SEPTEMBER 1976

# Orthotics and Prosthetics

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# Orthotics and Prosthetics



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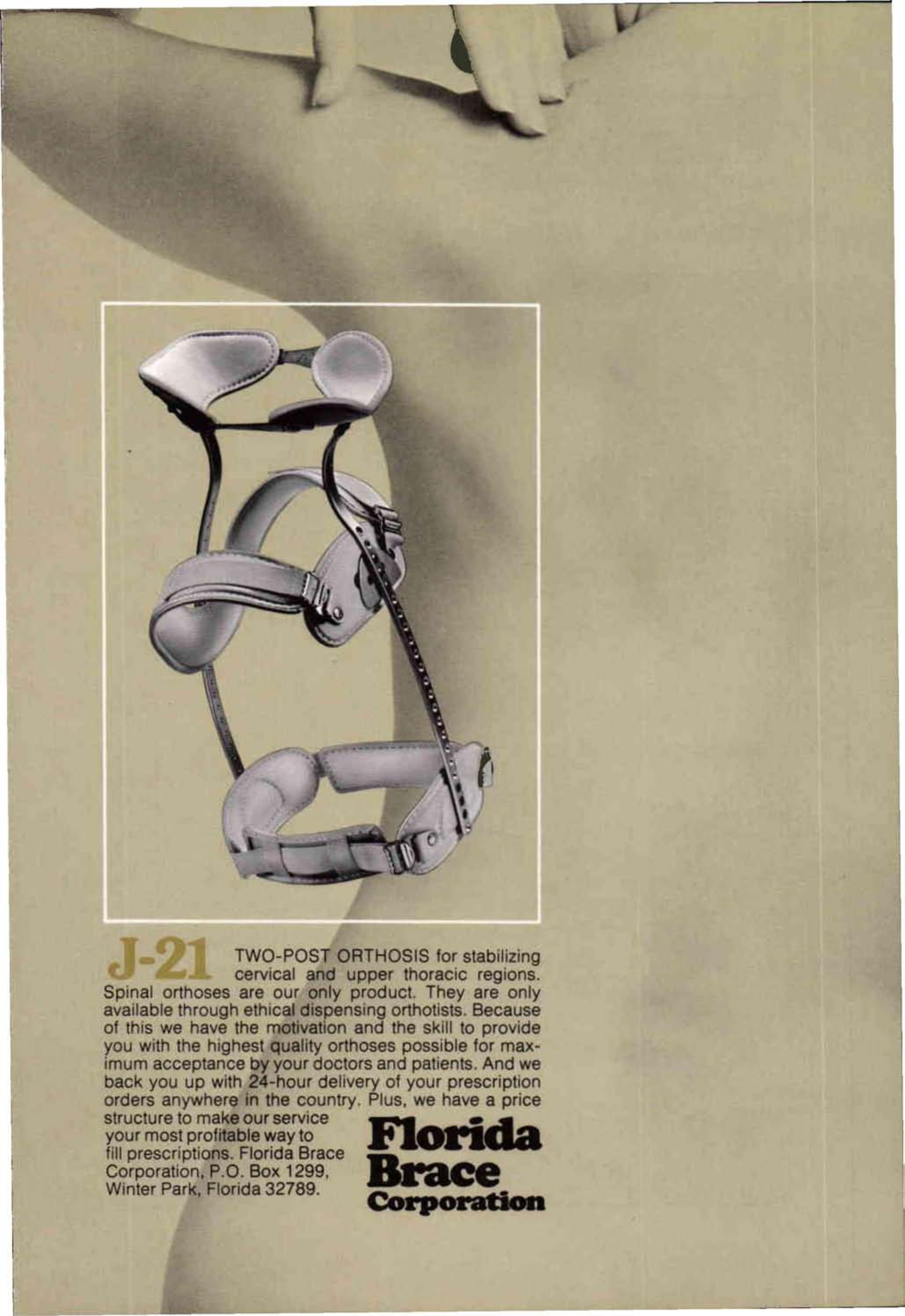
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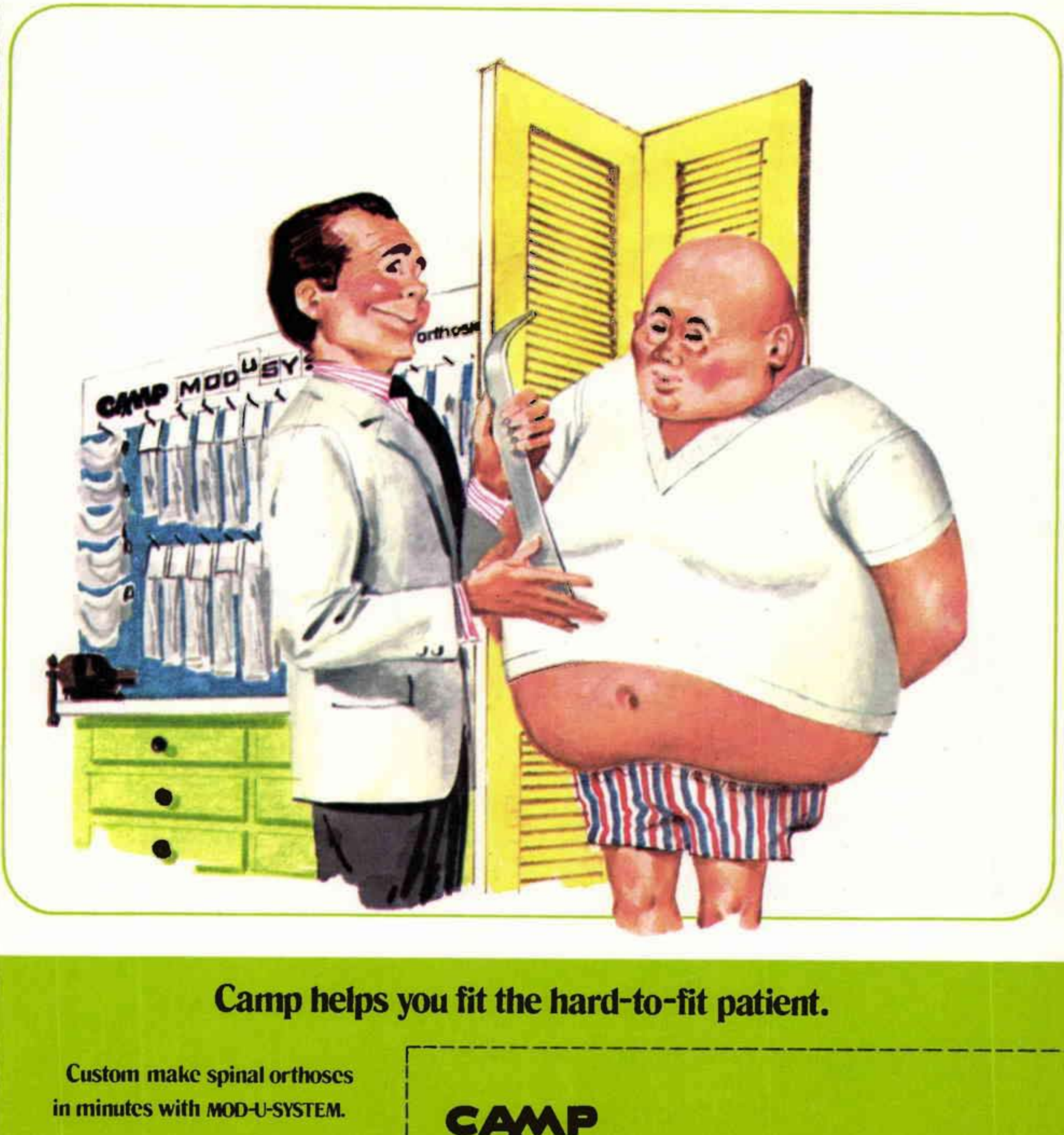


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# PROSTHETICS AND ORTHOTICS EDUCATION TODAY

As we find ourselves on the eve of the combined INTERBOR and AOPA Congress scheduled for Hollywood, Florida, I cannot help but wonder what would be the state of our profession today in the United States if government support had not been made available for research, development, and education over the years since World War II. I think you would all agree that a slightly different picture might exist. My remarks are in no way meant to minimize the effort and struggle put forth by prosthetics and orthotics practitioners in their continuing efforts to elevate our professional standing in the medical community. However, without the assistance of many federal, state, and private institutions, the degree of success we have achieved today would more than likely not have been possible.

In the June, 1976 issue of "Orthotics and Prosthetics", I read with interest the remarks made by Mr. André Bahler, President of INTERBOR, in his comparison of the professional training methods in Europe and the United States. I quote from his article, "whereas in Europe, the educational framework of orthopedic technology is still based largely on manual skills, and further education is necessary if one is to keep up with the developments". In the March, 1976 issue of "Orthotics and Prosthetics", one can also read that Mr. Howard Thranhardt, President of American Orthotic and Prosthetic Association, is troubled by both the present state of our basic education programs, as well as the future prospects for their continuation.

Well, just what are the problems being faced with our education system in the United States today?

Each year, for the past several years, our universities involved in teaching prosthetic and orthotic education have been facing a serious financial problem. The situation continues to worsen to the point that, at the present time, little or no financial aid is available. This year the situation was clearly recognized as a serious one by most prosthetists and orthotists when they reviewed the small offering of courses on the university schedules.

Many of you may ask why it is that the universities cannot conduct these courses without financial support and still show a fair return for their effort. Well, it appears that they can in a strictly lecture-type of course, but as soon as the prosthetic-orthotic laboratory and the patients needed for instruction are introduced, the expenses become quite high and must be dealt with either by a government subsidy or increased tuition and laboratory fees. The latter solution has already caused serious concern from both firms and individual students enrolling for the courses being offered this year.

Although I am personally against government subsidies in the operation of private industry, I can appreciate that all our medical educational institutions in the United States today could never have reached their current height of development and maintain their level of excellence without some direct federal government support. This, I understand, holds true for most medical colleges in the United States today. Our profession is very similar, and I'm afraid we must seek support if we are to maintain our basic educational schools at an acceptable proficiency level in the coming years. There is no doubt in my mind, that the American Academy of Orthotists and Prosthetists can continue to fill the need for continuing education amongst our certified practitioners. However, I do not foresee in the immediate future that we will be in a position to conduct basic university type of courses in prosthetics and orthotics.

This year, the American Orthotic and Prosthetic Association has clearly recognized the problems faced by our entire educational system and is actively seeking ways of making our position known to Congress and government agencies. From this individual's standpoint, as one who has been on the Washington scene for a number of years, I can assure all practitioners that the amount of support we are seeking would be considered infinitesimal when compared with other federally supported programs.

The support is here in Washington, it is just a matter of our people and organizations making our position known to our representatives in Congress. We have fortunately not been required to take this type of action in prior years because we have always had other organizations and institutions intervening in our behalf. However, times change and it is now imperative that we let the government know what our problems are and seek their assistance.

The American Academy of Orthotists and Prosthetists is extremely concerned about this situation from an educational standpoint and we recently notified the American Orthotic and Prosthetic Association that we stand behind them in their efforts to improve this declining situation with our universities. Although the "Academy" is primarily concerned with continuing education for practitioners, we recognize that without the basic educational programs, we would have serious difficulty in even maintaining the current number of certified practitioners in the United States at the current level.

What can you do as a private practitioner to assist with this serious and very real problem we are facing today? I personally hope that all Academy members, as well as other practitioners, assistants, and facility owners will write their representatives in Congress and explain to them the situation being faced in our field by the curtailment of Health, Education and Welfare funds to the universities. If all of us would take this action, I feel sure that the efforts being put forth by AOPA would pay dividends in a very short period of time.

J. M. Cestaro, C.P.O.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 3–8, September 1976

# AN ADJUSTABLE PLASTIC THIGH SECTION AND KNEE JOINTS FOR FEMORAL-FRACTURE ORTHOSES

Vert Mooney, M.D.[1], George Irons, C.P.O.[1]
Michael Quigley, C.P.O.[1]

The terms "fracture orthosis" or "cast brace" are used to describe a device that provides circumferential support to a segment of a fractured limb while allowing mobility of nearby joints, and thus early functional ambulation.

