March 1973

Orthotics and Prosthetics

Orthotics and Prosthetics

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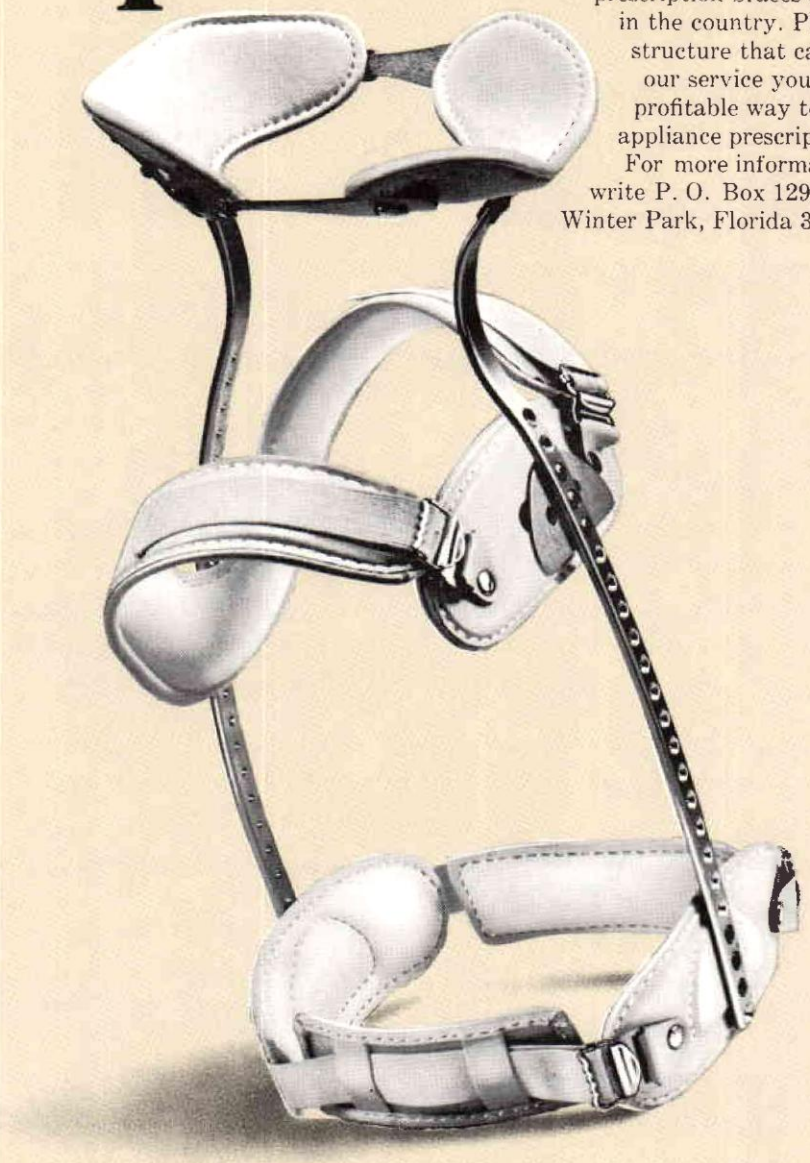

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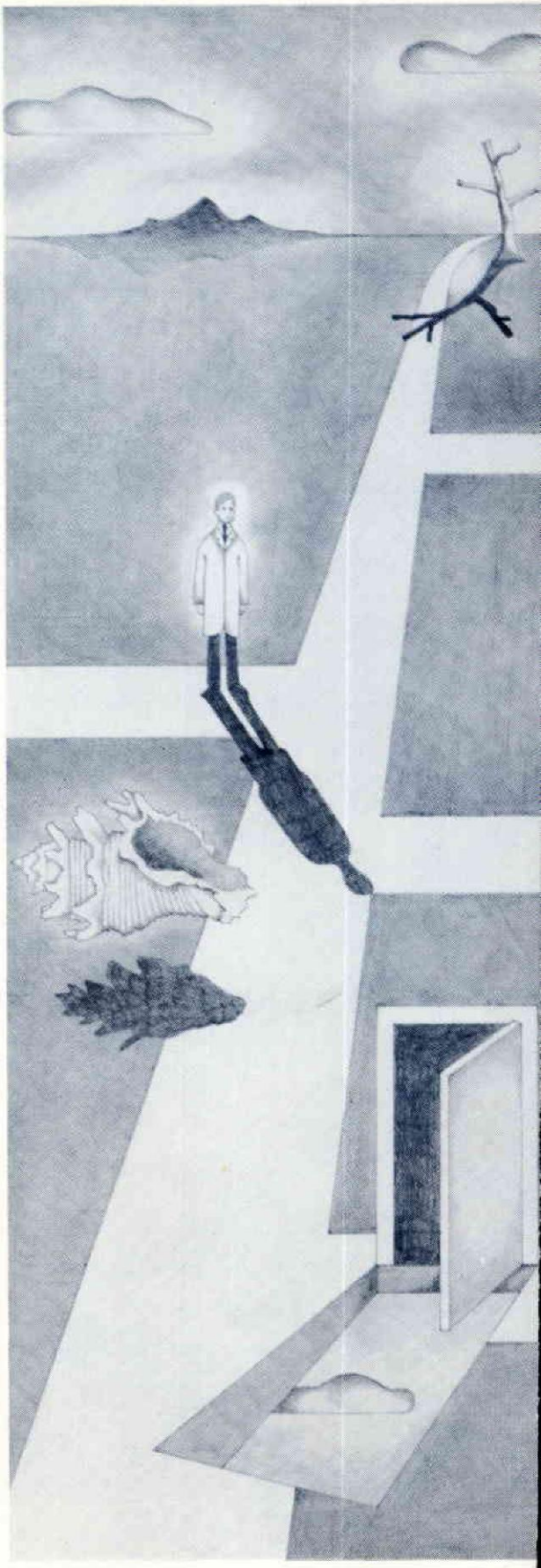

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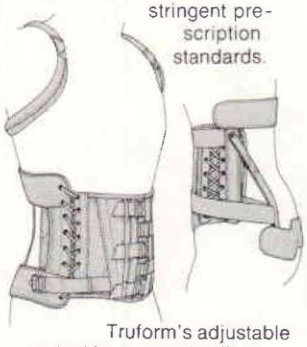

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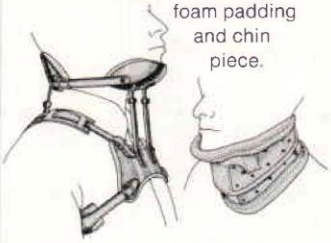

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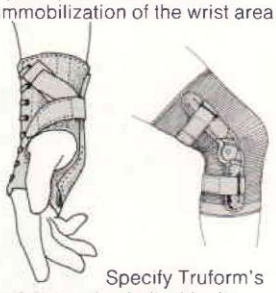

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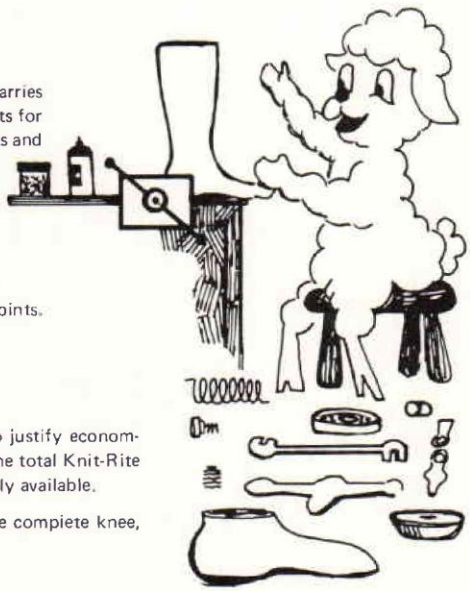

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Part No. 2L 188 Valve with plastic housing

Part No. 2L 189 Valve with stainless steel housing

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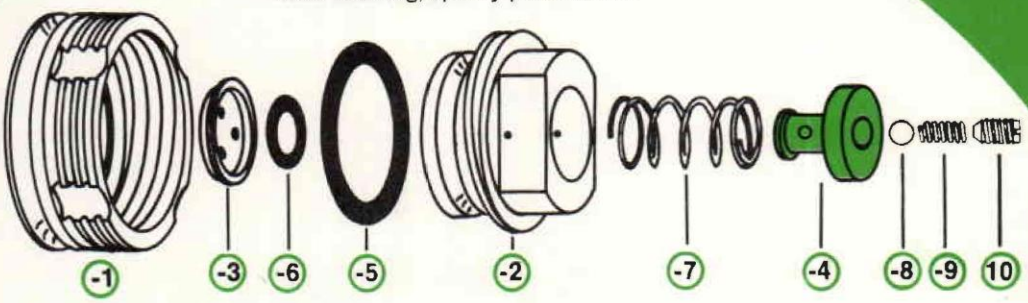

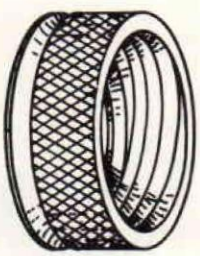

-1-A

-1	Plastic housing		"O" ring .375 O.D.
1-A	Stainless steel housing		Spring 7/16" dia.
-2	Valve body		Teflon ball 1/8"
-3	Cover	-4 Push button	Spring 1/8" dia.
-5	"O" ring .937 O.D.		Nylon screw 8-32

When ordering parts, specify part number and dash number.

Example No. 1 2L 188-1 / Example No. 2 2L 189-1-A

CLEANING INSTRUCTIONS.

Remove valve body from housing. Take a blunt object and compress the green button fairly hard until the plastic cover, with the three holes in it, pops out. Keeping the Green button pressed down, remove the small "O" ring with finger. Now the green button and spring can be removed from the body. To clean the adjustable expulsion valve, remove the nylon screw in the center of the green button. Then remove spring and teflon ball. Clean all parts and reassemble making sure the cover is replaced with the flat surface on the outside.

Corrosion resistant ● All parts interchangeable

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A NEW LOOK

Past readers of ORTHOTICS AND PROSTHETICS will recognize that page size, type style, texture of paper, and other design details have been changed in this issue, the first number of Volume 27. These changes have been made in an attempt to provide the optimum approach to dissemination of technical, scientific, and professional information in the field of orthotics and prosthetics.

The new size makes it easier for librarians of scientific collections to file and retrieve issues of ORTHOTICS AND PROSTHETICS; the new format enables the editors to integrate text and photographs more efficiently; and the texture of the paper guarantees better reproduction of the illustrations. In essence, we hope that the new format will be easier to read.

That there have been submitted to ORTHOTICS AND PROSTHETICS so many useful articles during the past year has been most rewarding to the Editorial Board and others who are concerned with providing information to members of AOPA and AAOP and all other professional personnel involved in prosthetics and orthotics that will help them to offer their patients the best health care available.

We urge all to continue to contribute to ORTHOTICS AND PROSTHETICS in order to make it the outstanding journal of its kind in the world. The Editorial Board invites every reader not only to submit contributions to ORTHOTICS AND PROSTHETICS, but also to submit letters to the Editor which register concrete criticism, request topics for presentation, or raise appropriate controversy.

We would like to suggest to authors that in preparing articles they keep in mind the need for the texts in education programs of the Social and Rehabilitation Service, the Veterans Administration, and the American Academy of Orthotists and Prosthetists. When it can be anticipated that the subject of an article will be appropriate for inclusion in an education program, much can be done by the author and editorial staff to make it more useful, and a sufficient supply of reprints can be ordered at the time of printing, thereby effecting a saving.

A. BENNETT WILSON, JR.

A Below-Knee Prosthesis With A Porous Socket

VIRGIL FAULKNER, C.P.O.[1]
CHARLES PRITHAM, C.P.[2]

Porosity in the socket wall of a limb prosthesis has long become recognized as a feature, especially in humid climates. Porosity permits constant interchange of air that cools the stump and helps eliminate the accumulation of water in the socket which in turn helps to retard the growth of unwanted bacteria.

The idea of a porous socket is not new. The Army Prosthetic Research Laboratory was experimenting with the idea of a porous socket before 1957. However, the technical expertise and time consuming procedures developed to date have caused most practioners to exclude this technique from their amputee management program.

In recent years much attention has been given to the so called "instant prosthesis", "adjustable prosthesis", and the temporary, or preparatory, prosthesis. To make these new type prostheses economically feasible, several modular, endoskeletal type systems have been developed. Many new types of plastics are being introduced to complement the new modular systems, and much attention has been focused on the polyethethlene and polycarbonate families of plastics.

Recently, the Prosthetics Laboratory at the University of Virginia was asked to participate in a program to study and find uses for the application of a system consisting of fiberglass impregnated with unsaturated polyester resin that has been treated so as to begin the first stages of polymerization. The polymerization of the product is interrupted before it is completed and it is packaged in an air tight container to prevent further polymerization until the material is ready to be used. The fiberglass, BETA type, was developed by NASA and is being marketed under the trade name of Lightcast®.[3]

Our attention was drawn to the material by the Navy Prosthetic Research Laboratory, Oakland, California in a preliminary report entitled *Lightcast Amputation Stump Socket* (1) which advocated application of the material over the stump. We made several sockets according to the instructions given in the preliminary report, but found that it was all but impossible to apply pressure properly to pressure tolerant areas and to relieve in areas that are pressure sensitive. These studies led us to the development of the procedure described in this paper which involves the use of a rectified model of the stump.

MATERIALS AND EQUIPMENT REQUIRED

1. Lightcast lamp®.
2. Four rolls of four-inch Lightcast®.
3. One, five-ply stump sock.
4. One, endoskeletal type modular system.
5. One SACH foot.
6. One cosmetic cover.
7. One complete channel assembly for a Fillauer-type removable brim. (2)
8. Two rolls of two-inch Lightcast®.
9. Two rolls of three-inch Lightcast®.

CASTING AND MODIFICATION

A negative cast is taken in the usual way using elastic plaster-of-Paris bandages. The cast is filled with plaster of Paris, and the positive model is modified in the usual manner (Fig. 1).

FABRICATION OF THE SOCKET

The step-by-step fabrication procedure for the Lightcast® socket is as follows:

[1] Lecturer in Orthopedics, University of Virginia.

[2] Prosthetist, University of Virginia, Department of Orthopedics, Division of Prosthetics and Orthotics.

[3] Solar Laboratories, Inc., Torrance, California.

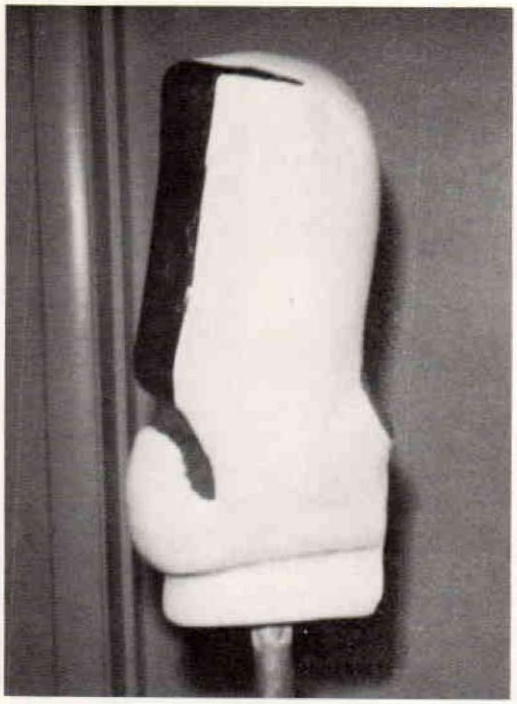


FIGURE 1

1. Coat the positive model with a parting lacquer.
2. Apply a five-ply wool stump sock to the positive model. Do not stretch the material. Staple to the proximal end of the mold (Fig. 2).
3. Apply two layers of four inch Lightcast® material over the five-ply wool stump sock (Fig. 3).
4. Place the mold in the Lightcast® light for three minutes (Fig. 4).
5. Make one longitudinal cut down the medial aspect of the socket with an electric cast saw. Do not cut the five-ply wool stump sock (Fig. 5).
6. Remove the socket from the mold.

APPLICATION OF THE SOCKET TO THE MODULAR SYSTEM

The VAPC adjustable pylon (3) is used routinely in our laboratory for below-knee prostheses, and the procedure for attaching this system to the socket is described here. However, the skilled practitioner will find that any one of the commerical available systems can be adapted to this socket.

1. Attach the correct size SACH foot to the VAPC shank and ankle plug assembly using the instructions provided by the manufacturer.

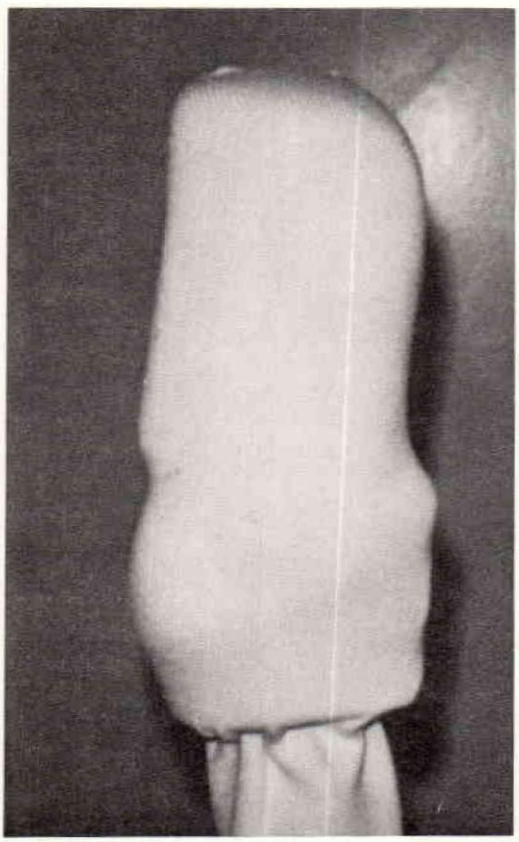
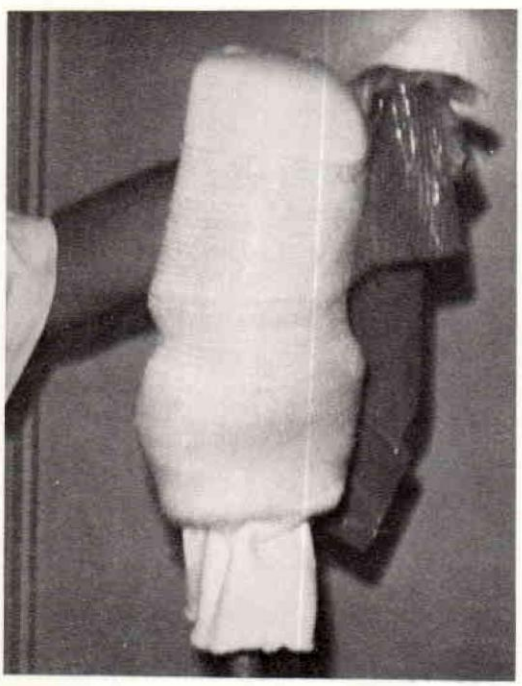


FIGURE 2



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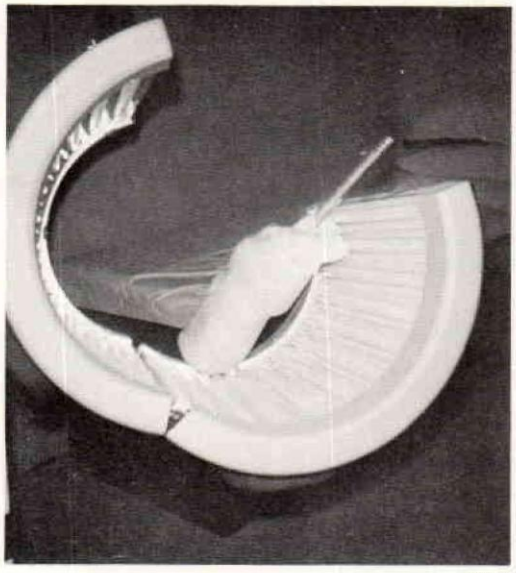

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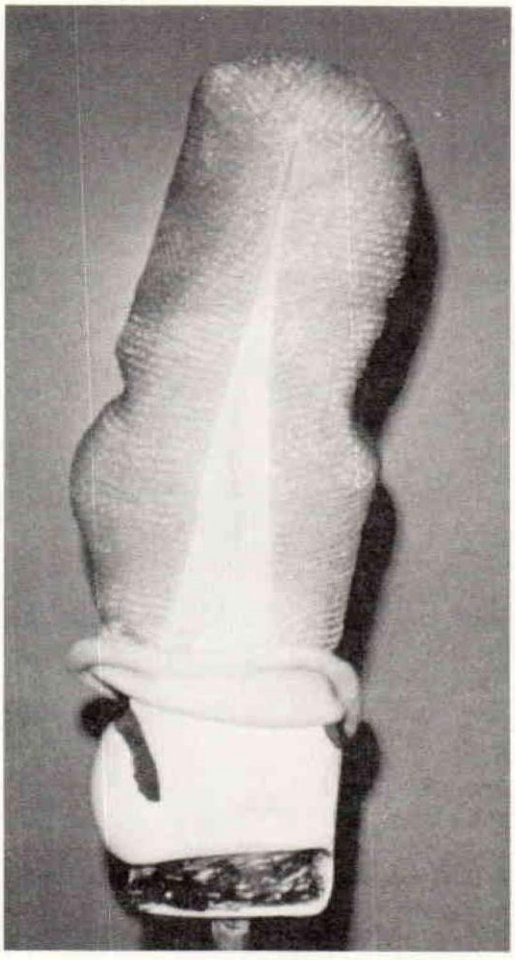

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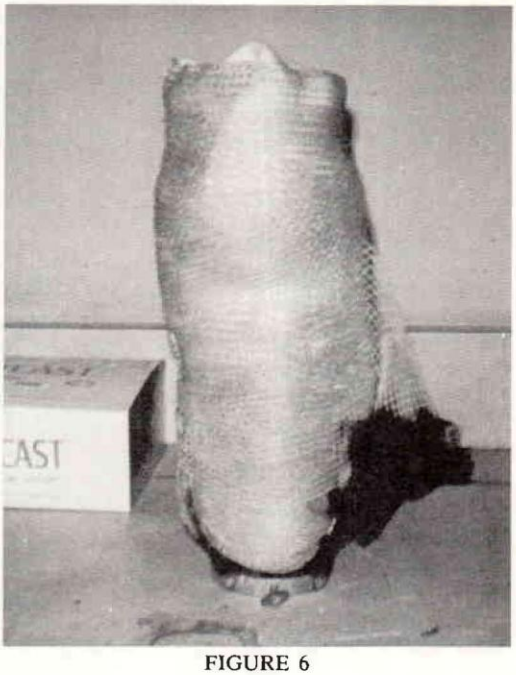

FIGURE 6

2. Set the Lightcast® socket into the socket attachment assembly and bend the four attachment straps to the contours of the Lightcast® socket.

