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Contents

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Audrey J. Calomino

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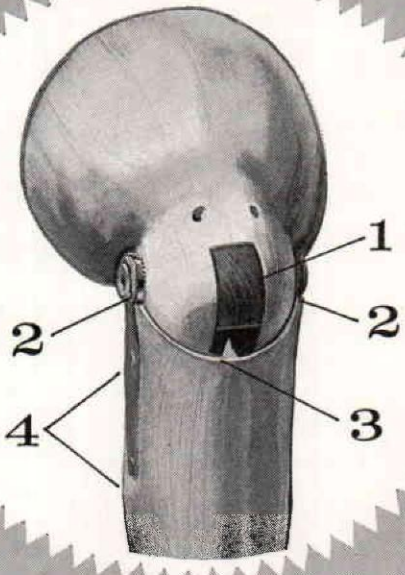

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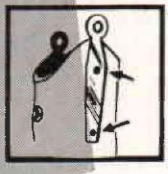

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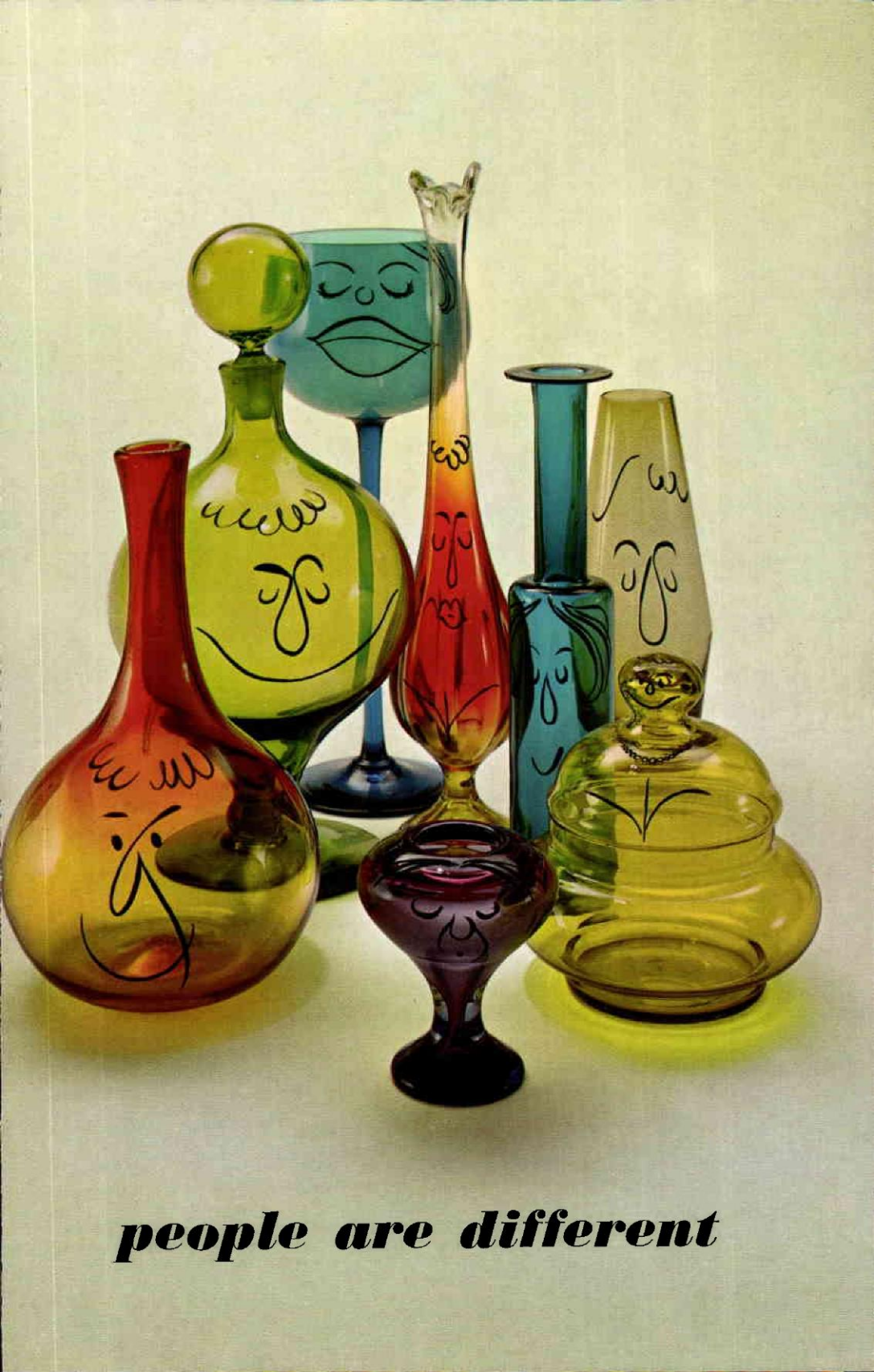

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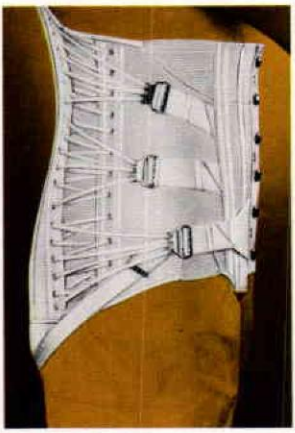

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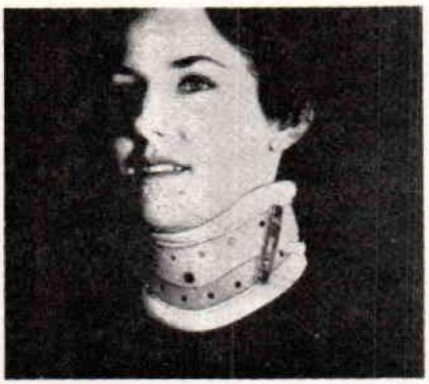

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A Fiberglas-Epoxy Drop-Foot Brace

by

James T. Hill and Allard L. Fenwick [1]

INTRODUCTION

It is generally believed that an appliance for the correction of drop-foot deformities should substitute for specific weakness only, and leave all other muscles free to act normally (1). A flexible appliance is preferred to a fixed or rigid device to allow as much freedom of motion of the forefoot and ankle as possible to enhance comfort, endurance and gait.

In this paper, the design and fabrication of a flexible plastic drop-foot brace which meet these criteria are described, and patient reactions are presented.

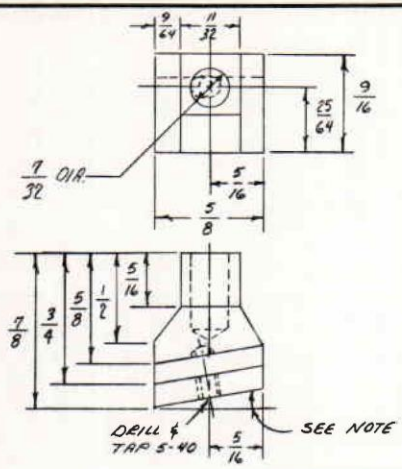
BRACE DESIGN

In the plastics drop-foot brace design, fiberglas-epoxy composites in rod form are used to duplicate the function of the metallic uprights and ankle joint in the conventional brace, thereby eliminating the need for a spring-loaded joint. By varying the diameter of these plastic rods, the degree of medial-lateral stability required may be accomplished with the accompanying change of force required for dorsiplantar flexion.

Toe lift is controlled by interchangeable shoe adaptors available with precut angles as required. The adaptors are coupled either to a shoe plate or a bar that fits into a conventional shoe channel.

The rods are connected at their proximal ends to a leather calf-cuff by insertion into sewn pockets. When maximum medial-lateral stability is required, a molded fiberglas-epoxy calf band is attached to the rods and covered

[1] U. S. Army Medical Biomechanical Research Laboratory Walter Reed Army Medical Center, Washington, D.C. 20012.



NOTE:
ONE PAIR 10° ± ½°
ONE PAIR 5° ± ½°
10° PAIR DRAWN

PART NO.	PART NAME	NO. REQ.	MATERIAL	REMARKS
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APPROVED JTH DATE 3-12-68	BRACE SHOE ADAPTOR		FIG. # 1

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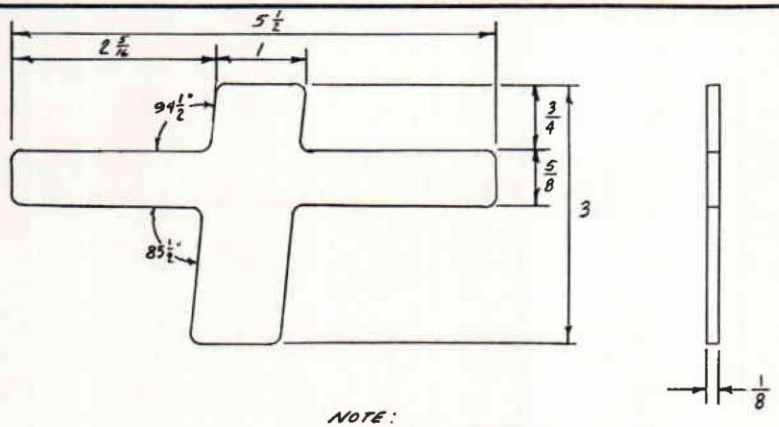

NOTE:
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PART NO.	PART NAME	NO. REQ.	MATERIAL	REMARKS
	SHOE PLATE	ONE	AL.	

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APPROVED JTH DATE 3-12-68	SHOE PLATE FOR PLASTIC BRACE		FIG. # 2

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with a leather cuff. Figs. 1–4 illustrate the brace and shoe attachment.

DESIGN MATERIALS

Since it is desirable to fabricate the strongest, lightest, and most durable brace possible, strength-to-weight ratio and flexural fatigue strength are the most important materials and design considerations.

Fiberglas-epoxy composites offer great flexibility of design due to their extremely high strength-to-weight ratio. The specific compressive strength of these composites is 2.8 times that of steel and 3.7 times that of aluminum (2). From a practical standpoint, this allows the brace designer a wide latitude for application. In a case where strength is of prime importance, a brace could be constructed almost three times the strength of steel with the same weight. At the other extreme, when reduction of weight is critical, a brace can be constructed as strong as aluminum but weighing only one-fourth as much. These comparisons show that braces may be designed both stronger and lighter than either steel or aluminum.

FIGURE 3—Lateral view of plastics brace at heel contact.

The superior fatigue performance of fiberglas-epoxy composites can best be illustrated by a comparison with the fatigue and flexural properties of steel and aluminum.

The fatigue strength of fiberglas-epoxy composites (unidirectional filaments) is 36,000 psi at 10 million cycles, while spring steel and aluminum are 33,000 psi

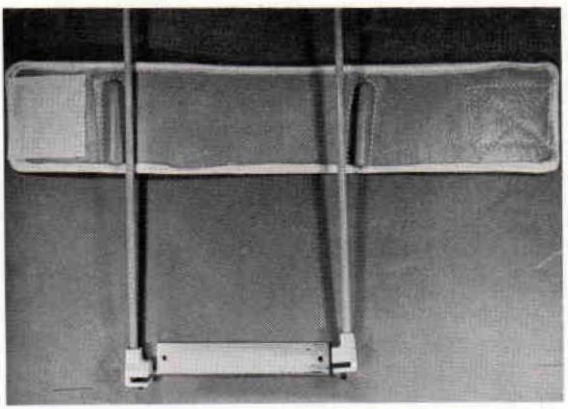


FIGURE 4—Individual plastics brace components. Top to bottom: 1) Calf cuff; 2) Fiberglass rods with adaptors; 3) Channel bar.

and 27,000 psi, respectively, at 10 million cycles. The specific fatigue of the fiberglas-epoxy composite is almost five times that of steel and over twice that of aluminum (3).

Notch sensitivity can be very important in fatigue applications, particularly with springs subjected to scratches and nicks which occur in brace application. These lead to premature fatigue failures. Reinforced plastics are far less notch sensitive than most metals. Aluminum retains only 37% of its original fatigue strength, steel 43–47% of its original strength, while fiberglas-epoxy composites retain 80–90% of their original fatigue strength.

The properties of fiberglas-epoxy composites compared to those of aluminum and steel make them an obvious choice for brace fabrication.

BRACE FABRICATION

Fiberglas-Epoxy Rods

The fiberglas-epoxy rods are fabricated from Scotchply* Reinforced Plastic, Type 1008. This is an epoxy-fiberglas preimpregnated material that comes in rolls up to 48" wide.

The rods are fabricated as follows:

Cut strips of prepreg tape 1 to 2 inches longer than desired length of rod. (19-inch rods would be long enough to fit 97.5% of the adult male population) (5). The number of strips of tape required is dependent on the desired diameter of the rods. Below are tabulated the strips of tape required per rod for a given diameter.

Desired Diameter	*Prepreg Required*
5/32 inch	1 strip—1½" width
3/16 inch	1 strip—1½" width plus 1 strip—¾" width
¼ inch	2 strips—1½" width plus 1 strip—¾" width

Take a strip of 1½" tape, remove the release paper and lay on a flat surface. Roll the tape longitudinally as tightly as possible (there should be no visible hollow core in the center) until a round rod is obtained. If necessary, roll additional pieces of tape around this rod until the desired diameter is obtained.

To insure maximum strength of the attachment of the rod to the shoe adaptor, the diameter of the rod is increased over the last 4½" to a maximum of ¼". This is accomplished by wrapping triangular pieces of prepreg around the ends of the rods.

For a 5/32" rod, cut a triangle 4½" long × 2¼" high.

For a 3/16" rod, cut a triangle 4½" long × 1½" high.

Remove the release paper and wrap the triangular prepreg around the end of the rod in such a manner that the 4½" side runs along the rod when you start to roll this piece around the rod. Roll the piece around the rod. Next, spirally wrap the rod with 1" wide release tape applying slight,

* Minnesota Mining & Manufacturing Co., St. Paul, Minn.

but uniform, tension during the winding. Tedlar,* polyvinyl alcohol and Scotchply† XP-242 release films have been successfully used.

After the rod has been wound, seal both ends to prevent leakage of the epoxy. Transparent adhesive-backed tape is excellent for this purpose. Lay the rods on a flat surface and cure for 2 hours at 160° ± 10° C. (320° ± 18° F).

Calf-Cuff

For those patients who don't require a rigid calf band, only a leather cuff is necessary. A 3" wide cuff should be adequate. It is made in the usual manner utilizing either straps or Velcro fasteners. Leather pockets are sewn on the medial and lateral aspects of the cuff to accommodate the rods.

Rigid Calf-Band

In those cases where a rigid calf band is required, the following procedure should be followed:

1. Determine the diameter of the patient's leg 1½" below the head of the fibula.

2. Using this diameter, draw a half circle on a 1⅝" to 2" thick piece of wood.

3. Draw a line parallel and equal in length to the diameter ¾" away from the diameter on the outside of the half circle. Connect the ends of this new line with ends of the diameter. Cut out the entire enclosed area. This will provide a form for the calf band.

4. Tape a strip of 1/16" thick, 40-70 durometer silicone sheeting over the perimeter of this form. Cover this sheeting with release film.

5. Stand the form on its flat edge. Cut a strip 1½" wide of prepreg tape to fit around the exposed perimeter of the form. Remove the release paper and lay onto the form. Next, cut pieces of 1½" prepreg 1½" long and lay the fibers 90° (perpendicular) to the fibers of first prepreg and fill in all around the band. Continue the buildup, alternating the fibers, until there are 9 layers.

6. Put a piece of release film over the final layer of prepreg tape.

7. Cover the release film with 1/32", 70 durometer silicone sheeting and place the entire layup into a polyvinyl alcohol vacuum bag.