Fracture orthoses were first mentioned in the literature in 1855, in Henry H. Smith's article entitled "On the Treatment of Ununited Fractures by Means of Artificial Limbs" (*8*). Delbert recommended the use of a spring loaded fracture orthosis around 1900, but the use of these devices for fracture treatment was not considered seriously until 1961, when Dehne (*2*) introduced the concept of ambulatory treatment for tibial fractures. The work of Sarmiento (*5, 6*) and Brown (*1*) was reported a few years later, and by 1970 the concept was extended to fracture treatment of the distal femur (*4, 7*).

In the 1970 article (*4*) we stated, "It is essential, we believe, that the cast brace be a total contact device which is applied as far proximal as possible on the thigh and is suspended so that there is no opportunity for distal displacement." Total contact is desirable in order to effectively enclose the tissues to prevent them from expanding when axial loading is applied to the bone and to offer additional stability by allowing the muscles to work against the firm enclosure.

The timing of ambulatory care for femoral fractures is a medical judgment and varies widely, based on the experience and attitude of the clinician relative to the purpose and function of ambulatory care. Many factors, such as the degree of bone comminution, shortening which occurs out of traction, inherent stability of the fracture site, and other associated injuries which limit ambulation, interweave so that no specific rule can be outlined as to timing of application. The fracture orthosis may even be used as an auxillary support system for fractures which already have internal fixation by either rod or plate but for various reasons are not sufficiently stabilized by the fixation.

There were various problems associated with application of the previously described technique (Fig. 1) (*4, 7*). Total contact was lost when thigh atrophy occurred, often necessitating a cast change. Proper placement of the metal knee joints and shaping of the joint uprights caused problems when an orthotist was not available. Another disadvantage arose in the use of a quadrilateral shape at the proximal cast, requiring a separate molded plastic brim, a forming device, or hand forming the mold. Usually, two hours or more were required to apply the fracture orthosis. The method described below attempts to alleviate these problems.

## COMPONENTS

The critical component is a pre-molded, quadrilaterally shaped thigh support (Fig. 2). Through experience it has been determined that three sizes (small, medium, large) will fit nearly all patients. Eighty percent of all patients can be fitted with the medium size. Because the brim is molded according to contours at the root of the leg, either a right or left thigh section must be used. The plastic chosen for this portion is polypropylene because of its well demonstrated toughness and resistance to tearing. Multiple perforations in the plastic are necessary, however, to allow the skin to "breathe." The tongue is of polyethelene which is slightly more pliable than the polypropylene. Three elastic-backed web belt straps are used. They are riveted to the posterior portion of the thigh section, but are fully adjustable to any position.

The knee joints are the flexible polyethylene joints originally designed at Tripler Army Hospital to substitute for the more expensive, and

[1]Amputee and Fracture Service, Rancho Los Amigos Hospital, 7601 East Imperial Highway, Downey, California 90242

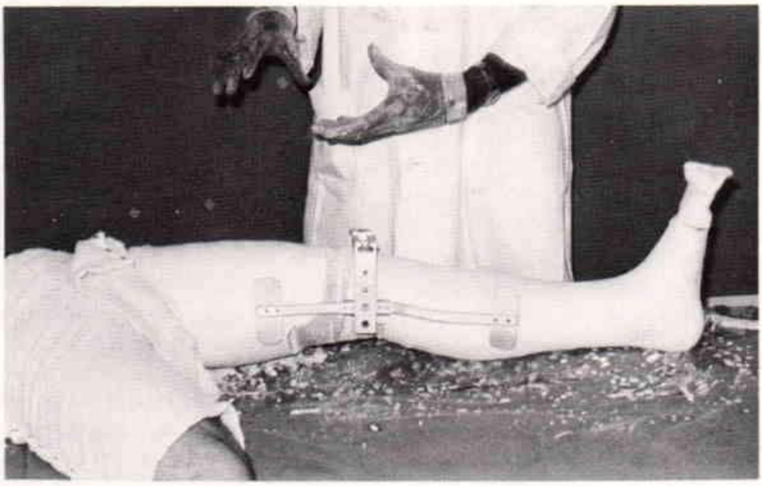

Fig. 1. The previously used technique (7) that required special alignment tools, bending irons, hose clamps, proximal brims, and a thigh cast.

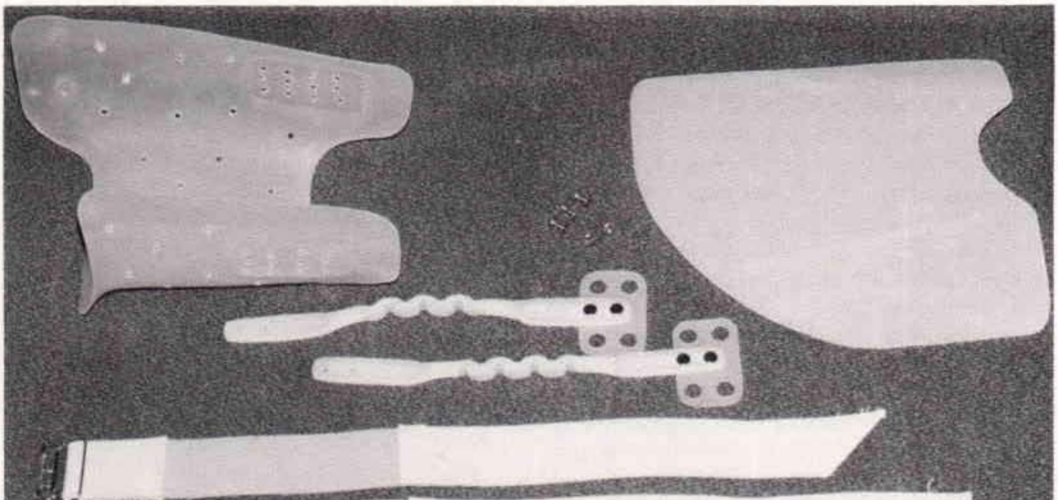

Fig. 2. Components unassembled. From top to bottom, quadrilateral thigh section with adjustment holes, polyethylene tongue, polyethelene knee joints, elastic-backed web straps. Components are delivered assembled.

often unavailable, metal polycentric hinges. For this particular application they are ideally suited because the multiaxis motion of these joints makes alignment far less critical. Furthermore, since motion is not biased in any direction, the joints will not bind if they are not parallel as is the case when the thigh portion is tightened to maintain total contact. The joints are fixed to the plastic thigh portion with screws. Considerable adjustability is available in terms of proximal-distal length as well as anterior and posterior positioning so that proper location in the coronal mid-line can be achieved.

Additional components are not critical and are quite variable. A Spandex cast sock is ideal for under both the plastic and plaster portions. In that no tissue compression need be achieved by the initial dressing, tube stockinet may serve just as well. Standard plaster-of-Paris bandage is used for the lower leg portion. There is no advantage in use of elastic plaster-of-Paris bandage here. When particularly hard wear or uncontrollable moisture is expected, the use of a plastic laminate is appropriate as an alternative for plaster-of-Paris.

In summary, the new components have the following advantages:

1. Total contact is always available because the straps and flexible brim can be tightened when atrophy occurs, thereby saving cast changes.

2. There is no need to use a separate brim which must be incorporated in the cast.

3. The orthosis is lighter than the earlier type, can tolerate draining wounds, and makes wound dressing easier.

4. No alignment tool is needed. Plastic joints do not need to be square when applied, and will not bind as they lose parallel orientation when the thigh portion is tightened.

5. No shaping of joints is needed. Flexibility of the plastic joints allows them to fit flush on the cast without use of bending irons.

6. Varus and valgus adjustments can be made. Extra holes on the thigh section where the joints attach allow angular and A-P adjustments simply by removal of two or four screws.

## THE TECHNIQUE

The following materials are needed:

4 each 4-in. standard plaster bandages

6 each 3-in. standard plaster splints

1 each Spandex cast sock or tube stockinet

1 each Molded plastic thigh section[2] with polyethelene hinges.[3]

These are available in three sizes: small, 7- to 11-in. length, 16- to 19-in. circumference: medium, 8- to 12-in. length, 19- to 23-in. circumference; and large, 9- to 13-in. length, 23- to 28-in. circumference. Length is measured from the adductor tubercle to a point one inch distal to the perineum; circumference is measured at the level one inch distal to the perineum.

Because minimal special equipment is necessary for the application of the fracture brace, the work can be accomplished in the emergency room, cast room, or even the patient's own room. The traction pin can be removed if the fracture has reached skeletal stability. If there is the expectation the fracture may need additional traction, the traction pin in the tibia should be left intact and incorporated in the plaster-of-Paris cast.

The stockinette or Spandex cast sock is rolled on the leg as high into the groin as possible. If drainage is present, bandages should be on the outside of the stockinette so that they may be changed as necessary (Fig. 3).

The plastic thigh portion is placed as proximal as possible up the leg so that the buttock fold matches the posterior portion of the folded thigh section (Fig. 4). The straps are tightened and the alignment of the plastic joints is confirmed to be parallel to the sagittal plane and slightly posterior to the mid-coronal plane.

For the lower leg and foot, Webril is applied over the cast sock or stockinet. Inclusion of the foot is not necessary if the traction pin remains intact since the pin will keep the system from sliding distally. An initial layer of 4-in. plaster-of-Paris bandage is applied. Several thicknesses of 3-in. plaster splints are overlapped around the distal portion of the plastic joints and these in turn are incorporated with additional rolls of

[2]U.S. Manufacturing Company, 623 S. Central Avenue, Glendale, California 91209; Orthomedics, Incorporated, 8332 Iowa St., Downey, California 90242

[3]Ultra High Molecular Weight (UHMW) Polyethelene

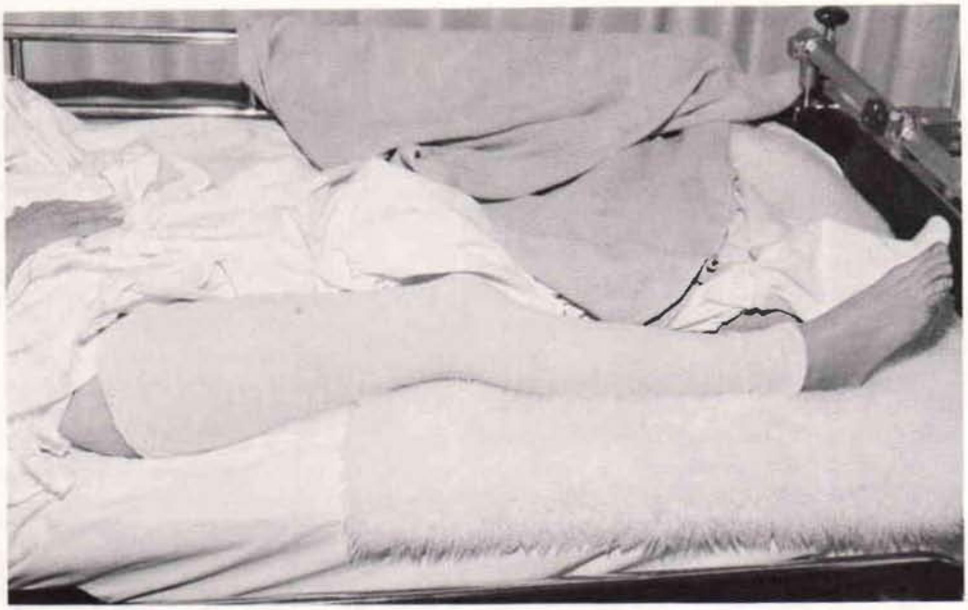

Fig. 3. Stockinet or a Spandex cast sock is rolled on the leg. Traction pins were removed in this instance but should be left intact if further traction is expected.