3. Adjust the straps until the socket is brought into good "bench" alignment.

4. Lay one strip of two-inch Lightcast® material under each of the four attachment straps.

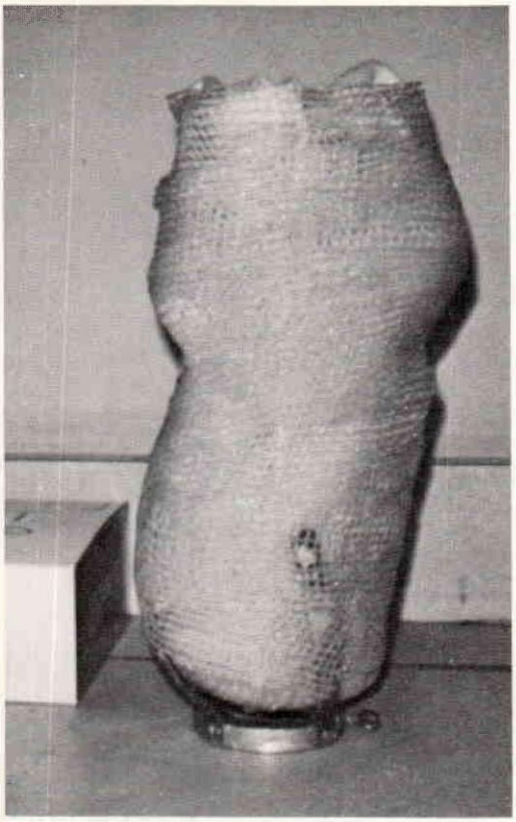

FIGURE 7

5. Starting one-half an inch above the proximal end of the attachment straps (Fig. 6), completely cover the straps by making circular wraps until two layers of Lightcast® material are laid down. This wrap should extend to the superior edge of the socket (Fig. 7).

6. Place the socket and socket attachment assembly into the Lightcast® light for three minutes.

7. Shape the Fillauer channel assembly for the socket.

8. Wrap a three-inch piece of Lightcast® material around the wings of the channel assembly.

9. Make one circular wrap around the proximal portion of the cast. Lay on the channel assembly and continue wrapping until four layers completely cover the channel assembly (Fig. 8).

10. Place the socket in the Lightcast® light for three minutes. Remove and make normal cuts with an electric cast saw to remove the medial brim.

11. Take 100 grams of 4110 polyester resin slightly thickened with sulfa flux and completely impregnate the material covering the four socket attachment straps and the Fillauer channel (Fig. 9).

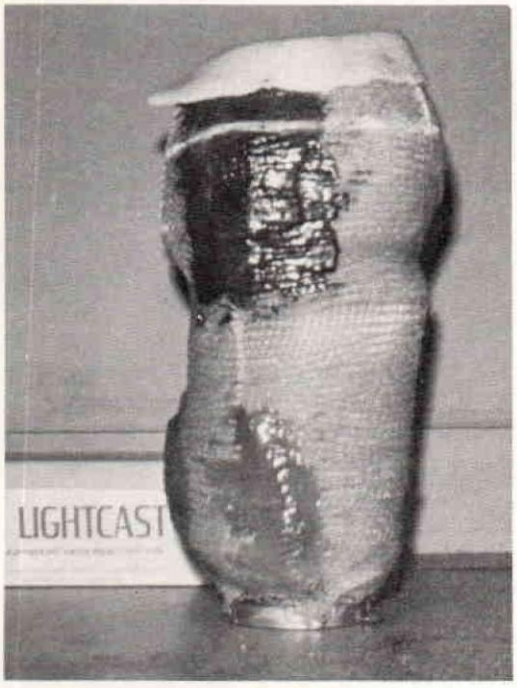

FIGURE 8

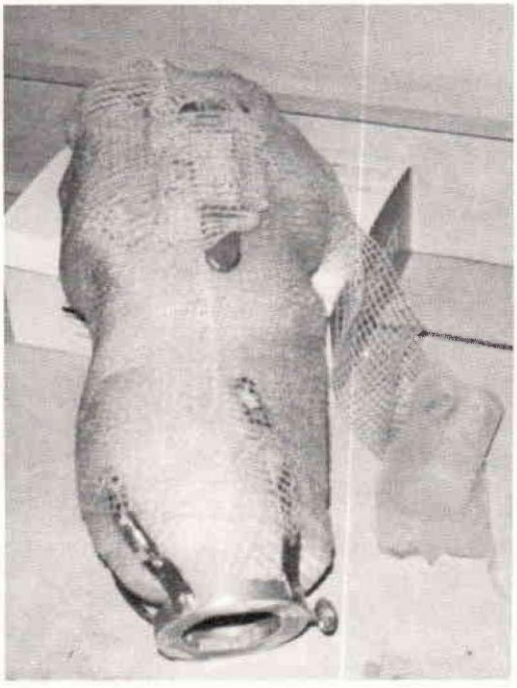

FIGURE 9

12. Remove the medial brim and trim the proximal edges.

13. Attach the socket and socket attachment assembly to the shank tube and the prosthesis is ready for dynamic alignment (Fig. 10).

After dynamic alignment a semi-rigid foam product called Koroseal can be fabricated to form a cosmetic cover as described by the United States Manufacturing Company[4].

DISCUSSION

The advantages of a porous socket made from Lightcast® are, the ability to permit the interchange of air and to some extent control the accumulation of fluids within the socket (Fig. 11). Lightcast® affords a light, porous, easy applied, and quickly cured material with a high resistance to impact for the construction of a prosthetic socket.

Many prosthetics facilities may find the application of this procedure to the patient prohibitive because of the high cost of the Lightcast® lamp.

The Lightcast® lamp uses a standard three-prong outlet, draws an average of nine amperes, and is designed to operate on a 120 volt system. A 15 ampere circuit is recommended. This device is necessary for curing Lightcast® because the polymerization process will not be completed unless the material has been subjected to ultraviolet light. However, it seems possible that most centers working with handicapped people will find this system of value because Lightcast® can be used in place of plaster of Paris in many instances including splints for immobilizing fractures. It also can be used to provide simpler splints in place of other plastics. Lightcast® is radio-transparent, and the patient is able to use it immediately after a three-minute curing period. A twenty-four hour waiting period before full weight-bearing is considered to be desirable but our experience has found it not to be essential.

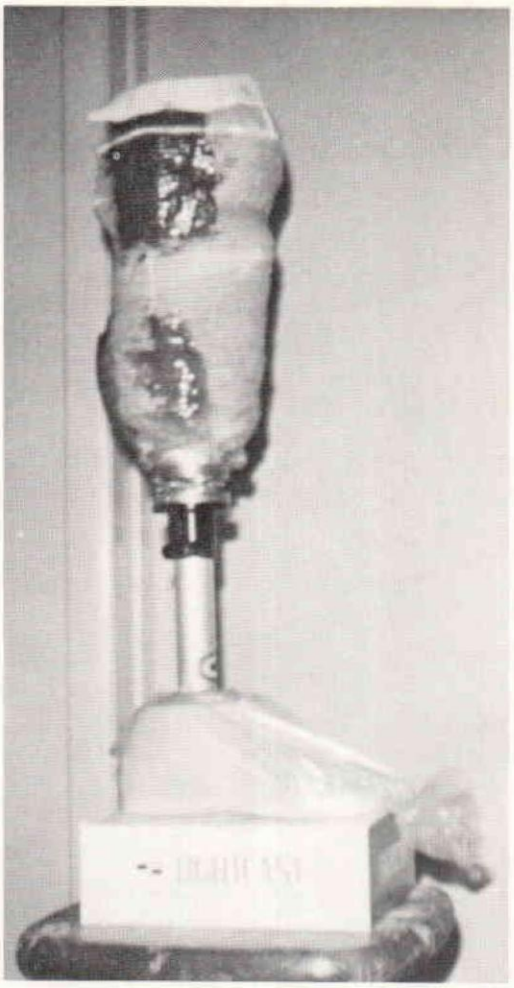

FIGURE 10

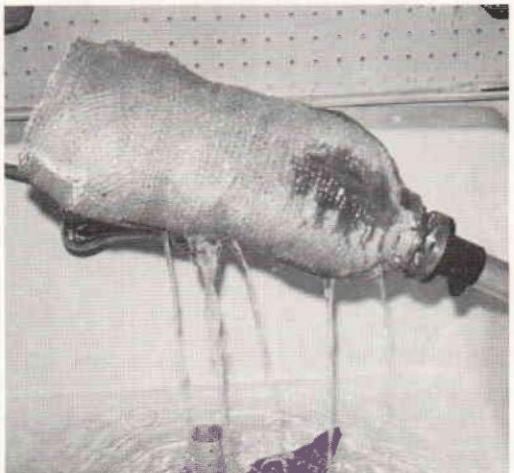

FIGURE 11

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The INAIL-CECA Prostheses[1]

HANNES SCHMIDL[2]

In 1964, the research department, of the INAIL Prosthesis Center at Vigorso di Budrio, began the development of a myoelectrically controlled prosthesis for upper-limb amputees. During that year, prostheses were applied, experimentally, to one unilateral amputee and to one bilateral amputee in order to investigate the practicality of the controls and acceptability by the patients.

After evaluating the initial research, we set up a small scale production run of fifty sets of parts to determine the problems associated with the application of myoelectric prostheses, and, above all, for the purpose of eliminating any weak points that might come to light during their use.

This first phase of practical application lasted over a year. In 1965, when it was certain that the myoelectric prosthesis we had developed provided a definite advantage to the upper-limb amputee, we began mass production of this device.

At first, our myoelectric prostheses could be applied only to below-elbow amputees. If, at this initial stage, we had not foreseen the possibility of applying this principle to higher level upper-limb stumps, our program would have been terminated. In our opinion, the true value of myoelectrically controlled prostheses is found in the replacement of two, three, or four articulations.

For this reason, concurrently with mass production of the below-elbow myoelectric prosthesis, the research team began a program to develop a multichannel myoelectric control system that would make it possible and feasible to utilize more than six myoelectric signals in a high level upper-limb amputee.

Research in this difficult task extended over a period of three years and was conducted with financial support from the European Coal and Steel Community. The research was completed in 1970, and large scale application of the multichannel, myoelectrically controlled prosthesis was undertaken. This made possible the production of prostheses with myoelectric control over eight movements.

A myoelectrically controlled prosthesis for above-elbow or shoulder-disarticulation amputees must provide at least three movements; that of the hand, wrist, and elbow. In patients of this type, however, there are a limited number of muscles available that can be used for control, and thus the task is complicated.

An on-off myoelectric control delivers a single movement from each muscle. This approach is impractical for amputations at high levels due to the lack of signal sites. Also, in fitting these high level cases with either mechanical or "switch control" electric power, the movements are not as harmonious as they are in the natural arm. Nor is amputee acceptance of the prosthesis as high.

The situation is quite different when it is possible to give the patient control over gripping force of the hand directly proportional to muscle contraction and also give the patient control over the speed of its movement. This possibility was offered by the development of an electronic device that obtains at least two controls from a single muscle while, at the same time, provides proportional control.

The most important "break through" in this program came with the development of a two-part amplifier that made it possible to utilize various levels of muscle potential; levels that are quite distinct and can be controlled by the amputee. The first part consists of an integrated circuit differential amplifier followed by a transistor which further amplifies the signal. Regulation of the output level is achieved through a potentiometer. The second part transforms the output signal of the first part into a two-level signal.

[1] This article was originally presented by the author as the Ninth Louis J. Horowitz Lecture. The Department of Rehabilitation Medicine, New York University Medical Center, April 25, 1972.

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The lower-level signal controls a power transistor connected directly to the power supply for motor movement. Control that is proportional to the contraction of the muscle itself is thus obtained. When the power transistor is saturated, the upper-level signal begins to function, activating a relay which reverses the polarity of the motor, and continues functioning as long as the muscle remains contracted. Upon rapid release of the muscle contraction, the motor again, momentarily, reverses polarity, thus achieving a braking effect. The two movements are controlled independently and do not interfere with each other.

This new amplifier enabled us to utilize one muscle for two-signal myoelectric control, and during practical tests we recognized that the instruments normally used for determining muscle sites for the myoelectric prosthesis were inadequate. It was further observed that for the application of multichannel control, the amputee requires special training. To assist in training it was necessary to develop an instrument capable of registering muscle potentials while at the same time indicating to the amputee through signal lights optimum control of the muscle used to operate the prosthesis.

The training instrument known as a myometer is used to measure muscle potential, and to determine electrode location sites (Fig. 1). In addition, it is used to conduct practical tests on the amputee in the use of proportional multichannel control. These tests also give the technician the necessary information for correct adjustment to the two-level muscle signal to meet the requirements of individual amputees.

Several warning lights of different colors correspond to the movements for opening and closing the hand and for wrist rotation. The varying intensity of the signal lights indicates to the amputee the proportionality of his own muscle contraction.

Such an instrument is indispensable. It was designed to give the amputee opportunity for adequate practice, so that he can obtain the utmost function possible from his myoelectric artificial arm with multichannel control.

At this point of our research we discovered it was possible to obtain eight signals by utilizing the pectoral, biceps, deltoid, and triceps muscles leading to the application of a myoelectrically controlled prosthesis on an above-elbow amputee (Fig. 2).

It is interesting to note that the first amputee to experiment with the above-elbow prosthesis learned to control independently the various movements within a few days. Thus, for the first time, a practical prosthesis with myoelectric control of the hand, wrist and elbow function was achieved (Fig. 3).

With the new multichannel amplifiers six signals are necessary. Two signals remain available for control of such additional movements as wrist flexion-extension, humeral rotation, or, in the case of shoulder disarticulation, for articulation of a shoulder joint. So, with the availability of this multichannel

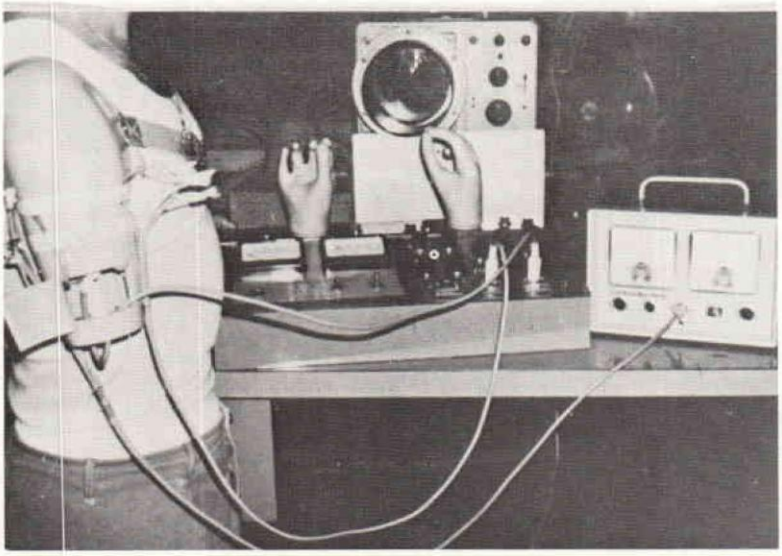

Fig. 1. The myometer used for training the patient in control of the prosthesis.

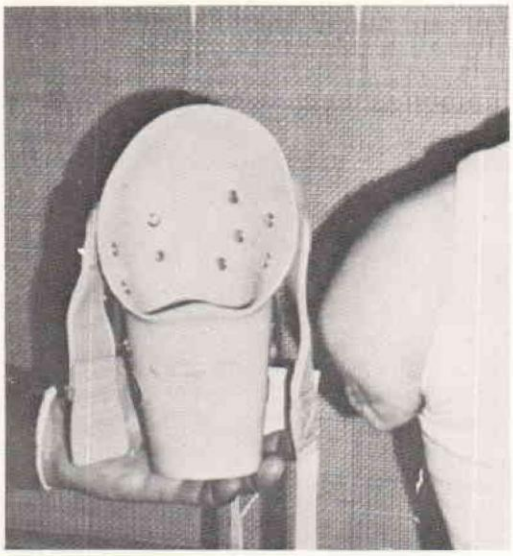

Fig. 2. View of an above-elbow socket for the myoelectric arm showing the positions of the electrodes.

myoelectric control system, it has now become possible to apply in a practical manner prostheses to all levels of upper-limb amputation (Fig. 4).

After the fabrication of a small quantity for experimental purposes, there began in 1971 the mass production of myoelectric prostheses for arm amputees. The INAIL Prosthesis Center of Vigorso de Budrio has, up to the present time, applied myoelectrically controlled prostheses to more than a thousand unilateral and bilateral upper-limb amputees.

None of the amputees encountered insurmountable difficulties in adapting themselves to the prosthesis. Many of them have been able to resume their former activity or take up a new kind of work, encouraged to do so by the excellent function of the prosthesis itself. The ages of the patients range from seven to seventy-five years.

At the present time, there is no type of prosthesis that can offer greater function. It should be emphasized that the application of a myoelectric prosthesis requires special techniques, training, and knowledge, and, above all, special equipment. It is only in this manner that maximum results can be obtained.

THE PROSTHESIS

The INAIL-CECA myoelectric system has been and will continue to be the object of continued improvement. Even today it is not possible to speak of a definitive system that is no longer capable of being improved.

It is extremely important that new technology be incorporated as it becomes available, but it must be recognized that continual

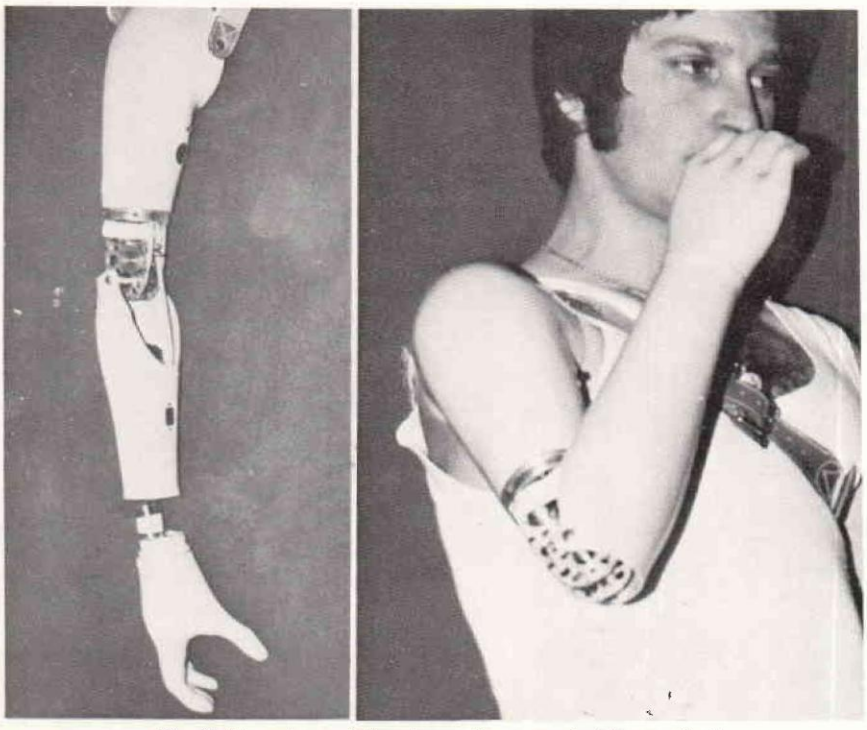

Fig. 3. Two views of the above-elbow myoelectric prosthesis.

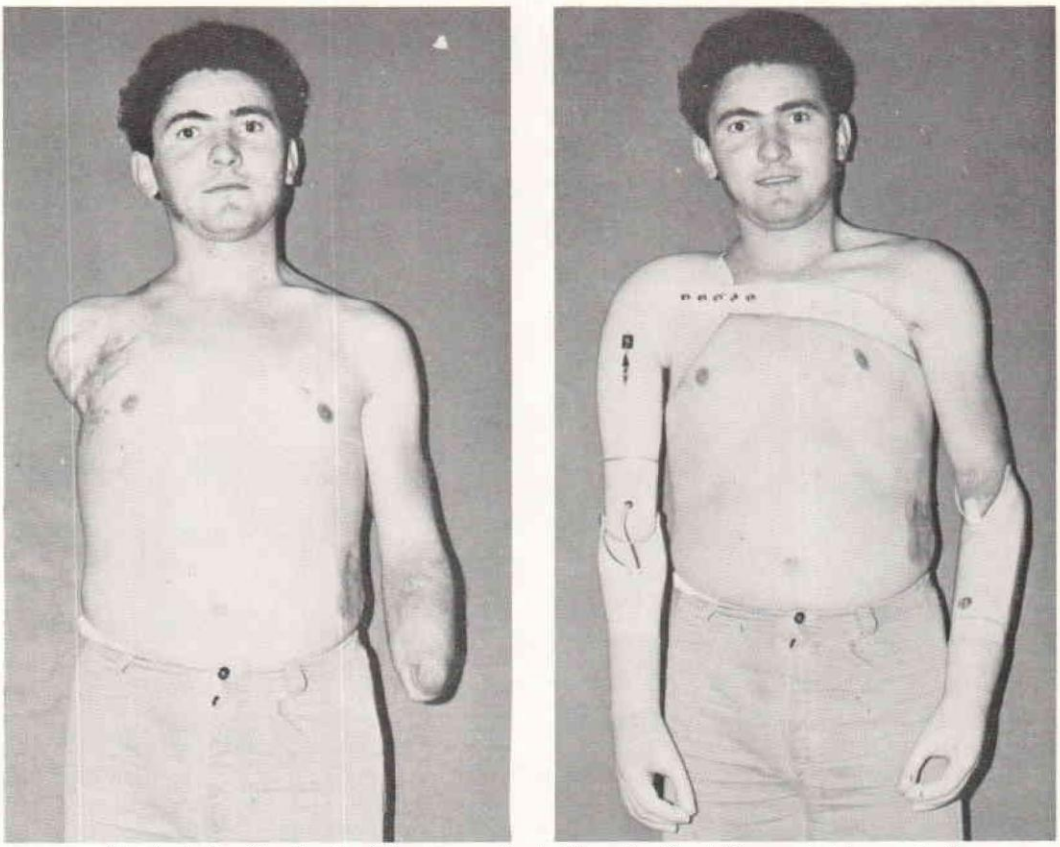

Fig. 4. A shoulder-disarticulation type patient with and without his myoelectric prosthesis.

changes create difficulties for the firms that manufacture components for prostheses. It may happen, in fact, that a product on the verge of being offered is rendered obsolete even before being placed on the market. Moreover, in order to control costs, it is necessary to produce large quantities. This is further complicated because the market is not now in a position to absorb large quantities. And, finally, it must be remembered, that, in the fabrication of a myoelectrically controlled arm, the prosthetist may, at times, be forced to develop his own special components for unique cases without being able to make use of commercially available parts all of which further complicates the problem.