8. Cure the layup for 2 hours at 160° ± 10°C (320° ± 18°F) under 25-30" vacuum.

9. Remove cured calf band from the form and sand the edges smooth.

Attachment of Rods to Rigid Calf Band

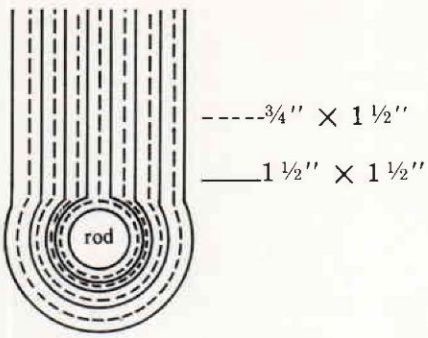
In order to attach the rods to the calf band, the ends of the rods must be first built up as described below:

1. From 1½" wide prepreg tape, cut 4 pieces 1½" long and 7 pieces ¾" long.

2. Lay up these prepreg tapes around rod as shown in the diagram. The fibers in the 1½" × 1½" pieces run in the same direction as the rod. The ¾" × 1½" fibers run 90° to the rod.

* E. I. duPont deNemours Co., Wilmington, Del. Film Dept.

† Minnesota Mining & Manufacturing Co., St. Paul, Minn.



3. After the prepreg tape is wrapped around both rods, lay the rods on a piece of wood that has been covered first with $^{1}/_{16}$" silicone sheeting, then a piece of release film. The rods should be placed so the buildups are facing in opposite directions. This will insure proper alignment of the rods on the calf band.

4. Cover the buildup with release film, then $^{1}/_{32}$", 70 durometer silicone sheeting, place in a vacuum bag and cure for 2 hours at 160° ± 10°C (320° ± 18°F) and 25–29" Hg vacuum. Sand the edges of the cured buildup.

5. Place the plastic calf band on the patient's leg and align the rods. Mark the location of the rods on the band.

6. Rivet or cement the rods to the calf band. If cement is used, the following formulation is very satisfactory:

Epon*815 . . .	4 parts
Curing Agent T-1	1 part
Filler	As required to make a paste

* Shell Chemical Co., New York, N.Y.

Shoe Adaptors and Shoe Plate

The design of the shoe adaptors is shown in Figure 1. Although this design is for a 5° or 10° lift, slots may be made for any angle of lift required. The shoe plate is shown in Figure 2.

Brace Assembly

The patient's shoe may have either a conventional flat channel attached, or a shoe plate similar to the one shown in Figure 2. The plate or channel may be attached with rivets or the cement formulation given above.

If the channel is used, cut a flat bar of aluminum to fit the channel and extend $^{7}/_{16}$" beyond the channel on each side. If the shoe plate is used, cut the extending medial and lateral pieces $^{7}/_{16}$" beyond the edge of the heel. Drill ⅛" holes ⅜" from the ends of the channel bar or shoe plate. Next, determine the proper rod length by actual measurement on the patient. Cut off ½" from the tapered end of the rod. Cement the tapered end into the shoe adaptor using Epon 815, 4 parts and curing agent T1, 1 part. Allow the cement to cure for 1 hour at room temperature plus 30 minutes at 50°–65°C (120°–150°F) or overnight at room temperature.

Attach the shoe adaptors to the shoe and have the patient put on the shoe. Determine the length of rods necessary (¾"–1" below head of fibula) and cut off the excess with a hacksaw.

If the flexible calf-cuff is used,

you need only insert the rods into the cuff, and the brace is ready for use. If the rigid band is being used, sew a leather cuff to slip over this plastic band.

EVALUATION

Plastic drop-foot braces have undergone an 8-month trial evaluation by two bilateral brace wearers. During this period, the following observations were noted (7, 8).

1. The plastic braces are lighter and more cosmetically attractive than metal braces.

2. There is little danger of the plastic braces interlocking and thereby creating a tripping hazard. The contouring of the metal brace away from the ankle presents this interlocking hazard with its potential of causing a serious fall.

3. Plastic braces are so light and flexible, one forgets he is wearing them. Metal braces cause fatigue when walking.

4. Plastic braces do not damage clothing, while metal braces cause damage to the trousers through tears and grease spots.

5. Plastic braces allow one to cross his legs comfortably, while metal braces do not.

6. Plastic braces require no upkeep, while the springs in the metal brace require occasional oiling and replacement.

7. Driving with plastic braces is almost as easy as it was without braces. Although metal braces are adequate for driving, they are heavy and clumsy to move from the accelerator to the brake pedal.

8. Plastic braces provide adequate medial-lateral stability yet are flexible enough to allow walking on uneven terrain.

9. Plastic braces are easier to put on than metal braces.

10. When one is inactive for an appreciable length of time, such as sitting at an office desk, the rigidity of metal braces prevents adequate flexing of ankle muscles causing severe aching in the calf and ankle regions. The flexibility of the plastic brace allows sufficient ankle motion to minimize this effect.

11. There is no noticeable numbness of feet since abandoning the rigid metal braces.

12. If it were a choice between metal braces and plastic braces, plastic would be preferred.

SUMMARY

A plastic drop-foot brace of simple design has been developed which provides toe lift and concomitantly insures comfortable and effective performance. To allow as much natural action of the foot and ankle as permissible and provide good gait and enhanced standing balance, it was found desirable to employ the dual spring action of flexible fiberglas rods to counterbalance dorsiflexion and plantar flexion (6).

The brace is easily removed, prevents clothing damage, and provides for an interchange of shoes. The plastic brace is lightweight, durable, cosmetically attractive, and corrosion resistant. Provisions have been made for

altering the degree of lift as necessary to compensate for progressive maladies.

Although the results discussed were obtained in a limited number of test cases, application of this appliance to a wide range of brace wearers appears warranted.

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The Unitized Below-Elbow Infant Prosthesis*

by

Carl Sumida, C.P. & O.[1], Julie Shaperman, M.A., O.T.R.[2], Wallace Sumida, B.S.E.E., C.P.[3], Yoshio Setoguchi, M.D.[4]

It has become accepted practice to fit a passive prosthesis to below-elbow infant amputees as soon as they have attained stable sitting balance. (1) In this way, the child's developmental experiences include the prosthesis from the beginning of his active life, for when an infant of this age is not sleeping or eating, he is actively exploring his own abilities in manipulating himself and his world. He rolls over, pushes up from the floor, leans on his arms while sitting, he crawls, climbs, and falls. He stabilizes and supports objects with his arms, clasps, pushes and holds a variety of shaped and textured objects. All of these locomotor and manipulative activities are enhanced or improved when the infant amputee has, and uses, a prosthesis. Early experience with a prosthesis has the additional advantage of preparing the child for spontaneous and skillful prehensile function when introduced to an active prosthesis, at approximately 24–30 months of age.

Arm motions of an infant tend to be gross due to the incompletely matured neuromuscular

This work supported by grant #00489 National Institute of Child Health and Human Development, National Institutes of Health, Department of Health Education and Welfare; grant #C-199 (R 2), Childrens Bureau, Department of Health Education and Welfare; and conducted at the Child Amputee Prosthetics Project, Departments of Pediatrics and Orthopaedic Surgery, University of California, Los Angeles, 1000 Veteran Avenue, Los Angeles, California 90024.

[1] Chief Research Prosthetist
[2] Research Occupational Therapist
[3] Research Design Prosthetist
[4] Medical Director

system. He flings his arms against objects; plays in sand, dirt, water, his food, and anything else available. His interests often lead him to lift loads which are very heavy in relation to his own body weight. The prosthesis which is actively included in the activities of an infant must stay on; must be easily cleaned; certainly must be durable; but above all must be valuable to him in pursuit of his interests and desires.

It has long been the practice at the UCLA Child Amputee Prosthetics Project to fit infant short and very short below-elbow amputees with a double wall polyester laminated socket and forearm which is pre-flexed fifteen degrees. The prostheses have a constant friction wrist unit, full polyethylene cuff attached to the socket with rivet insert hinges, a Dorrance 12 P or 10 P hook, and a figure-eight harness for suspension. In other words, the present standard procedure is to fit below-elbow infant amputees with a modified and miniaturized adult prosthesis.

Unfortunately the body proportions of an infant are not a miniaturization of the adult, and his activities and movement patterns are not the same. The adult may bend, lift, and reach, but never goes through the contortions of the infant. If the infant functions differently from the adult, and his body proportions are also different, his prosthetic requirements must also differ. It has long been recognized that miniaturization of adult devices is less than satisfactory, but recorded observation of activity patterns and requirements was necessary before the functional needs of the infant amputee could dictate a new design criteria. Data collected on a large number of infants with below-elbow limb deficiencies over a thirteen-year period have been analyzed in an effort to isolate the offending factors in the conventional components of the below-elbow prosthesis and to thereby determine a new design criteria.

These general observations together with specific functional criteria suggested a radically new approach in design. The standard prosthesis is constructed on an "exoskeletal" concept, in which the supporting framework is the exterior shell and the interior is hollow to the point where the terminal device is attached. In contrast, the unitized prosthesis which was designed is hollow only in the socket area which fits over the patient's stump, and the remainder is constructed on an "endoskeletal" concept in which the basic structure is internal and other materials fill the space around it to complete the arm.

The unitized prosthesis is constructed largely of plastics which have sufficient strength and durability to survive the approximately two years of infant wear. Plastics have additional advantages over conventional materials in that they are lightweight; have a soft appearance suggesting flesh turgor; require little maintenance as they are not subject to corrosion; and are less expensive. To fulfill the general design crite-

ria components were developed which, when combined, create a unitized arm (2). These components are not applicable individually to standard prostheses as they are inter-related to each other. Design criteria were established by considering each component separately.

TERMINAL DEVICE

Observed Needs and Problems

When the adult hook is reduced in size for an infant, some of its excellent features become distorted or reduced in effectiveness. The infant with his inexperienced eye-hand coordination cannot easily place objects precisely into the hook. Objects therefore often wobble precariously between the two unyielding "fingers" opposing each other in a scissor-type motion. This skill in eye-hand coordination will improve with maturation and of course can be improved by occupational therapy sessions, but until acquired its lack often results in frustration.

The "cant" of the hook is lost in the very small sizes and the stationary finger is designed to curve above the moving finger for no apparent reason, so that the only objects which can be held securely are long, thin objects which utilize the "thumb" for three-point stability.

Children prefer to play with large objects, but the opening of the hook is very small. Neuromuscular ability to open the hook is poor yet the need for prehension force is great. Plastisol covering on the small size models is helpful for gross non-prehension stabilization and for clasping, but the plastisol is too slippery to hold objects in the hook with security and confidence.

Design Criteria

The terminal device for infants should be capable of holding haphazardly placed objects securely and should resemble the cable controlled device to facilitate later transition from the passive to an active prosthesis (3). Rather than having a scissor-like action, it would be preferable for the terminal device to simulate the function of a two-finger-to-thumb grasp position of the hand. The device should have a large opening span and increased prehension grip without requiring a commensurate increase in strength on the part of the infant to operate it. This would be possible through the use of increased surface contact and resilient materials. Enough friction is required on the outer surfaces of the device to assist in stabilization.

Component Designed

A terminal device was designed consisting of a wide curved finger section which extends from a broad base and stud, and a thumb which is a narrow curved projection hinged to the base. The distal portion of the finger section is covered on both sides with neoprene crepe. The palmar surface is contoured. The thumb is covered with rubber tubing. The device simulates the grasp pat-

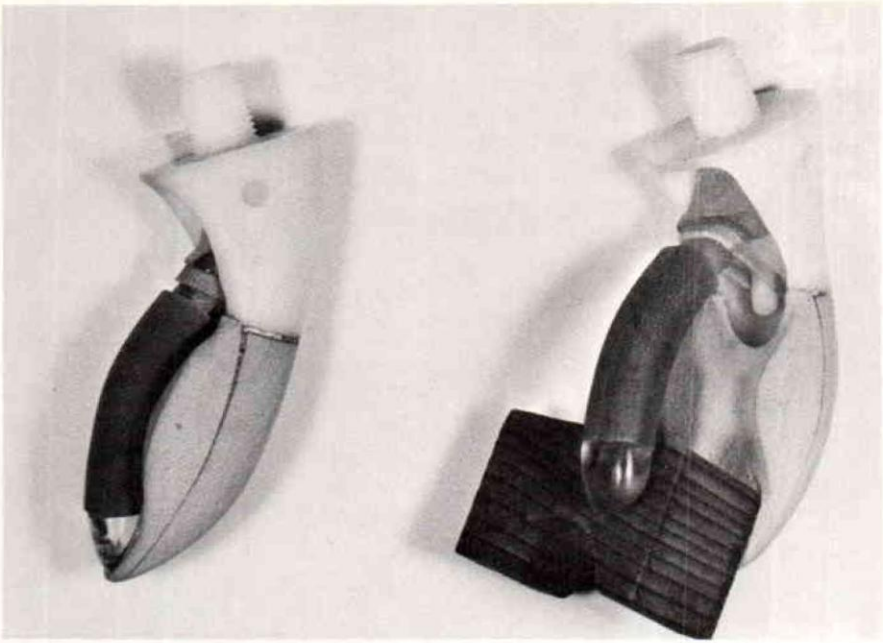

FIGURE 1—Terminal device.

tern of slightly flexed fingers opposed by a moving thumb. The width, resiliency, and friction of the neoprene covered finger section provides increased prehension stability. The narrow thumb allows good vision of objects grasped. The size of the opening is comparable to that of the Dorrance 99 X hook while the total terminal device length is only as large as the Dorrance 10 X hook. The basic device is made of Delrin* and, when inserted into the wrist unit, is the same circumference as the distal forearm, creating a continuous flowing line to the arm. In addition, this terminal device can be used on either the right or left side (see Figure 1).

* Delrin, E. I. DuPont De Nemours and Co., available at plastic distributors.

WRIST UNIT

Observed Needs and Problems

Although pre-positioning is beyond the skill and understanding of the infant, parents and therapists occasionally desire to position the terminal device for specific activities. This little used component adds terminal weight to the prosthesis, but some connection is necessary between the terminal device and the forearm to permit exchange of terminal devices.

Design Criteria

A lightweight constant friction wrist rotation unit without complex external adjustment is needed, made of a material that is inexpensive and requires minimal maintenance.

Component Designed

A wrist unit was made from a rod of nylon having a 1" outside diameter. When the thread was cut internally into this rod the final ¼" was diminished in depth to provide resistance. As the terminal device stud goes into the wrist unit it meets the section with less thread depth. Because the nylon of the wrist unit is a softer material than the Delrin of the terminal device, the threads yield under pressure, providing the desired amount of friction.

SOCKET

Observed Needs and Problems

When a very short below-elbow stump flexes beyond 90° it tends to retract into the antecubital fold, thus reducing the effective leverage area, permitting the stump to pop out of the socket. Rivet hinges which permit hyperextension can also result in loss of stump hold. Sometimes a figure-eight epicondyle strap will hold the stump in the socket, but this device markedly limits the range of forearm motion.

Adjustments are needed to maintain proper fit on the changing size and shape of the growing infant's stump. From six months to two years of age, stump circumference increases only 0.5 cm., but changes in shape and contour occur due to loss of baby fat and definition of musculoskeletal structures. Only limited modifications are possible with a conventional prosthesis to accommodate these changes. Thicker laminated areas may be ground out, but this is inexact, because the position of the stump in the socket changes with growth.

