Children's Rehabilitation Center, 936 Delaware Ave., Buffalo, N. Y. 14209. *Bernie P. Davis, M.D.*

Child Amputee Prosthetics Project, University of California at Los Angeles, Rehabilitation Center, 1000 Veteran Ave., Los Angeles, Calif. 90024. *Yoshio Setoguchi, M.D.*

Clinic for Amputees, Orthopedic Department, University of Oklahoma Hospital, 800 N.E. 13th St., Oklahoma City, Okla. 73104. *Gael Frank, M.D.*

Kernan Hospital Amputation Clinic, Windsor Mill Road and Forest Park Ave., Baltimore, Md. 21207. *Edward Wenzlaff, M.D.*

Child Amputee and Congenital Deficiency Clinic, Children's Orthopedic Hospital, Seattle, Wash. 98105. *Ernest M. Burgess, M.D.*

Child Amputee Clinic, Florida Crippled Children's Commission, 61 W. Columbia, Orlando, Fla. 32806. *Joseph G. Matthews, M.D.*

Amputee Clinic, Home for Crippled Children, 1426 Denniston Ave., Pittsburgh, Pa. 15217. *William R. Eaton, M.D.*

Child Amputee Clinic, Crippled Children's Hospital, 2009 Lamar Ave., Memphis, Term. 38114. *Robert E. Tooms, M.D.*

Child Amputee Clinic, State Hospital for Crippled Children, Elizabethtown, Pa. 17022. *James M. Hunter, M.D.*

Juvenile Amputee Clinic, Crippled Children's Hospital, 200 Henry Clay Ave., New Orleans, La. 71011. *Edward T. Haslam, M.D.*

Sunnyview Hospital and Rehabilitation Center, 1270 Belmont Ave., Schenectady, N. Y. 12308. *William E. Gazeley, M.D.*

Kessler Institute for Rehabilitation, Pleasant Valley Way, West Orange, N. J. 07052. *Ki Ho Kim, M.D.*

Juvenile Amputee Clinic, Handicapped and Crippled Children's Service, D.C. General Hospital, Washington, D.C. 20003. *Charles H. Epps, Jr., M.D.*

Amputee and Prosthetic Clinic, Ontario Crippled Children's Centre, 350 Rumsey Rd.. Toronto 17, Ontario, Canada. *John E. Hall, M.D.*

Child Amputee Clinic, Rehabilitation Institute of Montreal, 6300 Darlington Ave., Montreal, Quebec, Canada. *Maurice Mongeau, M.D.*

Child Amputee Center, Children's Hospital, 1056 E. 19th Ave., Denver, Colo. 80218. *William F. Stanek, M.D.*

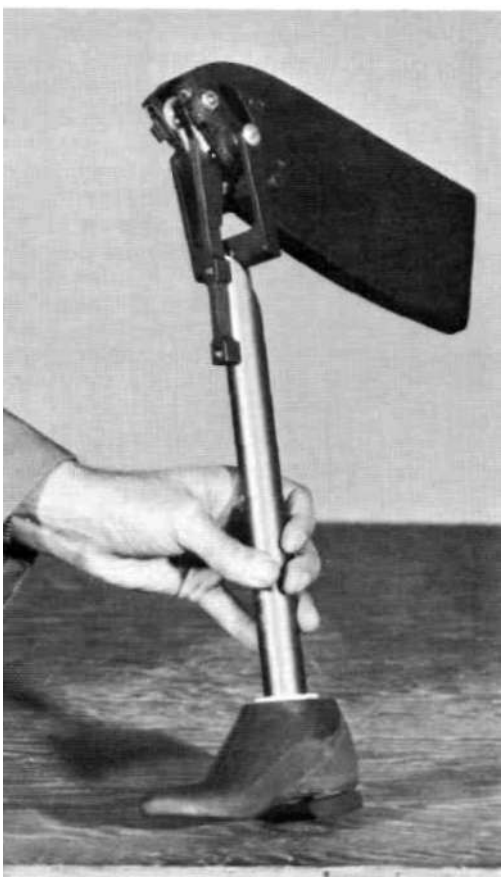
Children's Amputee Clinic, St. Louis Unit, Shriners Hospitals for Crippled Children, 2001 S. Lindberg Blvd., St. Louis, Mo. 63131. *Virginia M. Badger, M.D.*