Presently, we have available the following components (Fig. 5):

electrically powered hands
passive wrist units
stump activated wrist units
electrically powered wrist units
outside locking elbow hinges
electrically powered elbows
amplifier for single-channel systems
amplifier for multichannel systems
batteries for insertion in the prosthesis
batteries for external application

At the present time, we use an electric hand that is mass produced to our specifications by the firm of Otto Bock. It is known as the 8E8 Model Z6 hand. It provides a maximum opening of the fingers of more than ten centimeters (approximately 4 inches) and a gripping force of over 10 kilos (approximately 22 pounds). To date approximately 750 amputees who are engaged in widely varied activities have been fitted with this type of hand. In practice, the hand has given proof of durability and maximum function. The 8E8 hand does not lend itself, ideally, to proportional control because of its special gearing which transforms speed into force at the moment the fingers encounter resistance. While waiting for the new hand, created especially for proportional control, we use the older model.

PASSIVE WRIST UNIT

The passive wrist unit allows passive rotation of the hand through a range of 360°.

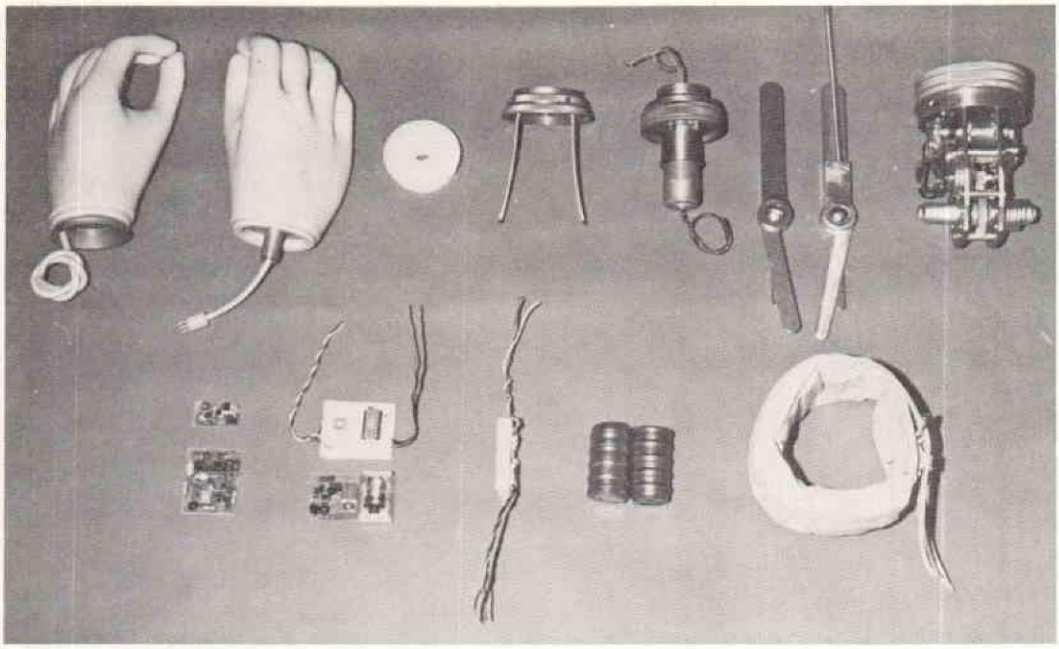

Fig. 5. The compenents available for use in fabrication of myoelectric prostheses.

Various friction designs maintain the hand in the desired position.

STUMP-ACTIVATED WRIST UNIT

The stump-activated unit is especially suitable for long stumps that have residual active pro- and supination of the wrist. These units are made in such a way that they require very little space and can, therefore, be used in fitting prostheses to very long stumps, even disarticulation of the wrist, since they do not create problems of excessive length in the artificial limb.

ELECTRICALLY POWERED WRIST UNIT

An electrically powered wrist unit that uses the same motor and gear reduction unit used in the hand was developed. In order to have a rational, yet economical approach, we feel it is important to standardize on the same drive system for all three components—hand, wrist, and elbow. This wrist unit is small enough to be used in prostheses for very long below-elbow stumps. It requires only 55 mm (2 1/8 in.) of linear space. The weight is about 180 gr. (6 1/4 oz.), range of motion is 360°, and the speed can be regulated by the proportional control. An additional feature is automatic locking.

OUTSIDE LOCKING ELBOW HINGES

Hosmer outside locking hinges are used for long above-elbow stumps especially in cases utilizing short sockets and hydrostatic pressure for suspension. Because locking and unlocking of this joint requires only minimal cable travel (3 mm) all control movements appear natural and harmonious.

ELECTRICAL POWERED ELBOW

The electric elbow uses the same type of motor as the hand and the wrist but with different gearing. The possibility of utilizing the reduction gear of the hand and wrist for the elbow is being explored. Presently, two motors are used in this elbow to drive the mechanism that makes possible a maximum speed of 1.2 sec. for the complete extension and flexion of the elbow. The lifting capacity is 1.5 kilograms (3.3 lbs.) with a lever arm of 45 cm (18 in.). The total weight of the elbow unit is 400 gr. (14 oz.). It can be used with proportional control.

AMPLIFIERS FOR SINGLE AND MULTICHANNEL CONTROL

The amplifier for both the single and the multichannel systems consists of two elements. One of the elements is the same for both systems, except for a slight modification in the final stage of the multichannel system.

The first element consists of an integrated circuit differential amplifier, with selective response on the frequencies between 200 and

500 Hertz. The gain at maximum sensitivity is about 50,000 times the input signal.

The single-channel system of the second element uses a miniaturized final amplifying unit that can be inserted into the hand itself. In the multichannel system, this element provides an output signal at two levels. The lower-level signal controls a power transistor directly connected to the power supply of the motor for movement, thus providing a control proportional to the contraction of the muscle itself. At the upper level, when the power transistor is in a state of saturation, a relay that inverts the polarity of the motor is activated and remains in operation as long as the muscle is contracted.

When the muscle is relaxed quickly, the motor passes, for an instant, to the preceding condition and a braking action is thus obtained. The two movements are controlled independently of each other and therefore do not interfere with each other. For example, in the myoelectric control of a hand the first level is utilized for proportional control for closing the hand and the second level for the opening of the hand.

BATTERIES

At present we use two types of batteries, the 225 DKZ type and the 450 D type manufactured by Varta. Both types are nickel-cadmium rechargeable batteries. The 225 DKZ type supplies 12 volts at 225 mA/h and weighs 130 gr. It is small enough for it to be contained wholly within the prosthesis. The 450 D type supplies 12 volts at 450 mA/h and weighs 280 gr. It is used with a battery case outside the prosthesis and can be worn anywhere on the body.

With the availability of the prefabricated components it is possible to provide a myoelectric prosthesis for all levels of amputation, ranging from a disarticulation of the wrist to disarticulation of the shoulder.

For the fabrication of the myoelectrical prosthesis, a special technique that uses a thin-wall inner socket has been adopted. Foam plastic is used to give the arm its final shape, and a thin lamination is applied over it. The foam plastic is then removed and the resulting hollow space is used to house electronic components and batteries.

By means of this technique, it has been possible to insert all electrical components into the prosthesis, with benefits not only from the point of view of appearance but also from that of practicality. Because all critical components are protected, they are not subject to damage as if they were mounted externally.

At the present time we have fitted over 1000 amputees with myoelectrical prostheses. These patients have returned to work and utilize their prostheses in manual activities that would not be possible without the use of their prostheses. It is interesting that there are only a very small number of persons who have had difficulty in adapting themselves to this type of prosthesis. The results of a study carried out on the first five hundred amputees we have fitted with the myoelectric prosthesis are presented here.

The levels of amputation for the first 500:

- 254 Right below-elbow amputees
- 168 Left below-elbow amputees
- 21 Bilateral below-elbow amputees
- 1 Bilateral above-elbow amputee
- 2 Bilateral amputees (right BE and disarticulation at left wrist)
- 1 Bilateral amputee (left BE and disarticulation at right wrist)
- 1 Bilateral amputee (right BE and left AE)
- 29 Right above-elbow amputees
- 18 Left above-elbow amputees
- 1 Disarticulation at right elbow
- 4 Disarticulation at left elbow

The results of the study are as follows:

- 54 amputees were able to resume their previous occupations
- 273 were able to take up another occupation
- 65 are students who wore their prostheses continuously
- 12 are housewives who use their prostheses in their housework
- 33 wear the prosthesis although they are so seriously disabled that they do not carry on any activity
- 28 wear the prosthesis occasionally
- 32 amputees do not wear the prosthesis because of the frequent need for repairs and the noiseness. These are disabled persons whose prostheses were constructed before 1969, when the first type of hand was perceptibly noisy. This hand is now being replaced with a new one.
- 3 amputees have not been able to wear their prosthesis because of allergic reactions.

The results reported are extremely encour-

aging if one considers that the study was carried out on the first five hundred amputees fitted with myoelectric prostheses, including a small number of amputees fitted in 1965-66 with the products of our first research efforts. If a comparison is made between the data of investigations carried out by this Centre and investigations by other countries in which the fitting of myoelectric prostheses has begun, one immediately notes the higher percentage of disabled who make effective use of the INAIL-CECA prosthesis. It is not arbitrary to state that such a high percentage has never before been attained.

I think that throughout the world the total number of amputees fitted with myoelectric prostheses must be between 5,000 and 6,000. The results have become more satisfactory since it became possible to apply myoelectrically controlled prostheses to above-elbow amputees and to disarticulations at the shoulder. The utilization of eight myoelectric signals has broadened the possible application of myoelectric prostheses, and, I am sure we still have many possibilities for further development.

SENSORY FEEDBACK

After completion of the research program for multichannel myoelectric control and, after the start of mass production of this type of prosthesis, research was directed towards the feasibiity of incorporating sensory feedback into the myoelectric prosthesis. When text books on limb prosthetics and the proceedings of various congresses are consulted, it will be found that they always refer to mechanical function with little, if any, reference to sensory function. It is a fact that an arm amputee has lost not only mechanical but also sensory function. Consequently, the research team at the Budrio Prosthesis Centre began research to give an artificial hand sensory feedback.

Much information can be transmitted to the human body through the visual and auditory senses. We know quite well how indispensible these senses are in daily life. It is not possible to restore eyesight, but we can restore the sense of touch, to a degree, by transmitting information through the skin using both mechanical vibrations and electrical stimuli.

According to experiments in physiology the skin is most sensitive to vibrations at 200 Hertz, but, vibrations over 150 Hertz are too intensive. For this reason, sinusoidal vibrations around 100 Hertz were tried first, the amplitude of which was modulated by the input signal.

At the beginning consideration was given to a mechanical vibrator. Later this was replaced by an electrical stimulator, because the mechanical devices were excessively bulky, and we began to realize that the information given by electrical stimulation was better perceived than were the mechanical vibrations.

Consequently, an electronic device was developed for sensory feedback information by means of skin stimulation. It consisted of an amplifier section and a final section. The power supply is 12 volts D.C.

The stimulator functions automatically and only when resistance between the fingers is encountered during prehension. As resistance increases, the frequency changes, and the patient is able to evaluate the degree of gripping force. The voltage output of the terminal section is about 75-80 volts. Frequency varies from 0-50 Hertz. The information time lag is so small as to present no problem.

The arrangement avoids excessive irritation of the skin and conserves power.

The electrodes for sensory feedback cannot be incorporated in the socket where the electrodes for myoelectric control are located, because skin stimulation may cause a muscle fibrillation that will upset the use of the control. The sensory feedback electrodes do not create any disturbance of this kind when they are applied on an arm band mounted on the outside of the socket.

The first application of a prosthesis with sensory feedback was on a blind woman with bilateral below-elbow amputations. It has given satisfactory results. Even we research workers were astonished at the accuracy of the information received by this patient, and at the precision with which she controlled the prosthesis.

This research is now at the stage of practical application. There is no doubt that, during this phase, the need for changes will become evident in order to obtain maximum function. One thing is certain, and that is that sensory feedback for an artificial arm must be given consideration because we know that a myoelectrically controlled prosthesis with sensory feedback information renders the amputee prosthesis relationship closer and more perfect, so that even from a purely functional point of view, it is possible to achieve the maximum that can be expected.

The PTB Supracondylar-Suprapatellar Air-Cushion Socket

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The patallar-tendon-bearing supracondylar, suprapatellar air cushion (PTS-AC) below-knee (BK) prosthesis is a special modification of the supracondylar, suprapatellar suspension and the air cusion socket. The air cushion principle in combination with the patella tendon-bearing (PTB) prosthesis was introduced by Lyquist, Wilson, and Radcliffe (Fig. 1), (2) in 1965. It requires more accurate casting, fitting, and fabrication, and importantly, final reduction of stump edema. We have added a refinement to the fitting and the final fabrication of prostheses for BK amputees— the "check-socket" technique, whereby during the fitting period the patient walks in the prosthesis approximately two hours, has periodic examinations for areas of irritation or excess pressure on the stump, and has adjustments made immediately.

Lyquist, Wilson, and Radcliffe (2) documented the results of improved comfort and decreased proximal pressure in a controlled clinical study of five amputees, each of whom for an initial period had worn a PTB hard socket prosthesis and then each was fitted with a PTB-air cushion prosthesis. They highlighted their report with evidence that a former ulceration located at the distal end of a stump healed while the patient was wearing the air cushion socket. Our experience supports their observations. We are now using the air cushion prosthesis with the supracondylar, suprapatellar suspension mechanism with increasing success. We propose that our modifications of fitting and fabrication with the check-socket technique are responsible for this success. The purpose of this paper is 1) to review the concept of the PTB, and the PTB-air cushion prosthesis 2) to describe the fitting and fabrication of the PTS-AC prosthesis, 3) to compare the responses of a group of patients wearing PTB

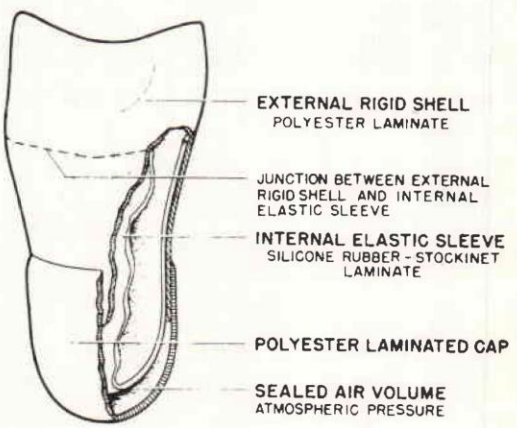

Fig. 1. The Air-Cushion Socket, From (2).

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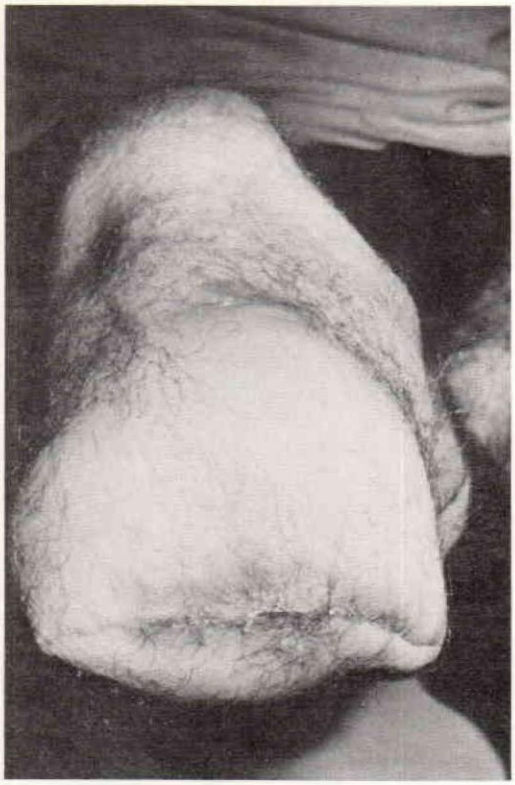


Fig. 2. Condition of below-knee stump prior to fitting with a PTS-AC socket.

prostheses to a group wearing the PTS-AC prostheses, and 4) to point out with illustrative cases the individual indications and accomplishments of the PTS-AC prosthesis. We have two examples of additional modifications. The emphasis of this paper will be on the fabrication of the PTS-AC prosthesis and the value of the "check-socket" technique.

PRINCIPLES OF THE PTB PROSTHESIS AND THE AIR CUSHION SOCKET

Lyquist *et al* had noted the advantages of the PTB prosthesis, particularly the elimination of the thigh corset thus permitting greater freedom of movement for the wearer, and total contact between socket and stump to improve circulation and help control edema. Even with the improvements, they noted that some patients still had edema or excessive pressure on the distal end of the stump. Lack of stump contact, as explained by Foort and Johnson[3], is the cause of most cases of stump edema. Elimination of this type of edema is accomplished by having an ideal stump-socket pressure relationship. This, of course, was the designers' premise for the development of the air-cushion feature of the PTB prosthesis. It was their intention, as well, to reduce proximal stump constriction. A tighter, more secure proximal fit had been used in order to avoid distal friction but had resulted in an increased risk of distal edema.

The PTB air-cushion socket (without the cap) was constructed somewhat shorter than the natural stump length so that when the stump moved distally in the socket with weight bearing, the elastic sleeve would stretch and provide increased pressure on the distal end of the stump, simultaneously contracting around the stump. The effect is similar to that which occurs when a finger is pushed against a suspended sheet of elastic material. An additional source of stump socket pressure is the sealed-in air volume distal to the elastic sleeve. Since volume and pressure are inversely proportional at constant temperature, the pressure in the closed air volume chamber increases as the elastic sleeve stretches under load. Recordings of this air pressure made by the original designers showed that maximum values of 60 to 100 mm Hg (positive) and 30 to 60 mm Hg (negative) pressures are produced during the stance and swing phases of the walking cycle. An air-cushion effect is generated in the stance phase and the negative pressure contributes to the suspension of the prosthesis during the swing phase.

In principle, the ability of the elastic sleeve to conform to the stump increases the area of the distal stump in contact with the socket, and therefore a better functioning, more comfortable, partial end-bearing socket is the result. Reduction of the pressure difference between the distal and the proximal portion of the stump further minimizes the risk of edema. The possibility of skin damage over the distal end of the tibia or fibula is reduced because the elastic sleeve moves distally and encompasses the stump during weight-bearing. It is probable that the gradually increasing pressures exerted around the stump by the action of the elastic sleeve contributes to the reduction of skin movement relative to bony structures.

PRINCIPLES OF THE PTS-AC PROSTHESIS

Our concept and principles of the PTS-AC

prosthesis can be stated in four points which are refinements to the fitting and fabrication process of the PTB air cushion prosthesis of Lyquist *et al*[2]: 1) We have a different method of suspension—supracondylar, suprapatellar. 2) The Silastic sleeve inside the PTS-AC prosthesis is not as long, i.e., it is not brought up inside the socket as high as the fibula head or the tibial tubercle. Thus, any possible irritation of these critical areas is avoided. 3) The socket is contoured to fit exactly the patient's stump, in contrast to the PTB air-cushion prosthesis where the socket is constructed shorter and then covered by an external cap that creates a space to allow for stretch of the Silastic sleeve. We provide this space by adding wax and then melting it out, thereby providing room for the distal movement. Thus the principle proposed initially is fulfilled, but with a more critical fit of the patient's stump. 4) Our check-socket technique permits altering the fit of the prosthesis before final fabrication. In essence, the first negative model without the Silastic sleeve is fitted to the patient, and he walks in this model for approximately a two-hour period. Frequent examinations are made of the stump to determine areas of rubbing or incomplete contact. After these are relieved, or built up, as necessary, a new positive is made. The Silastic is added during the final fabrication process.

FABRICATION

Materials and Equipment:

- Cast sock
- Plaster, 4-inch rolls
- Water
- Model plaster
- Surface-forming files
- Plaster knife
- Abrasive screen
- Wet-or-dry sandpaper
- Parting lacquer
- PVA film
- Dacron felt, ½ oz.
- Nylon stockinette
- Polyester resin: flexible 4134 rigid 4110
- Wood shin block
- Clay, oil-based
- Masking tape
- Cast cutter
- Silastic, 384 Dow Corning
- Paraffin wax
- Glass cloth
- Styrofoam
- Adjustable walking leg
- Vertical fabrication jig
- Vacuum machine
- Hot plate and pot
- Sanding machine
- Hose clamp
- Foot (i.e. SACH foot)
- Ankle block and bolt
- Scissors
- Plastic knife
- Hammer

PROCEDURES

The primary dissatisfaction that prosthetists have had with the air-cushion prosthesis has been the problem encountered in making adjustments to the socket. With the check-socket technique we attempt to make all the necessary changes before adding the Silastic and completing the air-cushion design. An illustrative example is a patient who had been wearing a PTB prosthesis which no longer fitted. His stump showed distal induration, edema, and local cellulitis (Fig. 2). We prescribed for him a PTS-AC prosthesis.

The steps of fabrication are:

Step 1. Place stockinette on the stump. Hold the knee in approximately 20 degrees flexion. (This will vary with the stump length—as length decreases, the amount of flexion increases.)

Step 2. Wrap plaster on the stump. Define the head of the fibula, the proximal border of the medial femoral condyle, and the patellar tendon.

Step 3. Remove cast from the patient's stump. Fill with model plaster. Modify positive model, as required. Paint with parting lacquer and cover with polyvinyl alcohol film under vacuum.

Note. Usually the prosthetist fabricates the air-cushion socket at this point, relying on the cast and his measurements to obtain the proper fit. By using the check-socket technique, we believe a more accurate socket fit can be obtained and distal pressures will be within the tolerance of the patient. With the check socket a true mold of the distal stump is obtained under weight-bearing while walking over approximately a two-hour period (between Steps 3 and 13 of the fabrication process).

Step 4. Place three layers of one-half-ounce Dacron felt and two layers of nylon stockinette on the model (Fig. 3, left).

Step 5. Place a PVA bag under vacuum over model.

Step 6. Impregnate a mixture of 60 percent rigid, 40 percent flexible polyester resin into this material (Fig. 3, right). After the resin is cured, break out the plaster model and trim the proximal border of the socket.

Step 7. Install the check-socket into a wooden block and statically align it on an adjustable BK leg with a foot of the correct size (Fig. 4).