Design Criteria

Maximum loading is tolerated at 90°, but flexion beyond 90° results in loss of purchase on the stump. The socket was therefore designed for total contact to retain maximum purchase on the stump by taking the cast at 90° of flexion. To insure that the stump does not flex beyond 90° a high cut rolled trim line was designed to act as a stop to further flexion.

Lexan,* a completely transparent thermoplastic material which can be heated and formed directly over a male cast, was the material chosen for the socket. It is semi-flexible with strong resistance to impact and is very lightweight. The transparency provides the prosthetist with the opportunity to visually examine the fit of the socket over the stump and to directly observe the pressure areas. Adjustments can be made by heating and contouring the offending location. Lexan is commercially available in tubular shape $^{1}/_{16}$" thick with inside diameters of 1⅞" or 2⅛".

FOREARM

Observed Needs and Problems

It is difficult for an infant to stabilize objects against the

* Lexan, Chemical materials division, General Electric Co., Pittsfield, Mass.

smooth, hard texture of a laminated plastic forearm. Pushing up from the floor or leaning on the forearm while crawling are precarious activities for infants with these slippery forearms and it is uncomfortable to roll over or fall on a hard surfaced prosthesis. Leaning and crawling on hard surfaces with a hard forearm causes wear on clothing, and the prosthesis clatters when placed on a table.

During the period from six months to two years of age, forearm length increases 5.0 cm. (4), but the conventional prosthesis can be lengthened only by substituting a larger terminal device for a smaller one.

Design Criteria

Some form of friction surface is needed for the forearm. Polyvinylchloride (PVC) and leather gauntlets which are sometimes applied are unattractive and do not stay in place. It is desirable to have a pliable surface that yields and allows contouring when the child falls or clasps an object, functioning similarly to the skin of the human arm. Soft, textured forearms would also reduce impact noise and wear on clothing.

Observation of infants with unilateral below-elbow amputations has led us to believe that they need to flex their forearm beyond 90° but do not necessarily need a full 135° of flexion. Similarly, full extension and even hyperextension which are sometimes permitted by conventional protheses are not needed as these children have free shoulder motion and another sound arm. The forearm should therefore be placed in the most optimal functional range which will allow the child to manipulate his terminal device within a range of approximately 20 to 120° of flexion.

Component Designed

The forearm consists of a 1″ inside diameter central tube of Lexan which is covered with a foam filler and an outer sleeve. The Lexan tube is flared to conform to the rounded socket end. The forearm is bonded to the socket at approximately 30° of

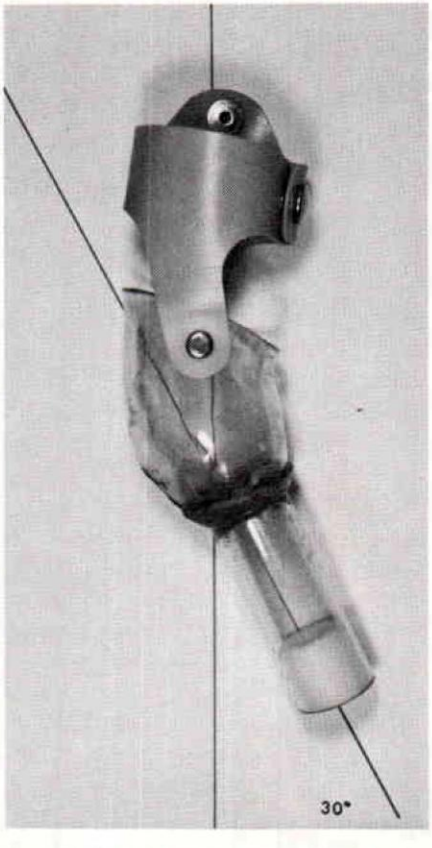

FIGURE 2—Pre-flexion angle of the forearm.

pre-flexion (see Figure 2). The length of tube added to the socket determines the length of the forearm section of the prosthesis, as the wrist unit fits inside the tube rather than adding to its length.

A foam material made of a lightweight compound of polyurethane is used to fill the space around the forearm tube and to complete the contour of the forearm from the socket to the terminal device. It is wider at the proximal end, as it needs to conform to the girth of the widest part of the socket, and it tapers at the distal end to the desired girth of the wrist. Made with a hole in the middle, it slips over the Lexan forearm tube. The prosthetist cuts it to the correct length and shapes the proximal end so it fits up against the base of the socket. He also shapes the outer contour to taper to the desired size. It is not necessary to provide any permanent attachment of the foam to the arm, for it stays in place well. It can be slipped off easily for repairs.

Covering the foam filler is a sleeve made of RREP.* Sleeves of PVC were tried and although they were flexible, gave good friction, and had an attractive appearance, they tore and stained easily. RREP is slightly less frictional and flexible but it is more stain resistant and does not tear as easily. Replacing the sleeve is a simple and relatively inexpensive procedure. However, even RREP stains more than is desirable. The sleeve covers the entire prosthesis up to the socket brim, giving the arm a finished appearance. It conceals and encloses all parts and provides a look of continuity (see Figure 3).

CUFF

Observed Needs and Problems

No change in design was felt to be necessary as the only objection to the polyethylene material is that it tends to elongate at points where hinges are attached.

Design Criteria

Cuff material which is more cohesive than polyethylene but otherwise has similar properties should be explored.

Component Designed

The cuff is a standard full cuff with a snap in the front. It is made of Ortholen† which is $^{1}/_{16}$" thick, flesh colored, medium- to high-density plastic. Polyethylene opaque of medium density can be used but Ortholen was found to have less tendancy to elongate and therefore rivets do not tend to pull through as easily. The cuff was made so it did not extend high into the axilla area. It was attached to the socket by rivet hinges which lie just inside the cosmetic covering.

* Plastic formula developed by the U.S. Army Medical Biomechanical Research Laboratory, Walter Reed Medical Center, Forest Glen Section, Washington, D.C.

† Kellogg Corset Company, Jackson, Michigan.

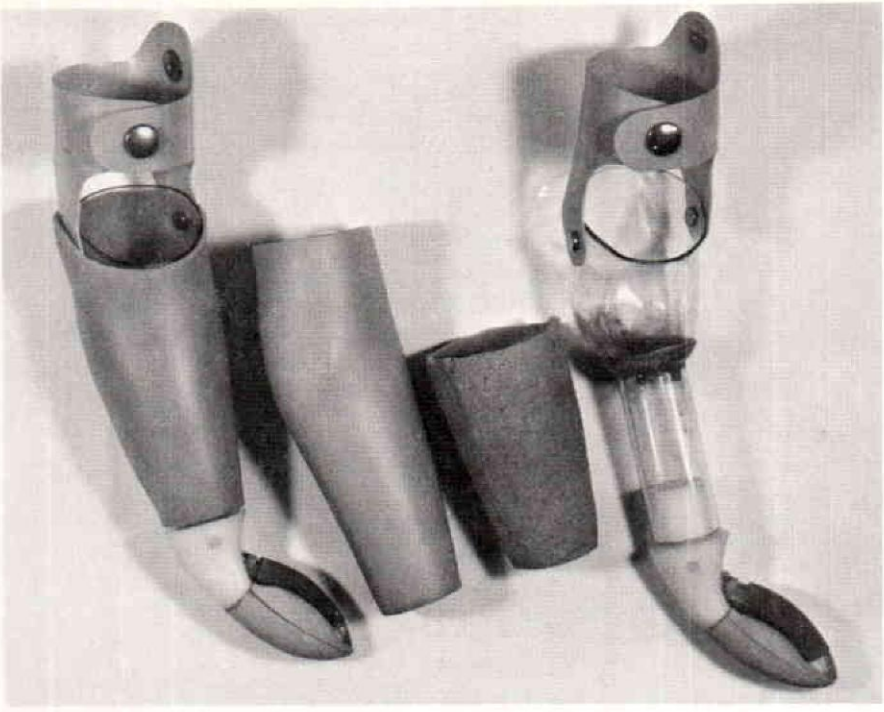

FIGURE 3—Forearm: Components separated and together.

SUSPENSION

Observed Needs and Problems

For infants, the primary function of the harness is to suspend the prosthesis, but because of the soft, rounded body contours, effective suspension is difficult to achieve. The soft body tissues of an infant are easily marked and displaced by tight straps, so that when the conventional harness is tight enough for good suspension, redness and eventually a callous formation on the skin may result. To avoid these undesirable effects, the harness must be fitted carefully which is difficult to achieve due to the lack of well-defined skeletal and muscular structures; and difficult to maintain due to the constant physical changes resulting from growth.

Since an infant's first motions are primarily lateral and initiate from the shoulder, the front support strap of the conventional harness often slides off the shoulder. A front chest strap therefore must be added to prevent this from happening. Unfortunately the extra strap restricts freedom of arm movement.

Because of the rapid increase in the chest circumference of an infant, the conventional harness requires frequent adjustment. From the age of eight months to the age of two years, chest circumference increases approxi-

mately 5.0 cm (5). Data on a group of twelve infants showed that all together they required fifty-one harness releases in one year from their first to their second birthdays, averaging a harness adjustment approximately every 2.5 months (6).

Design Criteria

A harness should be made which will provide direct counterforces to supply vertical suspension of the prosthesis. The harness should not fall off the shoulder, cause axilla pressure, require critical fit in the chest area, or need frequent adjustment. It should allow a maximum of shoulder motion and, as with any harness, a minimum of straps would be desirable.

Component Designed

The harness was made of ½" dacron tape and nylon utility cord (see Figure 4). It consists of a chest strap, over-shoulder strap, axilla cord and swivel tab. The over-shoulder strap and axilla cord together form a suspension loop which fits on the prosthetic side. The swivel tab slides on the axilla cord permitting free arm movement. This tab snaps to the cuff and is the only attachment point by which the harness suspends the prosthesis.

Vertical load forces are carried by the suspension loop which is fitted tightly enough to prevent the socket from being pulled off the stump. The nylon axilla cord must not bind the axilla and usually lies about one inch below it, allowing freedom of arm flexion and extension. It does not limit abduction if the swivel tab is set high enough.

The function of the chest strap is to retain the suspension loop on the shoulder and take care of lateral extended loads that are placed on the prosthesis. Therefore, it need not be fitted snugly around the chest and can allow enough slack for several fingers

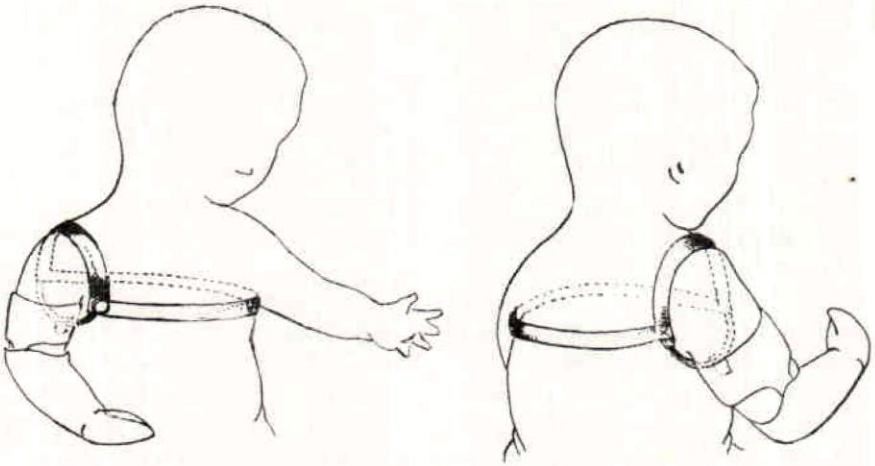

FIGURE 4—Harness.

to be placed under it and still be effective.

CARE AND REPAIR

Observed Needs and Problems

Care and repair required for an infant's prosthesis is considerable due to the growth and varied activities. Adults who used their prostheses similarly would be considered extremely heavy duty users of their limbs. Care of a conventional prosthesis requires regular cleaning and lubrication of the metal parts to prevent rusting and wear on moving components. Repair of the conventional prosthesis may involve relamination; repair of the broken part; or refabrication of the entire prosthesis.

Design Criteria

Simplified, low-cost repair procedures for infant prostheses are needed. Inexpensive, standardized parts which would permit simple replacement would be desirable. Materials which clean easily and would not require lubrication would simplify care problems.

Components Designed

Other than the socket, cuff, and harness, all parts of the unitized prosthesis were designed to be prefabricated and thus simple and inexpensive to replace. All metal except for three snaps have been eliminated. This extensive use of plastics allows the prosthesis to be scrubbed well without fear of rust or corrosion.

PROFESSIONAL REQUIREMENTS

Observed Needs and Problems

Infants, with their soft rounded contours, small size, and activities which demand extreme ranges of motion, make prosthetic fitting a challenge to the prosthetist. Often prostheses are not uniform in quality, resulting in unpredictable wear and use patterns. These limbs take as much time and more skill to fabricate than adult prostheses, yet families are reluctant to pay large sums of money for infant prostheses. A prosthesis is needed which will eliminate as much human error in its fabrication as possible.

Design Criteria

With less need to individualize prostheses, more uniform results could be expected. If standard production techniques were employed and individual components prefabricated, assembly should be easier, require less time and skill, and be less expensive.

Components Designed

The unitized arm has been designed to utilize primarily prefabricated plastic parts. The primary skill still required is in taking the wrap and shaping the socket. Simplified methods for cast taking and harness manufacture have been designed to further provide quality control. These methods are being tested but results to date are inconclusive.

WEIGHT

Observed Needs and Problems

One objectionable characteristic of the standard below-elbow prosthesis has been its weight (7). Although some of these little prostheses weigh only eight ounces, they often seem very heavy in relation to the musculature available to manipulate them. With experience and maturation, infants fitted with these prostheses have been observed to adapt well to them and to eventually become proficient users. Nevertheless it would be helpful if the weight of infant prostheses could be reduced and redistributed so that it is not concentrated at the distal end. When 33% of the prosthetic weight is in the terminal device, a potential hazard exists from the random banging of the infant's arm against furniture, windows, and people.

Weight demands an unnecessary amount of energy expenditure; does not contribute to efficiency of motion; is non-functional and actually decreases functional range. It does not seem to assist in balance and appears to inhibit use of the arm.

Muscular development of the affected side seems to be influenced by use and activity. Mere resistance against gravity is not effective. With lighter weight, use should increase and muscle development become more symmetrical.

Design Criteria

A lighter weight prosthesis with a more natural distribution of the weight should be developed.

Component Designed

Total weight of the unitized prosthesis has been approximately six ounces, representing an average 25% reduction in weight as compared to the conventional prosthesis.

APPEARANCE

Observed Needs and Problems

Conventional prostheses appear metallic and clumsy to parents of infants. The hook shape may frighten other children and appear to be a weapon. The prosthesis has a "hard, cold" appearance and texture. Parents are concerned at first about the potential danger of the hook, wear on clothing, and damage to furniture. Acceptance of the prosthesis is usually in relation to the functional advantages of having any prosthesis over none at all.

Design Criteria

Prostheses for infants should appear more pleasing, less weapon-like, and eliminate the suggestion of "Captain Hook". Rather than appearing mechanical they should appear aesthetically pleasing with flowing lines. They need not attempt to look like real body parts, but should not attract attention. They should have a pleasing color without incurring the expense entailed in precisely matching the color of human skin.

Components Designed

The unitized prosthesis appears to be one continuous piece. It does not have an obvious separation between the terminal device and the forearm. The terminal device is not hook shaped. It does not look like a body part, but does not attract attention. Except for the snap on the cuff and two snaps on the harness, all metal has been eliminated, thus making the prosthesis appear less mechanical and more pleasing for babies. The forearm cover is made in standard arm and leg color and does not attempt to match each child's skin. The softness and pleasing color also reduce the metallic appearance, making this arm more appropriate for this age group.