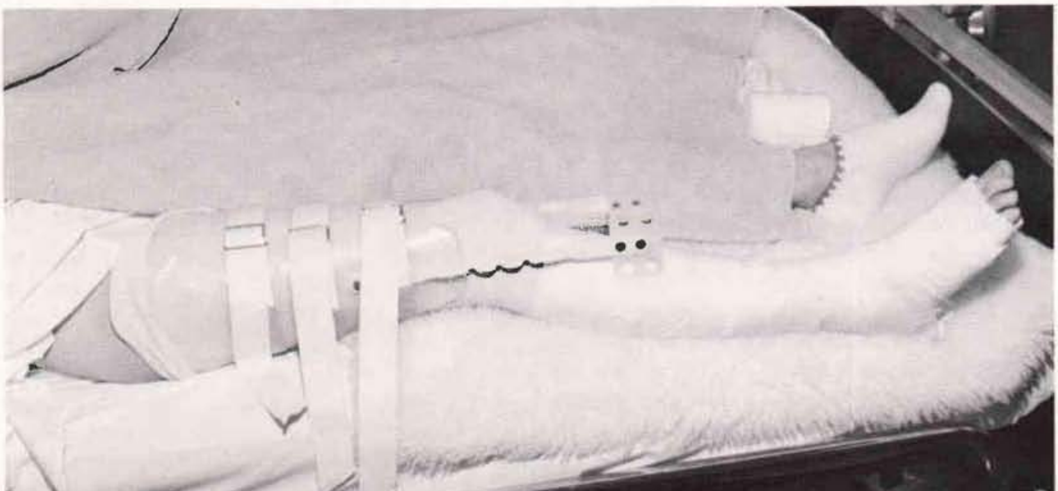

Fig. 4. Plastic thigh section is located properly on the thigh and the straps are tightened.

plaster-of-Paris bandage into a complete total contact plaster (Fig. 5). The foot should be held in neutral position both from the standpoint of plantar and dorsiflexion as well as varus and valgus so that weight-bearing to the limb is with the foot in its physiological position.

Excessive belt material is cut off. The patient is instructed in tightening and loosening the thigh section, and lines are drawn along the buckles with a ball point pen to identify the tightness. In time, these can be tightened as atrophy occurs.

Fig. 5. Plaster is applied to the leg with the foot maintained in a neutral position. Once the initial wrap is set, the attachment plates of the joints are wrapped in the cast.

The patient can start to ambulate on the fracture orthosis the following day, once the plaster has had sufficient time to harden (Fig. 6). When the patient is recumbent at bed rest the belts may be loosened to allow the skin to "breathe" and dressings to be changed as necessary. Whenever the patient uses the limb to stand or become ambulatory, however, the belts should be tightened to the previous degree of tension.

## EXPERIENCE

This method of adjustable fracture bracing has been used for over fifty patients at Rancho Los Amigos and Martin Luther King Hospitals (*3*). It has been applied both as a suspension method while the patient is in traction and as a definitive fracture orthosis when the patient becomes ambulatory following the discontinuation of traction care. In five patients it has been used as an auxillary support system when internal fixation was felt to be inadequate or fracture healing lethargic. When the fracture has become nearly completely healed, the lower portion of the orthosis can be removed, leaving the plastic thigh section to give additional support during the final fracture maturation.

Fig. 6. The patient is instructed to keep the straps tight whenever he uses the limb, and is allowed to ambulate the following day.

## SUMMARY

A simplified method of fracture orthosis care has been described. It is lightweight and adjustable, and can be applied with less specialized equipment and knowledge than with previous designs.

## ACKNOWLEDGEMENT

Mr. Hank Meyer, Engineer and Plant Superintendent, U.S. Manufacturing Company, was of great help during the development of the components by searching out new materials and providing prototypes of the knee joints.

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Orthotics and Prosthetics, Vol. 30, No. 3, pp. 9-11, September 1976

# A PROSTHESIS FOR FOOT AMPUTATION NEAR THE TARSAL-METATARSAL JUNCTION

Karl Fillauer, C.P.O.[1]

The human foot is one of the most elegant examples of physiologic engineering. The skeletal framework and the muscular forces acting upon it are so arranged that they are held in delicate balance throughout standing, walking, and running. When an amputation is carried out through the foot, this balance is lost. As we realize the complexity of the human foot and the tasks required of it in standing and walking, we begin to appreciate the crippling effect caused by the loss of any weight-bearing segment.

A common result from an amputation near the tarsal-metatarsal junction is either an equinovalgus or equinovarus deformity that presents an almost impossible condition for satisfactory prosthetic fitting and normal gait.

## PURPOSE AND GOALS

A prosthesis can serve as a functional replacement for the lost portion of the foot. It can also prevent deformities and provide cosmetic restoration.

The primary goal is distribution of weight-bearing loads to areas of the remaining portion of the foot in such a manner that the blood flow is unimpaired. Another goal is to prevent spreading of the remaining portion of the foot; that is, to hold the foot in proper anatomical alignment while free motion is allowed at the ankle along with subtalar motion of inversion and eversion. In addition, adequate retention of the foot in the socket is very important.

Currently most prostheses consist of an ankle joint and corset, a molded socket with a flexible felt or rubber toe section. While this design does satisfy some of our goals, the fabrication time is considerable and the result is a bulky and ill-appearing prosthesis.

## A NEW DESIGN

Over the past several years, we have been using a prosthesis that meets the functional requirements given above and one that is cosmetically acceptable.

This prosthesis is designed similar to a UC-BL shoe insert except for the trim lines which are more proximal. The socket is an acrylic lamination using banlon and fiberglass cloth reinforcements on sole, sides and heel areas. The toe piece is SACH-foot heel-cushion material[2] (soft for children, medium for youths and firm for adults). The posterior portion of the socket extends proximally to the retrocalcaneal bursa, and anteriorly to the most proximal portion of the instep around the mid-malleolar level. The posterior midline is parted to a point on the plantar surface. This is done for easy donning of the prosthesis. The socket and toe section are bonded together with epoxy resin.

## FABRICATION PROCEDURE

Because this prosthesis fits snugly in the shoe, the plantar interface must match the interface of the shoe sole. To duplicate this, a plaster wrap is taken while the patient is standing on a casting form[3].

- Before wrapping the plaster, two layers of tubegauze are placed over the foot and 1/4-in. rubber tubing is inserted along the anterior portion of the foot to facilitate removal of the cast.
- An elastic plaster-of-Paris wrap is applied.
- The foot is placed on the casting form (Fig. 1). Particular attention is given to leg-foot alignment. The ankle should be in a neutral position; not in plantar or dorsiflexion. With the thumb and long finger, the retrocalcaneal dimples are palpated

[1]Manager, Patient Services Division, Fillauer Orthopedic, Chatanooga, Tennessee.

[2]Kingsley Manufacturing Co., Placentia Ave., Costa Mesa, California.

[3]Foot Casting Boards (Available from Fillauer Orthopedic).

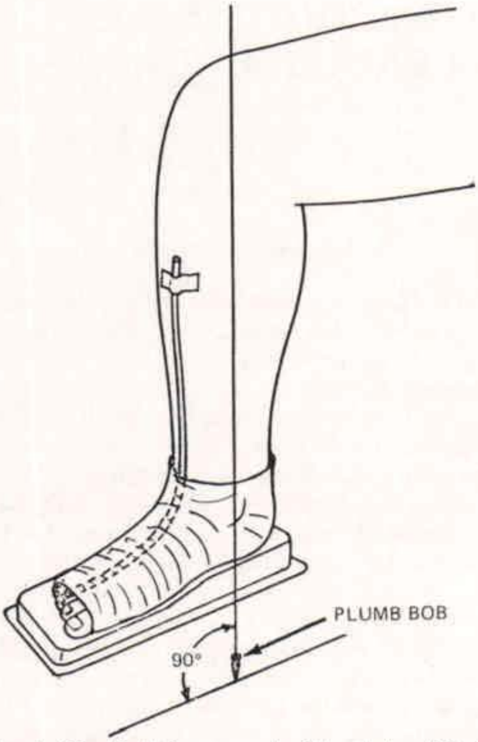


Fig. 1. The foot is wrapped with plaster-of-Paris bandage and placed on the proper casting form.

with moderate compression. With the other hand the forefoot is held firmly on the casting form.

· After the plaster has set and alignment marks have been drawn across the parasagittal cut line, the wrap is removed by cutting along a line over the tubing.

· The negative mold is poured. The cast modifications consist of plaster build-ups over bony areas and mild removal of plaster over the retrocalcaneal dimples which serve as the suspension mechanism (Fig. 2).

· A PeLite[4] distal pad is formed over the cast to provide for growth and relief when necessary. In several cases we have used a PeLite pad on the plantar surface to provide relief from excessive pressure.

· The cast is now ready for lamination with 4 layers of tricot, 2 layers of fiberglass cloth, sandwiched between the tricot layers covering the plantar surface and up the posterior aspect. A 70% rigid, 30% flexible acrylic resin laminate with vacuum is recommended.

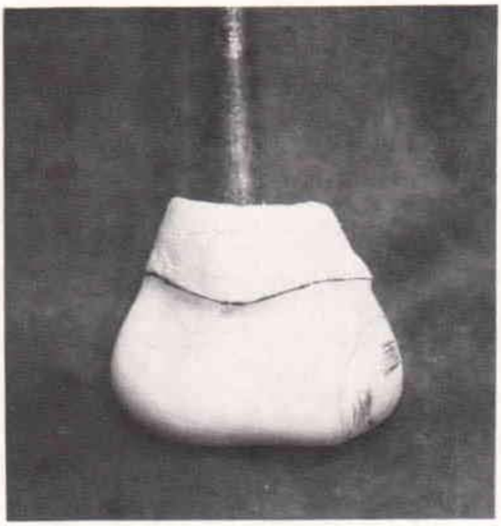

Fig. 2. The positive model.

· The socket is removed from the model. The proximal trim lines are left outsized. The posterior midline is parted to a point just on the plantar surface to make possible easy removal and donning of the prosthesis (Fig. 3).

· The profile of the toe section is outlined on a Kingsley SACH-Foot foam block, cut out, and shaped as needed (Fig. 3).

· The socket and shaped toe section are bonded together with epoxy resin (Fig. 4a). The two sections are placed in the shoe to insure proper alignment (Fig. 4b).

· Mannequin spray is used to finish the prosthesis, before fitting trials (Fig. 5).

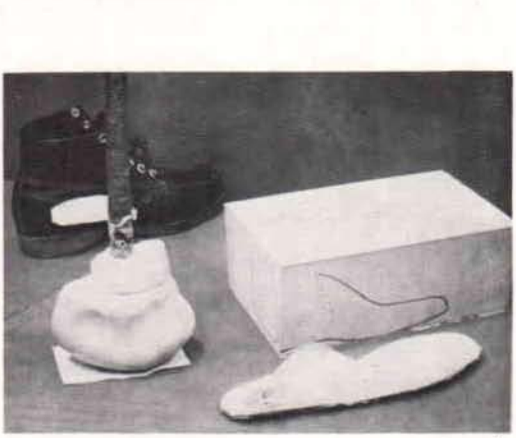

Fig. 3. The profile of the toe section is outlined on the SACH-Foot foam block.

[4]A polyethylene foam distributed by Fillauer Orthopedic.

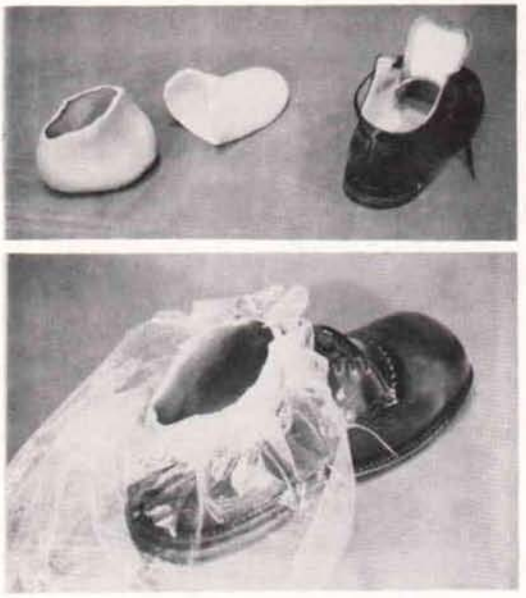

Fig. 4. Two photographs showing relationship of the components of the prosthesis.

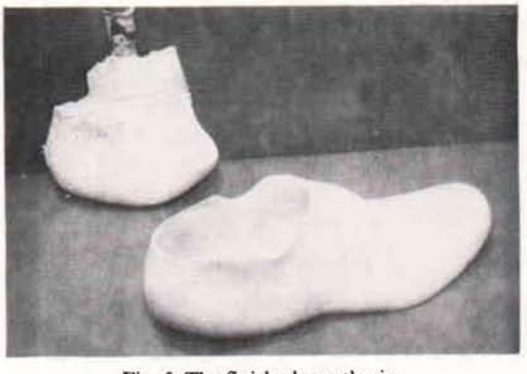

Fig. 5. The finished prosthesis.