Step 8. Place the prosthesis on the patient and check the fit of the socket.

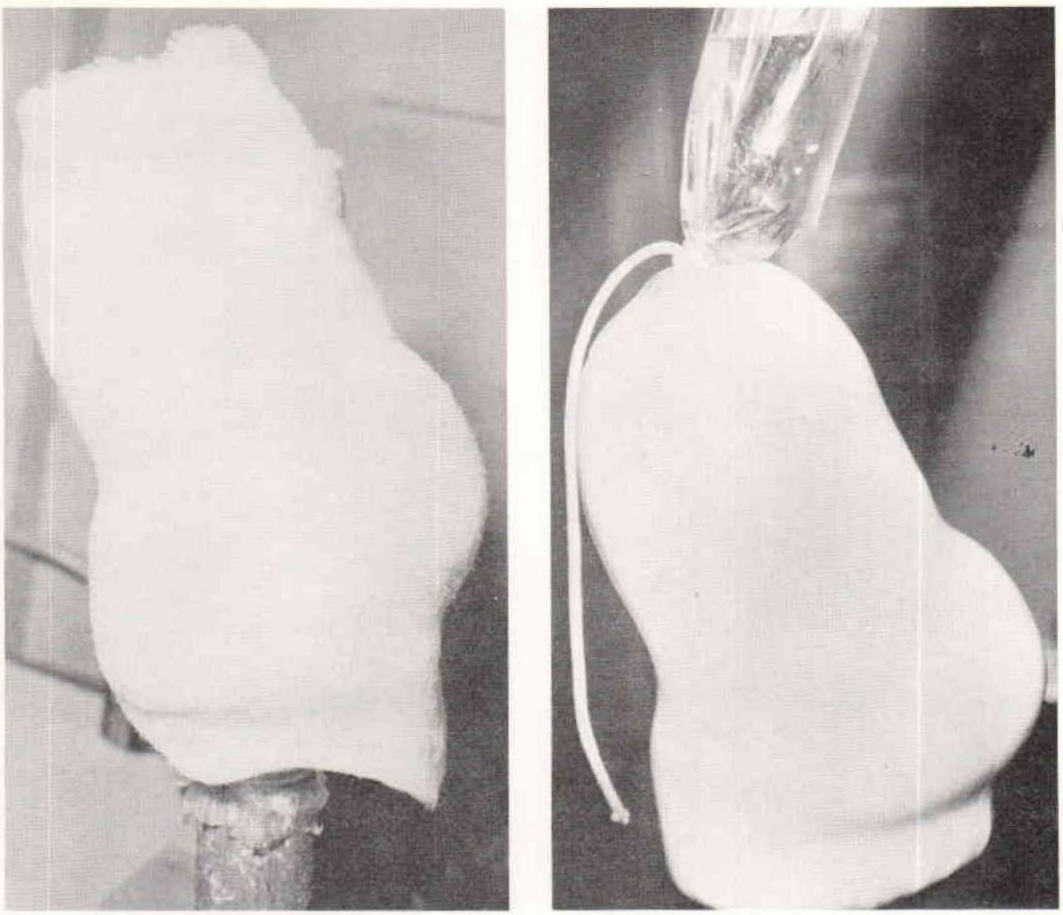


Fig. 3. Left, Model of stump covered with three layers of one-half ounce Dacron felt and two layers of nylon stockinette; Right, Impregnation of Dacron felt and nylon stockinette with a 60-40 mixture of rigid and flexible polyester resins to provide the check socket.

Step 9. Align the prosthesis dynamically (Fig. 5). Examine the stump for signs of irritation or pressure.

Step 10. Adjust the socket by adding leather fill to correct loose areas (Fig. 6) or by grinding where there appears to be pressure sites.

Step 11. Add small amounts of oil-based clay to the distal portion of the socket (Fig. 7).

Step 12. After the patient has walked on the prosthesis, ideally for two hours, check the distal socket for imprint, and add clay until firm, even contact is obtained. Examine the stump at intervals of time for areas of irritation or excess pressure. Analyze the impression of the stump sock in the clay. When the fitting is complete, the clay in the distal portion of the socket will show sock marks evenly, and will blend into the plastic walls of the socket (Fig. 8).

Step 13. Mark the foot-socket relationship, remove the foot, and place the prosthesis in the vertical fabrication jig.

Step 14. Extend the proximal border of the socket with tape, and fill the socket with plaster.

Step 15. After the plaster has hardened, remove the adjustable leg and wood block, and bivalve the socket (Fig. 9).

Step 16. Remove the socket from the plaster model and sand the model with fine wet-or-dry sandpaper.

Step 17. Paint the prepared model with a parting agent. Now the fabrication of the air cushion socket begins.

Step 18. Place three layers of nylon stockinette over the model.

Step 19. Pull a PVA bag over model. Place under vacuum. Attach a hose clamp lined with rubber, and tighten (Fig. 10). This will determine the proximal border of the Silastic sleeve.

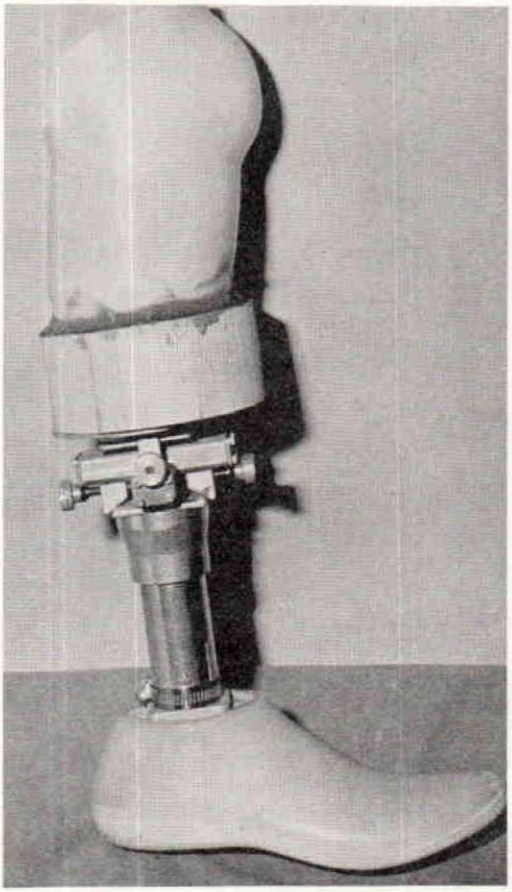


Fig. 4. The check socket mounted on a below-knee adjustable leg.

Step 20. Impregnate into the nylon a mixture of Dow Corning Silastic #384.

Step 21. When the Silastic has hardened, build up the distal end of the model with paraffin wax to the thickness of one-half inch and taper it up the walls (Fig. 11).

Step 22. Cover the wax portion with glass cloth to add strength.

Step 23. Add three more layers of stockinette and laminate with a mixture of 35 percent flexible and 65 percent rigid polyester resin.

Step 24. When the plastic has set, cut a hole in the end of the socket (Fig. 12).

Sept 25. Place the socket in hot water to melt out the wax. Seal the wax escape hole to create the sealed air chamber.

Step 26. Place the socket back in the vertical fabrication jig in the same position that the final dynamic alignment was obtained. Replace the adjustable leg with Styrofoam.

Step 27. Shape the prosthesis.

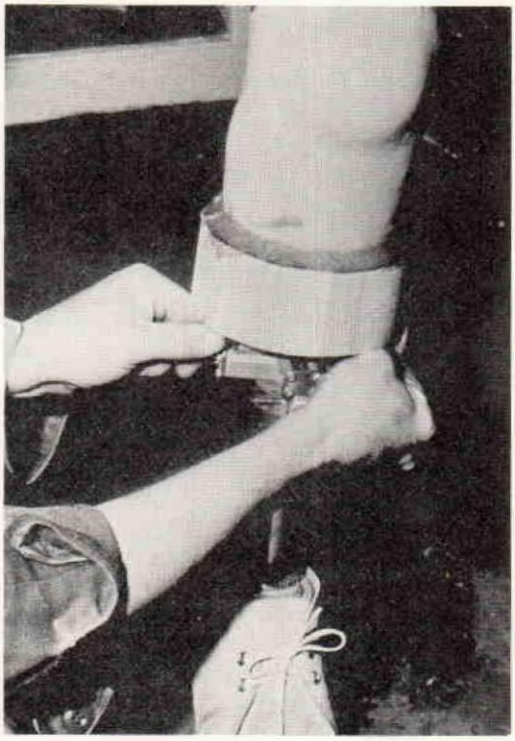


Fig. 5. Dynamic alignment of the temporary prosthesis.

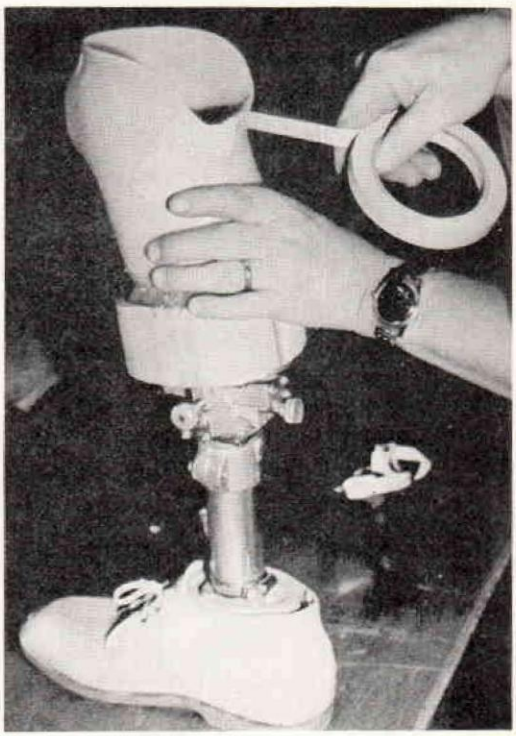


Fig. 6. Leather is used to fill loose areas when they exist.

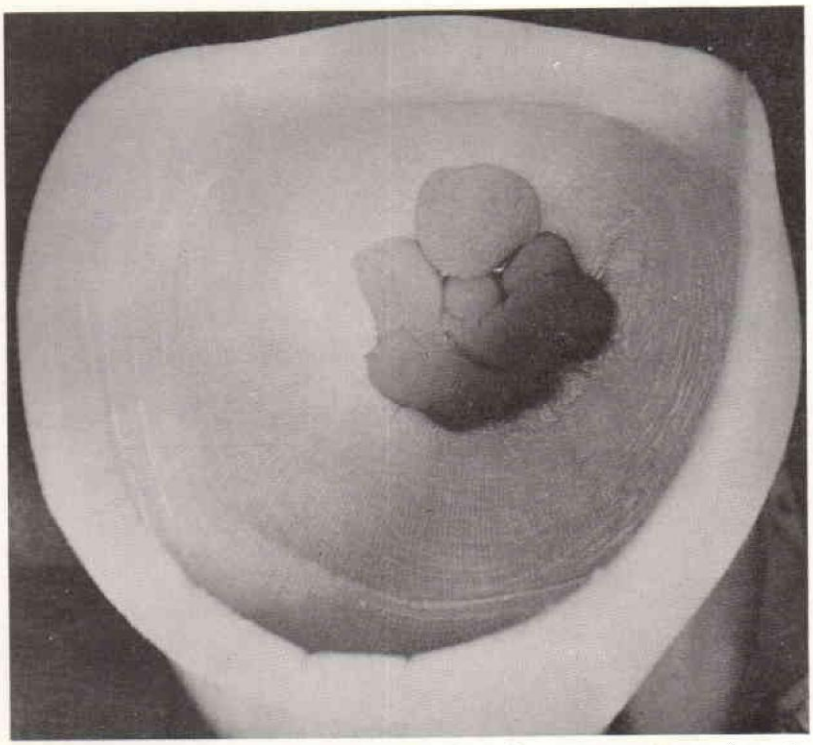


Fig. 7. Oil-based clay is placed in the distal portion of the socket to determine adequacy of socket fit in the distal portion.

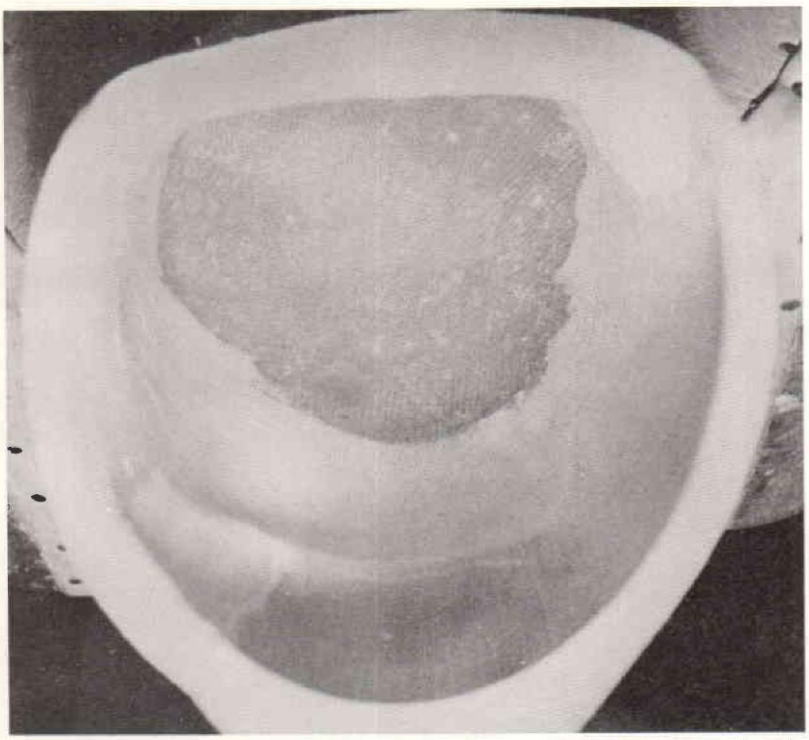
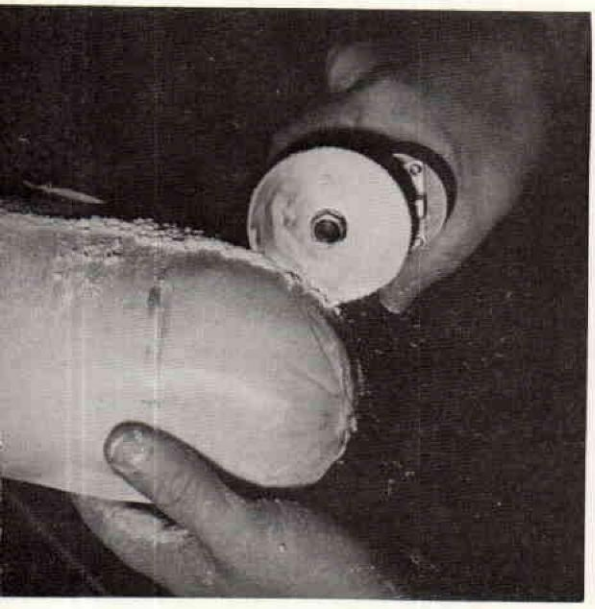


Fig. 8. Oil-based clay is blended into socket wall by pressure from the stump. A perfect mold is attained when sock marks are even.

Fig. 9.



Bivalving the "check-socket" from the new positive mold.

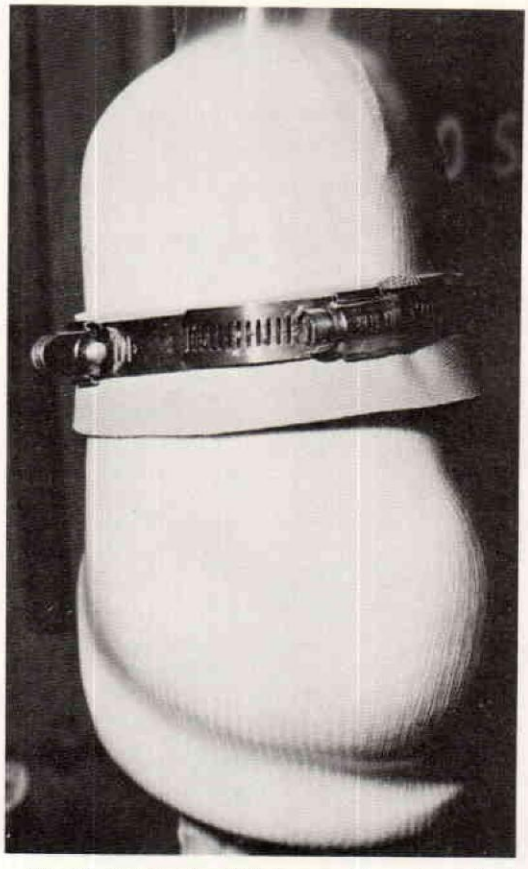


Fig. 10. Stockinette, PVA bag, and hose clamp lined with leather in place in preparation for adding Silastic. Hose clamp defines proximal border of Silastic sleeve.

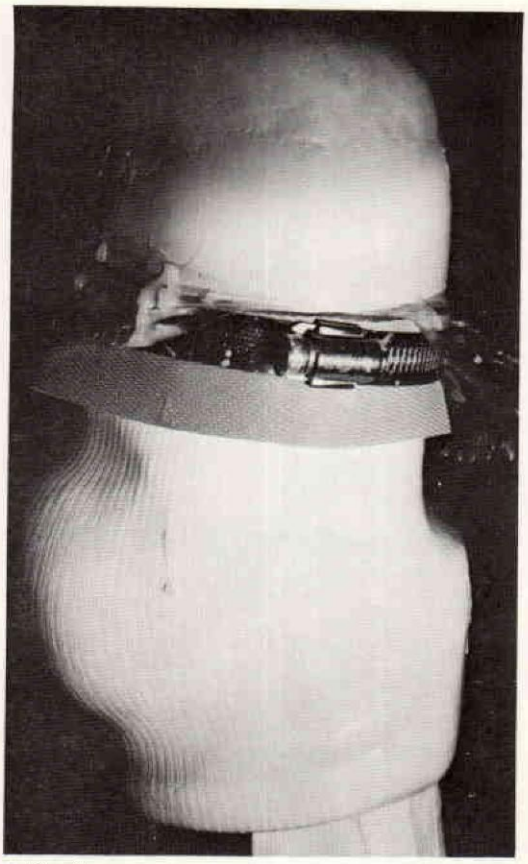


Fig. 11. Addition of wax to build up the distal socket area to create the space for the sealed-in air chamber.

Fig. 12. A hole is cut in the socket end to provide escape for melted wax.

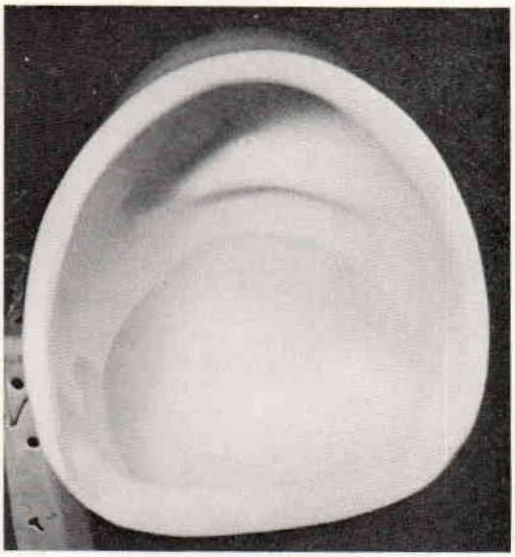


Fig. 13. Finished prosthesis smoothed and ready for delivery. Internal contour of Silastic sleeve can be seen.

Step 28. Complete the lamination with two more layers of nylon stockinette.

Step 29. Break out the plaster model and smooth the proximal edges of the socket. The Silastic sleeve, which can be seen from the inside, is contoured to the patient's stump (Fig. 13).

Step 30. Place the foot on the prosthesis and deliver the prosthesis to the patient (Fig. 14.).

Figure 15 shows the patient's stump after he had been wearing the PTS-AC prosthesis for three weeks.

The check-socket technique requires two to three hours additional fabrication time, but we believe that this is well worth the effort because a more satisfactory fit is obtained, and the number of return visits for readjustment are reduced.

RESULTS

We were able to canvas 21 of our amputees —10 wearing PTS-AC prostheses and 11 wearing patellar tendon-bearing prostheses. Although the numbers were not within the realm of statistical analysis, we can present the results of our questionnaire (Table 1) and comment on the trends of function, durability, and cosmesis. Cases are reported to illustrate the trends.

THE SURVEY

Each group of amputees disclosed in the questionnaire the number of hours the prosthesis was worn daily. The PTS-AC and PTB wearers claimed approximately the same

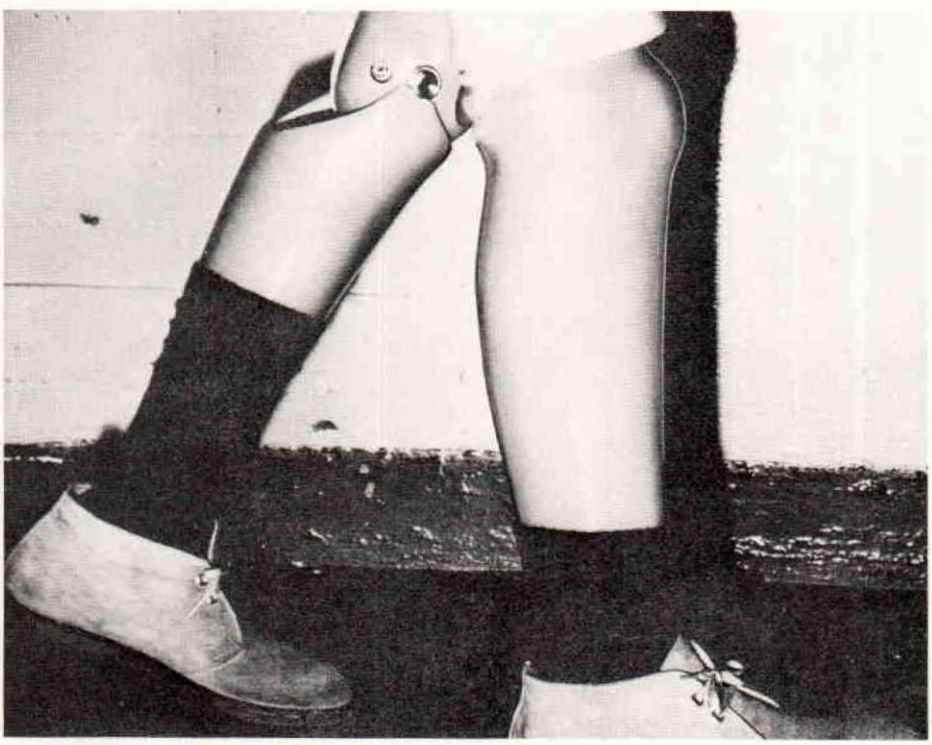


Fig. 14. Patient walking with a PTS-AC prosthesis.