EVALUATION

To determine the extent to which the unitized prosthesis satisfies the design criteria, six infants were fitted with unitized arms at the Child Amputee Prosthetics Project. They wore unitized prostheses from their initial fitting at between eleven and fourteen months of age until the evaluation period of from fourteen to twenty-one months was completed. At that time all were converted to conventional prostheses with Dorrance 10 X hooks.

All six infants in the study had unilateral congenital below-elbow limb deficiencies. All were seen monthly on an out-patient basis through the period of their participation in the study. Parents requested additional appointments for repairs if they were needed. Detailed data was maintained on all components and on the fit and function of the prostheses throughout the study period.

Terminal Device

Much as they do with a conventional plastisol covered hook, the infants used the terminal device primarily for stabilizing objects and themselves in manipulative and locomotor activities. There was natural use of the prosthesis and terminal device for holding down objects and clasping objects between the terminal device and sound hand.

The curve of the experimental terminal device was often used effectively when it served to prop long protruding objects held under the arm and it also curved over the objects it held down, causing a very effective hold. When supinated, it was used as a scoop and in any position was leaned upon for crawling, sitting, and pushing up from the floor.

Several of the children manually opened the terminal device and, when it was held open by an adult, they would place objects into it. This was gross placement, but objects stayed in place and did not fall out. Only 1/8" of rubber band tubing was used, providing 3/4 of a pound of prehension force. The fact that objects did not fall out of the terminal device when held with such minimal rubber band loading was highly promising for function in the period when the terminal device will be actively operated.

Material used in the terminal

device appeared generally satisfactory. The major problem occurred with the neoprene crepe covering which absorbed dirt and wore through along the edges and at the tip of the dorsal surface where the children used them a great deal. Records showed that the neoprene crepe covering needed to be replaced at an average rate of once every three and a half months.

Three of the terminal devices broke at the stud when the children fell on them. This weakness is felt to be due to the individual fabrication technique and it is believed they will be less likely to break when design changes are made in this area by increasing the thickness.

Wrist Unit

For most activities the terminal device was left in a position part way between pronation and midposition. Although the wrist unit was not used a great deal, it was considered an important component for those few activities in which parents or therapists needed to position the terminal device. In all six cases, the wrist unit held friction well and the terminal device turned smoothly throughout the range of motion. The wrist units required no maintenance, but it was considered important to carefully insert the terminal device into the wrist unit to avoid stripping the threads. In no case were the threads stripped.

Socket

Design of the socket effectively prevented stump pop-out. No accessory straps were required even though five of the six infants fitted have very short stumps with an average length from epicondyle to tip of stump of only 2¼". The longest stump was 2¾" and the shortest was 2". All six infants wore stump socks. Two, who tried periods of wearing the prosthesis without a sock, found the Lexan uncomfortable when they began to perspire. Transparency appeared to be a distinct advantage in determining fit of the socket during use and in assuring that load tolerance was as desired.

Cracking appeared in the material with two prostheses after eight months of wear and with one prosthesis after sixteen months of wear. Strangely, the cracking did not occur at points of obvious stress. The exact cause of this cracking at present has not been identified.

Forearm

All of the children were fitted with approximately 30° of forearm pre-flexion. They could all straighten their arms enough to lean on the terminal device while sitting on the floor, and they could all flex their forearm well past 90° to clasp objects.

Shrinkage, twisting, soiling and tearing occurred with the use of PVC as a forearm covering, and all were replaced with RREP coverings which were commercially molded and stayed in place well. RREP coverings resisted tearing well, did not shrink or soil as much as the PVC and were gen-

erally more satisfactory. Three RREP coverings were replaced due to soiling although the most persistent soil problem came from ball point pen ink.

The core of Lexan tubing cracked on one prosthesis after eighteen months of wear. As the crack was near the wrist unit its cause is probably related to the same factor that caused the Lexan sockets to crack.

Foam forearm fillers were durable and functional. Four of the six patients wore their prostheses at the beach or into the swimming pool. The covering needed to be pulled back so the foam could be cleaned and dried after these activities to prevent rotting and deterioration from moisture. The foam fillers used initially were considerably softer than the ones that were produced later. The firmer fillers provided less resilience but were more durable.

Function of the textured resilient forearm was excellent. Friction provided by the texture and resilience aided secure clasping of objects, provided security when children leaned on the forearm, and gave excellent grip on furniture or other obstacles which children climbed upon. Contouring of the forearm was not visually apparent except when a firm grip was made on objects or when body weight rested upon it as in climbing.

Suspension

The harness was stable and comfortable for the children. Freedom of arm movement was especially evident when dressing the child and when the child was engaged in play activities. Infants were at various stages of crawling, walking and climbing. Their postures for these very gross body movements place the arms in extreme positions of abduction, external rotation, and circumduction. They place considerable body weight on their arms in all these positions and at no time was any restriction in arm motion observed, nor was there any shifting of straps during these strenuous activities.

At times, the chest strap appeared tight when it was not. Fit was checked by placing two fingers under the strap while the child was seated. The chest strap needed adjustment more frequently than any other part of the harness due to rapid increase in chest circumference during this period. Chest strap release was done on the average of once every 3.9 months during the study. It was felt that the interval could be lengthened if the prosthetist initially made the chest strap as loose as possible without letting the over-shoulder strap fall off the shoulder.

The main suspension loop appeared tight but did not seem uncomfortable for the children. It was found that correct fit of this loop was important to the fit and comfort of the harness, as it determines the angle at which the chest strap rides. It was also important to comfort for the axilla cord to begin low on the back and yet be long enough to allow full range of flexion and extension.

Placement of the swivel tab was important to good comfort and function. Dacron tape proved more comfortable than polyethylene

and was easier to adjust. The tab needed to be short and high enough to allow free abduction, but not so short that it was uncomfortable. Adjustments to the suspension loop and/or tab were done approximately every six months and not more often than three times for each child.

Stump pop-out was effectively prevented with these children. The combination of forearm preflexion with high brim socket design plus correct fit of the harness made it unnecessary to have any accessory straps. Pop-out occurred when the main suspension loop was too large or when its attachment through the swivel tab was too loose. The tendancy to pop out was easily detected by pulling the supine baby to a sitting position by holding the terminal device and sound hand. This placed a vertical load on the harness and showed whether or not the socket was held tightly enough on the stump. Intimate socket fit plus freedom of arm movement allowed by this harness may also have been responsible for the almost immediate spontaneous arm movements observed in these children.

Care and Repair

All of the infants in the study wore their prostheses for a long enough period of time to present a clear picture of repairs required. Seven socket adjustments were required. Three of these were aided by the transparency of the socket material while the other four were to flare or re-shape the trim line, procedures which were not influenced or aided by transparency. All of these adjustments were accomplished by heating and contouring the Lexan. It was during the heating of the Lexan that two forearm coverings which had been folded back were cracked. Following this experience, coverings were completely removed before heating the Lexan.

Time required for repairs related to the availability of prefabricated parts. Re-covering the neoprene of the terminal device took most time and would have been considerably shortened if these had been commercially available during the study. Many harness adjustments done with this particular group of infants were of an experimental nature and would not be repeated with another group.

Care of the prosthesis did not seem to be a problem to the families. At first, some families had difficulty cleaning the inside of the base of the terminal device, but they found that a small brush and water did the job well. Some of the families complained about the need to wash the forearm covering frequently, but this was reduced by the use of the RREP which seemed easier to keep clean. One mother placed the prosthesis in the washing machine with the family clothes, but was advised that hand washing was preferable. In general, those families that would take the trouble to clean any prosthesis, found the unitized arm no problem to keep clean.

Professional Requirements

The lack of prefabricated components in a large part of this study made professional require-

ments difficult to evaluate. The prosthesis appeared simple to fabricate when made by the designer who clearly understood the purpose and design. A second group of infants is currently being fitted by another prosthetist and details of fabrication as well as professional requirements for fitting and repairs are being documented.

Weight

It was hypothesized that added weight decreased use and control of the prosthesis. Observation of the six infants with the unitized arm tends to uphold this theory. Most of these children were observed to move and include their prostheses in activities very soon after fitting. For evaluation of prosthetic inclusion a small control group wearing conventional prostheses was used. It was found that the infants wearing the unitized prosthesis included the prosthesis in the early phases of wearing more frequently and with more natural mannerisms than the infants wearing the conventional prostheses. It was felt that the lighter weight was an important contributing factor to the earlier usage.

Appearance

Families accepted the unitized prosthesis very well. They felt it looked better than the conventional prosthesis. Some commented that the terminal device seems to be part of the whole arm rather than a separate piece. Several reported being with relatives or friends who watched the baby play for some time before realizing that it was wearing a prosthesis. The unitized arm did not appear hard and cold and seemed quite appropriate for a baby. All of the families commented on the fact the they did not experience remarks from strangers until the children were fitted with conventional prostheses.

CONCLUSIONS

A comprehensive study of the needs and problems of infant below-elbow amputees has resulted in development of a new prosthetic device based upon an endoskeletal concept. Criteria established for the design of components and the current status of the device designed are summarized:

Design Criteria

Terminal Device:

1. Weigh no more than two ounces.
2. Have an opening range equal to that of the Dorrance 99X hook or larger.
3. Have improved prehension qualities related to shape and friction surface and simultaneously reduced amount of force required to open the device.
4. Require less precision for object placement.
5. Should resemble the prototype designed for cable control.

Wrist Unit:

1. Weigh less than the conventional wrist unit.
2. Require less maintenance and care.
3. Be less expensive to manufacture.
4. Maintain constant friction con-

cept of the conventional wrist unit but utilize a simplified mechanism.

Forearm:

1. Provide friction surface with resilience.
2. Pre-flex the forearm to transfer the 90 to 100 degrees of stump motion to the more functional range of 30 to 120 degrees of forearm motion.

Socket:

Design to allow a stump range of motion from 90 to 100 degrees without permitting the stump to pop out.

Cuff:

Use a plastic which will not elongate at the rivets.

Suspension:

1. Design a harness that will provide adequate counter forces against vertical pull and horizontal loading without restricting range of motion or binding soft tissue.
2. Design a harness with straps that do not fall off the shoulder and that does not require critical fit in the chest area.

Care and Repair:

1. Utilize materials that are easily cleaned, do not require lubrication, can be spot molded to accommodate changes in stump contour.
2. Design components for simple interchange or replacement.

Professional Requirement:

Design a prosthesis which will be easier to assemble, fit, adjust and maintain.

Weight:

Overall weight should be six ounces or less, with the weight evenly distributed rather than concentrated distally.

Appearance:

1. Design a prosthesis which is aesthetically pleasing and not conspicuous.
2. Eliminate the weapon-like, "Captain Hook" appearance.

Current Status of Unitized Prosthesis

Terminal Device:

1. Weight is 1.54 ounces.
2. Opening is 2¾ inches as compared to the 3-inch opening of a 99X hook.
3. Due to the textured resilient cover and special shape, prehension stability is good. Pinch force has been reduced from the average 1½ pounds of the 10X hook to ¾ of a pound.
4. Because the "finger" section is 1¼ inches wide in contrast to the usual width of only ⅜ inch of the conventional infant hook, there is a much greater margin of error permitted in the placement of objects within the hook.

Wrist Unit:

1. Smooth constant friction is provided and the unit proved sufficiently durable to require no replacement during the infant period.
2. Unit is made of plastic which is inexpensive to manufacture.
3. Unit utilizes the simple friction principle of changing thread depths.

Forearm:

1. The textured, resilient forearm covering contours around objects, providing a secure grip.
2. Pre-flexion of the forearm provides a range of forearm motion from 30 to 120 degrees.

Socket:

1. Socket design effectively prevented stump pop-out and allowed 90 to 100 degrees of stump motion.
2. Material used requires additional study.

Cuff:

Elongation at the rivets did not occur in cuffs made of Ortholen as the material apparently does not stretch as rapidly as polyethylene although it has the other satisfactory properties.

Suspension:

1. Harness design satisfactorily met the criteria.
2. Harness adjustments for growth were required at approximately four-month intervals as opposed to adjustments required every two and a half- to three-month intervals for the conventional figure-eight harness.

Care and Repair:

1. The prosthesis is easily cleaned and requires no lubrication as it is all plastic.
2. A longer forearm tube, filler and cover could be applied to compensate for linear growth on the normal side without remaking the entire prosthesis.

Professional Requirements:

1. Prefabricated forearm and fillers have led to uniformity in quality. Other parts should provide the same uniformity when commercially produced.
2. Socket and harness fabrication continue to require skill on the part of the prosthetist.

Weight:

1. Unitized prosthesis weighs approximately six ounces, representing a 25% reduction from the conventional limb.
2. Weight at the distal end has been reduced.

Appearance:

1. Parents, without exception, prefer the appearance of the unitized arm to the conventional arm. There is, of course, always room for improvement.

Under a grant from the U. S. Children's Bureau, Research Division, arrangements are in process to produce the terminal device from molds. When completed the devices will be clinically evaluated at selected child amputee centers throughout the United States.

A detailed procedure for fitting and fabricating the socket and harness is currently under preparation, together with a description of the techniques involved in making socket and harness adjustments to compensate for growth. The method of socket casting and harness fitting differs from methods used for the conventional prosthesis. Analysis of these procedures is currently under study and an article on fabrication and assembly will be prepared for publication when these studies are completed so that the below-elbow unitized prosthesis will become available for wider application.

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Critique of Lower Extremity Bracing

by

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INTRODUCTION

Lack of progress and poor results in the field of lower-extremity Orthotics are evident today. Braces have changed little in fifty years, yet the introduction of new materials and new concepts should make possible the improvements which have failed to materialize.

This is even less tolerable since engineers now tell us that an exoskeleton is mechanically more effective than an endoskeleton, and that spacemen will someday wear artificial exoskeletons which will multiply the available power while reducing energy cost (1).

1. *The Key Problem: Knee Stability*

Regardless of diagnosis or condition, all paralyzed individuals today are provided with the same kind of brace, reason for this being that the key problem remains knee stability. Lower-extremity ortheses are nothing but glorified knee locks, slightly modified to suit individual tastes.

2. *Ambulation Without Brace*

During normal human locomotion, muscles of the lower extremity act upon the segments to accelerate and decelerate them; they also provide the "holding" power which locks the joints and provides stability. When the muscles are totally or partially paralyzed, "weakness" endangers stability and impairs control (2) (3).

2.1. When paralysis is not complete, the patient may sometimes continue to ambulate without a brace, by modifying his method of obtaining joint stability. For ex-

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ample, he can use recurvatum to lock the knee.

2.1.1. This method of ambulation increases energy consumption; when the patient "throws" forward his paralyzed limb, the shin accelerates rearward abruptly. It is deprived of power for deceleration and/or counter-acceleration, so that gravity itself must bring the shin forward past the vertical. This also results in considerable slowness of displacement (see 3.1).

2.2. Vertical displacement of the center of gravity of the body is influenced by the increased length of the lower extremity on the stance side. Once the knee is hyperextended, the patient must hoist his body mass higher than when the knee remains slightly flexed as physiologically desirable (see 10.3). Again, the result is higher energy cost (4).

3. *Ambulation with a Brace*

More severe cases need to wear a brace with a locked knee. The orthosis restores stability, but it creates complex problems of utilization.

3.1. During normal ambulation, forward acceleration of the thigh segment is counter-balanced by rearward motion of the shin. This bi-pendular action reduces the amount of muscle work by transfering angular momentum from thigh segment to shin segment, and vice-versa.

3.2. As soon as the brace is locked at the knee, the lower extremity is changed from a double pendulum into a single pendulum. The increased length resists the initiation of forward motion, according to the law of resistance to angular acceleration (5) (see 4.1).