## FITTING DETAILS

The patient should wear the type of sock he generally uses. In the case of children, a heavy wool sock is desirable because thinner socks can be used to prolong the life of the prosthesis as the child grows. In adults a heavy cast sock or a three-ply cotton sock is preferred.

Entry of the foot into the socket is facilitated when the posterior opening can be spread apart. When donning is difficult, the anterior trim line can be lowered to accommodate the foot more easily.

Socket retention is accomplished by tightness over the heel section when the prosthesis is in the shoe. Therefore, well built shoes having rigid counters should be worn.

When acrylic resin is used, trim line pressure areas may be relieved by either grinding or by post forming under heat.

## CONCLUSION

Each patient will undoubtedly present a new problem and require special precautions. Good weight-bearing areas as well as tender points that need protection should be noted. Under no circumstances are scar tissues or grafted skin areas to be subjected to pressure or friction.

This type of prosthesis should encourage active dorsiflexion, and thus avoid, or at least delay, heel cord lengthening procedures while providing an acceptable prosthesis, both cosmetically and functionally.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 13-16, September 1976

# CLINICAL EXPERIENCE WITH THE "SOLID-ANKLE" ORTHOSIS[1]

Melvin Stills, C.O.[2]

The conventional method of bracing the child with myelodysplasia at the L3-4 level seems to be by use of metal and leather in the form of hip-knee-ankle-foot orthoses (HKAFO) as illustrated in Figure 1, in spite of the availability of polypropylene "solid-ankle" orthoses described by Glancy in the March 1972 issue of *Orthotics and Prosthetics* (*2*).

Fig. 1. Conventional hip-knee-ankle-foot orthoses provided for children with myelodysplasia at the L3-4 level.

The intent of this paper is to describe the experiences gained in utilizing Glancy's technique at the Krusen Center for Research and Engineering.

Patients with paraplegia at the L3-4 level have intact hip flexors and knee extensors, but the muscles below the knee are flaccid. When HKAFO's are used the unaffected musculature is not permitted to control function when the mechanical knee and hip joints are kept in the locked position. Functional activity while wearing the conventional orthosis is reduced, not only because of the rigidity of the total system, but also because of the additional burden caused by the weight of the orthosis.

Another problem seen in the myelodysplastic child is deformity of the bones of the lower limb. All children seen in the myelodysplasia clinic during the past year had worn conventional orthoses, and had deformities varying between slight and extreme (Fig. 2 & 3). It seems to be self evident that conventional orthoses do little to prevent deformity, and that alternate means of bracing, such as the molded plastic techniques need to be investigated with respect to their potential ability to control and prevent deformity.

In his paper Glancy (*1*) discusses the rationale for the "solid-ankle" orthosis, and goes on to describe completely the procedures necessary for fabrication and fitting.

The only change made at this center, except to ignore the need for shoe modification, was to use vacuum-forming techniques, rather than hand layups.

## FABRICATION AND FITTING

The casts were taken and positive models made according to Glancy (*1*). Polyethylene foam was

[1]This work was carried out with partial support of the Rehabilitation Services Administration, Department of Health, Education and Welfare under Grant #23P-55518

[2]Director of Orthotics, Krusen Center for Research, Moss Rehabilitation Hospital, 12th Street and Tabor Road, Philadelphia, Pennsylvania 19141

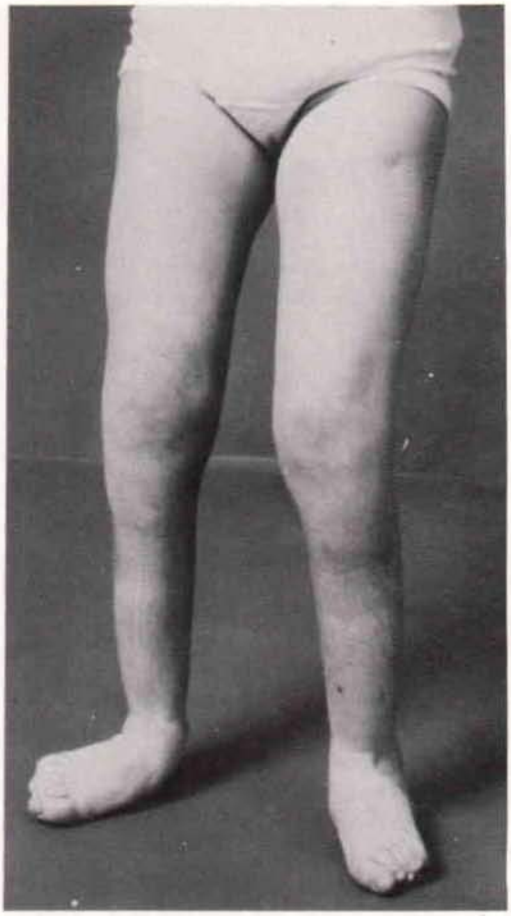

Fig. 2. Typical condition of patient with myelodysplasia at the L3-4 level.

formed over the anterior section of each positive model, and both models then were placed on the platen of the vacuum-forming machine so that the orthoses can be molded simultaneously (Fig. 4)

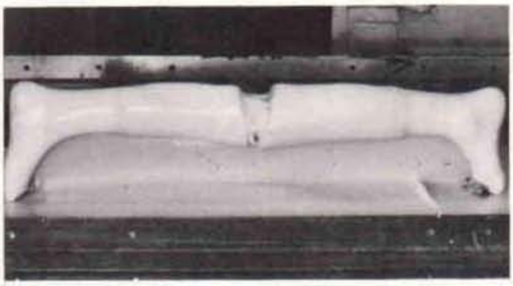

Fig. 4. Right and left orthoses molded simultaneously.

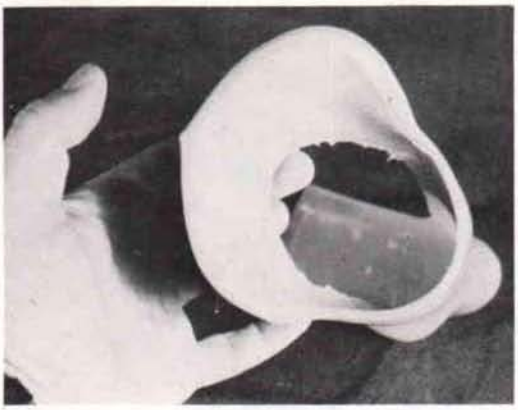

Fig. 5. Top view of "solid-ankle" orthosis.

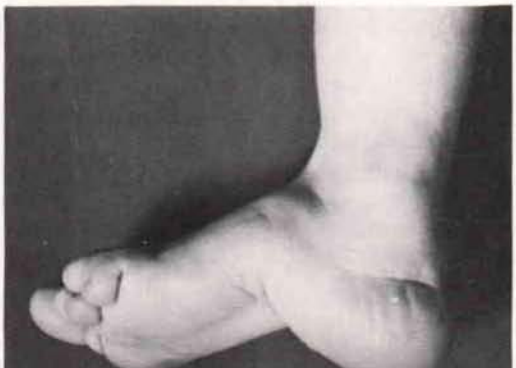

Fig. 3. Severe foot deformity seen in patient with myelodysplasia at the L3-4 level.

(*3*). Polypropylene, 1/4-in. thick, was used so that after the forming process the thickness of the anterior wall of a finished orthosis is at least 3/32-in. and the thickness of the posterior section is approximately 1/8-in. In vacuum forming, a weld is produced vertically in the anterior section as the heated material on each side is brought together. Breakage in this area has occurred when adequate thickness was not maintained. When breakage does occur, Kydex 1/16-in. thick can be used to provide reinforcement to the anterior section.

Shoe modifications have not been found to be necessary. The patients have been fitted with ordinary shoes as illustrated in Figures 7 and 8. Ankle position is maintained and a knee extension moment is provided as well.

Fig. 6. Anterior view of a pair of "solid-ankle" orthoses.

Fig. 7. Typical myelodysplastic patient with "solid-ankle" orthoses.

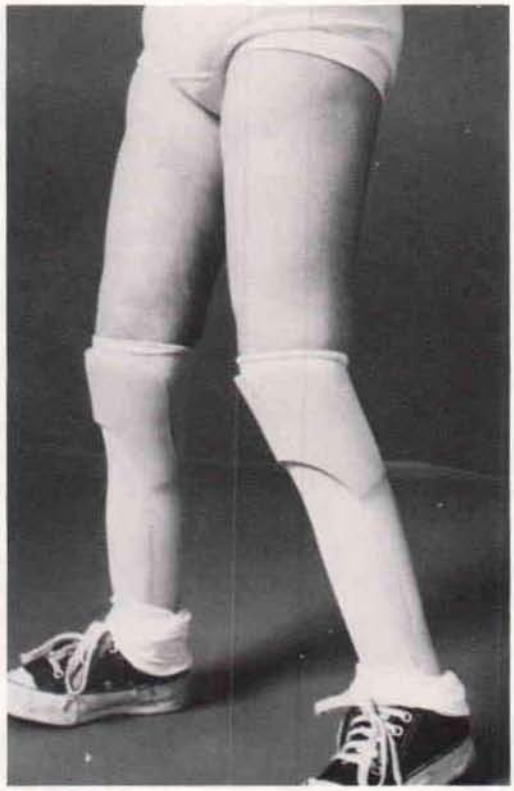

Fig. 8. Typical myelodysplastic patient with "solid-ankle" orthoses.

## SUMMARY OF EXPERIENCE AND DISCUSSION

When marked knee instability was present in the medio-lateral plane, children provided with the "solid-ankle" orthosis presented a number of orthotic problems that have not been overcome completely. The techniques described by Foort in reference to the CARS-UBC knee orthosis (*1,4*) are being applied to the "solid-ankle" orthosis in an attempt to control M-L motion better without restricting knee flexion and extension.

Probably the most serious problem confronted while fitting the "solid-ankle" orthosis has been skin breakdown. In most cases of myelodysplasia there is loss of sensation, and condition of the skin must be followed closely. Careful instruction must be given to the child and to the parent to check skin condition frequently during the period immediately following initial fitting and at least three times daily as long as a molded orthosis is used.

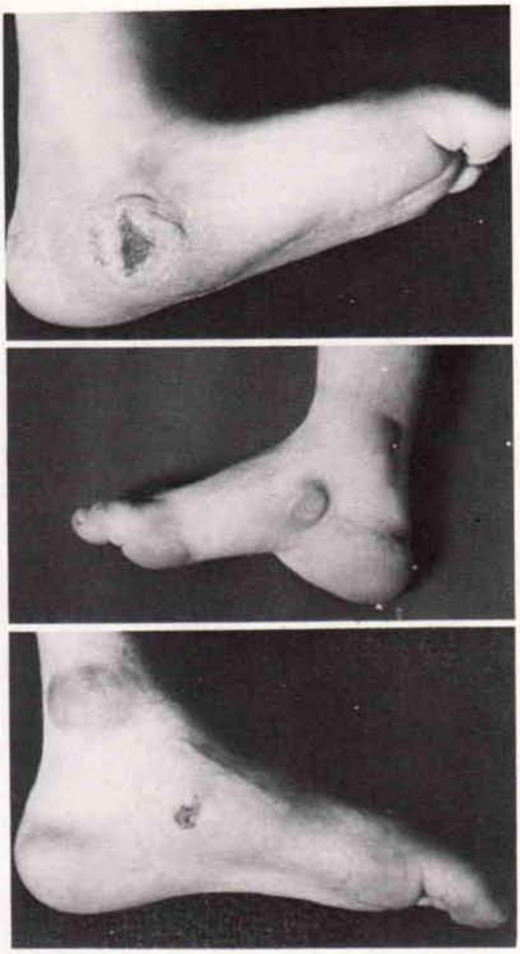

Fig. 9. Lesions caused by inadequate fit and lack of attention on the part of the patient and parents.

Skin breakdown that could have been avoided easily if simple instructions had been followed are shown in Figure 9. Fortunately, simple modification of the orthosis and good medical management permitted these pressure areas to clear up without severe complications.

To date 21 children with myelodysplasia have been fitted with "solid-ankle" orthoses, and a total of 47 "solid-ankle" orthoses as described by Glancy have been fitted to 24 children. Our experience to date has been mostly very satisfactory, failure as a result of use of the orthoses having been observed in only a small portion of the cases treated.

This work has been carried out with the assistance and cooperation of the staff of the A. I. duPont Institute, Wilmington, Delaware.