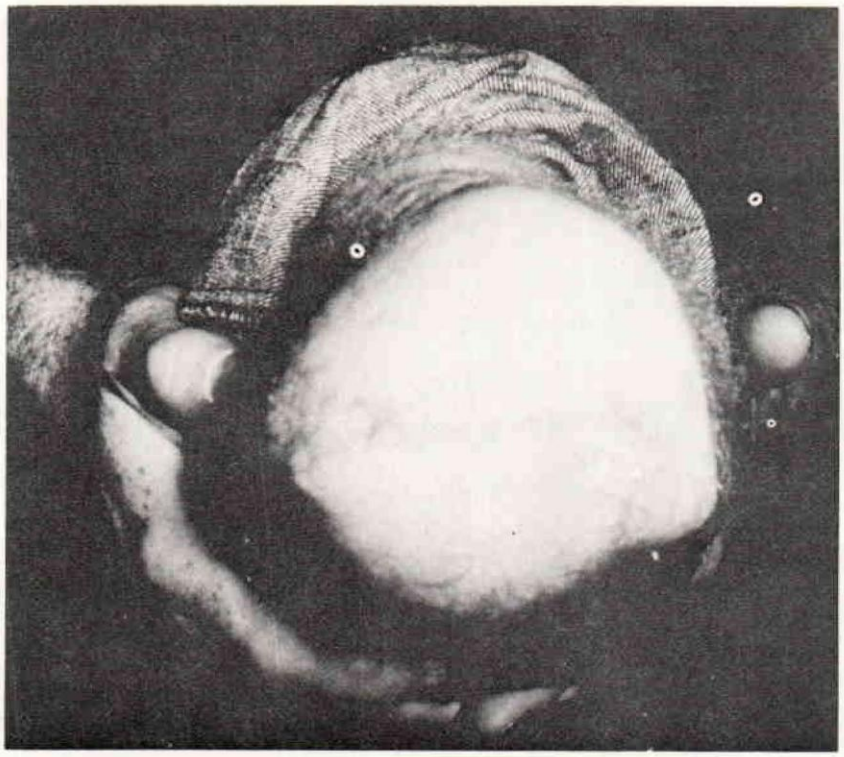


Fig. 15. Patient's stump after wearing PTS-AC prosthesis for three weeks. Inflammation and edema have resolved.

length of time per 24-hour period (averages, 14.1 and 15.2 hours).

Seven PTS-AC wearers had no distal stump breakdown. Three reported "slight" breakdown. Two of these decreased the size of their stump breakdown while wearing the PTS-AC prosthesis. One of these had formerly worn a conventional below-knee prosthesis with slip socket, external joint, and thigh lacer. A two centimeter distal ulcer had been a chronic problem for him. A PTS-AC prosthesis was prescribed following control of infection; no further ulceration has occurred. Of the PTB prosthesis wearers, four reported no breakdown, two slight, and five reported ulcers greater than one-inch in diameter.

Each group reported good to excellent results on hard flat surfaces and each had comparable results on irregular hard surfaces. On irregular soft surfaces and hilly terrain, the PTB user is slightly less stable. The supracondylar suprapatellar suspension, as well as the air cushion principle, probably is the explanation for the increased stability.

The PTS-AC prosthesis was not as accommodating to the wearer when kneeling. Many had discomfort and some felt considerably unstable. In comparison, no PTB wearer reported that he was unable to kneel, and only a few indicated discomfort.

A minor difference in the patients' satisfaction with cosmesis existed in these two groups. Two of eleven PTB wearers desired an improvement. Although this is not a significant concern of most of our patients, the supracondylar, suprapatellar construction does improve appearance, a simple but important esthetic consideration.

There appeared to be minimal difference in restriction of knee motion in these two groups during ambulation and stairclimbing; and there was little variation in the patients' reports of prosthetic security.

Only one patient in the PTS-AC group expressed a desire for additional suspension. All patients in the PTB group used a supracondylar suspension strap, and seven expressed either a desire for additional suspension or a dissatisfaction with the strap. After accidental falls several patients in the PTS-AC group reported dislodgement of their prostheses. Whether this is an advantage or a disadvantage could not be determined.

The PTB wearers required more adjustment to their prostheses than did the PTS-AC group (16 visits compared to 8).

ILLUSTRATIVE CASE REPORTS

Fig. 16. *Case No. 1.* Note the pleasing cosmetic appearance.

Case No. 1

A 44-year-old housewife sustained a left below-knee amputation in an automobile accident when she was a teenager. She was fitted with a conventional below-knee prosthesis with thigh lacer and external knee joints. After a long period of use, two attempts to convert her to a PTB prosthesis wearer were unsuccessful. Local stump pressure, secondary to bony prominence, was a problem. She was fitted successfully with a PTS-AC prosthesis approximately eight months ago. She is highly satisfied with this prosthesis as to comfort, weight, and cosmesis (Fig. 16). Since wearing the PTS-AC prosthesis she has had an increase in the size of the quadriceps muscle to the extent that when she donned her former prosthesis, a thigh roll was created proximally (Fig. 17).

Fig. 17. Added quadriceps muscle bulk demonstrated by Case No. 1 (Fig. 16) when she donned her former conventional BK prosthesis after wearing the PTS-AC for eight months.

Case No. 2

A 21-year-old Vietnam veteran sustained an open devitalizing wound approximately one year before he was fitted with the PTS-AC prosthesis (Fig. 18). Chronic distal

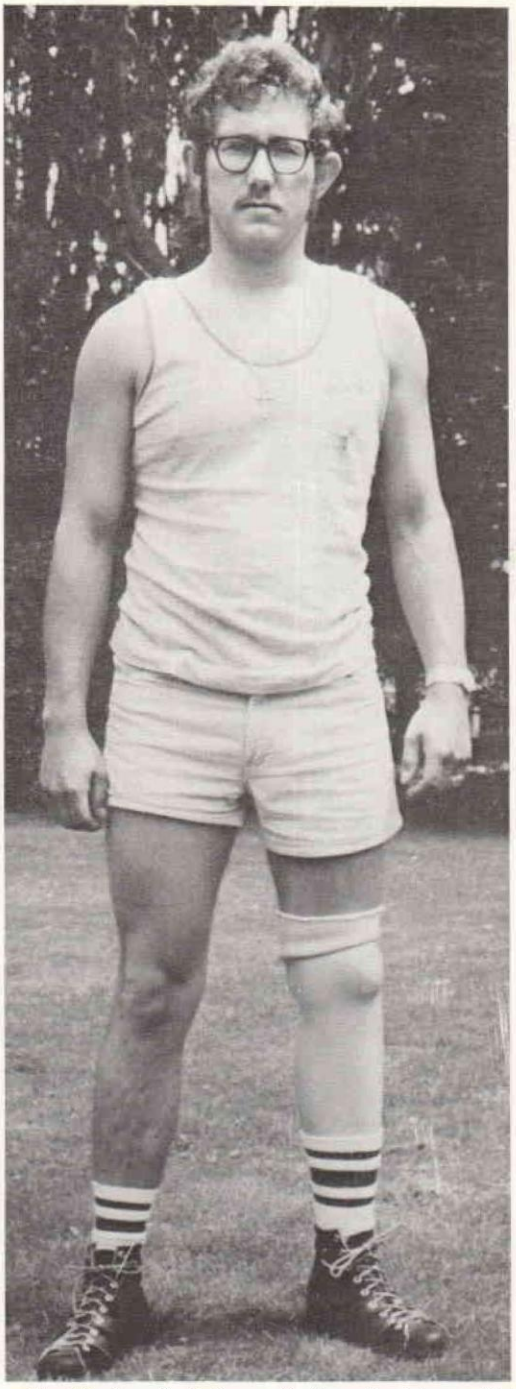


Fig. 18. *Case No. 2.* Secure fit of prosthesis is provided for demanding activities.

induration has subsided while wearing the PTS-AC prosthesis. He is employed now as a salesman, wears his prosthesis an average of ten hours per day, and drives daily a pick-up truck with a standard shift transmission.

Fig. 19. *Case No. 3.* Teacher and part-time interscholastic coach who has excellent functional result.

Case No. 3

A 24-year-old soldier sustained an open devitalizing wound secondary to a land mine explosion. After using a PTB prosthesis for three months he was fitted with a PTS-AC prosthesis. This patient has demonstrated an excellent gait pattern for a below-knee amputee. Presently, he is a teacher of biology and mathematics, and works two to four hours per day on the athletic field as an interscholastic football coach (Fig. 19).

Case No. 4

This patient has a left above-knee and a right below-knee amputation resulting from injuries suffered during combat in Vietnam. He is presently fitted with a left above-knee prosthesis with suction socket, hydraulic "SNS" knee unit, and SACH foot. His four-

Fig. 20. *Case No. 4.* Above-knee and below-knee amputee wearing a PTS-AC prosthesis.

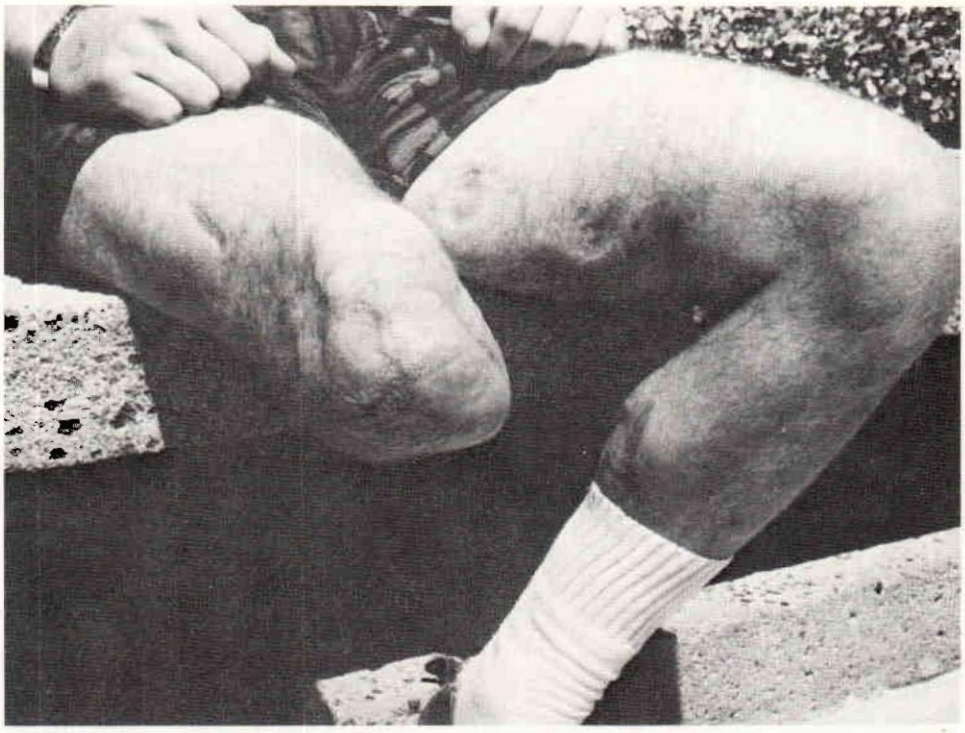


Fig. 21. *Case No. 5.* Three-inch right BK stump which is nearly totally covered by split thickness skin graft.

inch BK stump is fitted with a PTS-AC prosthesis (Fig. 20). Initial distal stump edema and breakdown associated with infection has completely resolved while wearing this PTS-AC prosthesis (Figs. 1 and 15).

Case No. 5

This 24-year-old Vietnam veteran has a two-inch below-knee amputation which is nearly completely covered with split thickness skin graft (Fig. 21). Several areas of breakdown complicated by *Staphylococcus aureus* infection protracted his fitting and ambulation. A PTS-AC prosthesis improved the condition of his stump and allowed longer periods of walking, but his requirements for this labile stump to stand up to the pressures

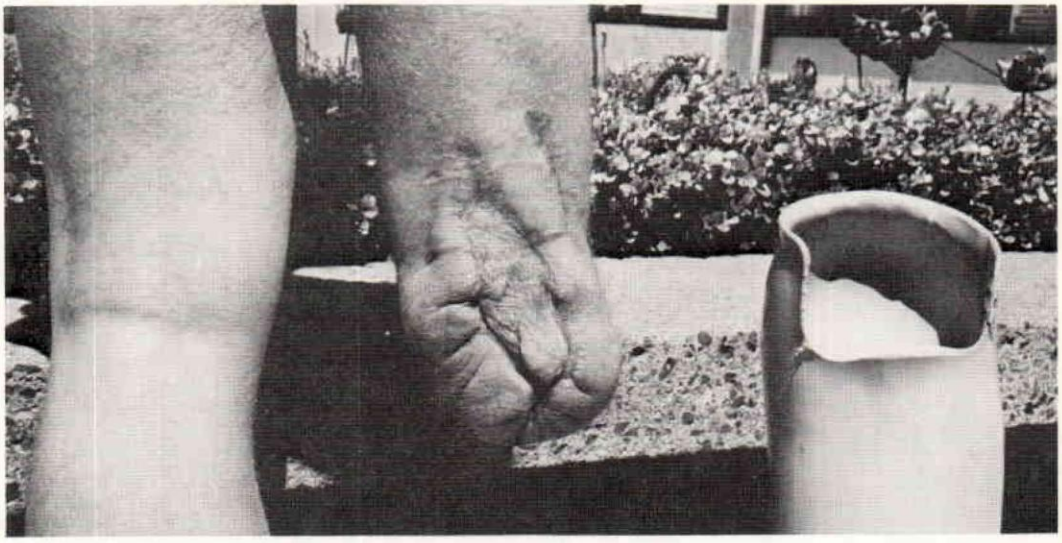


Fig. 22. *Case No. 6.* Popliteal scarring in a short BK stump with posterior and distal split thickness skin graft.

in snow and water skiing, golf and camping were not completely resolved with the latter prosthesis. The patient is presently wearing a patellar tendon-bearing supracondylar, suprapatellar soft socket BK prosthesis with KEMBLO liner. This has been the most satisfactory prosthetic prescription for him to date. The various legs provided by several fittings can be used on demand.

Case No. 6

This 23-year-old Vietnam veteran had the complications of popliteal scarring and of split thickness skin graft stump coverage (Fig. 22). He was bothered by scar sensitivity and a prominent distal fibula in his PTB prosthesis. A PTS-AC prosthesis was more comfortable and allowed longer periods of ambulation, but further modification was necessary. A patellar-tendon-bearing supracondylar suprapatellar soft socket BK prosthesis with KEMBLO liner has resulted in a more satisfactory prosthetic prescription.

COMMENT

A specialized below-knee prosthesis which combines the modifications of supracondylar, suprapatellar suspension and the air-cushion socket design has been described.

The theoretical function and basic design, along with our present method of fabrication and alignment, stressing the importance of the check-socket technique is presented. A comparison of two groups (10 wearing PTS-AC prostheses, 11 wearing PTB prostheses) in eleven categories provides a generalized evaluation of the clinical applications of each type. Several successful fittings have been presented, highlighted by the control and elimination of ulceration and edema and by the conversion of two patients from conventional BK prostheses (thigh lacer and external joints). Two cases required additional prosthetic consideration—the first because of complete stump split thickness skin graft coverage coupled with his active avocational interests; the second because of scar sensitivity.

Although we use the prosthesis most successfully in the short BK stump, when bony prominences exist and when skin cover is less than ideal, we see no reason why the PTS-AC prosthesis could not be used for any BK amputee.

TABLE 1
Results of a Questionnaire

QUESTIONS	PROSTHESIS	
	PTS-AC*	PTB**
Average number of hours worn in 24-hour day	14.1	15.2
Stump breakdown		
None	7	4
Slight	3	2
Ulceration (1" in diam.)	0	5
Usability		
Hard flat surfaces	G to E	G to E
Less hard flat surfaces	G to E	G to E
Irregular hard surfaces	F to G	F to G
Irregular soft surfaces	F to G	
Hilly surfaces	F to G	F
Stair climbing	.....No difference.....	
Kneeling	Discomfort Unstable	Les discomfort Less unstable
Cosmesis	. . .Minor difference. . .	
Restriction of knee motion	.....No difference.....	
Security		
When straight	E	G to E
When bent	G	G
Suspension	G	Desire more
Prosthesis came loose after accidental fall	Several "yes"	Less than PTS-AC disadvantage?/ Advantage?
Revisits for adjustments to prosthesis	8	16

*PTS-AC = patellar-tendon-bearing-supracondylar, supra-patellar air cushion prosthesis, 10 patients surveyed.
**PTB = patellar-tendon-bearing prosthesis, 11 patients surveyed. Other abbreviations: E=Excellent, G=Good, F=Fair.

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2. Lyquist, E., L.A. Wilson, and C.W. Radcliffe, *Air-cushion socket for patellar-tendon-bearing below-knee prosthesis: Principles and fabrication procedures,* Tech. Memo., Biomechanics Laboratory, University of California, San Francisco and Berkeley, April 1965. (Reprinted in *Bull. Pros. Research,* Fall 1968.)

Dynamic Control of Abnormal M-L Motion of the Os Calcis: The Cushion Heel Wedge— A Possible Solution

JOHN GLANCY, C.O.[1]

The efficient control of excessive mediolateral motions of the foot has been a problem for many years. Opinions remain mixed as to the best approach to managing the variety of problems which result from foot imbalance. The growing child's feet continue to be of particular concern, since the restoration of foot balance at an early age is generally considered to be the best way to prevent bony deformation later.

At present, the mechanical methods of controlling abnormal varus or valgus motion of the os calcis may be said to share a common feature—the control they offer is static. That is the os calcis is pre-positioned in the position of "correction" or "balance," and held there forcibly. The most frequently used method of control for children is the heel wedge, in conjunction with, or without, a "Thomas heel." The practice of prepositioning the os calcis to achieve foot "balance," by wedging, forces the os calcis to rest on an *inclined surface* that is *not parallel* to the floor. It is, therefore, difficult to accept the claim that such a practice does indeed restore foot balance. The analysis that follows led to a search for a dynamic yet simple means of controlling abnormal M-L motion in the subtalar joint. The cushion-heel wedge appears to be a practical solution for either varus or valgus conditions involving the hind foot.

THE STANDING POSITION

The downward arrow in Figure 1 represents the body's weight passing through the midline of the os calcis, shown in dotted outline in a standing position. The circle in the center of the os calcis represents its axis, arbitrarily located. The rectangular block represents the heel of the shoe. The two upward arrows represent the floor reaction force, and the assumption is made that the

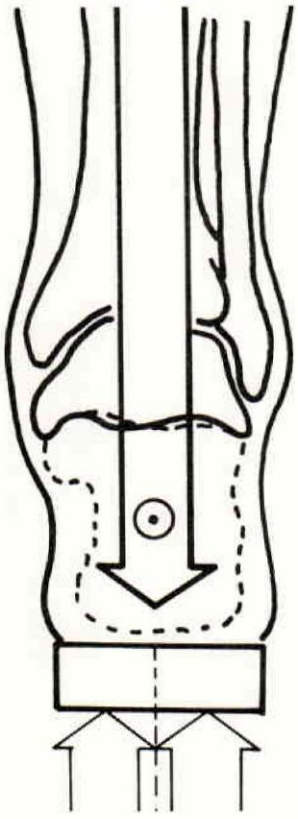

Fig. 1. Schematic posterior view of the right heel in balance in the standing position.

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distribution of the reaction force has to be equal on both sides of the midline of the heel of the shoe in order to maintain normal foot balance.

BETWEEN FOOT-FLAT AND STANDING

Figure 2 attempts to show the os calcis at an arbitrary point in time between the instant that foot-flat has occurred and the time the tibia reaches the standing, or upright posi-

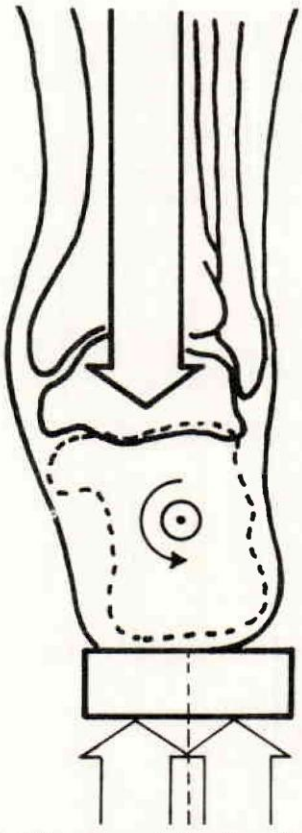

Fig. 2. Schematic posterior view of the right heel showing eversion of the os calcis when the foot is pronated.

tion. As the tibia rotates forward during this period, the weight-bearing line moves medially and the amount of the body's weight that is being borne by the foot increases and reaches its peak as the full weight of the body is registered at midstance of the walking cycle, or one-half the body's total weight, when both tibiae reach the standing position.

The standing position subjects the foot to a *constant* load for much longer durations. The assumption is made that the constant load of long duration during periods of standing is the major destructive force to normal foot balance when structural weakness is present.

In the common situation where inversion motive power and/or the ligament system on the medial portion of the foot is inadequate, there appears to be no internal means of resisting the counterclockwise (eversion) rotation which occurs, except the configurations of the superior articulating surface of the os calcis and the inferior articulating surface of the talus. It should be noted that, at this time, the fatty pad and skin under the bottom of the os calcis has not fully flattened, yet the counterclockwise rotation of the os calcis has begun.

It is suggested that the os calcis will be "prepositioned" in pronation before the vertical loading has reached its peak, and that the final "settling," as the standing position is reached, is probably compression of soft tissues. At the point when even a small fraction of the vertical load makes contact with the articulating surface of the os calcis, the body's weight might be described as being in "free-fall"—there being negligible known resistance until full "bone-to-bone" contact serves as a positive mechanical stop. The pronated foot then becomes "locked" between two equal and opposite forces—the body's weight from above and the floor reaction force from below.