3.3. A comparison of energy cost in ambulation with a free knee, then with a lock added, has been made (6) on a subject fitted with a conventional, unilateral brace without ischial seat. Energy cost remained constantly higher with the knee locked, the average increase being placed at +25.4%.

4.1. *Swing Phase on the Affected Side*

Once the foot is fixed in a neutral position, the lack of dorsiflexion results in a hip-toe distance greater than hip-heel. The lengthened limb forces the subject to elevate his pelvis to clear the ground; in this case, the amount of work performed increases in proportion to the height to which the C. G. is raised (7). Moreover, the pendular axis finds itself elevated with regard to the ideal position of positive Tredelenburg (4) (8). Any amount of foot drop makes the situation worse.

4.2. *Swing Phase on the Sound Side*

During normal ambulation, at heel strike the C. G. of the body is descending and must be reversed (8). If the lower extremity is rigid, the forward momentum carries the body over the foot; this slows down the body and fails to provide the knee flexion preparatory to contralateral swing phase. This increases energy cost (6) (9) (10).

4.3. A rational brace, making possible a near-normal swing phase while providing stability during

stance phase is offered by the UCLA functional long leg brace (2). Thus it becomes possible to combine stability and joint freedom.

5. *Weight of the Brace*

The weight of the orthosis is concentrated in the distal part (knee and ankle joints, foot plate). This lowers the center of gravity of the pendular segment.

5.1. Whenever the center of gravity moves away from the rotational axis (in this instance the hip joint), the moment of inertia increases by a quantity proportional to the square of the distance mass-axis.

5.2. Time of swing of a pendulum is proportional to the square root of its length ($1/\sqrt{L} \cdot L = \sqrt{L}$ where L = pendulum length). The "unipendular" segment created by the knee lock slows down the motion of the braced extremity.

5.3. Once in motion, the braced lower extremity is submitted to increased pendular acceleration. The energy accumulated must be dissipated by the patient who, if paralysis of the gluteus maximus is present, performs some abnormal braking motion.

6. *Location of the Center of Gravity of Limb*

Weight itself is not as important as its distribution. This was demonstrated by a study concerned with prosthetic substitution of the lower extremity (11).

6.1. Lowering the center of gravity of a limb results in an increase of inertia (see 5.1). Inertia is the most significant obstacle to muscle action, in deceleration as well as acceleration. Using a weight sliding on a calibrated stick, Bresler (11) was able to record the influence of the location of limb center of gravity. He showed that energy cost increases in proportion to lowering of the center of gravity, inertia becoming proportionally greater.

6.1.1. The same author showed that inertia had more influence on energy cost than specific location of the center of gravity of the limb. Thus, the presence of a knee lock is a worse factor than is the weight of the brace itself.

6.2. Starting from theoretical premises, Staros (4) could calculate that, if the weight of the foot and shin sections in a prosthesis were halved, inertia would be reduced by 60% and energy cost by 40%. This should apply to orthotics as well. Even if lower extremity braces retain the same shape, their weight, at least, can be reduced through use of light alloys and plastic materials.

7. *Influence of Proper Alignment on Weight of Brace*

Steel uprights and thick leather corsets could be justified if the weight of the patient's body were borne by the brace. This is not true (2) (12). The weight is borne—with few exceptions—by the patient's own skeleton. The brace acts only to maintain the knee in an extended position; this is achieved by A-P contention.

7.1. Correct alignment is the

result of pre-tibial pressure combined with reactions at the heel and upper thigh levels. As long as the lower extremity is properly held, stress on the brace is minimal. But if vertical alignment is poor (loose pre-tibial band, for example), the subject "sits" in his brace and quickly destroys it.

7.1.1. Perfect alignment has made it possible to create the "Attelle Monotubulaire" (12) and the "Unibar" brace (13) (14), each utilizing a single external upright. As regards to weight, this should be getting close to Staros' hypotheses (Para. 6).

7.2. In addition to weight reduction, another advantage of the single upright brace is better alignment of mechanical knee and ankle joints, including tibial torsion along the vertical axis (see 9.2.2).

8. *Movement in Lower Extremity Joints*

Several complex movements take place in the joints of the lower extremity. As for mechanical joints, until now they allow only flexion-extension by rotation around a single axis (double-axis joints are rare, and again permit only flexion-extension).

8.1. *Hip Joint*. During ambulation, the femur articulates with the ilium:

8.1.1. in flexion, 35°+.

8.1.2. in extension, about 5°.

8.1.3. the amount of angular rotation has not been precisely measured, but it is essential to normal ambulation.

8.1.4. mechanical joint: flexion-extension only.

8.2. *Knee Joint*. The following movements take place (15) (16):

8.2.1. Flexion ± 70°.

8.2.2. Extension: neutral position is almost never reached, the knee remaining slightly flexed (5° to 20°) during stance phase.

8.2.3. Tibial rotation around vertical axis:

—at heel rise, 9° of internal rotation;

—at heel strike, 8° of external rotation.

8.2.4. Mechanical joint: knee locked in hyperextension, no flexion possible, no tibial rotation.

8.3. *Ankle Joint*. During ambulation, the following takes place (16):

8.3.1. At heel rise, ± 10° of dorsiflexion.

8.3.2. At heel strike, ± 20° of plantar flexion.

8.3.3. Movements of the subtalar joint:

—just before heel rise, 6° of inversion;

—just after heel strike, 5° of eversion.

8.3.4. Mechanical joint: dorsiflexion and/or plantar flexion either free or limited. No inversion or eversion possible.

9. *Problems of Alignments With Mechanical Joints*

Any alignment of a mechanical joint is a compromise. However, one must guard against gross oversimplifications, such as considering the knee joint as a hinge.

9.1. The anatomical knee joint combines hinge and sliding motions, the rotational axis changing in relation to the position of the fem-

oral condyles. This constant migration was evidenced by an X-ray series, the conclusions to be later controlled on cadavers (17).

9.1.1. Ideal placement of the mechanical rotational axis at the knee joint is impossible today; the orthotist has no alternative but to make use of the best possible compromise.

9.2. At ankle level, the rotational axis is approximately perpendicular to the path of the body center of gravity; in this fashion the ankle is free to plantarflex and dorsiflex during lateral oscillation (18) (19).

9.2.1. The mechanical ankle joint should be aligned with the transverse axis of the anatomical joint. Placement of this axis is secondary to the amount of tibial torsion and not related to the more or less everted position of the foot.

9.2.2. Tibial torsion takes place during maturation of the individual, going from 2° in the newborn to 20° or 30° at about 7 years of age. This torsion on the longitudinal axis places the ankle joint in a position suited to bipedal locomotion (20). A measuring board makes it easier to correct for ankle alignment (21).

10. *Interference of the Brace with Basic Components of Gait*

Going through the basic components of gait, it becomes easy to note the abnormal factors introduced by lower extremity bracing of the conventional type (7) (22).

10.1. Pelvic rotation around the vertical axis, total of 8°: made impossible by pelvic bands and locked hip joints.

10.2. Pelvic tilt on the swing side, 5°: made impossible by the invariable length of the brace with knee lock.

10.3. Knee flexion at stance phase, about 15°: made impossible by the knee lock.

10.4. Knee and ankle action whose coordinated action eliminates brutal ascent and descent of body center of gravity—impossible because of the rigidity of the brace.

10.5. Lateral displacement of pelvis, bringing center of gravity of the body on top of the foot: often compromised by improper brace alignment which keeps the foot too far away from mid-line.

10.5.1. Importance of lateral motion of the pelvis: during swing phase the center of gravity projection comes very close to the center of the heel on the stance side (23).

10.5.2. The need to reduce the lateral excursion of the pelvis explains the presence of knee valgus in the architecture of the normal lower extremity.

10.5.3. It is imperative to respect this alignment, and not to force the lower extremity to conform to a brace built straight, as is often the case.

CONCLUSION

Braces for the lower extremity have changed little since the days of Von Hessing (1839–1918). By contrast, prostheses are improving rapidly. Thus it seems that Orthotics should borrow as much as

possible from Prosthetics and make use of the same scientific evidence. It is, after all, a paradox to see that a man who has lost his leg can walk with only a trace of limp, when a paralyzed person whose skeleton, at least, is intact is immediately conspicuous in a crowd and wastes considerable amounts of energy to ambulate.

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Application of External Power in Orthotics*

by

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The application of external power in orthotics has been explored for a number of years and has, in certain instances, progressed to routine clinical applications. This paper represents an attempt to delineate the practicability of external power versus conventional power and control sources. Both modes of power have been employed on a research and clinical level at the Institute of Rehabilitation Medicine, New York University Medical Center. The results of this experience are presented in this paper.

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In choosing one power source over the other, one must bear in mind that external power in itself does not necessarily bring about improved function. It is generally agreed by most investigators that, if at all feasible, and if there is no appreciable difference in the functional end-result, body power should be chosen in favor of external power. The reasons for this choice are obvious. At the present state of the art, body powered devices are less complex and require, therefore, less maintenance and are likely to be of lighter weight. More important, the basic design of a prehension orthosis, for example, is identical in the type of pinch provided, whether external or body power is used. This means that there is no inherent improvement in the terminal function of externally powered devices. Ex-

ternal power is, however, indicated when body power is insufficient to activate the orthosis. Control of an externally powered device requires little force and range of motion at the control site, but does not provide feedback in an anesthetic limb. Control of body powered orthoses, on the other hand, provides some means of feedback because of a certain proportionality of excursion and force between the control site and the actuating device. So far, most of the research in externally powered devices and their routine clinical application has concentrated on the upper extremity. There are, however, a number of interesting lower-extremity developments which offer some functional improvements over conventional braces.

LOWER EXTREMITY

External power applications in lower-extremity orthotics have been confined to a relatively small number of designs. A likely reason for this lack of sophistication is that a paralytic lower extremity can be stabilized by very simple means, i.e., limiting or eliminating ankle, knee, or hip motion with a conventional brace. Although this enables the patient to ambulate, most conventional braces produce an abnormal gait pattern and an increase in energy consumption by blocking joint motions. This may not be of great consequence in unilateral involvements but imposes severe limitations in locomotion when the patient is paraplegic.

This problem has been attacked by Dr. Liberson by motorizing the hip joints of bilateral long leg braces and pelvic belts. The electric torque motors are designed to alternately drive the braces in opposite directions, thus producing hip flexion on one side while producing hip extension (push-off) on the opposite side. Control of hip motion is obtained through switches placed in the patient's shoes. This development is still in the experimental phase, but holds great promise in future developments in providing the paraplegic with greater mobility.

Another approach eliminating the brace entirely has been proposed by Liberson, et al., as well as Moe and Post. It is a muscle stimulator used in drop foot conditions, due to upper motor neuron disorders. An electrode placed over the peroneal nerve area provides a stimulus to pull the foot into dorsiflexion and eversion. A switch placed in the heel portion of the shoe is used to interrupt the muscle activating pulse in the stance phase of gait. The power pack of the muscle stimulator is carried around the waist.

An alternate solution to providing more nearly normal locomotion with a brace is the incorporation of a hydraulic stance and swing phase system between the ankle and knee joints, rather than the application of external power (Figure 1). Developed at the Institute of Rehabilitation Medicine, this system is in the experimental fitting phase at the present time. It is designed to provide stability at the knee during the critical period from heel strike to mid-stance and at the

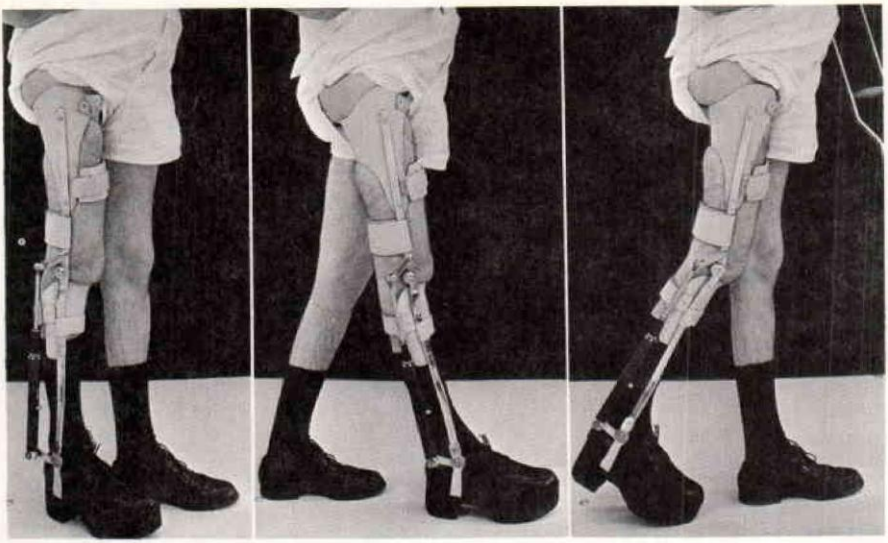

FIGURE 1—Above knee brace with Hydra-Nu-Matic cylinder for coordinated knee-ankle control.

FIGURE 2—The Hydra-Nu-Matic cylinder offers controlled fluid resistance to plantar flexion.

FIGURE 3—Controlled knee flexion offers a more nearly normal pattern of gait.

same time offers controlled fluid resistance to plantar flexion (Figure 2). Plantar flexion causes the hydraulic fluid in the cylinder to be displaced upward, resulting in an extension moment about the knee joint. This reciprocating action also comes into play in the swing phase, where knee flexion produces dorsiflexion of the foot. A 90 degree dorsiflexion stop is used for standing stability and to substitute for push-off. In allowing controlled knee flexion, a more nearly normal pattern of gait as well as a reduction in energy consumption is achieved (Figure 3). This design is still in the early stages of development and more clinical applications are needed to determine the practicability of such a design in terms of mechanical wear and maintenance.

UPPER EXTREMITY

The task of providing useful hand and arm functions is much more complex than that of providing ambulation by orthotic means. Hand and arm functions are generally much more important in the activities of daily living and vocational pursuits than lower extremity function. One may, in fact, consider the hand an extension of the brain, as we employ our hands not only for physical activities but also to lend greater expression to the spoken word and to enhance the effectiveness of speech.

The application of external power usually depends on the patient's residual motor power, i.e., if it is insufficient to activate a body-powered orthosis, external power is indicated. There are, however, conceivable exceptions to this rule. The use of external power

should be explored in certain applications even when sufficient body power is present. For example, a wrist-driven prehension orthosis is commonly used to provide pinch when the patient has residual wrist extensor strength. Although adequate body power is available, activation of the device involves motions not only at the wrist but also compensatory motions at the elbow and shoulder joints in order to maintain the hand over the object to be grasped. It would seem that with further development of external power a more efficient mode of finger prehension could be obtained. On the basis of clinical experience at the Institute of Rehabilitation Medicine, patients fitted with gas or electrically-driven prehension orthoses tend to use their devices more regularly than patients fitted with wrist-driven prehension orthoses. This tendency may not only be ascribed to the fact that externally powered orthoses are fitted to patients who sustained a cervical cord lesion above the C 6-7 level and are, therefore, more dependent on orthotic devices. Rather, it is very likely that functional performance is improved because compensatory motions, needed with wrist-driven prehension orthoses, are not necessary for prehensile activities.