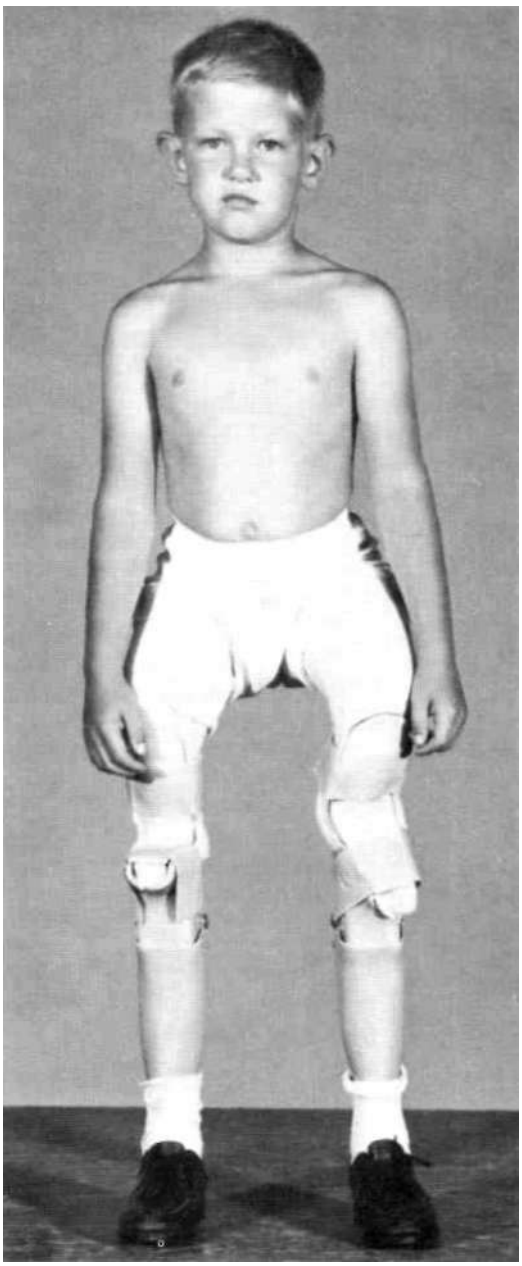
Juvenile Amputee Clinic, Children's Hospital, Buchtel Ave. at Bowery St., Akron, Ohio 44308. *Walter A. Hoyt, Jr., M.D.*

Child Amputee Clinic, Springfield Unit, Shriners Hospitals for Crippled Children, 516 Carew St., Springfield, Mass. 01104. *Leon M. Kruger, M.D.*

Juvenile Amputee Clinic, Greenville Unit, Shriners Hospitals for Crippled Children, 2100 N. Pleasantburg Dr., Greenville, S.C. 29609. *Leslie C. Meyer, M.D.*

Child Amputee Clinic, Portland Unit, Shriners Hospitals for Crippled Children, 8200 N.E. Sandy Blvd., Portland, Ore. 97220. *Arthur L. Eckhardt, M.D.*





Meeting of Committee on Prosthetics Research and Development

The 19th meeting of the Committee on Prosthetics Research and Development was held in Washington, D.C., October 29, 1968.

Mr. Anthony Staros, the newly appointed Chairman of the Subcommittee on Design and Development, reported on meetings of the Panel on Upper-Extremity Prosthetics Components and the Panel on

Lower-Extremity Prosthetics Fitting and stated that during the remainder of the fiscal year the following meetings were scheduled to be held under the auspices of the Subcommittee:

Symposium on Below-Knee Prosthetics
Workshop on Knee-Disarticulation Prostheses
Workshop on Hip-Disarticulation Prostheses
Workshop on Orthotic Knee Joints
Workshop on Fracture Bracing
Workshop on Braces for Spina Bifida Cases
Workshop on Braces for Cerebral Palsy Cases
Medical Orientation for Research Prosthetists, Orthotists, and Engineers

Mr. Howard Thranhardt, Chairman of the Subcommittee on Evaluation, stated that the clinical evaluation of the Engen Plastic Hand Orthosis and the Veterans Administration Prosthetics Center Patellar-Tendon-Bearing Brace, started as part of the pilot program in evaluation, was nearly completed and that reports would be forthcoming shortly. Plans were being made for the immediate initiation of a clinical evaluation program for below-knee sockets formed from synthetic balata directly over the stump by application of pneumatic pressure as developed at VAPC. Mr. Thranhardt also reported that every effort is being made to coordinate as closely as possible the activities of the Evaluation Subcommittee with the Education Program in order to accelerate effective introduction of new devices to the field.