The force moment created by the body's weight in "free-fall" about the axis of the rotating os calcis can be assumed to be of a high magniture. For example, as the os calcis rotates laterally from under the limb above, the weight line will then fall *medial* to the axis of the os calcis. If the patient weighs 50 lbs. and the weight line passed ¼ in. medial to the axis, the moment would be 6.25 inch-pound about each foot in the *standing position*. The "trigger" that sets the counterclockwise direction of the os calcis in the beginning appears to be the normal valgus accommodation that the foot makes to the normal internal rotation of the tibia in the transverse plane which occurs between heel-strike and midstance. In the previous example, the moment about the axis of the os calcis during *midstance* (the full weight of the body now being borne upon a single foot) increased to 12.5 inch-pounds[2]. Thus, the abnormal

[2] The fact that, at midstance, the weight bearing borne by the foot (especially the longitudinal arch) is *twice* that which is borne during standing is significant to arch supporters and/or shoe inserts designed to support the longitudinal arch when it is structurally weak. Since the

valgus position of the foot during standing may be said to be predetermined by conditions in effect *before* the standing position is reached.

The floor reaction force appears to be equalized on both sides of the midline of the heel. Its role now would seem to be that of maintaining the "overall" M-L balance of the body, rather than foot balance, to accommodate the changed circumstances.

THE MEDIAL HEEL WEDGE

As the foot receives its peak load during the standing position, clinically, the wedge appears to restore M-L balance to the foot. The wedge "checks" the counterclockwise moment by prepositioning the os calcis on an inclined plane in relation to the floor (Fig. 3). This prepositioning is achieved by rotating the os calcis clockwise in the direction of inversion as it rests on the inclining wedge. In effect, the medial portion of the articulating surface of the os calcis is jammed upward to become the first point of contact with the medial portion of the inferior articulating surface of the talus as the body's weight is received. The heel counter of the shoe is expected to maintain the os calcis in its unnatural position. The "free-falling" of the vertical load from above appears to be effectively checked. However, the effect is more apparent than real.

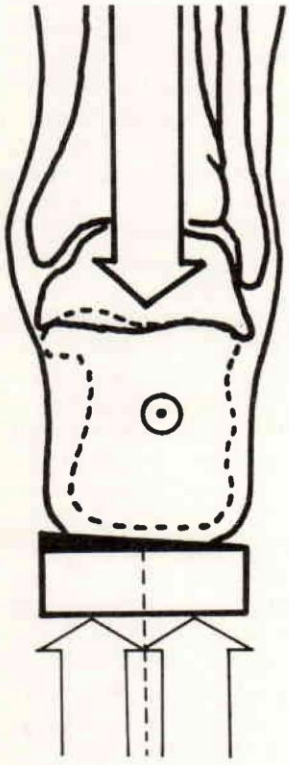

Fig. 3. Schematic posterior view of the right heel in the standing position when a conventional medial heel wedge is used.

The weakness of the arrangement provided by the medial heel wedge is that the os calcis rests on an *inclining plane* which slopes downward in the direction of *eversion*. By forcibly prepositioning the os calcis thus, its articulating surface is also inclined downward in the direction of eversion (Fig. 4). Under such circumstances, it would seem valid to conclude that as the talus receives the

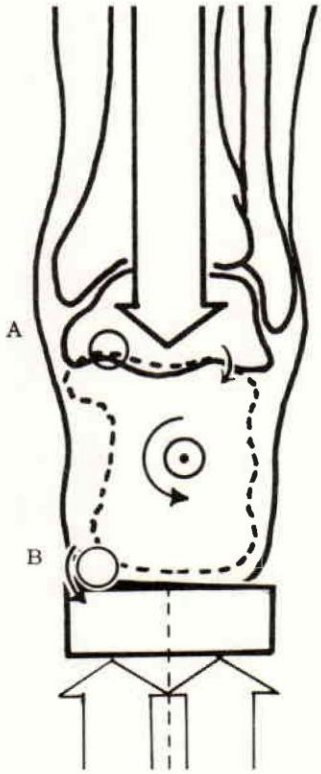

Fig. 4. Schematic posterior view of the right heel showing the relatonship between the os calcis and talus during the early part of the stance phase when the medial heel wedge is used.

application of the body's full weight is inexorably applied *prior* to reaching the standing position and can be lowered only when both feet share the load, it appears that an important element of the dynamics of the normal foot is not being considered in the design of current arch supporters. Assuming that the orthotic goal should be to achieve as close a mechanical simulation to nature's dynamics as possible, the ideal longitudinal arch supporter should yield sufficiently, up to and at midstance, to permit the normal amount of valgus to occur and then *return* to sustain the arch in a normal standing position.

body's weight its first point of contact with the inclined articular surface of the os calcis must be its medial edge. It would appear to follow that this point of contact then becomes an instant center, A, about which the talus must rotate. Since the os calcis position is fixed, the moment about the instant center will be lateral, where it may be assumed there exists little, or no, resistance. The direction of rotation of the talus is now inevitable—clockwise and downward in the direction of inversion. The vertical load might be described as "free-falling," until the abnormal spacing within the talocalcaneal joint has been reduced by full contact of the two surfaces. The rapid increase of body weight upon the talus "locks in" its inclined position as it rests upon the inclined os calcis.

In such a situation, it appears that one of two things takes place:

1. Assuming that the mortise formed by the malleoli of the tibia and fibula does not permit mediolateral rotation of talus within the ankle joint, the leg must then follow the talus as it rotates clockwise within the talocalcaneal joint, as previously described. Thus, the line of the body's weight as it is borne through the talus to the os calcis and down to the floor would no longer be a true vertical line. The weight-bearing line would now be oblique and medial to the axis of the os calcis. The net effect is that all forces are so aligned that conditions are "go" for the foot to revert to the pronated position. It is merely a matter of time, for no shoe can resist the forces being generated under the conditions just described.
2. The final "settling" mentioned earlier, i.e., the compression of the fatty pad under the os calcis, may be sufficient to allow the entire leg (fixed os calcis, talus, *et al.*) to "right" itself and bring the weight-bearing line to a true vertical line. The contact point between the soft tissues of the heel and the innersole of the shoe becoming the center point, B, of rotation. It appears reasonable to assume that there would be some translatory motion, along with the rotation occurring tween the heel and innersole. The key factor that would determine which of the two above events would occur would seem to be that the soft tissue under the heel must be able to compress (when the peak load is reached) an amount equal to the height of the thickest portion of the wedge, thus enabling the weight-bearing line through the leg to "right" itself to a true vertical line in relation to the floor. The net effect, in the latter case, is that the os calcis *immediately* reverts to a position of eversion taking the foot with it.

In addition, regardless of which event occurs, the propensity of the tibia to rotate internally between foot-flat and the standing position, accompanied by the valgus accommodation of the foot, continues to apply a counterclockwise moment about the axis of the os calcis with each step. The os calcis, as it rests on the sloping wedge, it in an extremely weak position to resist what can be assumed to be a counterclockwise moment of considerable magnitude. With surprising rapidity, the lower, lateral portion of the os calcis will force the lateral heel counter to bulge outward. The outward bulging of the lateral counter permits the os calcis to slide down off the wedge and resume its pronated position. A shoe that fits poorly at the heel, or a loosely laced shoe, expedites the process. Throughout the process described, the floor reaction force seems to be a "negative factor" in terms of foot balance *per se*. It appears to be constantly equalizing (either side of the midline of the heel) to accommodate to each change as it occurs, thus maintaining M-L balance of the body, whatever the foot's position.

THE CUSHION-HEEL WEDGE

Figure 5 attempts to depict the action of the cushion heel wedge the instant that the foot-flat position has been achieved and the leg is beginning to rotate forward over the foot now "fixed" to the floor to receive the weight of the body. As the leg rotates forward, there is a rapid increase in the amount of body weight passing through the leg to the foot and downward to the floor. This increase continues until the peak of significance to foot balance is reached, that is, when the tibia is at right angles to the foot and/or floor.

The cushion heel wedge must begin to depress immediately as the body's weight is received by the os calcis. *Immediate depression is essential* in order to create a clockwise moment in the direction of inversion about the axis of the os calcis. Should there be a delay in the depression, as would be the case if the material used in the cushion portion of

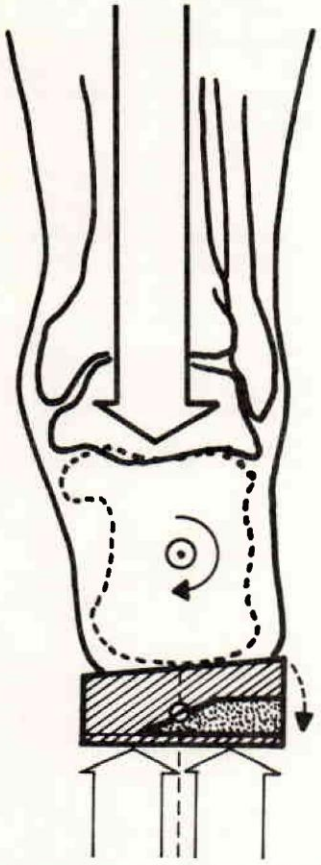

Fig. 5. Schematic posterior view of the right heel at the first instant of the foot-flat positon. A suggested new type of heel wedge is shown.

the heel was too firm, the weak medial structures, coupled with the *inward* sloping heel, would cause the body weight to place a counterclockwise moment about the axis of the os calcis. Thus, the combination of the body's weight from above, and the instantaneous depression of the cushion-heel wedge below dynamically prepositions and maintains the os calcis in normal alignment for the midstance and/or standing positions.

To accommodate for the standing position, the cushion heel wedge should be made so that the amount of depression has been sufficient to bring the whole heel parallel to the floor. The os calcis will rest on a level horizontal plane, parallel to the floor. The floor reaction force "shifts" to the medial half of the heel so that it is substantially greater under its firm medial portion.

The body's weight might still be described as "free-falling," since there is very little initial resistance to it as it is being absorbed by the cushion wedge. It is assumed that the cushion wedge stores an unknown portion of the vertical load before it can make contact with the floor. During this short period, an unknown portion of the vertical load is passing through the firm medial portion of the heel (note its extreme medial displacement) to the floor without interruption (Fig. 6). The floor reaction force must equal what appears to be a proportionally greater vertical load from above, passing through the medial

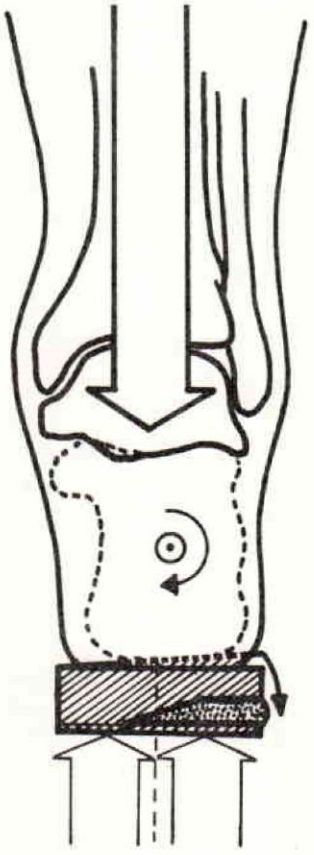

Fig. 6. Schematic posterior view of the right heel in the standing position with the cushion-heel wedge.

half of the heel, than that which is making contact with the floor through the cushioned lateral half.

DESIGN CONSIDERATIONS

The amount and type of cushion material must be such that, as the peak of the vertical load is achieved, the cushion must not "bottom out" when the heel becomes parallel to the floor. If the cushion wedge were to "bottom out", the effect would be similar to a solid medial wedge. Thus, the lateral portion of the os calcis is somewhat "floating" on the

remaining thin layer of air within the cushion. The combination of the body's weight having been forced to create a clockwise moment, and a preponderance of the floor reaction force being "shifted" to the medial side, under the medial half of the os calcis owing to the absorption of forces by the cushion wedge on the lateral side, results in what appears to be a constant force couple about the subtalar joint which maintains foot balance whenever the limb is in the standing position.

SUMMARY

This analysis of conventional heel wedges raises doubts as to their effectiveness in controlling abnormal mediolateral rotation of the os calcis. Heel wedges do not appear to be an effective means of restoring foot balance in the young. A cushion-heel wedge is suggested as a possible solution for young patients. The cushion heel wedge is not intended to be used in situations where bony wedging or abnormal articulations, or both, are present.

Inflatable Wedge Suspension System

TIMOTHY STAATS, M.A., C.P.[1]

In the past decade, the introduction of supracondylar suspension systems for below-knee prostheses has given the prosthetist a wide selection of variants from which to choose (2). Recently, at the University of California, Los Angeles, Prosthetics-Orthotics Program, an ingenious variation of the supracondylar hard wedge suspension system was introduced by inventor and amputee, Lincoln Baird. The new system involves an inflatable bulb (Fig. 1) placed superior to the medial femoral condyle in a position similar to the Fillauer hard wedge (1). The prototype models of the inflatable wedge fitted to amputee subjects showed that it provided secure, comfortable, reliable suspension as well as built-in adjustability not available in other suspension methods.

FIGURE 1

The inflatable wedge system was fitted to five below-knee amputees. Because the systems were prototype models, the fittings were not considered to be definitive.

In all cases, the system provided suspension with comfort. In an early version the system was quite prone to leakage, but this problem has been eliminated.

One amputee continues to wear his prototype inflatable wedge, preferring it over his previous prostheses. Another subject is having his inflatable wedge modified into a peg-leg for beach and swimming wear. Other fittings were conducted only in the adjustable state due to the lack of available suspension units. Suitable results were obtained to warrant further investigation of this technique, using production parts.

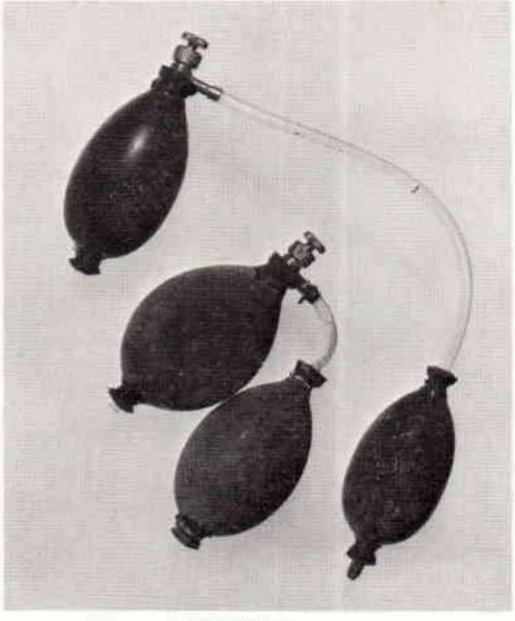

FIGURE 2

[1] Director of Prosthetics, University of Illinois at the Medical Center, Chicago, Illinois. Formerly Senior Instructor, UCLA Prosthetics-Orthotics Program.

Photography by Mary Louise Histon

The interesting possibilities offered by the inflatable wedge are the adjustability that allows tightening or loosening suspension characteristics that vary from day to day due to stump edema and varying stump sock thicknesses, virtual elimination of piston action, and comfortable supracondylar suspension, even with bony stumps.

Although minor design changes are planned for the inflatable wedge, most of the present technical aspects have been shown to be adequate. Originally, the inflatable wedge system used two blood pressure bulbs (Fig. 2), with a simple needle valve between them. The early prototypes, while crude, were comfortable and durable. However, they were not in the least cosmetic. The original bulb system has evolved to a suspension wedge shaped similar to the hard wedge variant. The models are hand fabricated from black neoprene (Fig. 3). At present, further development of the fabrication technique of the wedge is still being conducted.

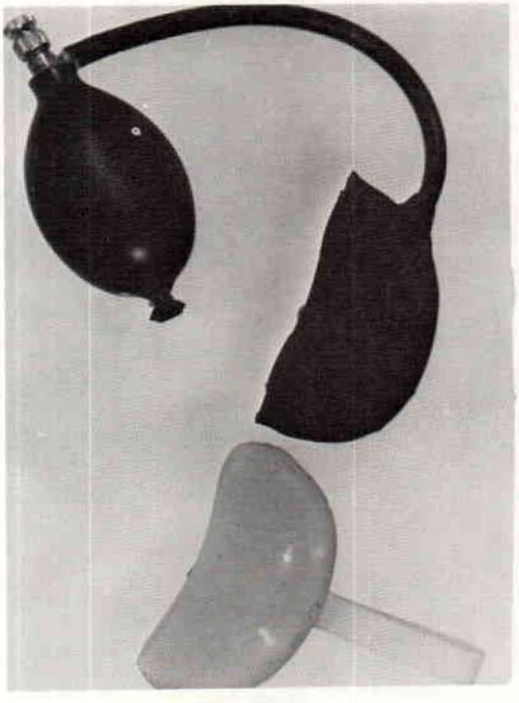

FIGURE 3

One problem of supracondylar-type suspensions has been a lack of inate adjustability. The inflatable wedge adjustability can be seen to range from 5mm (.2 in.) deflated, to 25mm (1 in.) inflated (Figs. 4 and 5). Figure 6 illustrates the location of the inflating bulb in the gastrocnemious bulge area. The wedge

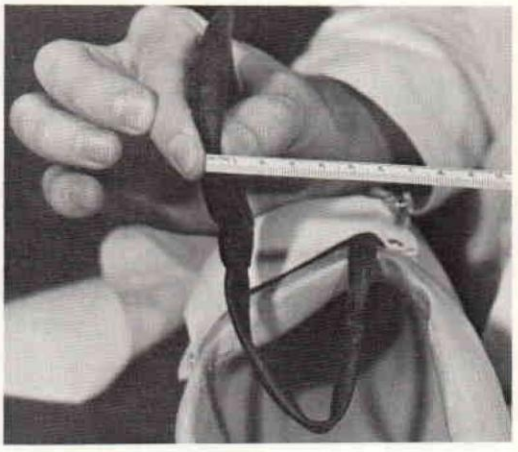

FIGURE 4

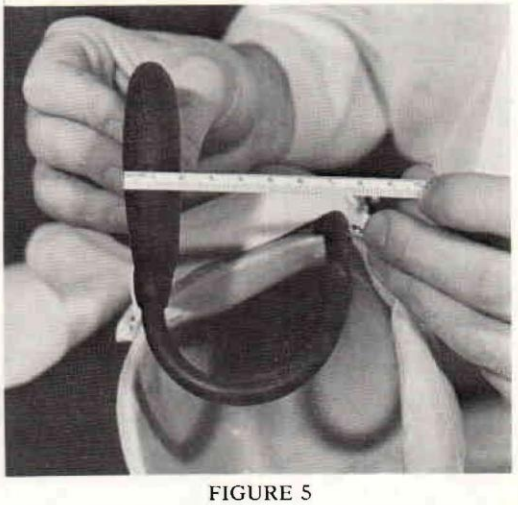

FIGURE 5

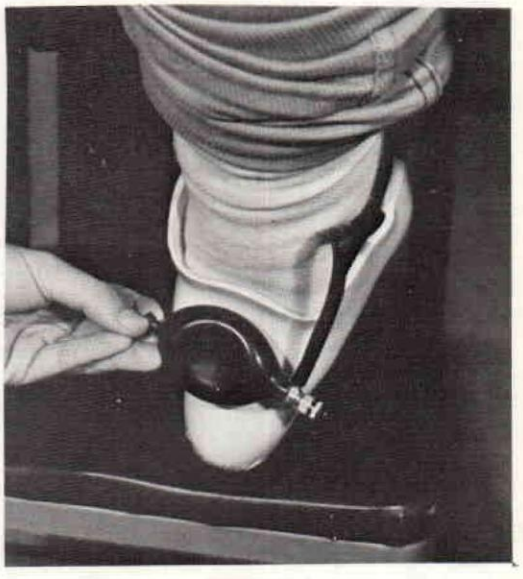

FIGURE 6

is inflated either by hand pressure, or as shown in Figure 7, by placing the inflating

FIGURE 7

bulb on the sound leg. The needle valve is opened to allow sufficient air for adequate suspension pressure and then is closed. The suspension portion of the system varies in thickness from 5mm (.2 in.) at the superior

FIGURE 8

edge where a rigid upper lip is necessary. The portion of the wedge next to the soft tissues of the stump about the knee is 2.5mm (.1 in.)

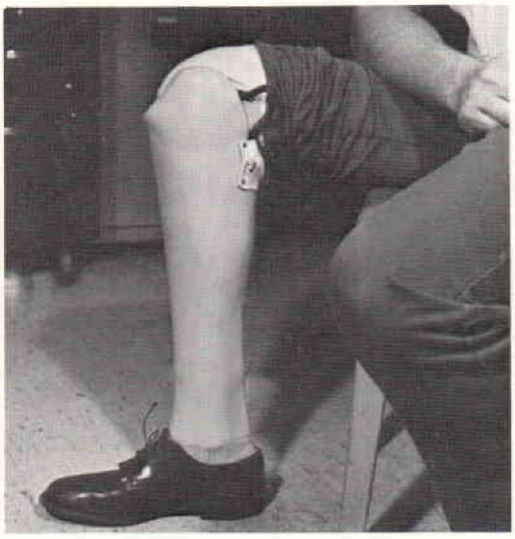

FIGURE 9

FIGURE 10

thick, making it thin, soft and pliable for comfort and conforming characteristics. The connecting hose is laminated to the suspension wedge. The wedge-and-bulb system is filled with permanent type anti-freeze to pro-

vide year round service in all climates. The medial ear lip, which retains the suspension bulb in place, is now approximately ¼ in. wide. In early models, the overhanging lip was quite wide to retain the awkward shape of the blood pressure bulb (Fig. 8). Both medial and lateral ears of the socket are heavily reinforced with wire screen and fiberglass. The added strength is necessary as the hydraulic force of the inflating wedge had the tendency to spread the ears of the socket.

Mr. Baird is presently in the development and manufacturing stages of the inflatable wedge. Production units are to be available in the near future. Production models will be in kit form, in Caucasian and dark brown hues.

It appears that the inflatable wedge may provide some of the answers to problems that many prosthetists have found in fitting other type of supracondylar suspensions for below-knee amputees. Further information concerning the inflatable wedge can be obtained by writing Mr. Lincoln Baird, c/o UCLA Prosthetics-Orthotics Program, 1000 Veterans Avenue, Los Angeles, Califor ia 90024.

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1. Fillauer, Carlton, *Supracondylar wedge suspension of the P.T.B. prosthesis,* Orthotics and Prosthetics, Vol. 22, No. 2, June 1968.

2. Wilson, A. Bennett, Jr., *Recent advances in below-knee prostheses,* Artificial Limbs, Vol. 13, N. 2, Autumn 1969.

Technical Notes

The Below-Elbow Inversion Harness

The below-elbow inversion harness (Fig. 1) is a modification of the conventional figure-eight harness, and is applicable in most cases where a conventional harness is prescribed. The triceps pad or cuff is eliminated, greater harness and socket stability are provided, and power transmission is more efficient. The inversion harness can be fabricated easier and in less time than the conventional. It is also more cosmetic and more durable. The various components of the inversion harness are shown in Figure 2.