Over the past three years electrically-driven prehension orthoses have been used for patients who lack wrist extensor strength (Figure 4). The power pack consists of a nickel cadmium battery, a permanent magnet 12 volt motor, and a charger, permitting the patient to recharge the battery from a regular household outlet. An adjustable slip clutch (Figure 5) is used to selectively adjust the pinch force as well as to provide a safety mechanism in case of switch failure. The orthosis is activated by an unidirectional microswitch usually

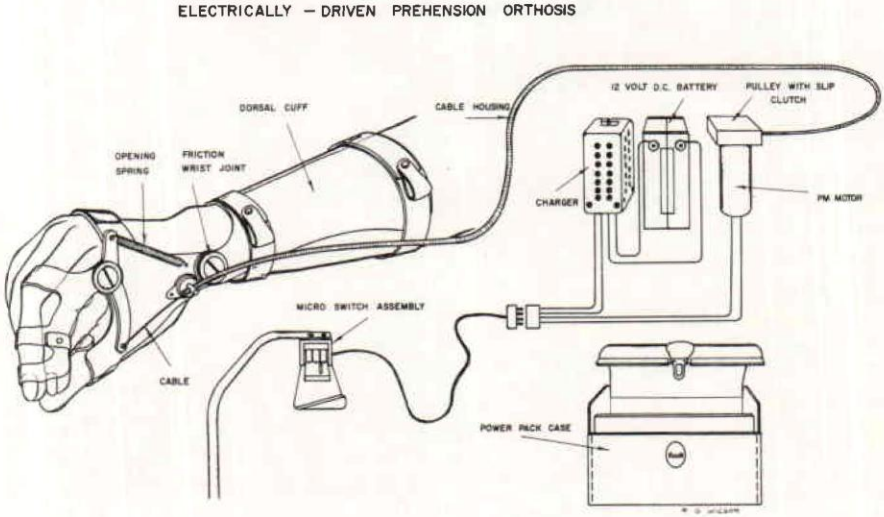

FIGURE 4—Electrically-driven prehension orthosis.

placed superior and posterior to the contralateral shoulder (Figure 6). This control site has been found to be most effective in leaving the fitted extremity free to move without inadvertently operating the orthosis. It provides a more nearly synergistic control motion since one normally uses a certain amount of "body English" when reaching for objects. Thus, a sound kinematic mode of control is achieved which requires little or no patient training.

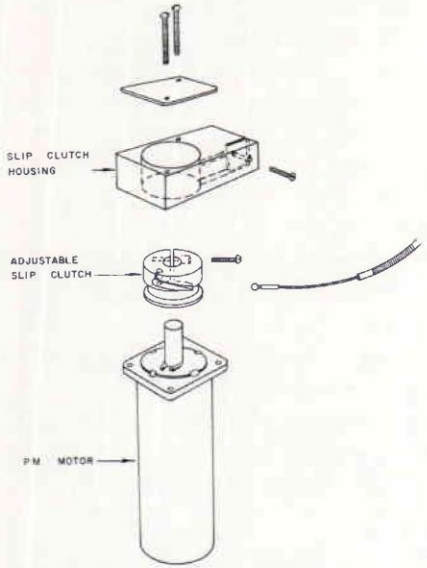

FIGURE 5—Exploded view of adjustable slip clutch for electrically-driven prehension orthosis.

The application of external power would seem useful even in patients who have good wrist extensors and who would conventionally be fitted with a wrist-driven prehension orthosis. Thus, there would be no need for compensatory motions of the other arm joints for terminal device operation. The orthosis would not have to extend above the wrist, leaving the upper extremity with the optimum degree of freedom. Although such fittings are possible, they are not practicable at the time with the conventional mode of orthotic control. Further development of

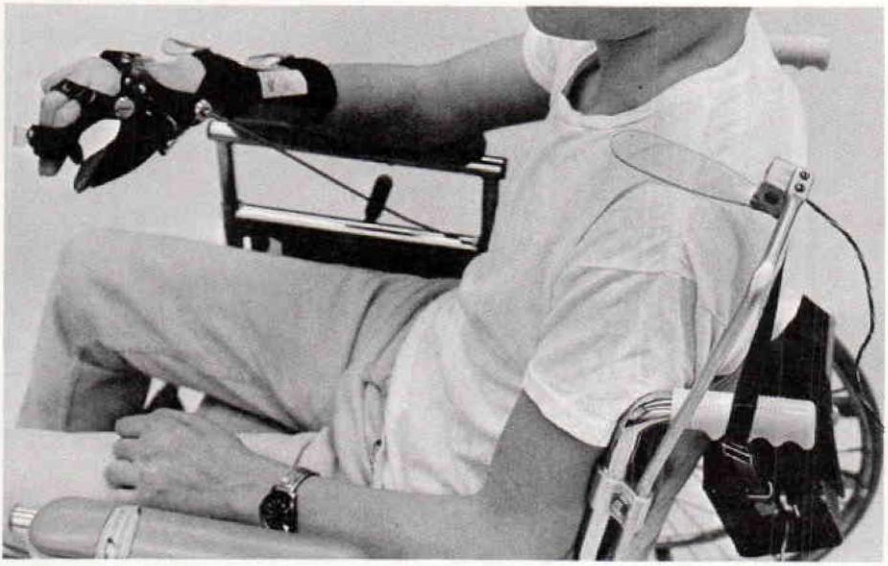

FIGURE 6—Unidirectional microswitch is placed superior and posterior to contralateral shoulder.

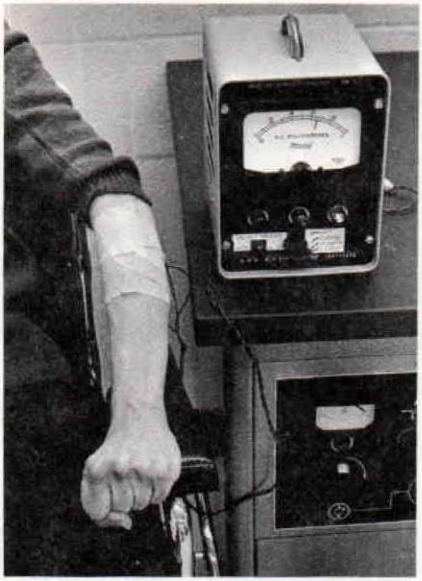

FIGURE 7—Three state myoelectric trainer unit.

myoelectric control systems may, however, lead to development of a prehension orthosis for the C 6-7 quadriplegic patient which would permit freedom of the wrist. This is possible by picking up myoelectric control signals from suitable forearm musculature. The three state myoelectric trainer developed at the University of New Brunswick is utilized to test the feasibility of such a system (Figure 7).

The results of external power applications in ambulatory patients has not proved to be as satisfactory as it has for wheelchair-bound patients. The additional weight of the power source and actuator which the patient has to carry are difficult problems to overcome. Furthermore, ambulatory patients possess greater mobility which may increase the frequency of inadvertent operation. It has been our experience that for ambulatory patients body powered and/or manually controlled orthoses are preferred because of their simplicity and lighter weight. Patients who have no hand function can be successfully fitted with a shoulder-driven prehension orthosis (Figure 8), providing voluntary opening and spring closing (Figure 9). The spring closing feature necessitates the incorporation of a pressure relief mechanism to avoid skin breakdown of the thumb, index and middle finger pads during periods when the orthosis is not actively used. This is of utmost importance in patients with a brachial plexus or similar lesion, resulting in both motor and sensory losses. The pressure relief control consists of a spring-loaded push-button below the MP joint

FIGURE 8—Shoulder-driven, cable controlled prehension orthosis.

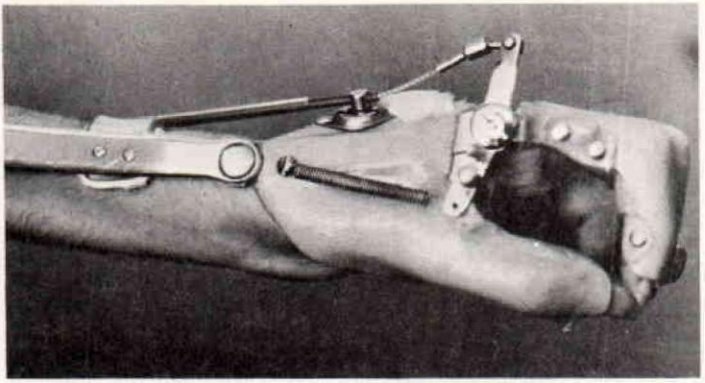

FIGURE 9—Voluntary opening-spring closing to three jawed chuck prehension.

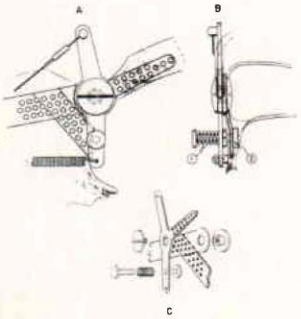

FIGURE 10—A. Finger flexion joint with pressure relief control assembled. B. Transverse view; the spring-loaded pressure relief button (1) engages in recess (2) when pressure relief is desired. C. Exploded view of pressure relief mechanism.

which is pushed into a semicircular recess. This prevents full finger closing. Increased cable tension causes the pushbutton to automatically retract, unlocking the pressure relief (Figure 10).

Manual control of arm braces for ambulatory patients, i.e., positioning and locking of the elbow at the desired angle of flexion, has proved more satisfactory than both external power or cable-controlled body power. Obviously, the force requirements for body-powered arm orthoses are infinitely greater than for an equivalent prosthesis because, in addition to the weight of the orthotic device, the weight of the patient's arm must be considered. While for unilateral arm braces manual control is most practical, in patients with bilateral arm involvement, external power is the only alternative to providing useful arm and hand functions. Here, CO_2 as a power source seems to have advantages over electric power because of the greater simplicity of the system. A CO_2 piston and cylinder elbow actuator obviates the mechanical elbow lock needed with the McKibben muscle substitute (Figures 11A and 11B). In the piston, the carbon dioxide is confined in a rigid-walled container, which affords sufficient resistance against elbow extension when forearm loads are applied. Activation of the control valve requires little force which can usually be obtained by harnessing residual arm motions (Figure 12). The wheelchair-bound patient requiring an arm orthosis poses again another problem. In this applica-

tion electric power sources have been found more useful than carbon dioxide. Electric power actuators provide a more definite control when compared to CO_2 actuators, especially the McKibben muscle substitute. In the latter, the problem of rebound is difficult to overcome when loads of various magnitude are placed in the terminal device. The electric arm orthosis developed at the Institute of Rehabilitation Medicine is designed to provide the high level quadriplegic patient (C 4-5) with voluntary arm and hand functions (Figure 13). It allows the patient a total of five degrees of freedom, four of which are motorized. The power actuators are 12 volt permanent magnet motors located at the back of the wheelchair. A twelve volt battery serves as the power source for both the arm orthosis and the wheelchair. Power transmission from the electric motors to the moving orthotic arm segments is provided through Bowden cables (Figure 14). The motor used for powered prehension is equipped with an adjustable slip clutch designed to vary the force of prehension and to act as a safety device. The other motors are provided with limit switches. Motions motorized are:

1. Finger opening and closing to a jaw chuck type of pinch (Figure 15).
2. Pronation and supination obtained through a spiral shaft running in a nylon sleeve (Figure 16). A linear pull on the sleeve causes forearm rotation (Figure 17).
3. Elbow flexion and extension

FIGURE 11 (A & B)—Arm brace with CO_2 piston actuator to provide elbow flexion.

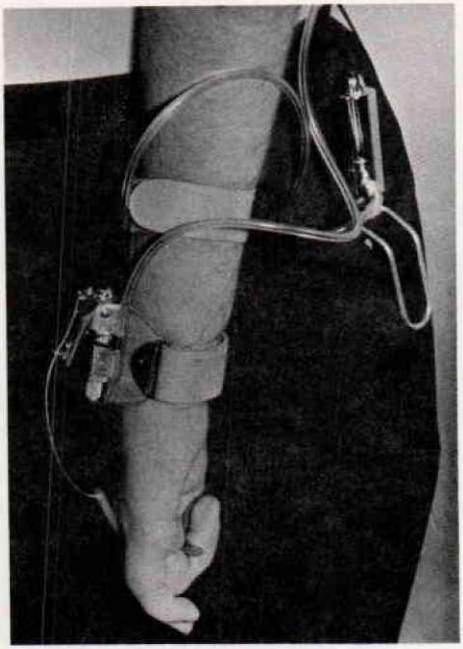

FIGURE 12—Residual wrist and finger flexion is harnessed to activate CO_2 control valve.

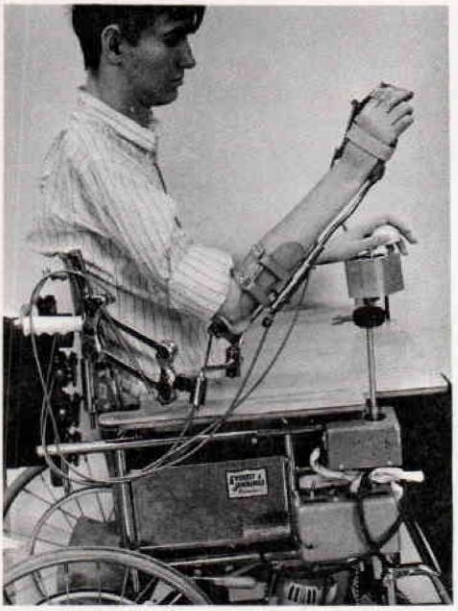

FIGURE 13—IRM electric arm orthosis.

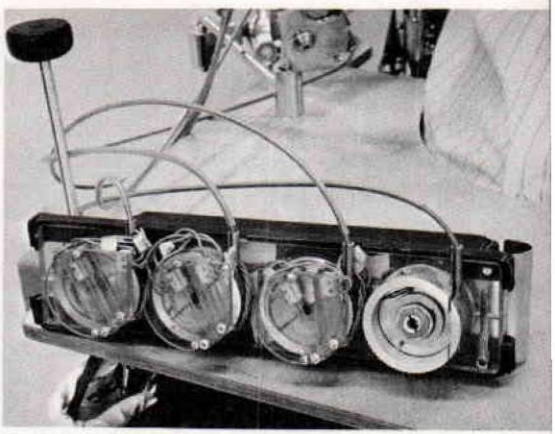

FIGURE 14—Power transmission from the electric motors to the orthotic arm segments is provided through bowden cables.

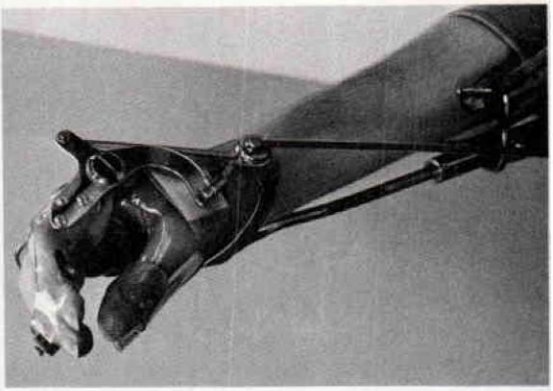

FIGURE 15—Finger closing to a three jawed chuck type of pinch.

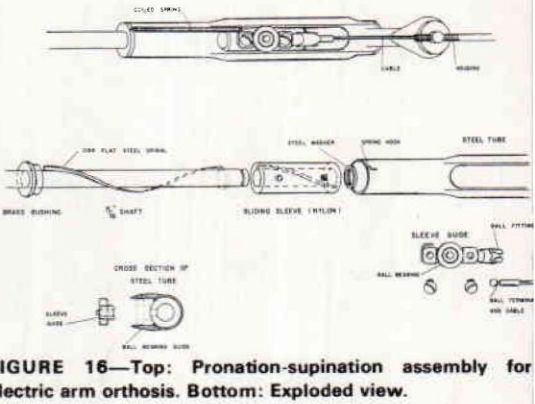

FIGURE 16—Top: Pronation-supination assembly for electric arm orthosis. Bottom: Exploded view.