Mr. Colin McLaurin, in reporting the activities carried out by the Subcommittee on Child Prosthetics Problems in the absence of the Chairman Dr. George T. Aitken, noted that there are 28 child amputee clinics participating in the program and others are being considered. It is estimated that the 28 clinics participating now serve 65% of the population. He reemphasized the need for specialized fitting centers to serve the severely handicapped child amputee.

Dr. Arthur Lesser of the Children's Bureau stated that the Bureau was in accord with this recommendation but with the budgetary limitations in effect no new centers could be started immediately. However, he felt that the activities of some existing centers might be expanded in the near future to undertake some of the work that had been proposed at the previous meeting of CPRD.

There was a general discussion on fundamental research that indicated a need for stimulation of basic studies concerning the function of the spine. It was recommended that a workshop on the spine be held during the year.

The meeting was concluded by a roundtable discussion of improving further the introduction of results of research to clinical practice.

Pilot Course in Spinal Orthotics for Physicians and Surgeons Given at New York University

Following an extended period of planning and preparation, a pilot course in Spinal Orthotics for Physicians and Surgeons was given by the New York University Post-Graduate Medical School from November 7-9, 1968. The course, which was attended by 20 physicians and surgeons from 15 institutions, provided a "trial run" of the course schedule and content, as well as an opportunity to obtain reactions and suggestions from those who attended. Following the pilot offering, the first formal course was given from November 25-27, with an enrollment of 24 physicians and surgeons.

The course content included material on nomenclature and components, design, fitting, and functions of braces and corsets for the lumbar, thoracic, and cervical spine in relation to various pathologic conditions, as well as a review of basic biomechanics, kinesiology, and pathomechanics of the spine. The third day of the course was devoted to orthotic management of the scoliosis patient with emphasis on the Milwaukee brace.

The course in spinal orthotics was organized by Dr. Sidney Fishman, coordinator of the teaching and research programs in prosthetics and orthotics at the NYU Post-Graduate Medical School, with the collaboration of Mr. Norman Berger,

associate coordinator. The following served as consultants and lecturers in the planning and presentation of the course: Drs. Jacob J. Graham, David Levine, Ralph Lusskin, John McCauley, Jr., Peter Rizzo, Henry Sprague, Howard Thistle, Walter A. L. Thompson, and Nicholas Tzimas, and Messrs. Clauson England, Robert Fannin, Carlton Fillauer, and Warren Springer.

The next session of Spinal Orthotics for Physicians and Surgeons will take place from April 24-26, 1969. This course now takes its place as a regular offering in the NYU prosthetics and orthotics teaching program and it is anticipated that it will be offered several times during each succeeding academic year.

Meeting of the Subcommittee on Evaluation

The Subcommittee on Evaluation met in Miami, Fla., December 7, 1968, to ascertain the status of various evaluation projects and make recommendations for the future.

Reports were received from the CPRD staff concerning the Engen Plastic Hand Orthosis, the VAPC Patellar-Tendon-Bearing Brace, and below-knee sockets formed of synthetic balata (Polysar X414) directly over the stump using pneumatic pressure. Clinical evaluation of the Engen Plastic Hand Orthosis and the VAPC Brace was nearly completed. The clinical study involving the below-knee socket technique was just being initiated. The pneumatic components developed by the American Institute for Prosthetic Research will be evaluated as soon as they are ready. They were scheduled for delivery early in March.

Mr. William Bernstock reported that the Veterans Administration clinical study involving the Henschke-Mauch Model A (Swing-and-Stance-Phase Control) was nearly completed and that a similar study of the University of California Pneumatic Swing-Control Knee Unit was to be initiated almost immediately.

Dr. Edward Peizer reported on a number of items being subjected to laboratory evaluation at Veterans Administration Prosthetics Center, including externally powered terminal devices, the Blatchford Knee, and modular lower-extremity prostheses.

Dr. Sidney Fishman reported on the New York University evaluation program involving devices and techniques for child amputees. Items under study include the electric arm developed at the Ontario Crippled Children's Centre, the electric cart developed at the Child Amputee Prosthetics Project, UCLA, the Winnipeg cable recovery unit, and synthetic balata sockets for upper-extremity amputees developed at VAPC.