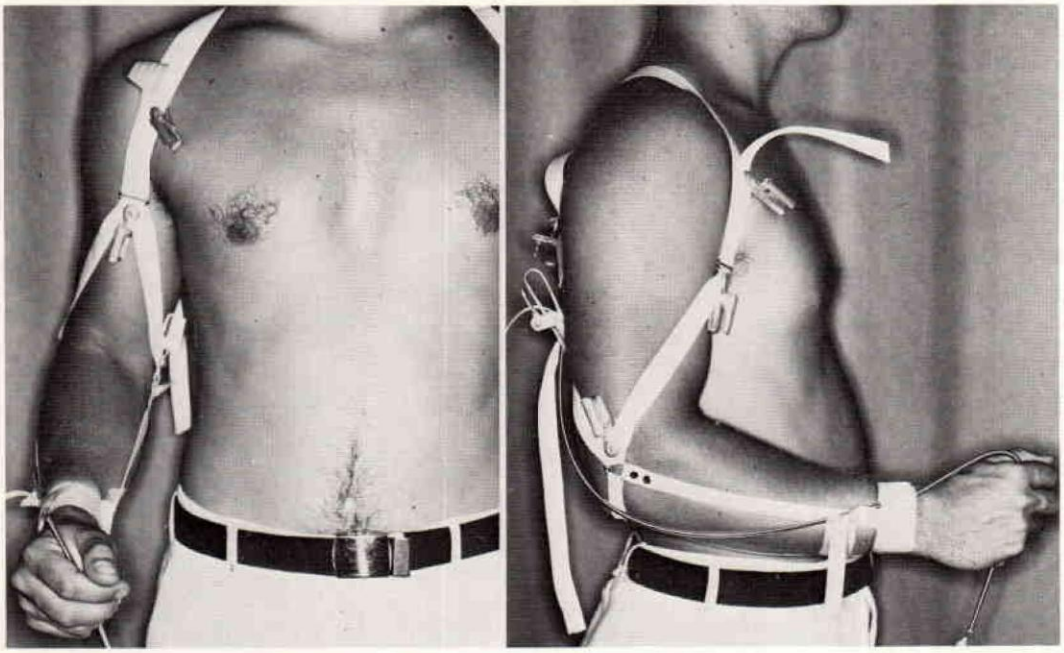


Fig. 1. The inversion harness laid up to demonstrate the principles involved.

The basic functional precepts of the below-elbow harness are not altered. Rather, the modification alters the means by which these precepts are realized.

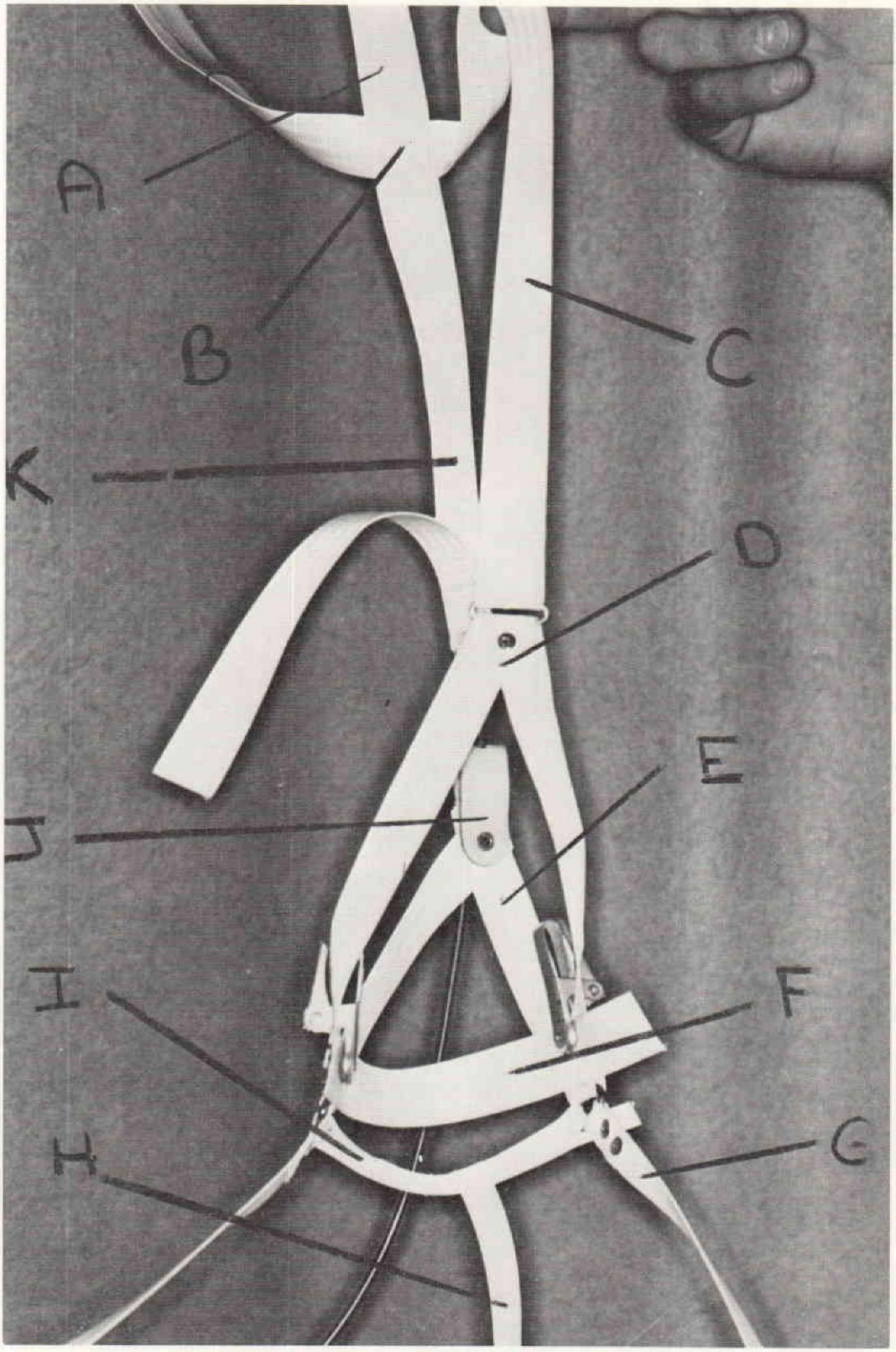
Before attempts are made to replace the triceps pad or cuff with a series of straps, the functional characteristic of the pad should be understood thoroughly. The function of the pad is basically threefold.

First: to stabilize the below-elbow socket against vertical loading as the elbow is extended by transmitting a counterforce from the anterior support strap to the flexible hinges.

Second: to stabilize the socket against a distally directed longitudinal force during elbow flexion by transmitting a counterforce from

Fig. 2. Parts of the Inversion Harness:

A. Axillary Loop
B. Harness cross
C. Anterior Support Strap
D. Inverted "Y"
E. Inversion Strap
F. Supra-olecranon strap
G. Flexible hinges
H. Anchor Strap (optional)
I. Ulnar Strap
J. Cross Bar Leather Loop
K. Cable Control Strap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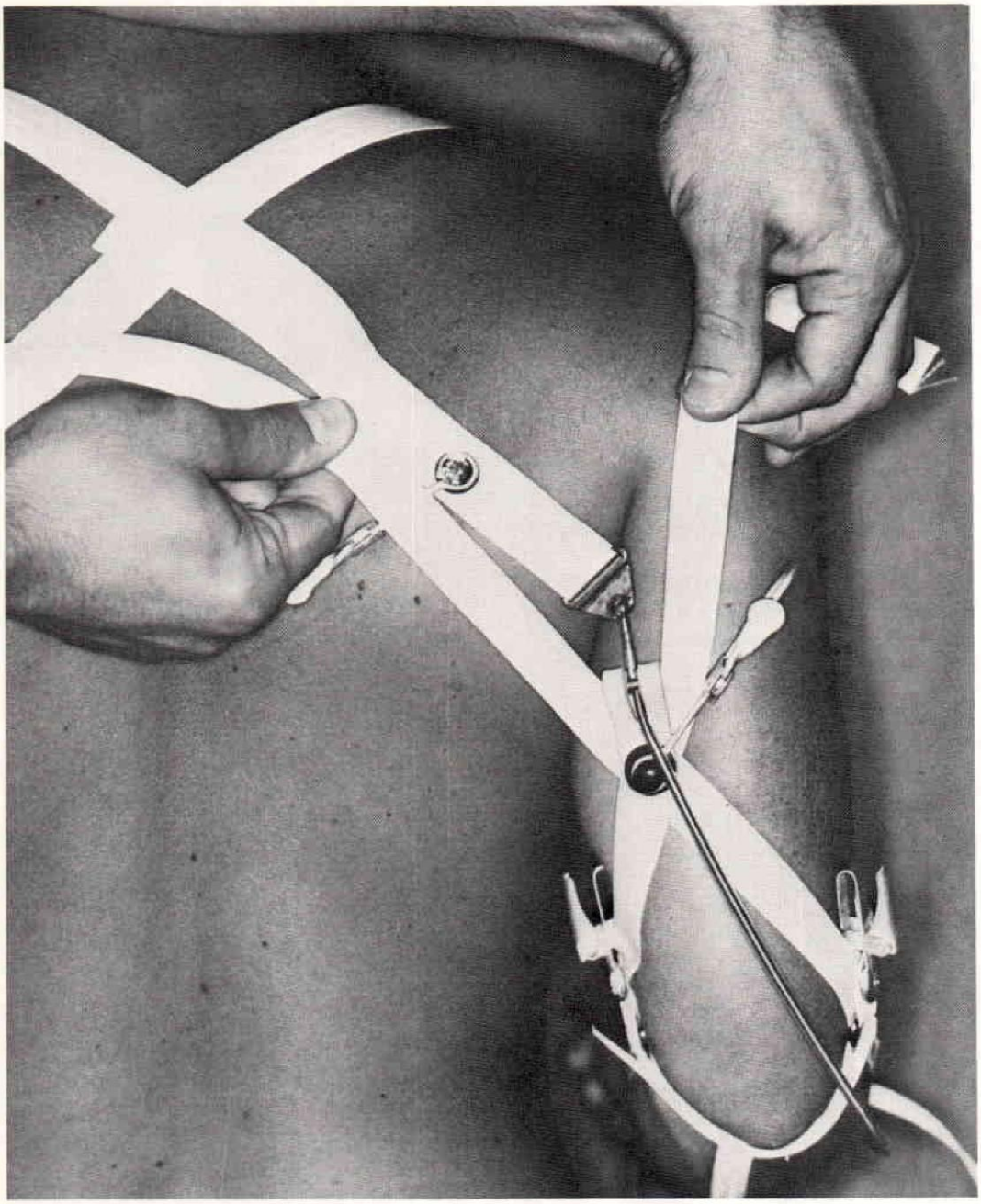

Fig. 3 Assembly of the posterior of the inversion harness.

the upper arm to the flexible hinges.

Third: to serve as a cable control reaction point by providing a location for the attachment of the crossbar assembly.

These three functions of the pad are fully realized with the application of the below-elbow inversion harness. Vertical loading is counteracted more efficiently by the linear continuum of the inverted "Y" strap and flexible hinges (D and G, Fig. 2), via the "D" rings. During elbow flexion, longitudinal forces are counteracted by the supraolecranon strap (F, Fig. 2). A base for attachment of the cable reaction point is provided by the intersection of the medial and lateral members of the inverted "Y" strap on the

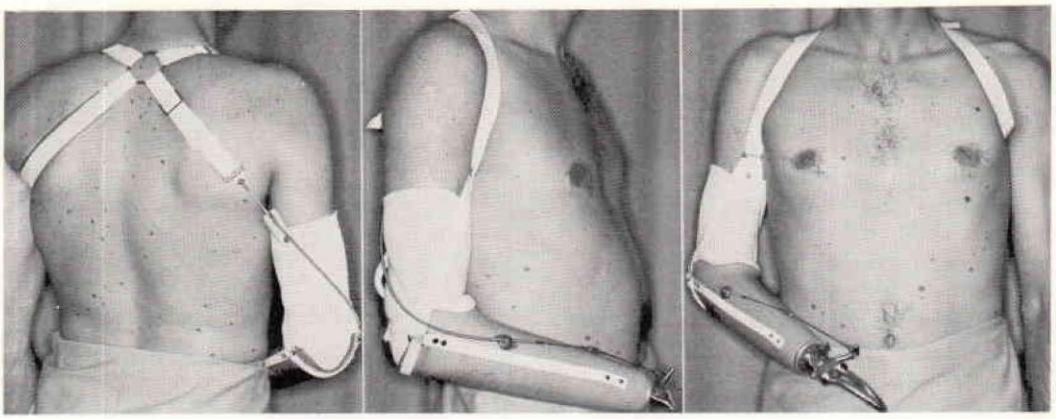

Fig. 4. Three views of the inversion harness.

posterior surface of the upper arm. Because all forces are transmitted through a balanced system, there is no buckling, chafing, or shifting of any of the harness components throughout the entire range of shoulder, elbow, and forearm motion.

FABRICATION

The first step is the preparation of the axillary loop, anterior support strap, and cable attachment strap in the conventional manner. Either the Northwestern ring or harness cross can be used. Insert a 4 ft. length of ¾ in. dacron webbing through a four-bar buckle to the midway point. Permanently secure this strap just below the buckle with a divergent angle of about 30 degrees. The two diverging members will serve as the inverted "Y" strap (D, Fig. 2). Insert the two members through the 5/8 in. "D" ring to 8 in. of the four-bar buckle. Twist and invert these two members, and secure with a harness clamp just superior to the "D" rings (Fig. 2).

Permanently attach ½ in. dacron flexible hinges to the "D" rings with rivets (Fig. 2). The hinges should be about 14 in. long to facilitate attachment to the socket as distally as possible. The ulnar strap should be inserted through the "D" rings posterior to the flexible hinges, doubled over on itself, and clamped with a harness clamp. Finally, the supraolecranon strap (F, Fig. 2) should be cut about 6 in. long. In the final stages of fitting, the olecranon strap will be attached to the inverted strap immediately superior to the "D" rings.

FITTING

At this time, the harness, as well as the entire prosthesis (i.e., cable system, etc.) is ready for fitting. The countralateral arm should be extended throughout the axillary loop. The anterior support strap should be attached for the four-bar buckle immediately superior to the belly of the biceps (Fig. 3). Flex the elbow to 90 degrees. Secure the flexible hinges with tape as far distally on the prosthesis as possible and adjust the ulnar strap so that the "D" rings coincide exactly with the humeral epicondyles (Fig. 3). From the back of the patient, draw the inversion straps through the "D" rings so that they will resist an equal amount of tension. Intersect these two straps at a point about 1½ in. below the middle of the arm. Secure the intersection as well as the cross bar leather loop with a harness clamp (Fig. 3). Secure the supracondylar strap to the inverted strap immediately above the "D" rings. Make sure that this strap provides the tension necessary to maintain properly the coincidence of the "D" rings and the humeral epicondyles.

The harness is now ready for final fitting. Attach all components of the cable system in the conventional manner (Fig. 4). Adjust all straps attached to the "D" rings in such a manner that the tension is equal, and there is no buckling.

Before removing the harness clamps for sewing, staple all strap unions, to insure that the harness is not altered during sewing. If the ulnar strap rides up and around the olecranon process, an anchor strap (H, Fig. 2) may be sewed to it and attached on the ulnar, distal side of the prosthesis.

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An AK Prothesis with Tilt Socket

The patient was a white, 22-year-old male, 6 ft. 4 in. tall who sustained severe injuries in a motorcycle accident which eventually resulted in a high above-knee amputation of the right lower limb. Sight in the right eye was lost, but the left leg was not involved.

The examination revealed a 4½ in. amputation stump, with a femur 3½ in. long. The circumference at the level of the ischium is 25 in., at the 2 in. level, 21 in. Extensive and deeply seated scars covered the anterior and lateral aspects of the stump (Fig. 1, upper).

The stump motion was limited. Max. extension was 26 deg. (Fig. 1, center); Max. flexion 49 deg. (Fig. 1, lower) for a total range of motion of 23 deg.

In consultation with the patient's physician, it was decided to fit him with an above-knee quadrilateral, wood socket, (conventional type, since the extensive scar tissue prohibited the use of a suction socket) in which the anterior and lateral walls were to be 3½ in. above the ischial seat to provide stability. The socket was to be fitted in approximately 5 deg. of abduction in order to gain space at the medial aspect. A Silesian belt was to be provided for suspension. The Coleman "Safety Knee" and the Bock 1H20 foot were prescribed.

Because of the extremely limited range of motion present, additional provisions had to be made to allow the patient to sit. Therefore a joint at the distal aspect of the stump, about 4 in. from the ischial seat, and a locking mechanism to assure stability while standing and walking were incorporated.

The necessary components consisted of two upper joint sections with screws, two lower joints with extension and flexion stops, and the lock mechanism (Fig. 2), and were obtained from Habermann[1]. The fitting was done in conformance with established methods, using the Staros-Gardner coupling (Fig. 3). After static and dynamic alignment were considered to be satisfactory, the assembly was transferred into a definitive prosthesis in the usual fashion.

The swivel points were then established at the midline of the socket, medially and laterally, perpendicular to the line of progression and parallel to the knee axis at the level of the end of the stump. (If more than 50 deg. tilt is desired these points should be moved

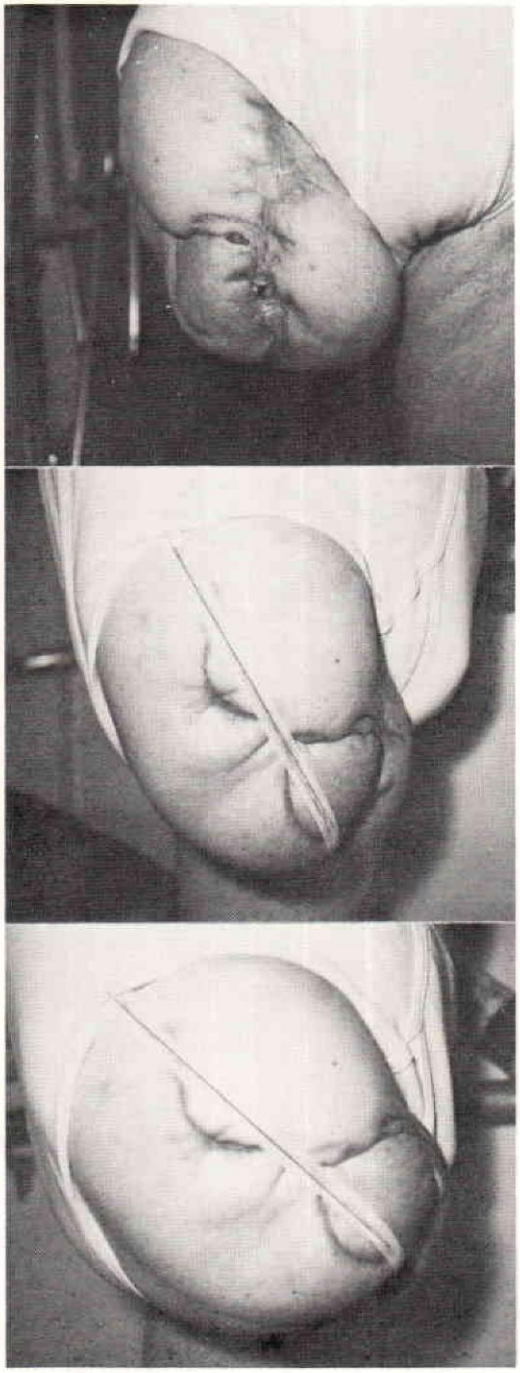

Fig. 1. Three views of the patient fitted with the tilting socket.

[1] Habermann, Hellmut, Orthopaediewerk Habermann GmbH, Chief Orthotist and Prosthetist, Marienburgstrasse 5-7, Frankfurt am Main, Germany.

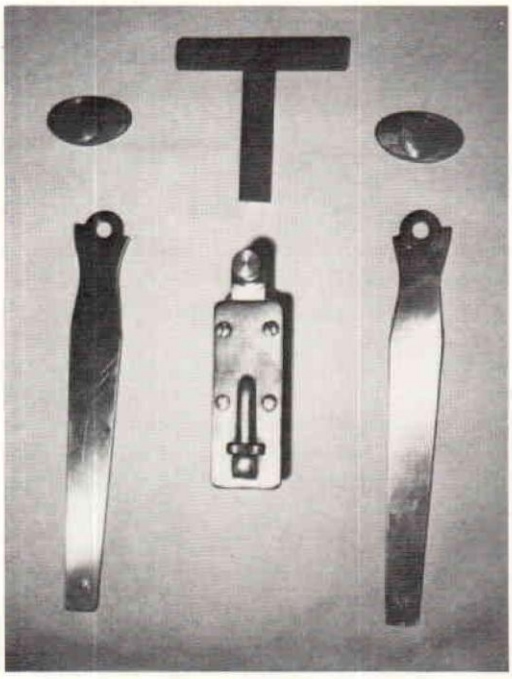

Fig. 2. Components required for the tilting socket. These were obtained from Habermann of Frankfurt, Germany.

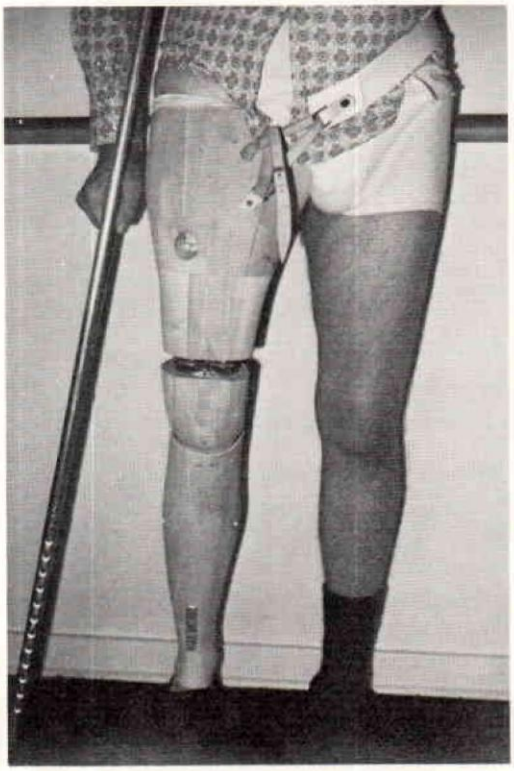

Fig. 3. Leg "in-the-rough" with the "Staros-Gardner" coupling.

slightly anterior of the midline.) After these points were marked they were drilled on the lathe between centers with a 3/8-in. drill, and with a 1¼ in. cylindrical counterbore; two flat surfaces were obtained which were parallel to each other as well as perpendicular to the established axis.