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Orthotics and Prosthetics, Vol. 30, No. 3, pp. 17–24, September 1976

# GAIT OF UNILATERAL BELOW-KNEE AMPUTEES

James Breakey, M.Sc.[1]

The below-knee amputee has lost part of his locomotive system, not only the static supporting structure, but dynamic function of the foot-ankle complex as well. Lower-limb prostheses easily provide static structural supports, but not dynamic functions that correspond to muscle activities that have been lost. Consequently, good locomotion by the BK amputee requires adaptation in the joints of the remaining lower limb and, thus, compensation by the remaining musculature of the lower limbs.

Five active male unilateral BK amputees were used in a study of BK amputee gait (*1*). The subjects wore modular prostheses, with the supracondylar suspension variation of the PTB socket (*2*) and a SACH foot. Electrical switches were used on both shoes to measure foot position during stance phase. On the amputated side, knee motion was measured by an electrogoniometer. Muscle activity was recorded by use of an electromyograph and surface electrodes. The muscles investigated were the gluteus maximus, the quadriceps groups, and the hamstring group on the affected side; the gluteus maximus and vastus lateralis on the unaffected side.

Among the factors influencing the gait pattern of a unilateral BK amputee are the lost foot-ankle function and alteration of the normal knee-ankle mechanism, and the effect of the ground reaction upon the foot of the BK prosthesis.

As a result of the lost foot-ankle function, compensation in the amputee occurs at the knee joint on the amputated limb; in the foot-timing on the normal limb and in the muscle activity in both lower limbs.

## TERMINOLOGY USED IN HUMAN LOCOMOTION STUDIES

A complete gait cycle (Fig. 1) is the period of time from which the heel of one foot contacts the ground to the next heel-contact of the same foot. Two distinct phases occur in this cycle: the stance phase and the swing phase.

Stance phase begins at the instant the heel of one foot contacts the ground and ends when contact with the ground is lost at toe-off of the same foot. Shortly after heel-contact, the sole of the foot makes contact with the floor. This event is referred to as foot-flat. The period between heel-contact and foot-flat is referred to as early stance. Following foot-flat is a period called mid-stance which occurs between foot-flat and when the heel loses floor contact at heel-off. During mid-stance the body weight advances directly over the supporting limb. Following heel-off, a period of forward body propulsion or push-off occurs prior to the foot leaving the ground at toe-off. This period between heel-off and toe-off is referred to as late stance phase.

The sequence of events in stance phase that have just been described will, in this article, be referred to as *foot-timing*.

Swing phase begins at toe-off and continues until the heel contacts the ground. Swing phase is divided into two equal periods, early and late. Three events occur in the swing phase: acceleration, mid-swing, and deceleration. Acceleration begins at toe-off and continues to mid-swing when the swinging leg passes directly beneath the body. Deceleration occurs after mid-swing until the heel of the forward moving limb makes contact with the ground.

During walking, alternation from stance phase to swing phase of each leg results in a period when both feet are in contact with the ground simultaneously. This period is known as double-support or double-stance.

## THE EFFECTS OF LOST ANKLE FUNCTION ON NORMAL KNEE MOTION (Fig. 2)

In normal subjects the knee flexes 15 deg. (*4, 8, 11, 17*) to maintain the center of gravity of the body level as it moves (*15*). Because this amount

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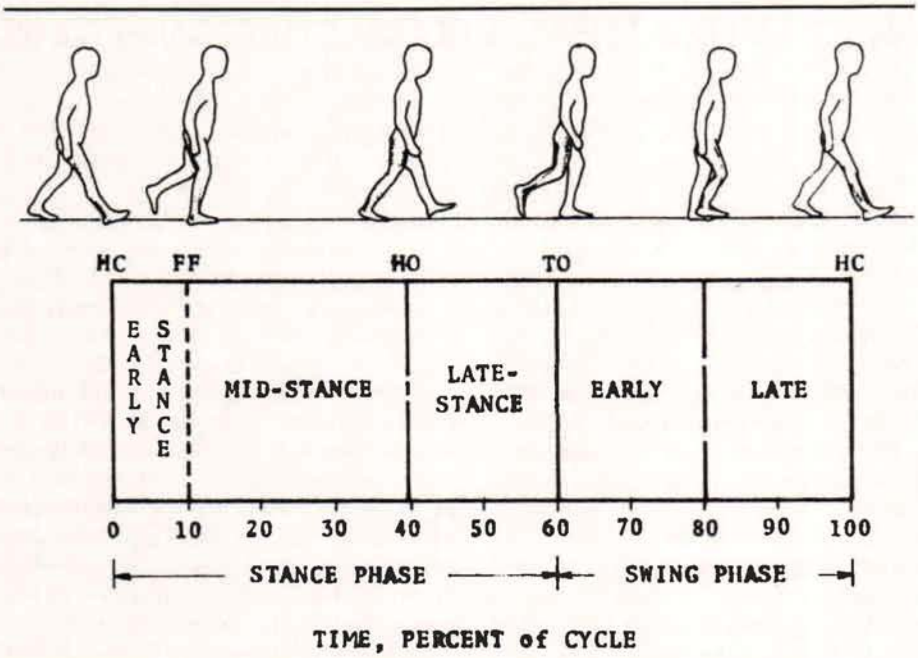


Fig. 1. Components of a gait cycle. HC - Heel-contact; HO - Heel-off; FF - Foot-flat; TO - Toe-off.

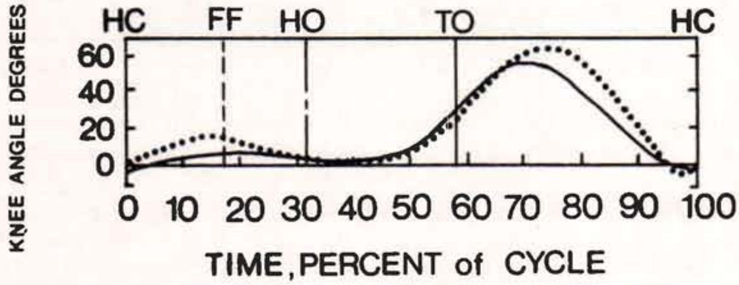


Fig. 2. Knee motion for the prosthetic side. Solid line (______): Knee motion curve for amputee; Dotted line (. . .): Knee motion curve for normals.

of controlled plantar-flexion does not occur in the SACH foot, the amputee must compensate by reducing the amount of knee flexion on the affected side to approximately 7 deg. during stance (*3*). If the amount of knee flexion were not reduced, but the knee were flexed as much as a normal subject (approximately 17 deg.) a noticeable limp would occur during stance phase. The limp would result in the dissipation of more energy because the center of gravity is raised and lowered an excessive amount (*15*).

In late stance phase the ankle begins to plantarflex in the normal subject, and at toe-off has reached an angle of approximately 10 to 15 deg. (*4, 8, 11*) (Fig. 2). The lack of plantar-flexion in the SACH foot at this period of the gait cycle presumably accounts for a mean knee flexion of about 30 deg. occurring at toe-off on the amputated limb (*3*). This value is lower than that found in studies of normals who have shown average knee flexion values of 37 to 43 deg. at toe-off (*4, 8, 11, 17*).

The interaction of a normal knee and ankle during late stance phase assists in lowering the body's center of gravity smoothly with a vertical displacement of about 5 cm. (*7*). If the amputee's knee flexed at toe-off to the same extent as a normal subject, his center of gravity would descend more than the usual 5 cm. and result in a noticeable limp. The amputee's reduced knee flexion at toe-off compensates for his lost ankle plantar-flexion in late stance phase, and thus helps to prevent an exaggerated vertical excursion of his center of gravity.

During swing phase, peak knee flexion on the amputated side was approximately 57 deg. and occurred at 70 percent of the gait cycle (*3*) (Fig. 2). Studies of normal subjects have shown peak knee flexion of a normal knee in swing phase to occur between 70 and 75 percent of the cycle and vary between 61 and 68 deg. (*4, 8, 11, 17*). These studies also showed that the plantar-flexion attitude of the normal ankle, on the average, was from 7 to 10 deg. at the same instant when peak knee flexion occurred. A SACH foot does not attain a plantar-flexion attitude, as does a normal ankle, during knee flexion in swing phase. Thus a lesser degree of knee flexion is sufficient on the prosthetic side to keep the artificial foot clear of the ground during swing phase.

## THE EFFECTS OF LOST ANKLE FUNCTION ON FOOT-TIMING (Fig. 3)

In the study (*3*), the foot-timing for the BK amputees was asymetrical. The stance phase on the affected side occupied 57 percent of the cycle and the swing phase 43 percent, while on the normal limb, stance phase lasted 63 percent and swing phase 37 percent. This disharmony in the duration of stance could be explained as being the result of an early toe-off by the amputated limb owing to loss of the push-off function of ankle plantar-flexion.

In normal subjects the gait pattern between the two limbs is symmetrical. Stance phase has been found to occupy 60 to 62 percent of the cycle and swing phase to occur between 38 and 40 percent (*3, 4, 6, 9*).

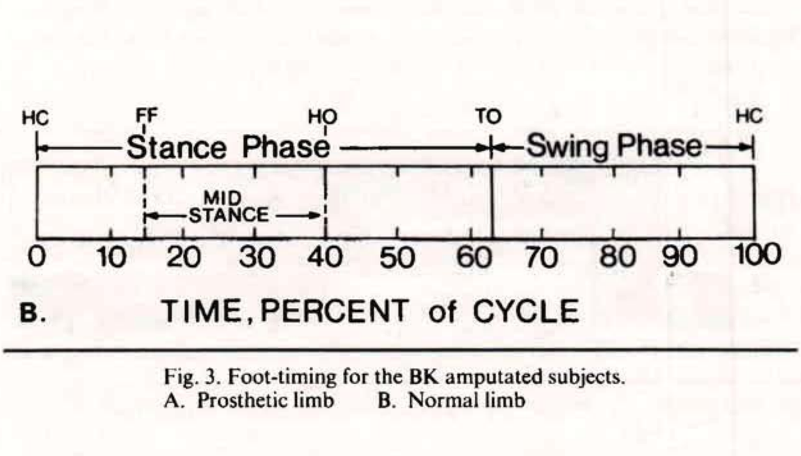


Fig. 3. Foot-timing for the BK amputated subjects.
A. Prosthetic limb B. Normal limb

In the BK amputee, foot-flat occurred on the affected side an average of 17 percent of the gait cycle, and in the contralateral limb an average of 15 percent (*3*). Foot-flat in normal subjects has been reported taking place at approximately 9 percent of a normal walking cycle (*3, 4*). Compression of the heel cusion of the SACH foot seems to be responsible for an elongation of the period between heel-contact and foot-flat in the amputated limb.

Lack of true ankle function in the prosthetic limb affected foot-timing of the normal limb (*3*). According to Elftman (*5*) and Inman (*7*) ankle plantar-flexion provides energy for forward motion of the body. This loss of energy from the prosthetic limb appears to result in a longer duration before foot-flat occurs on the normal limb, which probably assists in movement of the body over the foot. If foot-flat on the contralateral limb were to occur at approximately 9 percent of the gait cycle, as seen in normal subjects, additional muscular effort would be required to move the body forward over the foot, since little assistance can be expected from the prosthetic limb. It appears that postponing foot-flat on the contralateral side until a later period in stance phase (i.e. approximately 15 percent) allows the body to move forward, pivoting on the heel of the normal foot.

Absence of dorsi-flexion in the SACH foot affects the foot-timing on the amputated side resulting in early occurrence of heel-off (*3*). Heel-off occurred in the prosthetic limb at an average of 31 percent of the walking cycle (*3*), compared to 43 percent in normal subjects (*4*). In a normal subject, during midstance, the body moves forward over the foot resulting in approximately 8 to 12 deg. of dorsi-flexion of the ankle (*4, 8, 11*). Perry (*13*) stated that dorsi-flexion of approximately 10 deg. in mid-stance allows the body to move forward more easily over the foot. The earlier occurrence of heel-off in the gait cycle of the amputee would appear to be a result of the lost dorsi-flexion and permit the amputee to move his body forward over the artificial foot.