Mr. Howard Thranhardt stated that he had been contacted by representatives of Liberty Mutual Insurance Co. concerning evaluation of the so-called Boston Arm. Further talks were scheduled.

Seminar on Orthotics

The Fifth Annual Postgraduate Seminar on Orthotics by the University of Miami School of Medicine in cooperation with the National Academy of Sciences, the Veterans Administration, and Jackson Memorial Hospital, was held in Bal Harbour, Fla., December 9-11, 1968. The major portion of the program was devoted to orthotics but one morning was spent on reviewing the latest techniques of fitting limb prostheses immediately after surgery. The faculty consisted mostly of individuals deeply engaged in research as well as clinical practice.

The foundations for sound orthotics practice were set forth in the opening session by George T. Aitken, Newton McCollough, and Donald Pierce. Details of the latest practices in bracing the lower extremity were followed by management of patients with paralysis of the upper extremities. In both instances current research efforts were reviewed and the need for further research was emphasized.

Rationale for the design and use of the Milwaukee brace was covered thoroughly.

Some aspects of rheumatoid arthritis were included.

The latest results of the experiments in treating certain long-bone fractures with orthopaedic bracing to give the patient mobility and accelerate healing were reported by Vert Mooney, Augusto Sarmiento, and John Connolly.

Over 300 surgeons, physicians, therapists, orthotists, and prosthetists attended.

Faculty

George T. Aitken, M.D., Chairman, Committee on Prosthetics and Orthotics, American Academy of Orthopaedic Surgeons, Grand Rapids, Mich.

Richard G. Bidwell, C.P.O., President, House of Bidwell, Inc., Milwaukee, Wis.

Ernest M. Burgess, M.D., Clinical Associate Professor of Orthopaedics, University of Washington, Seattle, Wash.

John W. Campbell, Director of the Orthotic Research Project, Biomechanics Laboratory, University of California, San Francisco, Calif.

Mack Clayton, M.D., Associate Clinical Professor of Orthopaedic Surgery, University of Colorado, Boulder, Colo.

Clinton L. Compere, M.D., Professor of Orthopaedic Surgery, Academic Advisor, Prosthetic-Orthotic Center, Northwestern University, Chicago, Ill.

Herbert Elftman, Ph.D., Professor of Anatomy, College of Physicians and Surgeons, Columbia University, New York, N. Y.

Thorkild J. Engen, CO., Director, Department of Orthotics, Texas Institute for Rehabilitation and Research, Houston, Tex.

Charles M. Fryer, M.A., Director, Northwestern University Prosthetic-Orthotic Center, Chicago, Ill.

John Glancy, CO., Assistant Professor of Orthopaedics, Chairman, Division of Orthotics, Indiana University Medical Center, Indianapolis, Ind.

Verne T. Inman, M.D., Professor of Orthopaedics, Director of Biomechanics Laboratory, University of California Medical Center, San Francisco, Calif.

Robert Keagy, M.D., M.S., Associate, Department of Orthopaedic Surgery, Northwestern University Medical School, Chicago, Ill.

H. R. Lehneis, C.P.O., Director, Orthotics & Prosthetics, Institute of Rehabilitation Medicine, New York University Medical Center, New York, N. Y.

Newton C. McCollough, III, M.D., Assistant Professor of Orthopaedics, Associate Director of Rehabilitation, University of Miami School of Medicine, Miami, Fla.

Charles Lindsay McDowell, M.D., Chief, Hand Surgery, McGuire Veterans Hospital, Richmond, Va.

John Moe, M.D., Professor and Director, Division of Orthopaedic Surgery, University of Minnesota School of Medicine, Minneapolis, Minn.

Vert Mooney, M.D., Chief, Amputation and Fracture Service, Rancho Los Amigos Hospital, Inc., Downey, Calif.

Vernon L. Nickel, M.D., Medical Director, Rancho Los Amigos Hospital, Inc., Downey, Calif.

Herbert Pedersen, M.D., Professor of Orthopedics, Wayne State University, Detroit, Mich.

Edward Peizer, Ph.D., Chief, Bioengineering Research Service, VA Prosthetics Center, New York, N. Y.

Donald S. Pierce, M.D., Clinical Associate Professor of Orthopaedics, Harvard University, Cambridge, Mass.

Augusto Sarmiento, M.D., Professor of Orthopaedics, Director of Rehabilitation, University of Miami School of Medicine, Miami, Fla.