The upper joint sections were then installed, followed by installation of the lower joints. Extreme care must be taken to preserve the alignment already established.

After the joints were fastened with wood screws a cut was made through the thigh portion, approximately 1½ in distal of the joint center. A second cut 1 in. distal of the first one on the anterior half is necessary so that this wood may be removed to provide for clearance (Fig. 4). In addition to this more wood may have to be removed until the

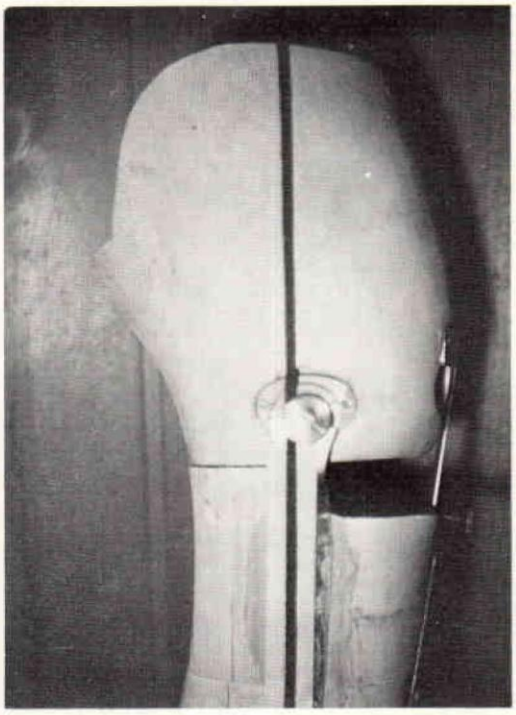

Fig. 4. Lateral view of thigh section showing the saw cuts that are necessary.

desired degree of tilt is obtained. After this the lock is installed, the distal part first, followed by the installation of the proximal "T" section (Fig. 5). To release the lock, the patient merely depresses the knob. Upon extension it locks automatically.

At this time the patient was asked to return for a second fitting, mainly, to assure that the amount of tilt was sufficient. It became

obvious that the Silesian belt was inadequate, and therefore a hip joint and a pelvic belt

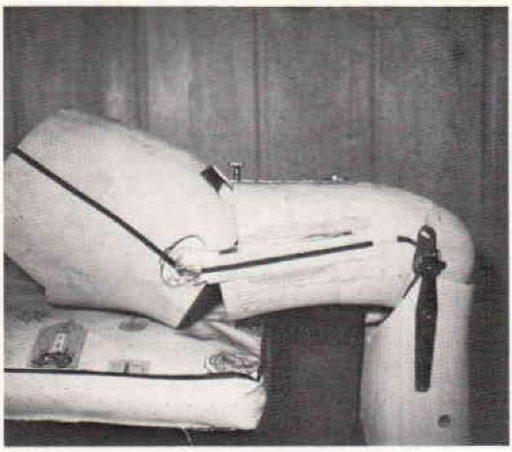

Fig. 5. The complete assembly. Note the locking arrangement.

were added. In addition to this a light webbing strap over the left shoulder was installed to give additional suspension.

The prosthesis was then finished. Upper and lower joints were fastened with N. 10 copper rivets, and the shin and thigh sections were laminated. After lamination the lock was installed. Then the upper "T" section of spring steel was aligned and riveted to the socket. The lower part has a height adjustment to assure a positive lock without any play. The finished prosthesis is shown in Figure 6.

This procedure was obviously time-consuming and, consequently, costly. However, the results were excellent. At this time the patient has worn the prosthesis routinely for 10 months, with only minor adjustments being necessary. I would like to add that this method is used quite often in Germany, not only for patients with a limited range of motion, but also for very short above-knee stumps to be fitted with suction sockets.

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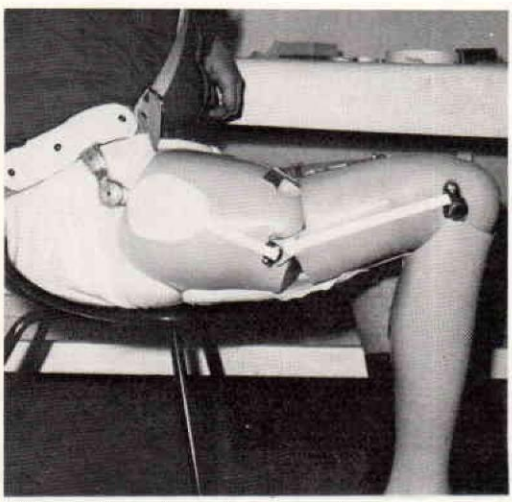
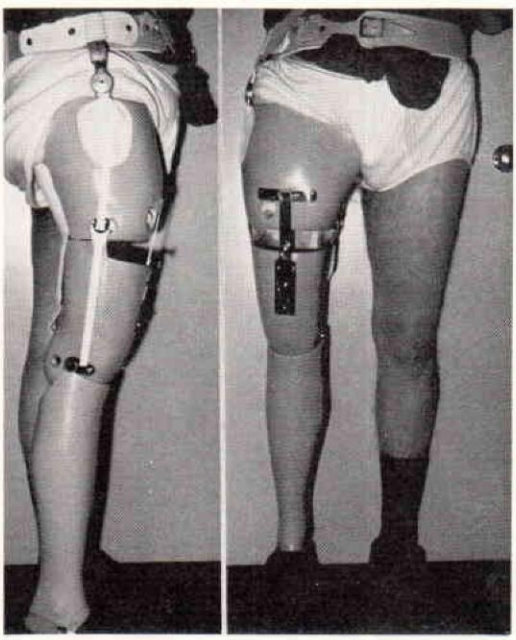


Fig. 6. Three views of patient with the tilting socket.



Proposed Method of Alignment for Knee Disarticulation Prosthesis

Among the disadvantages and problems of the knee disarticulation prosthesis cited by critics is that there is no adequate method to achieve proper alignment. In this brief article it is our intention to propose a method by which alignment of the knee disarticulation prosthesis can be accomplished by means of standard equipment.

We, at the University of Virginia, have not had the opportunity to employ this procedure, and therefore cannot comment on its efficacy. It is our intention only to suggest a possible means of alignment for the consideration and trial by other prosthetists.

We feel confident that if a number of prosthetic facilities throughout the country were to explore independently the possibilities of this procedure and compare experiences through this *Journal* and other means, a feasible technique can be developed.

This paper can, then, be said to propose not only a technical procedure, but also an informal way to accelerate development and evaluation.

Some may say that this is better left to centers and agencies set up for this purpose. While we support this viewpoint in cases of more complex and expensive research, we feel that the nature of this suggested procedure lends itself to a less formal approach and immediate implementation and trial. Moreover, as professionals it is our ever present duty to experiment seriously with new and innovative means of accomplishing our common goal.

THE PROPOSED TECHNIQUE

The proposed technique features the use of the standard above-knee adjustable leg and vertical fabrication and transfer machine, both available from Hosmer . We have been using the vertical fabrication and transfer machine routinely for over a year and have found it superior to the older horizontal transfer device in ease, speed, and accuracy of operation. It specifically allows for the use of foam as a structural material in the fabrication of prostheses. While the suggested procedure is no doubt possible in the horizontal machine, we do not consider it practical.

The individual steps involved are:

1. Using as thin a section of wood as possible, set the socket up on the above-knee adjustable leg in bench alignment.
2. Ignoring the obvious knee center discrepancy for the moment, carry out static and dynamic alignment. At the conclusion of the session correct knee center height is determined by measurement and observation to provide the best possible cosmetic result.
3. The prosthesis is transferred into the vertical fabrication and alignment machine in accordance with standard procedure.
4. Medio-lateral width of the socket at the proposed knee-center height is determined. The lower section of a pair of knee disarticulation joints are set into a block of wood as usual at this previously determined medio-lateral width.
5. The stand is then adjusted to obtain proper knee center height.
6. The proximal shank section is set into the knee center holding screws and is adjusted so that the joints lie on either side of the socket. The proximal shank section is joined to the ankle piece by foam or wood.
7. The upper joint sections are cut to length, contoured, and then assembled in lower joint sections. The upper joint sections are set on to the socket with Solka flox and polyester resin.

At this point transfer is complete and the prosthesis can either be finished or tried on the patient.

SUMMARY

A method is suggested for aligning and transferring the alignment to the knee disarticulation prosthesis using currently available equipment. It is proposed that this technique be developed and evaluated informally by prosthetists in the field.

VIRGIL FAULKNER, C.P.O.
CHARLES PRITHMAN, C.P.

New Publications

TIBIAL AND FEMORAL FRACTURES—BRACING MANAGEMENT, by Augusto Sarmiento and William F. Sinclair, University of Miami School of Medicine, Department of Orthopaedics and Rehabilitation, Miami, Florida 33152; 211 pages.

This publication is a manual for the application of orthotic principles to management of certain types of fractures of the long bones of the lower limb. It is the result of work sponsored in part by the Social and Rehabilitation Service of the Department of Health, Education, and Welfare. A critical review of this most timely text will appear in an early issue of *Orthotics and Prosthetics.*

FUNCTIONAL NEUROMUSCULAR STIMULATION—Report of a Workshop, April 27-28, 1972, Committee on Prosthetics Research and Development, National Academy of Sciences—National Research Council, Washington, D.C.; 161 pages; gratis.

As the title implies, this is a report of a workshop on the research activities devoted to electrical stimulation of muscle for useful purposes. Included are 20 technical papers by experts from the United States and Yugoslavia ranging from the simple problem of direct stimulation of the gastrocnemius to assist the patient with "drop-foot" to the use of electrical stimulation to alleviate spasticity. This report should prove to be useful to all who want to keen informed of the latest developments in management of patients with neuromusculoskeletal deficiencies.

AIDS FOR CHILDREN—Technical Aids for Physically Handicapped Children. ICTA Information Centre, FACK, S-161 03 Bromma 3, Sweden; 87 pages; $2.00.

This publication is an atlas, or catalog, that provides detailed information on various technical aids, equipment, and techniques that have been found to be useful in the treatment of children with physical handicaps. As one might imagine, many of the ideas are applicable to adults with similar problems. This most useful document was compiled by ICTA (International Committee on Technical Aids) Information Centre in collaboration with the International Cerebral Palsy Society with financial support from the government of Sweden.

THE CHILD WITH AN ACQUIRED AMPUTATION, Committee on Prosthetics Research and Development, National Academy of Sciences-National Research Council, Washington, D.C.; 162 pages; $5.50.

This soft cover book, edited by George T. Aitken, is a collection of papers given at a symposium sponsored by the Subcommittee on Child Prosthetics Problems of the CPRD and held in Toronto, Canada, June 9-11, 1970. It is a continuation of a series begun several years ago by the National Academy of Sciences, and it contains a great deal of information useful in the management of child amputees. A critical review of this publication is planned for an early issue of *Orthotics and Prosthetics.*

The table of contents is as follows:

BULLETIN OF PROSTHETICS RESEARCH, SPRING 1972 (BPR 10-17), 290 pages, published by the Prosthetic and Sensory Aids Service, U.S. Veterans Administration, and available for $1.25 from the Superintendent of Documents,

U.S. Government Printing Office, Washington, D.C. 20402.

Following past custom, the Spring 1972 issue of the *Bulletin of Prosthetics Research* consists of individual papers concerned with results of research coupled with reports of the status of research projects in limb prosthetics, orthotics, and aids for the blind and hard-of-hearing.

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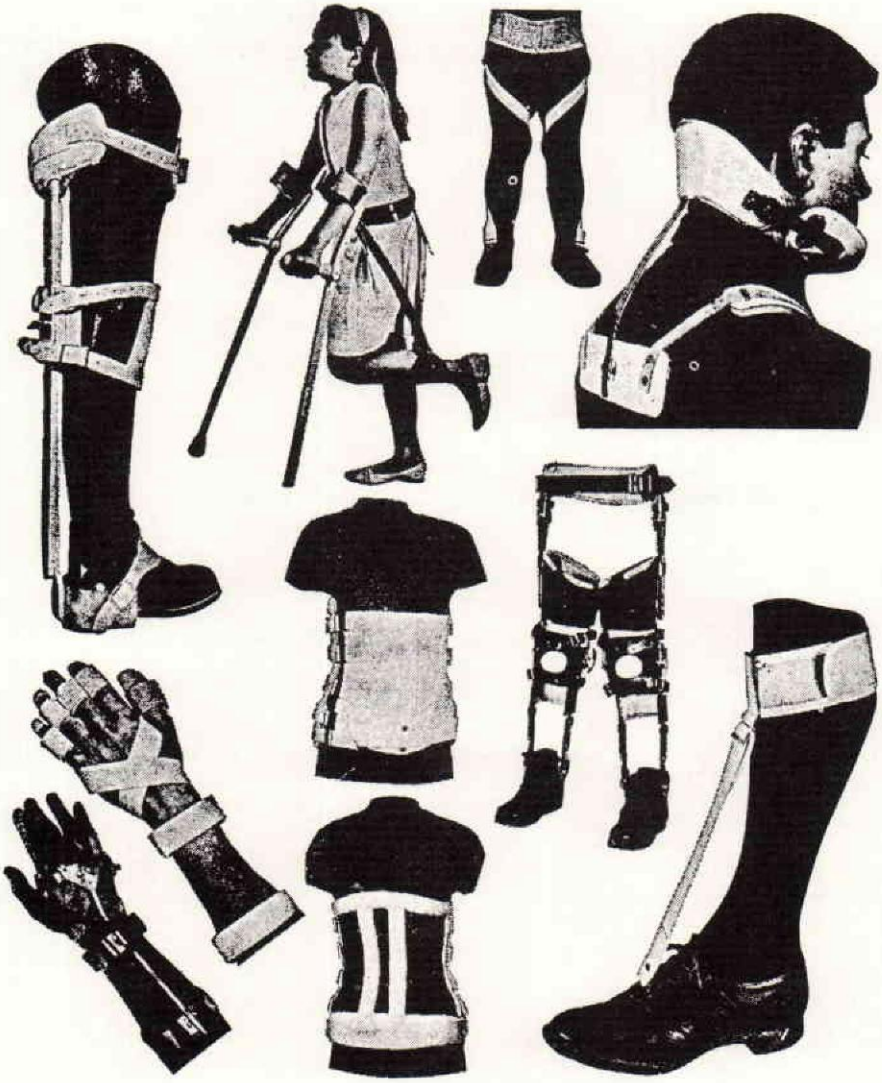

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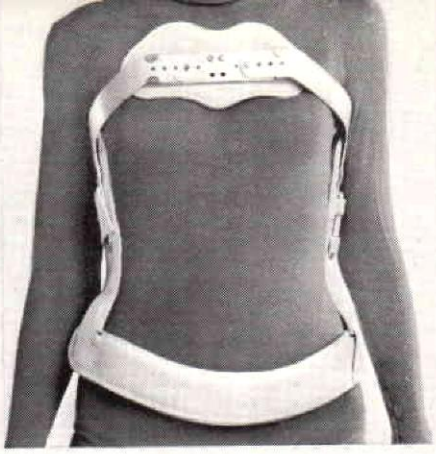

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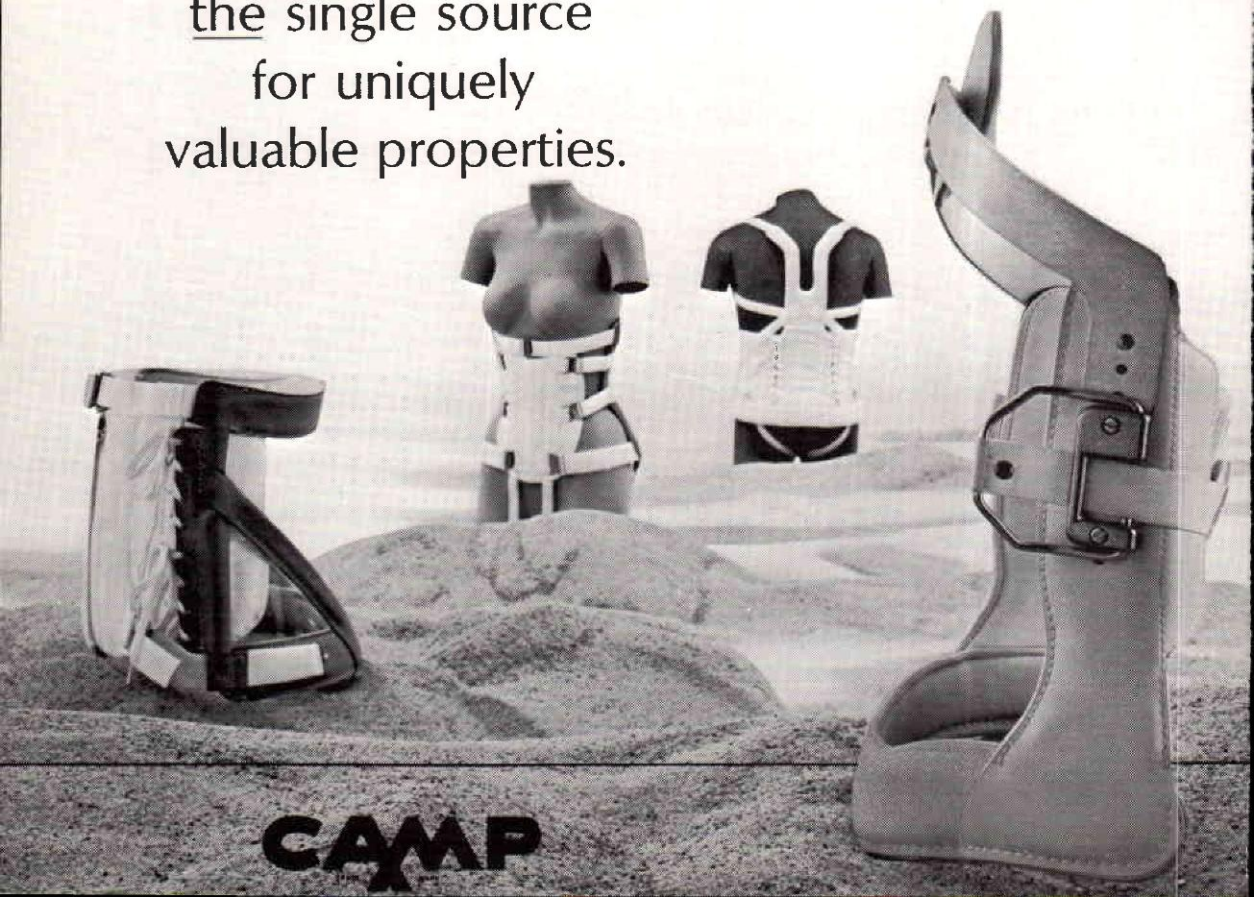

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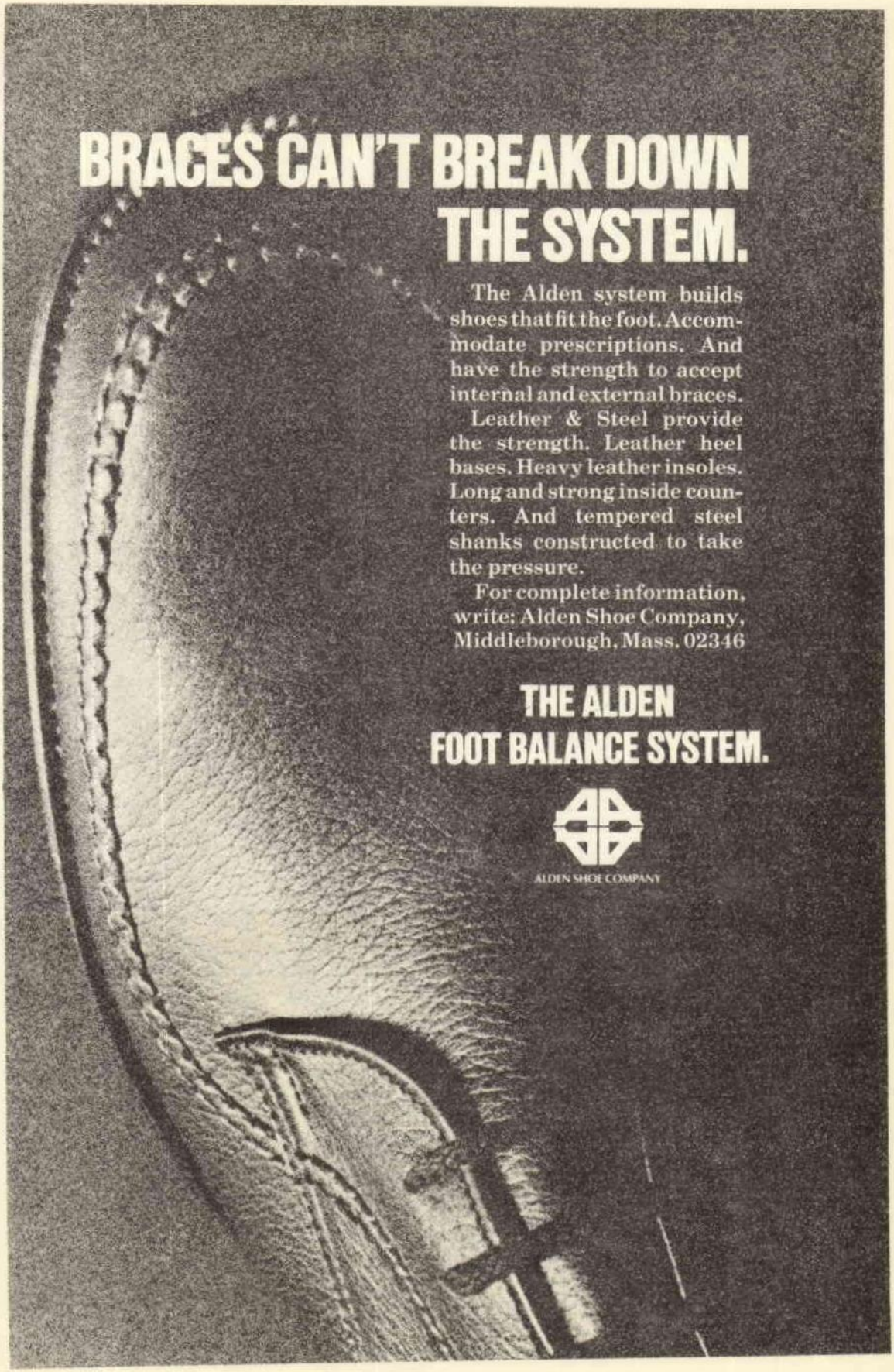
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