## DOUBLE SUPPORT PHASE (Fig. 4)

Double support of the below-knee amputee subjects showed an average duration of approximately 20 percent made up of approximately 9 percent occurring at heel-contact of the amputated limb and toe-off of the normal limb and approximately 11 percent occuring when the foot positions were reversed. This increased time period of 2 percent (i.e. 11 to 9 percent) appears to be related to the shortening swing phase duration of the normal limb. Between heel-off of the prosthetic limb and heel-contact of the normal limb, the amputee pivots forward through space on the fore part of the artificial foot. Early heel-contact on the normal limb shortens this tenuous period on the prosthetic limb and increases (i.e. by 2 percent) the duration of double support occurring between heel-contact of the normal limb and toe-off of the amputated limb.

## THE EFFECTS OF LOST ANKLE FUNCTION ON MUSCLE ACTIVITY

As the amputee walks, external forces (ground reaction) act upon the foot of the prosthesis. The artificial foot is thought of as a lever through which the ground reaction acts to produce moments occurring about the knee joint. Influenced by these moments, the amputee's musculature responds by attempting to balance the moments produced about the knee.

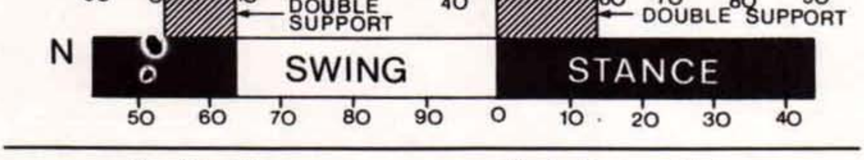


Fig. 4. Double support. Amp - amputated limb N - normal limb

The muscles in the affected limb investigated in the BK study (*3*) (i.e. gluteus maximus, quadriceps group, and hamstring group) were found to be active in early stance phase as has been reported by other investigators in studies with normal subjects (*1, 4, 5, 18*). However, the activity of muscles in the amputee subjects was found to be of longer duration than that found in normal subjects reported in the University of California Study (*18*).

A comparison of the myoelectric activity of the muscles investigated in the affected limb (*3*) and the activity found in the same muscles in the normal subjects in the University of California Study (*18*) referred to above are presented in Figure 5.

The activity recorded from gluteus maximus in the affected limb occurred from heel-contact to approximately 22 percent of the gait cycle (*3*) compared to heel-contact to 12 percent in normal subjects as reported in the University of California Study. Also, in the amputated limb, muscle activity was observed in the quadriceps group (i.e. rectus femoris, vastus medialis, and vastus lateralis) from heel-contact to approximately 27 percent of the cycle (*3*) compared to heel-contact to an average of 18 percent in the University of California Study. In the hamstring group (i.e. medial and lateral hamstrings) of the amputee subjects, muscle activity was present from heel-contact to approximately 25 percent of the cycle (*3*) in contrast to the University of California Study finding of heel-contact to an average of 8 percent of the gait cycle.

The increased durations of muscle activity in early and mid-stance phase found in the amputees (*3*) can be explained as being a compensation for lost ankle function. The calf muscles in normal subjects act from about 10 to 55 percent of the gait cycle (*18*) and are thought to have an important stabilizing effect upon the knee joint in mid-stance phase (*14, 16*). Sutherland (*16*) contends that the restraining action of the ankle plantar-flexors controlling dorsi-flexion in mid-stance allows the extrinsic forces (kinetic forces, gravity, and ground reaction) to extend the knee as the body moves forward over the fixed foot. Paul (*12*) mentions that from 10 to 25 percent of the gait cycle no muscle in the leg is suitably placed to resist flexion of the knee and hip simultaneously. It would appear that the indirect action of the ankle plantar-flexors plays an important role in stabilizing both joints in mid-stance. One

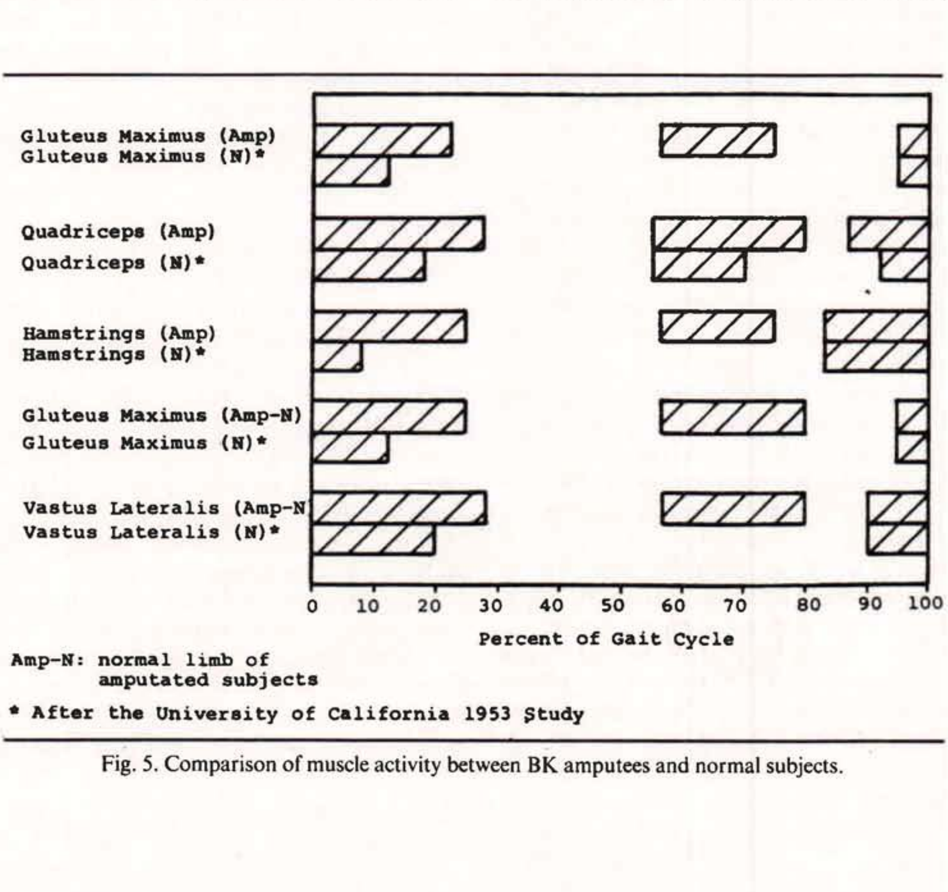


Fig. 5. Comparison of muscle activity between BK amputees and normal subjects.

could speculate that because an ankle plantar-flexion moment in normal subjects occurs from approximately 15 to 60 percent of the gait cycle (*16*) this moment causes a knee extension moment and, in turn, a hip extension moment. Because plantar-flexion action and, therefore, the plantar-flexion moment is lost in the below-knee amputee, the knee and hip would tend to flex in mid-stance if the amputee's muscle activity behaved as in a normal individual. Thus, hip and knee extensors showed a prolonged period of activity, presumably in order to counteract flexion moments at the knee and hip joints.

In the amputee subjects, muscle activity in the knee and hip extensors disappeared in late mid-stance. However, the ankle plantar-flexors in normal subjects cause a knee and hip extension moment. Because the amputated subjects have lost the ankle plantar-flexors, what prevents the knee from buckling during late mid-stance and early late-stance prior to the initiation of knee flexion in late stance? Sutherland (*16*) found in normal subjects that the body weightline falls in front of the knee and ankle joints at some point after 26 percent of the walking cycle. He used this finding to emphasize that gravity alone could not extend the knee as it would also tend to dorsi-flex the foot, which, in turn, would produce a flexion moment about the knee. He therefore hypothesized that the restraining action of the plantar-flexors on ankle dorsi-flexion contributes to knee stability. In the below-knee amputee the loss of ankle dorsi-flexion, making the artificial foot a somewhat rigid lever pivoting about the forefoot, reduces the need for plantar-flexion action in late mid-stance (i.e. 25 to 31 percent) to assist in stabilization of the knee. It is concluded that the effects of gravity acting in front of the knee joint at approximately 26 percent of the cycle (*16*) together with the effects of the ground reaction on the artificial foot act to stabilize the knee joint in late mid-stance and early late-stance phases without the requirements of plantar-flexion muscular activity which seems to be so necessary in normal limbs.

Plantar-flexion of a normal ankle in late-stance phase initiates hip and knee flexion to begin swing phase (*7*). In early swing phase in a normal subject, the swinging leg receives further energy from the hip flexors (*5, 7*). However, a below-knee amputee having lost the foot-ankle push-off mechanism requires increased action of the hip flexors to initiate swing phase. This results in an increased heel rise of the prosthetic limb during early swing phase. It is thought that the activity found in rectus femoris, vastus medialis, and vastus lateralis (*3*) served to control excessive heel rise on the affected side during early swing phase. By comparison, no activity has been reported in the same muscles during early swing phase in studies of normal subjects walking at a normal cadence (*1, 4, 18*). Rectus femoris in the amputated subjects (*3*) showed a longer duration of activity in early swing phase (between 55 to 80 percent of the cycle) as compared to 55 to 70 percent in the normal subject (*18*). This longer period of activity of rectus femoris is probably a result of a damping effect by quadriceps on the heelrise of the prosthesis to counteract the increased activity of the hip flexors as referred to above. Also, activity in rectus femoris may serve to accelerate the leg in swing phase after maximum knee flexion has occurred (i.e. at 70 percent of the cycle).

Myoelectric activity appeared in the gluteus maximus and the hamstrings of the affected limb during early swing phase (*3*). However, no activity has been seen in these muscles in gait studies using normal subjects. The activity in these muscles in the amputee is probably a counterbalance to excessive activity in the hip flexors and quadriceps muscles.

The electromyographic findings in gluteus maximus and vastus lateralis in the amputee's normal limb (*3*) differ from findings in the same two muscles reported in normal subjects studied elsewhere (*1, 4, 10, 18*). In stance phase, activity in gluteus maximus was present from heel-contact to 25 percent of the walking cycle (*3*) as compared to only 12 percent in the University of California study (*18*). Elftman (*5*) stated that an important source of energy in the initiation of forward swing of a limb came from push-off by the calf muscles. He postulated that additional energy is still required for forward swing is provided by the hip extensors of the supporting limb and the hip flexors of the swinging limb. The loss of push-off in the affected limb may account for the longer period of activity observed in gluteus maximus in the normal limb (i.e. supporting limb) while swing phase was occurring in the affected limb.

Vastus lateralis in the amputee's normal limb (*3*) was active in stance phase for an average of 8 percent of the cycle longer compared to normal subjects investigated elsewhere (*18*). This somewhat longer duration of muscle activity may be due to an increased energy requirement for forward motion as a compensation for lost energy input on the amputated limb.

In early swing phase, both gluteus maximus and vastus lateralis in the amputee's normal limb were active (*3*), a condition which has not been reported in normal gait studies. It is quite feasible that a more than normal amount of push-off occurs in late stance phase on the normal limb in order to provide additional impetus to assist forward motion of the body during early stance phase on the prosthetic limb. As push-off initiates hip and knee flexion to begin swing phase (*7*), the activity seen in gluteus maximus and vastus lateralis could serve to counterbalance the increased flexion moments in both the hip and knee joints produced by the increased intensity of push-off.

## SUMMARY

Compensations occur in the gait pattern of the below-knee amputee due to loss of the normal foot-ankle mechanism. The compensations are seen at the knee joint of the affected limb, in the foot-timing on the normal limb, and in the activity of the lower-limb muscles studied in both the affected and contralateral limbs. The loss of a normal foot and ankle results in the alteration of the normal walking pattern as follows:

1. A longer stance phase occurred consistently in the normal limb and a shorter stance phase duration in the amputated limb. Also, the foot-timing during stance phase of the two limbs varied the one from the other.

2. Knee motion in the amputated limb followed the same general pattern as seen in normal knee motion. However, the magnitude of knee flexion in the amputated limb was reduced in stance phase, at toe-off, and at peak knee flexion during swing phase.

3. The phasic myoelectric activity of the muscles studied in both the amputated and non-amputated limbs increased in duration as compared with the same muscles investigated elsewhere in normal (i.e. non-amputated) subjects.

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# A MODIFIED STANDING ORTHOSIS

Ron A. Cartaya, C.O.[1]

The importance of standing in rehabilitation of the paraplegic has long been controversial. Reasons have varied but primarily attaining upright posture has been for both psychological and physiological purposes.