FIGURE 17—Forearm rotation is caused by a linear pull on the sliding sleeve of pronation-supination unit.

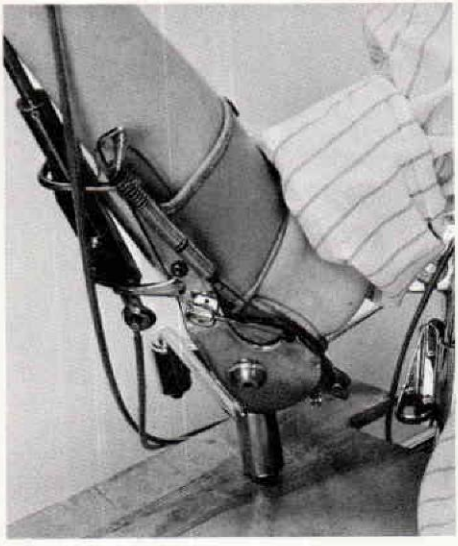

FIGURE 18—Motorized elbow flexion unit.

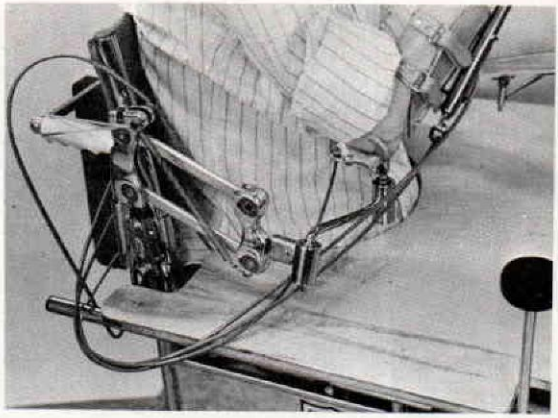

FIGURE 19—Parallel linkage lateral to humerus provides combined flexion-abduction and extension-adduction.

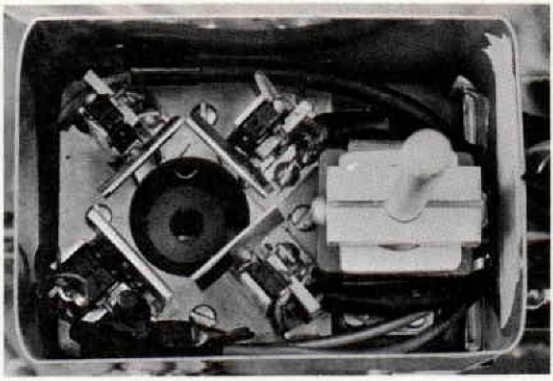

FIGURE 20—Double pole, double throw sequential microswitches for orthotic arm control (joy stick removed).

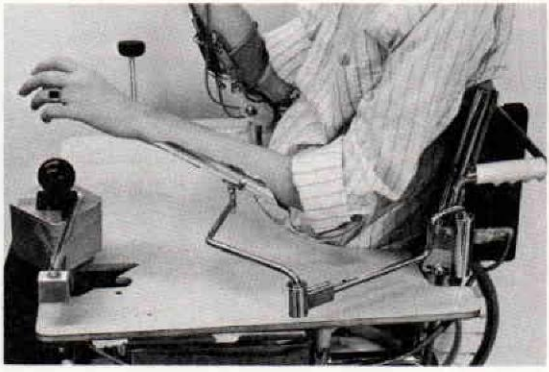

FIGURE 21—Balanced forearm orthosis on contralateral side permits ease of joy stick control.

through a spring-loaded pulley located medial to the elbow (Figure 18).

4. Combined humeral flexion-abduction and humeral extension-adduction through a parallel linkage, lateral to the humerus (Figure 19).

Horizontal adduction and abduction are not motorized since the type of patient requiring an electric arm orthosis is likely to have sufficient residual shoulder girdle control to produce such desired motion, once the effects of gravity are eliminated in a properly balanced linkage system. The electric motors are activated through specially designed double-pole, double-throw sequential microswitches (Figure 20). The patient's contralateral arm is supported in a balanced forearm orthosis to permit ease of switch control (Figure 21). This is possible by a shift of the center of gravity, induced by head motion, in combination with residual shoulder and arm motions.

SUMMARY

An attempt was made to delineate the practicability of external power in orthotic applications. In lower extremity orthotics the application of external power is, at the present time, restricted to relatively few, mostly experimental devices. Developments in this area which hold promise are motorized hip joints for paraplegic patients, electrical stimulation to evoke muscle contraction in upper motor neuron disorders, and as an alternate to external power, hydraulic controls to coordinate knee and ankle motions in the above knee brace. In upper extremity orthotics the application of external power depends, to a great extent, on whether the patient is ambulatory or wheelchair-bound and whether he is unilaterally or bilaterally involved. In the ambulatory, unilaterally involved patient, shoulder-driven, body powered and/or manually controlled orthoses have found greater patient acceptance and are of greater practicability in terms of weight, wear, and maintenance than externally powered devices. If, however, the patient is wheelchair-bound or an ambulatory patient with bilateral arm involvement, the indications for external power are definitely within the realm of practicability. It provides, in most cases, the only means of obtaining useful hand and arm functions.

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The PTS Knee Brace

by

Robert O. Nitschke, C. P., Rochester, N. Y., and Kurt Marschall, C. P., Syracuse, N. Y.

Early in 1966 we initiated fitting patients with the patellar tendon supra-condylar prosthesis. It was immediately evident that the amount of flexion and extension of the stump during locomotion could be controlled through the high brimline and the intricate fit of the socket superior to the patella where the three vastus muscles and the rectus femoris unite to form the quadriceps tendon. By coincidence at this time, a female patient, suffering from extreme recurvatum of the right knee as a result of polio fifteen years ago, asked to be fitted with a long leg brace. She complained bitterly that the braces she wore before were cumbersome to apply to the extremity, excessively heavy, and most unsatisfactory of all, unsightly cosmetically with their bulky uprights, bands, and sharp protruding joints. Immediately our curiosity was aroused as to whether the PTS principles were adaptable in this particular case. After securing the consent of the prescribing physician, we fabricated a light plastic knee brace, consisting of two half shells overlapping each other at their long axis, medially and laterally, and secured by four Velcro straps. This procedure, which has been duplicated since in many other cases, proved to be highly acceptable to this female patient cosmetically as well as functionally in preventing hyperextension of the knee with its accompanying pain.

THE CAST-TAKING PROCEDURE

Two women's nylon stockings and one cotton stockinette are applied to the patient's leg and secured with an elastic waist belt

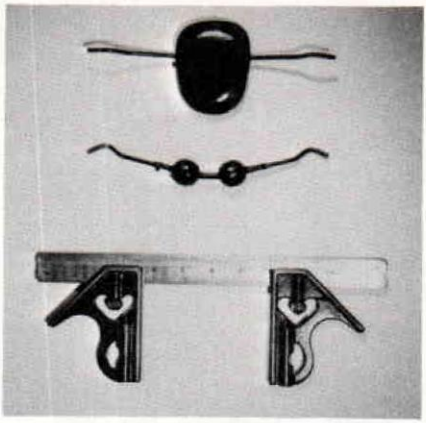


FIGURE 1—A.-P. tension clamp and M.-L. measuring device.

and clamps to prevent slippage. The following areas are then identified with indelible pencil: patella, level of tibial plateau, tibial tubercle, tibial crest, head of fibula, and the medial and lateral hamstrings. One M-L dimension is recorded at the level of the tibial plateau; a second one immediately above the femoral condyles. The proper A-P dimension is taken from mid-patellar tendon at a right angle to the popliteal area. For these measurements we use a special square consisting of one crossbar and two movable shanks (Fig. 1, Bottom). Three circumference measurements are taken for later reference on the positive mold starting with the first one just below the tibial flare. For casting, we use elastic plaster bandages only, beginning the wrap distally to the knee joint, continuing proximally to about two inches above the patella, applying only moderate tension to the bandage. Immediately above the patella, the bandage is twisted once 180° in such a fashion as to create an indentation and firm pressure at the quadriceps tendon. Care should be taken at this point not to displace the patella downward. The wrap is then completed distally, well past the bulge of the gastrocnemius, and the A-P tension clamp is applied before the plaster hardens. In our opinion, the use of the A-P tension clamp (Fig. 1, Top), which can easily be fabricated in any facility, or the use of a Coleman tension clamp, is an absolute must to assure even compression and to free your hands for proper molding of the cast, particularly in the area above the femoral condyles. At this stage careful attention should be given to the amount of flexion desired, ranging from 15° to 25° in most cases. After the plaster has hardened, the A-P tension clamp is removed, and the cast is reinforced with regular plaster of Paris bandages. With a cast cutter, cut along the medial and lateral hamstrings to a point of ½" above the level of the tibial plateau. Then fold back the posterior flap to allow knee flexion. It is at this early stage that it is determinable whether the brace will be effective for the patient by using the cast on a trial basis. After that the cast can be removed by cutting along the long axis of the extremity, medially and laterally. The resulting two halves should be joined together again by use of an additional bandage. The negative cast is then filled in the conventional manner.

CAST MODIFICATION

After removal of the negative wrap, the three circumference measurements of the positive mold are compared with the measurements taken previously. If the patient's subcutaneous tissue is average, these measurements should be reduced by ¼". The A-P and M-L dimensions of the positive mold should coincide with the recorded measurements. If they prove to be larger, plaster should be removed without destroying the overall contour. Care should be taken not to remove any plaster from the bony protuberances. The crest of the tibia is built up 1/16" to ⅛" in its entire length. A generous flare is created at the popliteal aspect. The entire cast is then smoothed and sealed.

FABRICATION PROCEDURE

The layup for the anterior shell is as follows: Two layers of nylon stockinette are pulled over the whole cast with 6 layers of fiberglass in between. This fiberglass reinforcement should cover the entire length of the half shell but should be staggered in width from ½" to 3½" so as to provide the greatest strength along the tibial crest. For the lamination, a mixture of 60% Laminac 4110 and 40% Laminac 4134 is used. After this lamination has cured sufficiently, the outer PVA bag is kept in place and the layup for the posterior shell is begun. Since the posterior shell overlaps the anterior one by about 1" on both sides, it should be somewhat stronger. Use, therefore, 3 layers of nylon stockinette and 8 layers of fiberglass. Again the fiberglass is staggered in width from 2" to 4". After this lamination has cured

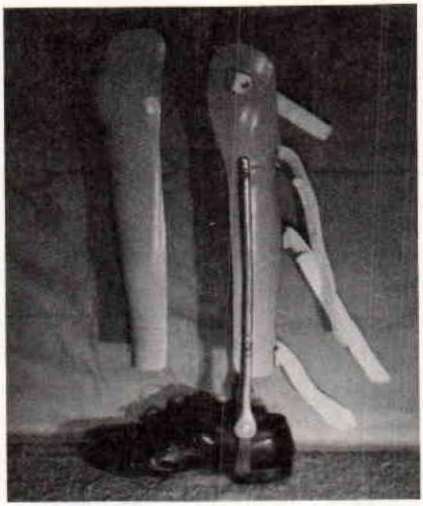


FIGURE 2—PTS Brace. anterior and posterior shell from a medial view.

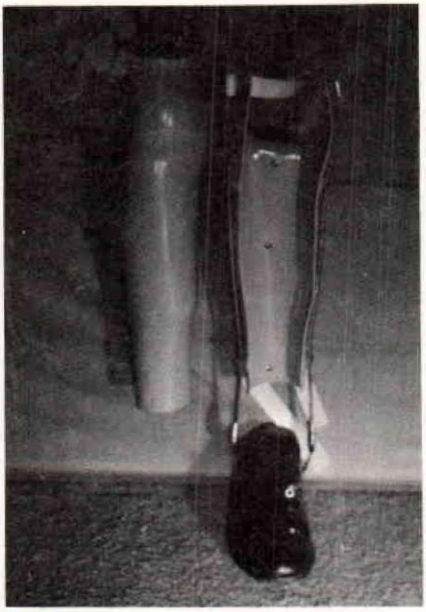


FIGURE 3—Anterior and posterior shell From an anterior view.

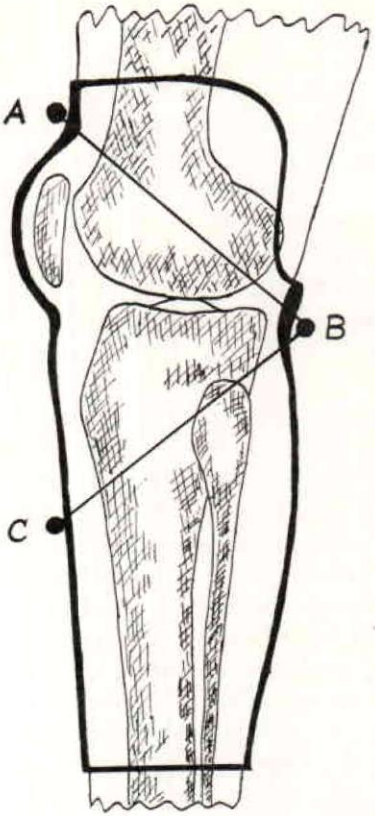


FIGURE 4—Pressure Distribution in the PTS knee brace.

sufficiently, the two halves can be removed from the master mold. In cases where the attachment of a short leg brace is necessary, the fiberglass reinforcement should cover more than half of the posterior shell (Fig. 2 and 3).

As previously mentioned, the two halves are trimmed so that they overlap each other by 1". The anterior superior brimline is kept purposely high in the beginning about 1½" above the patella. It is only lowered when it is established that no undue pressure is placed upon the quadriceps tendon. The posterior brimline is cut as outlined by the mold. The distal brimline should run well below the bulge of the gastrocnemius. Three or four Velcro straps are usually enough to keep the desired tension between the two shells. It should be emphasized that the PTS brace should not be worn directly against the skin. A nylon or cotton underhose, cotton stockinette, or wool sock, should be worn underneath. In cases where the PTS brace is not attached to a short leg brace, it must be suspended by a light waist belt.

SUMMARY

The PTS knee brace encloses the medial and lateral femoral condyles in their entirety, thus assuring maximum stability. The desired degree of flexion is controllable by the intricate fit superior

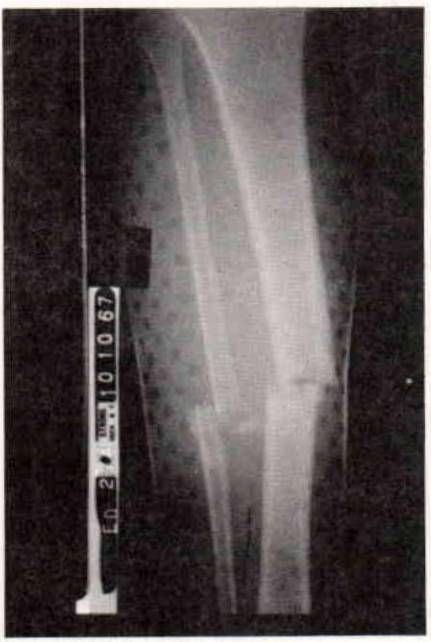


FIGURE 5—Fracture of tibia and fibula on 62-year-old male patient.