Shahan K. Sarrafian, M.D., F.A.C.S., Associate in Orthopaedic Surgery, Northwestern University Medical School, Chicago, Ill.

Bruce A. Scott, C.P.O., President, Scott Surgical, Inc., Denver, Colo.

William F. Sinclair, C.P., University of Miami School of Medicine, Miami, Fla.

Roy Snelson, CO., Project Director, Amputation and Fracture Research, Rancho Los Amigos Hospital, Inc., Downey, Calif.

Anthony Staros, Director, VA Prosthetics Center, New York, N. Y.

W. Dean Warren, M.D., Dean, University of Miami School of Medicine, Miami, Fla.

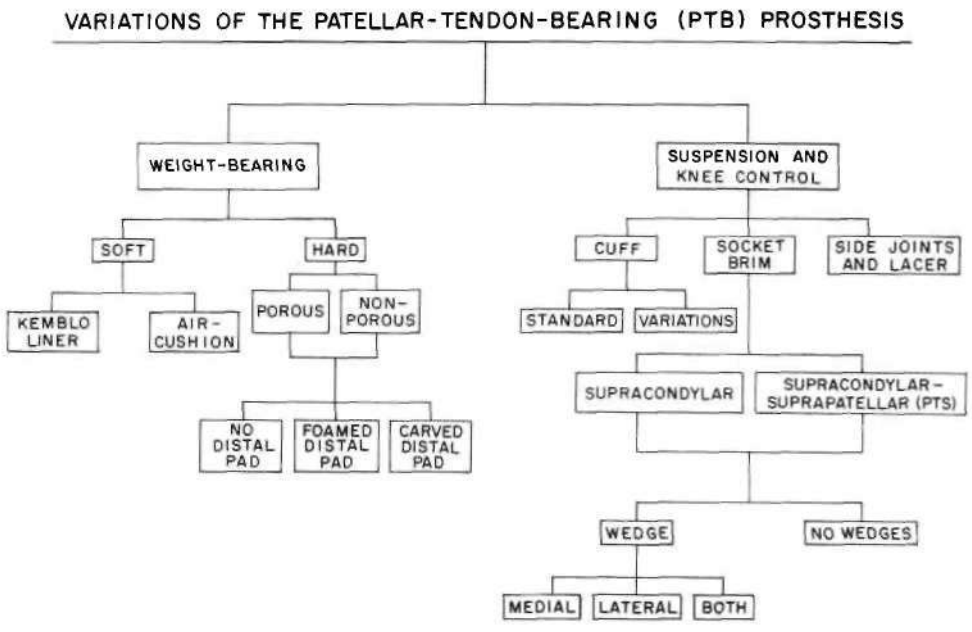
Edward M. Williams, M.D., Assistant Professor of Orthopaedics, University of Miami School of Medicine, Miami, Fla.

A. Bennett Wilson, Jr., Executive Director, Committee on Prosthetics Research and Development, National Academy of Sciences, Washington, D.C

Symposium on Below-Knee Prosthetics

On December 16-18 the Panel on Lower-Extremity Prosthetics Fitting, CPRD, conducted a Symposium on Below-Knee Prosthetics at the Veterans Administration Prosthetics Center, New York City. Participating in the symposium were prosthetists and engineers prominent in research, education, and clinical application. The primary mission of the symposium was to review and assess current practices and results of research projects, and make recommendations to the educational programs. At the same time research currently under way was reviewed and recommendations for future work were offered.

The symposium was begun by reviewing the curriculum concerning below-knee



prosthetics being offered at the University of California at Los Angeles, New York University, and Northwestern University. After this, clinical experience with variations of the patellar-tendon-bearing socket, such as the air-cushion socket, the prosthesis-tibiale-supracondylien (PTS), and wedge suspension method was reported, and fitting and fabrication techniques were discussed in detail. Experiences with new materials including synthetic balata, porous plastic laminates, and transparent plastics were presented, and progress reports were made by those projects which are attempting to map pressure distribution over the stump.

In the general discussion that followed the formal presentations, it was the consensus that:

1. There exists a body of knowledge in BK prosthetics that has been developed in recent years that should be made available to practicing prosthetists and other clinicians for use in everyday practice. (See chart.)

2. All education programs are urged to include this material in their curricula.

3. Postgraduate-type courses in these latest techniques for practicing clinic teams should be made available.

4. It is recognized that manuals and instructional materials are needed. This points out the need for a central group that would be responsible for the preparation and dissemination of technical information.