For the young congential paraplegic, or myelomeningocele patient, there is yet another problem. Achievement of standing balance as early as possible by the use of orthotic devices and standing frames helps either to prevent deformities or control existing deformities until the need for surgical intervention can be determined. Use of a standing orthosis tends to eliminate problems associated with non-ambulation such as the development of osteoporosis which may lead to pathological fractures of the lower limbs. Genitourinary problems also are often associated with non-ambulation. Another important reason for standing is "peer group recognition", a very important factor during this critical age of self awareness and identification.

The purpose of this paper is to describe one of the standing frames being used at the University of Miami Medical Center as a pre-Parapodium orthosis (Fig. 1). It was originally designed for the 11- to 20-month old myelomeningocele child in order to provide an orthosis that could be repaired, adjusted, and modified virtually anywhere, keeping in mind that patients who leave areas where sophisticated help is available may not be able to find it in their new location.

These statements are not in any way intended to discredit the use of the OCCC[2] Parapodium which we use with frequency here at the Mailman Center for Child Development. We feel that for standing orientation, the sitting feature is not needed at this early age, and therefore we have designated this design as a pre-Parapodium Orthosis. The primary needs at this time, as we see it, are ease of application, height adjustability, and the hinged knee block section to facilitate donning and doffing. This of course results in lesser cost.

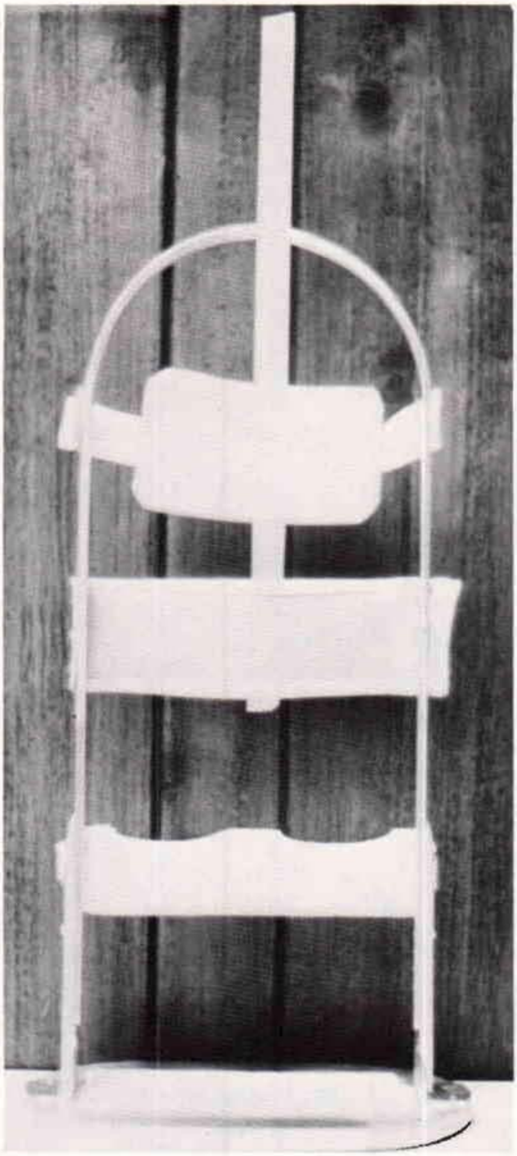

Fig. 1. The pre-Parapodium orthosis.

We urge the use of "cut out" tables where the child can stand and play with their toys. Proper

[1]President, Kendale Orthotic Laboratory, Inc., 7600 S.W. 57th Avenue, South Miami, Florida 33143

[2]Ontario Crippled Children's Centre, Toronto, Canada

placing of the child in an area where he can observe the most activity is important so that he may feel part of the normal daily living situation.

## FABRICATION

### MEASUREMENTS

A silhouette of the child with the legs extended and shoes on is made. Diameter measurements are taken at the knee center, trochanter, and axilla. Height measurements are taken from the shoe baseline to the knee center, trochanter, axilla, and shoulder level.

### MATERIALS REQUIRED

3/16-in X 10-in. X 14-in. 2020-T4 sheet aluminum

3/4-in. X 1/4-in. 2024-T4 aluminum flat bar stock

2-in. X 4-in. aluminum angle

One 1-in. stainless steel hinge

Approximately 20-in. length of 1-1/4-in. X 1/8-in. 2024-T4 flat bar stock

One 2-in. thick Styrofoam block.

Approximately 2-ft. length of Velcro hook and pile

One 1-in. roller buckle without prong

Two 1-in. square loops

Four 1/2-in. dia. self-tapping screws

### PROCEDURES

The pattern for the base plate is the same size as the extra large Parapodium base, approximately 14-in. X 10-in., and is cut from a piece of 3/16-in. thick 2024-T4 aluminum sheet.

The adjustable metal uprights are constructed of 3/4-in. X 1/4-in. 2024-T4 flat bar stock, and shaped to form an inverted "U" (Fig. 1) that, when completed will have a diameter 3-in. greater than the width of the patient at the level of the trochanter. The length is determined by the height from the base plate to a point 1-in. below shoulder level.

Two 4-in. X 2-in. sections of angle aluminum are fastened to the base plate (Figs. 1 and 2). To these sections are fastened uprights made from 2024-T4 bar stock and in which evenly spaced holes have been drilled and tapped with #8-32 threads for adjustment. The distal third of the inverted U has been drilled with clearance holes to match those in the uprights.

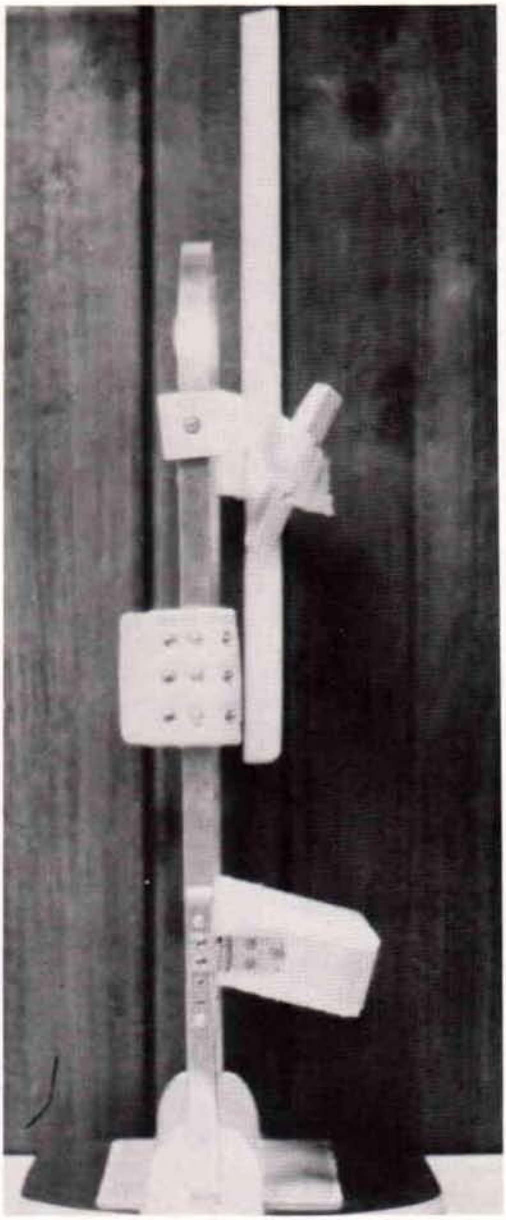

Fig. 2. Lateral view of the pre-Parapodium orthosis

Half-inch-thick neoprene rubber has been placed on the bottom surface of the base plate to provide a slip resistant surface.

A 4 to 5-in. wide buttock strap is made from Vibretta-covered elk hide. Four eyelets are placed at the edge, and the strap is located at the trochanter level and held in place by screws in the pre-drilled #8–32 holes. The length of the strap is 4-in. more than the diameter of the frame. The series of eyelets allow the application of tension required for posterior control (Fig. 2).

A 1-1/4-in. wide X 1/8-in. thick aluminum strap bent to shape is used to contain the Styrofoam knee block. It is attached laterally with a 1-in. wide stainless steel hinge and angulated approximately 10 to 15 deg. to conform with the flexed-knee balancing position. The knee block section is placed proximal to the tibial tubercle (Figs. 2 & 3).

On the superior aspect of the base plate, Velcro hook material, backed by a light piece of leather, is glued and fastened. The bottom of the shoes are provided with Velcro pile to assist in proper positioning of the feet on the base plate. Velcro instep straps are provided to control the feet upon weightbearing, and are held in place by 1/2-in. self-tapping machine screws (Fig. 4).

Various other modifications have been provided when necessary such as: 1) An adjustable member superiorly along with a wider base plate and an adjustable knee block section to accommodate for change in diameter; 2) A posterior superiorly directed bar to accommodate a head halter attachment for the hydrocephalic child.

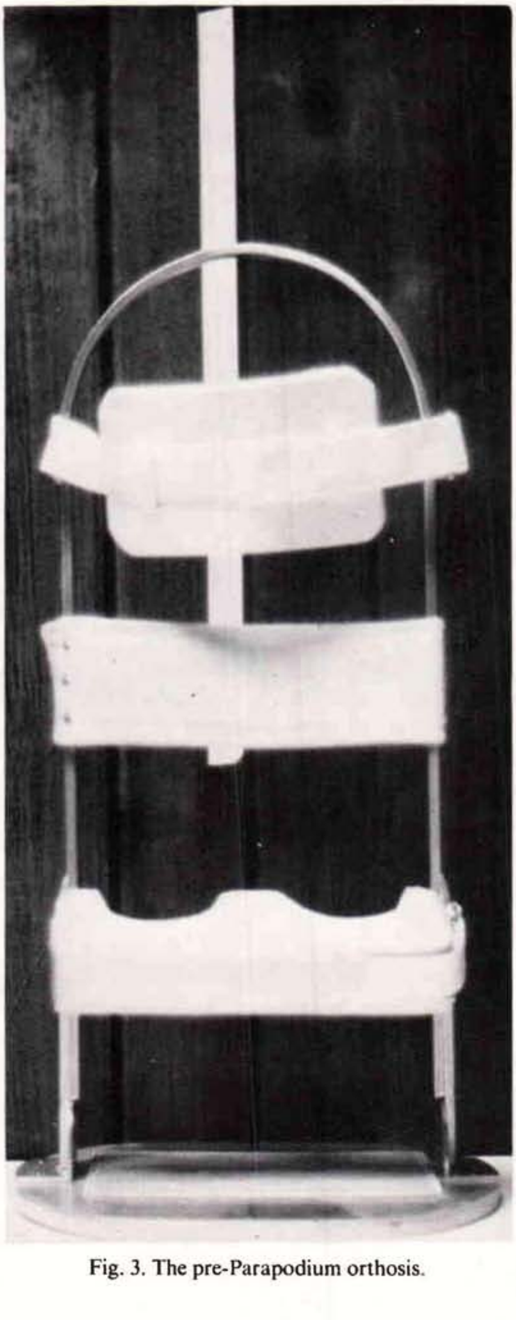

Fig. 3. The pre-Parapodium orthosis.

## ADVANTAGES

The advantages offered by this type of standing orthosis are:

1. It is adjustable
2. It is quite light in weight
3. It is durable
4. It can be used over and over on children as each user graduates to the Parapodium and other orthoses.
5. Any orthotic or prosthetic facility can make it easily.
6. The parts and materials are readily available.
7. Donning and doffing pose no problems.