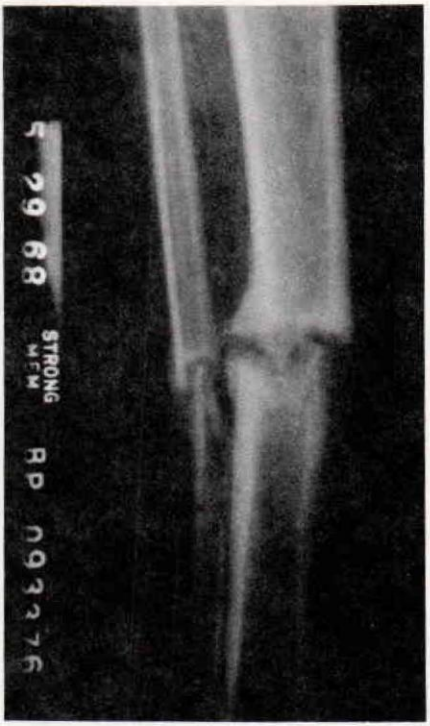
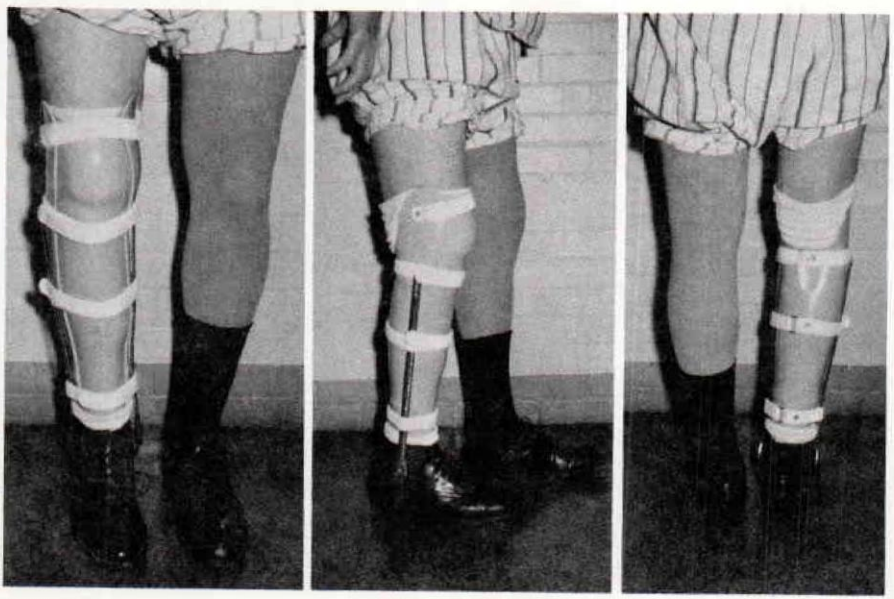


FIGURE 6—Same fracture after 7½ month.

to the patella at the beginning of the quadriceps tendon. To maintain the desired degree of flexion, three areas of pressure and counter-pressure are of the utmost importance: 1. Quadriceps tendon superior to the patella; 2. The entire area of the tibial crest; 3. The popliteal area (Fig. 4).

The PTS brace is fabricated in two plastic shells partially overlapping each other and secured with three to four Velcro straps. Since it is fitted intimately to the long structure of the extremity, its cosmetic appearance is superior to metal braces. If necessary, the brace can be easily connected to a short leg brace without increasing the weight appreciably.

Since 1966, the PTS knee brace, adapted from the PTS prosthesis, has been fitted in eight cases, each



FIGURES 7, 8, 9 PTS brace attached to short leg brace from anterior lateral and posterior view.

involving one of the following: fractures of the tibia and fibula immediately below the tibial flare; non-union or pseudarthrosis of the tibia; medio-lateral instability of the knee; and in cases of extreme recurvatum of the knee.

Figures 5 and 6 show you a 62-year-old male with fractures of the right tibia and fibula about 8″ below the tibial plateau. He remained in a cast from October, 1967 until May, 1968. He was then placed in a PTS knee brace with short leg brace attachments. He is now ambulating under full weight bearing without auxiliary aids, and X-rays show a marked increase in callus formation. (Fig. 7, 8, 9)

X rays and photographs courtesy of Dr. William Boger, Strong Memorial Hospital, Rochester, N. Y.

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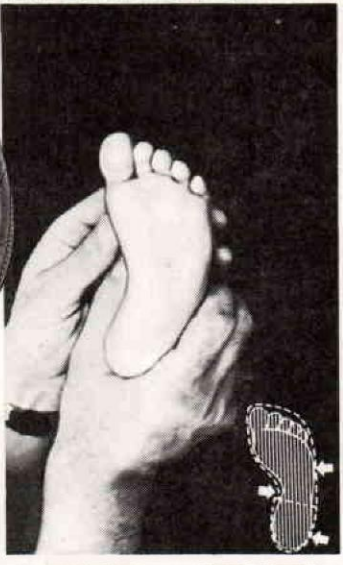

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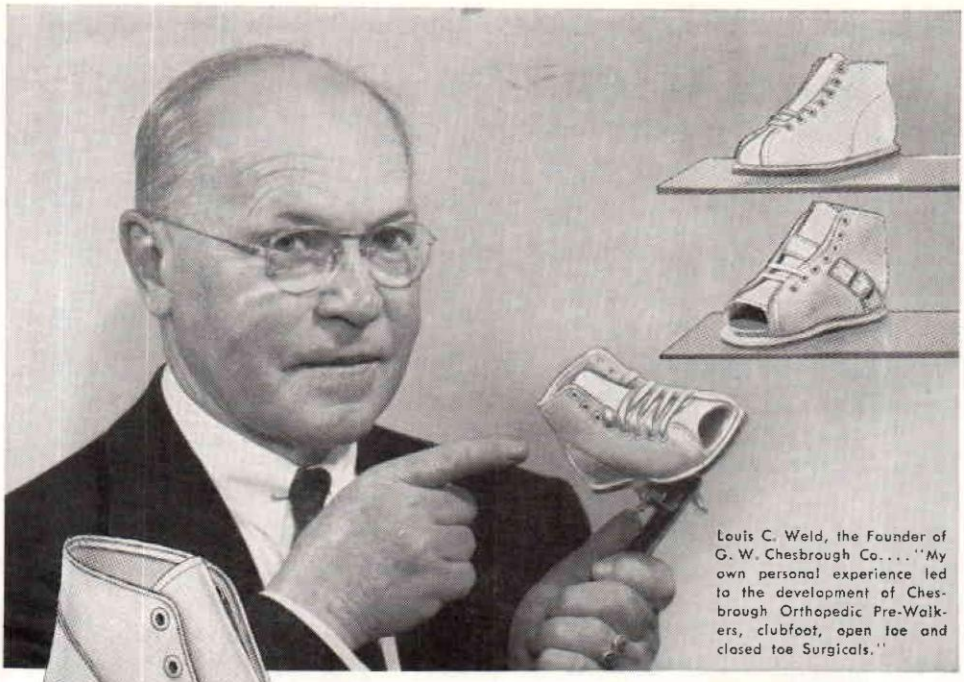

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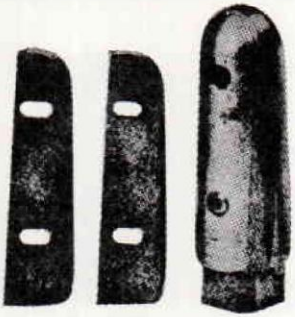

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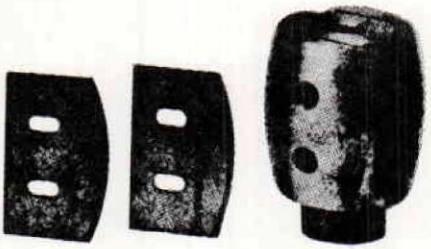

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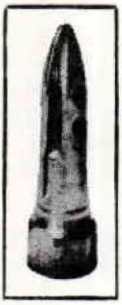

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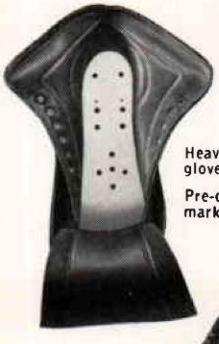

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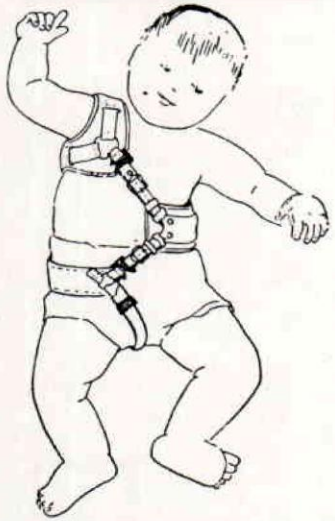

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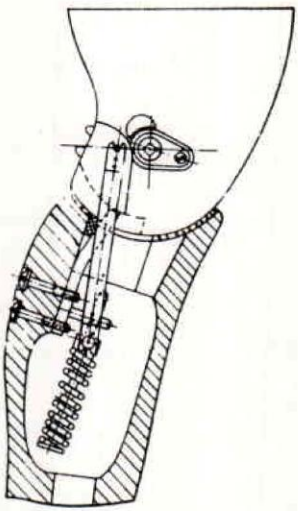
Model 319 new single axis knee with cosmetic features



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The Assembly is open to all who are interested in the rehabilitation of the orthopedically disabled

1968 NATIONAL ASSEMBLY PROGRAM

WEDNESDAY, SEPTEMBER 25

9:30 a.m. AOPA Board of Directors Meeting
Presiding, Alvin L. Muilenburg, C.P.O.

1:00 p.m. ABC Educational Standards Committee Meeting
Presiding, Bert R. Titus, C.P.O.

3:00 p.m. AOPA Education Committee Meeting
Presiding, Robert E. Fannin, C.O.

THURSDAY, SEPTEMBER 26

9:30 a.m. General Registration
Voters' Registration
Women's Auxiliary Registration
ABC Board of Directors Meeting
Presiding, Paul E. Leimkuehler, C.P.
North American Subcommittee, ICPO
Presiding, Anthony Staros

1:30 p.m. Joint Executive Council Meeting
Presiding, M.P. Cestaro

6:30 p.m. AOPA-ABC Reception-Buffet

FRIDAY, SEPTEMBER 27

8:30 a.m. The President's Breakfast
Toastmaster, Roy Snelson, C.O.

9:30 a.m. Formal Opening of Assembly
Presiding, Alvin L. Muilenburg, C.P.O.

10:15 a.m. Opening of Exhibits
Presiding, John R. Hendrickson

11:00 a.m. "Hey There! Almost a Doctor!"
Moderator,
Joseph M. Cestaro, C.P., J. E. Hanger, Inc.
Vert Mooney, M.D. Chief, Amputation Fracture Service, Rancho Los Amigos Hospital, Downey, California
Joseph E. Traub, C.P., Social and Rehabilitation Services, U.S. Department of Health, Education and Welfare, Washington, D.C.

11:00 a.m. Women's Auxiliary Meeting and Tour
Presiding, Mrs. John A. Metzger

1:15 p.m. Exhibitors' Presentation of New Products
Presiding, John R. Hendrickson

3:00 p.m. Amputee Gait Training for Prosthetists
Moderator,
Bert R. Titus, C.P.O., Department of Orthotics and Prosthetics, Duke University Medical Center, Durham, North Carolina
Bernard R. Strohm, M.A., Director, Prosthetics-Orthotics, University of California at Los Angeles

7:30 p.m. The Polysar Socket
Moderator,
Anthony Staros—Director, Veterans Administration Prosthetic Center, New York City

SATURDAY, SEPTEMBER 28

8:30 a.m. The Delgado College Bench Technician Program

Moderator,

George H. Lambert, Sr., C.P.O. Snell's Limbs and Braces, Baton Rouge, La.

Jimmie C. Steedley, M.A.

9:30 a.m. The Diabetic Amputee

Moderator,

Clinton L. Compere, M.D., Paul R. Meyer, M.D.

11:30 a.m. The Insurance Program

Moderator,

H. B. Warburton, Executive Director, AOPA—ABC, Washington, D.C.

Corie Hale, Jr., Washington Representative, Rathmell & Company, Washington, D.C.

Richard Erlandson, Assistant Group Manager, John Hancock Mutual Life Insurance Co., Washington, D.C.

11:15 a.m. Women's Auxiliary Luncheon

Presiding, Mrs. John A. Metzger

1:00 p.m. Orthotic Management of Low Back Pain

Moderator,

Arthur W. Guilford, Jr., C.O., Chief Orthotist, Rancho Los Amigos Hospital, Downey, California

Frank Raney, M.D.

Walter Koniuk, C.O.

1:30 p.m. University Council on Orthotic-Prosthetic Education Business Meeting

2:15 p.m. Report of Committee on Prosthetic Research and Development, National Research Council

Moderator,

A. Bennett Wilson, Jr., Executive Director, CPRD, Washington, D.C.

Charles W. Radcliffe

Colin A. McLaurin

Anthony Staros, Director, Veterans Administration Prosthetic Center, New York City.

Joseph E. Traub, Social and Rehabilitation Services, U.S. Department of Health, Education and Welfare, Washington, D.C.

4:15 p.m. Conference of Orthotists

Presiding, William L. Bartels, C.O., Wm. L. Bartels Inc., Portland, Oregon

SUNDAY, SEPTEMBER 29

9:00 a.m. Presentation by The University Council on Orthotic-Prosthetic Education

Participants,

New York University, Northwestern University, Chicago City College, Cerritos College, University of California at Los Angeles

12:15 p.m. ABC Certification Luncheon

Presiding, Paul E. Leimkuehler, C.P., Paul E. Leimkuehler, Inc., Cleveland Ohio

Guest Speaker, Dr. Carl S. Winters, sponsored by General Motors Corp.

2:00 p.m. ABC Annual Meeting

Presiding, Paul E. Leimkuehler, C.P.

3:15 p.m. The Supracondylar Tibial Prosthesis

Moderator,

Samuel E. Hamontree, C.P., Frees & Tyo Inc., Syracuse, New York.

Carlton Fillauer, Fillauer Surgical Supplies, Chattanooga, Tenn.

4:00 p.m. Conference of Prosthetists

Presiding, Basil Peters, Prosthetics and Orthotics, New York University Post-graduate Medical School, New York City

ABC Board of Directors Meeting

Presiding, Paul E. Leimkuehler, C.P.

6:30 p.m. Buffet and Dance

MONDAY, SEPTEMBER 30

9:00 a.m. Orthotic Management of Fractures

Moderator,

M. J. Benjamin, C.O., Los Angeles, California.

Vert Mooney, M.D., Chief, Amputation Fracture Service, Rancho Los Amigos Hospital, Downey, California.

Augusto Sarmiento, M.D., Professor of Orthopaedics, School of Medicine, University of Miami, Miami, Florida

George Irons, C.O.

William F. Sinclair, C.P., Jackson Memorial Hospital, Miami, Florida.

10:45 a.m. The Child Amputee

Moderator,

Robert C. Gruman, C.P., The Winkley Company, Minneapolis, Minn.

R. Jones, M.D.

Chester C. Nelson, C.P.

Daniel G. Rowe, C.O.

1:30 p.m. AOPA Annual Meeting

Presiding, Alvin L. Muilenburg

6:30 p.m. Concluding Reception and Banquet

TUESDAY, OCTOBER 1

9:00 a.m. AOPA Board of Directors Meeting

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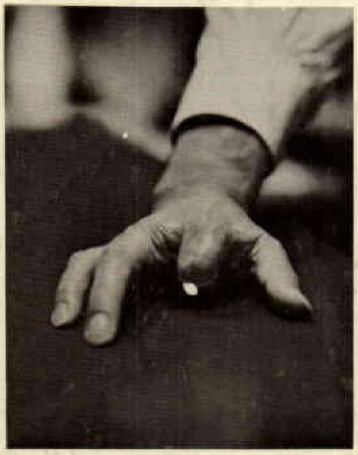

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