5. Current research efforts in BK prosthetics should be continued with emphasis on the development of a truly refined theory of fitting.

Meeting of CPOE Subcommittee on Orthotics

Members of the Subcommittee on Orthotics met at the Americana Hotel, New York, January 17, 1969. Dr. Jacquelin Perry, Chairman, reported that the second phase of a survey by orthopaedic surgeons on external support in treatment of low back pain was completed and that the data was being analyzed in preparation for a final report. Members of the Subcommittee reviewed with Dr. Perry the tabulations that had been developed.

Mr. Charles W. Rosenquist reported the results of a survey of 12 orthotics facilities

on the construction of orthotic appliances. The information, which reflected some discrepancies in measurements and methods of measurements, will be forwarded to the Conference of Orthotists for further study as indicated. Dr. Ralph Luskin and Mr. Warren Springer of the New York University Medical Center described the pilot course on Spinal Orthotics for Physicians which was conducted at New York University in December 1968. The Subcommittee expressed interest in NYU's program, and will attempt to obtain available syllabi, course outlines and similar materials on spinal orthotics from schools and other groups conducting programs in this field. The potential value of these materials for use in seminars and workshops will be considered.

Attending the meeting were E. Burke Evans, M.D., Hans R. Lehneis, Charles W. Rosenquist, Roy Snelson, Herbert E. Pedersen, M.D., Chairman, CPOE, and Barbara R. Friz, Executive Secretary, CPOE.

Meeting of CPOE Subcommittee on Special Educational Projects in Prosthetics and Orthotics

The Subcommittee on Special Educational Projects in Prosthetics and Orthotics of the Committee on Prosthetic-Orthotic Education met in New York January 18, 1969. Present were J. Warren Perry, Ph.D., Chairman, Jack D. Armold, Ph.D., Florence L. Knowles, Herbert B. Warburton, Sidney Fishman, Ph.D., Charles 0. Bechtol, M.D., Herbert Pedersen, M.D., Chairman, CPOE, and Barbara R. Friz, Executive Secretary, CPOE.

Most of the meeting time was spent reviewing basic tabulations developed from the Manpower Survey data collected from 236 prosthetics and orthotics facilities reporting on 1,374 personnel. The tabulations reflected the distribution and magnitude of manpower shortages and projected manpower needs and correlations between salary ranges and educational background, and salary ranges and experience.

The data will be thoroughly studied and pertinent information will be extracted for distribution and publication. It is expected that the findings from this survey will be most useful in developing recruitment and educational programs.

The advisability of a survey of graduates of prosthetics and orthotics degree programs was discussed, and it was decided to delay this project until such time as a more inclusive study could be conducted.

Meeting of *Ad Hoc* Planning Committee for the Geriatric Amputee Workshop

Members of the *Ad Hoc* Planning Committee for the Geriatric Amputee Workshop met at the Americana Hotel in New York on January 19, 1969, for the purpose of making final plans for the conduct of the workshop. The panels selected and the respective chairmen are:

The Care of the Patient with Vascular Disease: Richard Warren, M.D., Chairman

Amputation Surgery: Herbert Pedersen, M.D., Chairman

Postoperative Care: Augusto Sarmiento, M.D., Chairman

The Prosthesis: Frank Clippinger, Jr., M.D. Chairman

Prognosis of the Geriatric Amputee Following Surgery: Robert Mazet, Jr., M.D., Chairman

To Fit or Not to Fit: Claude N. Lambert, M.D., Chairman

The Workshop will be held after the papers are submitted by the chairmen and reviewed by the members of each panel. The purpose of the project is to provide an authoritative document that will be useful to all who are engaged in the rehabilitation of the geriatric amputee.

It was decided to retain the name, "Geriatric Amputee," in speaking of the Workshop; however, the manual will probably not carry this title.

Those attending the meeting were William M. Bernstock; Frank W. Clippinger, Jr., M.D.; Edward T. Haslam, M.D.; Herbert E. Pedersen, M.D., Chairman; Augusto Sarmiento, M.D.; Joseph E. Traub; and Barbara R. Friz.

AAOS Appointment

George T. Aitken, M.D., Chief of Orthopaedic Surgery, St. Mary's Hospital, Grand Rapids, Mich., was named President-Elect of the American Academy of Orthopaedic Surgeons at the Thirty-sixth Annual Meeting of the 4500-member organization in New York City, January 18-23. He succeeds Dr. S. Benjamin Fowler, Nashville, Tenn., who became President in installation ceremonies at the meeting.

Dr. Aitken is Medical Co-Director, Area Child Amputee Center, Grand Rapids; Consultant in Orthopaedic Surgery, Mary Free Bed Guild Children's Hospital and Orthopaedic Center, Grand Rapids; and Chairman of the Subcommittee on Child Prosthetics Problems of the Committee on Prosthetics Research and Development. Dr. Aitken was Chairman of CPRD from 1963 to 1965 and continued as a member until 1968.

He became a Fellow of the American Academy of Orthopaedic Surgeons in 1942 and served as Chairman of the Services Section of the National Health Program for Orthopaedics (announced January 20). The 264-page report recommended a broad program to improve orthopaedic health care for the American people.

Dr. Aitken is a member of the Committee on Orthopaedic Rehabilitation and an Associate Editor of the *Journal of Bone and Joint Surgery.* He was Chairman of the Academy Committee on Prosthetics and Orthotics.

He received B.S. and M.D. degrees from the University of Indiana. In World War II he was a Lieutenant-Colonel in the U.S. Army, serving as senior orthopaedic consultant in the European Theatre.

Dr. Aitken is a member of the American Orthopaedic Association, Clinical Orthopaedic Society, Michigan Orthopaedic Society, American College of Surgeons, American Medical Association, and the Michigan Medical Association. He is regional chairman for the Orthopaedic Research and Education Foundation.

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 co-operation with CPRD and published in 1954, originally under the imprint of the McGraw-Hill Book Co., has now been reprinted, with an updated bibliography, by the Hafner Publishing Co. A 10 per cent discount from the $17.50 price will be allowed if orders are sent directly to: Stechert-Hafner, Inc., 31 E. 10th St., New York, N. Y. 10003.

NATIONAL ACADEMY OF SCIENCES—NATIONAL RESEARCH COUNCIL

THE NATIONAL ACADEMY OF SCIENCES is a private, honorary organizat ion of more than 700 scientists and engineers elected on the basis of outstanding contributions to knowledge. Established by a Congressional Act of Incorporation signed by Abraham Lincoln' on March 3, 1863, and supported by private and public funds, the Academy works to further science and its use for the general welfare by bringing together the most qualified individuals to deal with scientific and technological problems of broad significance.

Under the terms of its Congressional charter, the Academy is also called upon to act as an official—yet independent—adviser to the Federal Government in any matter of science and technology. 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 and its activities are not limited to those on behalf of the Government.

THE NATIONAL ACADEMY OF ENGINEERING was established on December 5, 1964. On that date the Council of the National Academy of Sciences, under the authority of its Act of Incorporation, adopted Articles of Organization bringing the National Academy of Engineering into being, independent and autonomous in its organization and the election of its members, and closely coordinated with the National Academy of Sciences in its advisory activities. The two Academies join in the furtherance of science and engineering and share the responsibility of advising the Federal Government, upon request, on any subject of science or technology.

THE NATIONAL RESEARCH COUNCIL was organized as an agency of the National Academy of Sciences in 1916, at the request of President Wilson, to enable the broad community of U.S. scientists and engineers to associate their efforts with the limited membership of the Academy in service to science and the nation. Its members, who receive their appointments from the President of the National Academy of Sciences, are drawn from academic, industrial, and government organizations throughout the country. The National Research Council serves both Academies in the discharge of their responsibilities.

Supported by private and public contributions, grants, and contracts, and voluntary contributions of time and effort by several thousand of the nation's leading scientists and engineers, the Academies and their Research Council thus work to serve the national interest, to foster the sound development of science and engineering, and to promote their effective application for the benefit of society.

COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT
COMMITTEE ON PROSTHETIC-ORTHOTIC EDUCATION

The *Committee on Prosthetics Research and Development* and the *Committee on Prosthetic-Orlhotic Education,* units of the Division of Engineering and the Division of Medical Sciences, respectively, undertake activities serving research and education in the fields of prosthetics and orthotics, when such activities are accepted by the Academy as a part of its functions. Activities of the Committees are presently supported by the Department of Health, Education, and Welfare and the Veterans Administration. Information or reports developed by activities of the Committees are officially transmitted and published through the National Academy of Sciences.