## CONCLUSION

Parents are instructed initially to use the standing frame for a period of one to one-and-a-half hours BID for several days and slowly increase

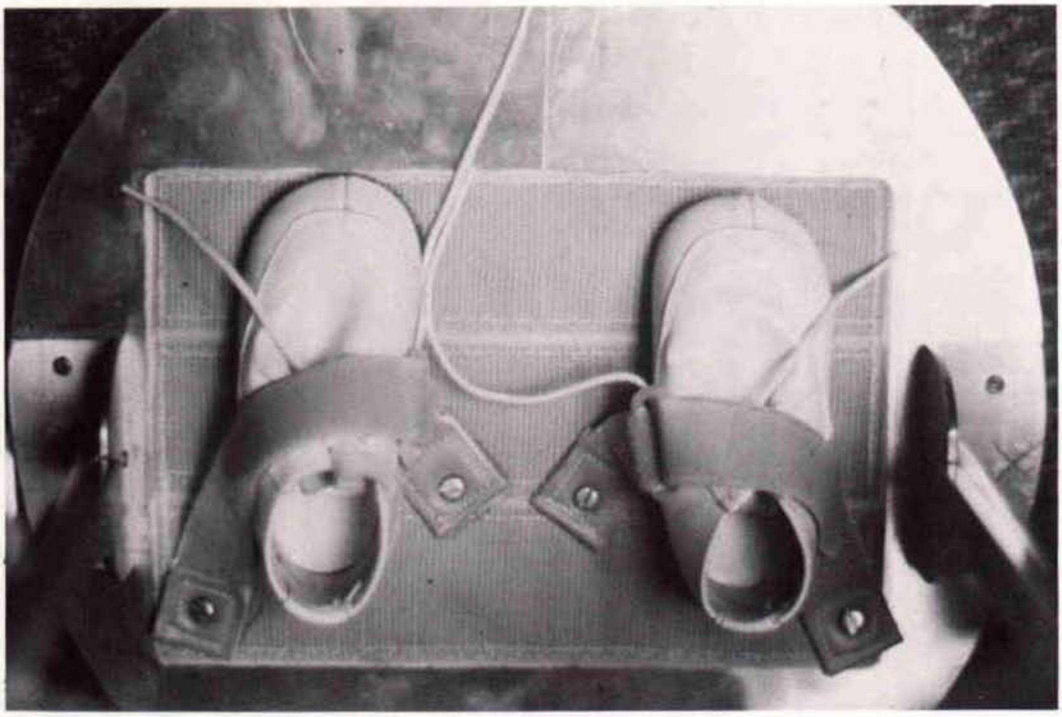

Fig. 4. Velcro straps are used across the insteps to control the feet during standing.

wearing time up to three to four hours TID with care always taken to make sure that the socks are not wrinkled and shoes are not fastened tightly.

After the child is removed from the standing frame, he should be checked for any pressure sensitive areas on the feet, instep, and knee areas. We encourage the parent to gently massage this area to promote circulation and thus prevent tissue trauma.

We discourage the use of baby lotions because they tend to make the skin too soft and more subject to pressure sores. In fact, we prefer alcohol slightly diluted with water to promote toughening of the tissue.

Standing is important for these children and it should be the goal of everyone of us to provide systems to better meet these needs. We tend to take for granted our standing and ambulating capabilities. To the young paraplegic child, it is a dream come true.

## ACKNOWLEDGEMENT

My sincerest thanks go to Newton C. McCollough, III, M.D., Director, Rehabilitation Center, Jackson Memorial Hospital and to Mr. Charles C. Thomas, C.O., J. E. Hanger Inc., Miami, Florida for their assistance in this project.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 29-31, September 1976

# ANOTHER USE FOR A STANDING ORTHOSIS

Maurice A. LeBlanc, C.P.[1]
Eugene E. Bleck, M.D.[1]
David P. Freligh[1]

This paper describes how a standing orthosis with a pivot walker attachment was used to assist a child with bilateral fibular centralizations and ankle disarticulations due to bilateral longitudinal deficiencies of the tibia.

## PATIENT HISTORY

M.L. is a three-year-old black girl born with bilateral longitudinal deficiencies of the tibia. At four months of age, she had bilateral fibular centralizations of the knee (Brown procedure) (*1*) and bilateral ankle disarticulations of her deformed feet. She has flexion contractures of the knees (right, 20 deg.; left, 40 deg.); weak hip musculature; and ranges of motion of the hips as follows: abduction, 90 deg.; flexion, 125 deg.; extension 0 deg.; external rotation, 90 deg.; internal rotation, 0 deg.

## PROSTHETIC TREATMENT

Initially, at about age two, she was fitted with bilateral Symes prostheses with hip joints and pelvic band. Her use of these prostheses was only part-time and were not successful in providing mobility.

Subsequently, she was fitted as shown in Figures 1 and 2. The intention was to provide knee stability and weight-bearing with the prostheses and to allow her to use and strengthen her hip musculature. However, because she had been sitting for most of her three years, her hip muscles were quite weak, and she could not stand without support.

Fig. 1. M.L. with bilateral Symes prostheses.

## USE OF STANDING ORTHOSIS

One possibility to allow her to gain mobility would have been to revert to use of hip joints and pelvic band. Instead, we chose to provide a standing orthosis as shown in Figures 3-5. The pivot walker attachment (*3*) was used to increase her mobility.

[1]Rehabilitation Engineering Center, Children's Hospital at Stanford, 520 Willow Road, Palo Alto, California 94304.

Fig. 2. Symes prostheses showing knee flexion contractures.

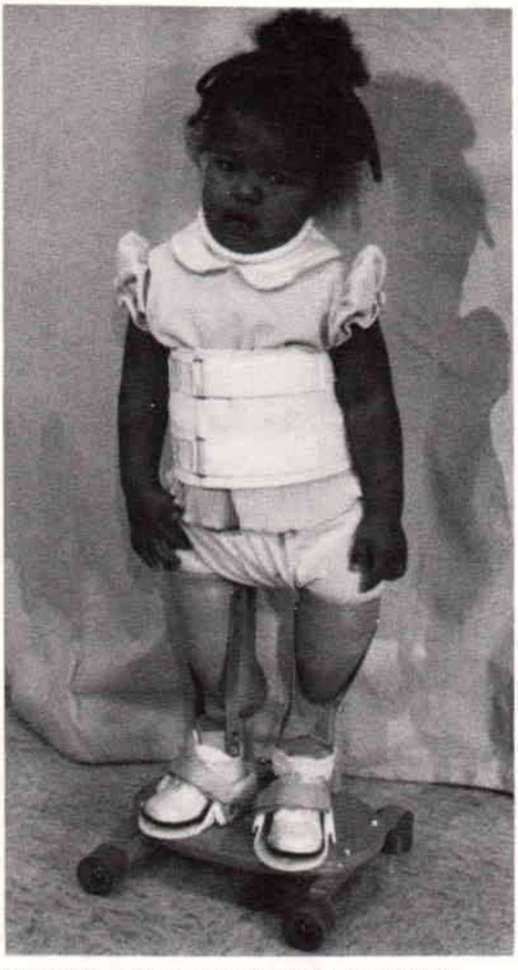

Fig. 3. M.L. with prostheses and standing orthosis.

The goal is to give her enough support to stand, and at the same time encourage the use of her own hip musculature. Once her muscles are strong enough, the plan is to discard the standing orthosis, and have her use only the prostheses, perhaps with a tie-bar (*2*), for ambulation.

In the meantime, the standing orthosis provides a good way of giving hip and trunk support as necessary and controlling external rotation of the hips without modification of the prostheses. Her therapists[2] report she is now wearing the prostheses and standing frame at school and at home, and is using the pivot walker around the classroom and outside on the playground. She puts it on and takes it off independently, and is practicing falling to the floor.

[2]Jeanie Cho, R.P.T. and Belinda Chan, Chandler Tripp Medical Therapy Unit, CCS, Santa Clara County, California.

Fig. 4. Standing orthosis showing A-P alignment.

Fig. 5. Standing orthosis showing M-L alignment and elastic trunk support.

The long range plan for this girl is to reduce her knee flexion contractures as much as possible by manual stretching, complete the correction surgically, and re-fit with prostheses.

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3. Motloch, Wallace M., *1970 research report*, Rehabilitation Eingineering, Ontario Crippled Children's Centre, Toronto, Ontario M4G-1R8, Canada

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 33-34, September 1976

# TO "WALK" OR TO RIDE?

Maurice A. LeBlanc, C.P.[1],
Fran Ford, R.P.T.[1],
Wallace M. Motloch, C.O.[1],
Eugene E. Bleck, M.D.[1]

This paper describes the prosthetic management of a young quadramembral amputee.

## PATIENT HISTORY

M.C. is a two-year-old boy born with transverse deficiences of both arms and longitudinal deficiences of both femurs (total) and tibias (partial).

## PHILOSOPHY OF TREATMENT

There are immediately apparent two approaches to the lower-limb management of this child: one, to assume that he will eventually elect wheeled mobility and therefore start at the outset with a wheeled vehicle such as the CAPP cart (*1, 4*), bypassing the often short-lived attempt at ambulation; and the other, to provide a swivel walker at the earliest opportunity, increase the height as the child grows, and hope that it will be beneficial for him. Experience has shown that patients usually will choose wheeled mobility in the end. The question, therefore, seems to be is it worthwhile to provide swivel-walker ambulation?

## RESOLUTION OF APPROACH

There was a great deal of discussion by our staff concerning which approach to take. The strongest influence was the desire of the parents for M.C. to "walk". Hence the decision was made to provide ambulation.

[1]Rehabilitation Engineering Center, Children's Hospital at Stanford, 520 Willow Road, Palo Alto, California 94304.

## PROSTHETIC TREATMENT

At 8 months of age we provided M.C. with conventional, bilateral, passive, shoulder prostheses for the upper limbs. At the same age, we drew on the experiences of Ontario Crippled Children's Centre (*3*) and provided a molded seat mounted on a snow disc to encourage trunk motion and balance (Fig. 1).

Fig. 1. M.C. with bilateral shoulder prostheses and snow disc for learning balance.

Also at the same age, for mobility, we provided another molded seat mounted to a "hippopotamus" crawler from a local toy store. We cut holes in the crawler and positioned the seat the correct height off the floor so M.C. could push along with his flipper feet (Fig. 2).

Fig. 2. M.C. with Hippopotamus mobility aid.

Fig. 3. M.C. with swivel walker.

These devices worked satisfactorily, but the patient quickly outgrew them. At 1-1/2 years of age we provided M.C. with a red, white, and blue "Bi-Centennial" swivel walker (*2*), which he is now happily using (Fig. 3). The molded seat is made of vacuum-formed white polypropylene; the rotational pylon units are covered with white Plastazote; the seat platform is wood painted red; and the "feet" are wood painted blue and lined with shoe-sole bottoms.

## DISCUSSION

Having had the experience of one case, we would now view a second such case with more direction, realizing of course that each situation must be considered individually.

M.C. will likely decide to use a wheeled vehicle for mobility when he gets older, because it is, by far, the more efficient mode of "transportation". However, in younger years the growth-development process is extremely important in shaping his future. If, in these years, we can provide a tool/toy/device to assist that process, then we have helped to shape a person. There is not much better physical assistance we can give than to provide the means for a child to stand, "walk" and to "grow" in height with his peers.

## REFERENCES

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3. Rehabilitation Engineering Center, Colin A. McLaurin, Director, Ontario Crippled Children's Centre, 350 Rumsey Rd., Toronto, Ontario M4G-1R8, Canada

4. Sumida, Carl, C.P.O., Yoshio Setoguchi, M.D. and Julie Shaperman, M.A., O.T.R., *"The CAPP electric cart: recent developments,"* Artificial Limbs, Autumn 1971, pp. 11-15.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 35-36, September 1976

# MODIFICATION OF THE POSTERIOR LEAF-SPRING ORTHOSIS

John Sabolich, C.P.O.[1]

For two years we fitted many posterior leaf-spring ankle-foot orthoses with the idea that they were contraindicated when medial or lateral instability was present. The types of cases where this instability was present are numerous, and when stroke cases are added, the number of cases with medial or lateral instability are enormous. We felt surely that some type of modification could be made to provide adequate control so that the many thousands of patients who have this common combination of ankle pathomechanics could take advantage of the plastic posterior leaf-spring ankle-foot orthosis (AFO).

Fig. 1 shows the ideal location of the resultants of the forces needed to correct varus of the ankle joint. Obviously, it would not be wise to bring pressure to bear directly on the lateral malleolus; however, from a mechanical standpoint, the closer the force vector is to the ankle joint the better.

Our first approach was to attach an aluminum extension to the plastic orthosis. These lateral extensions worked so well that we experimented with simply extending the plastic on the lateral side in the area superior to the lateral malleolus, as shown in Figure 2.

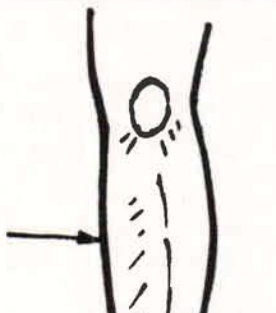

Fig. 1. Ideal location of the resultants of the forces needed to correct varus of the ankle joint.

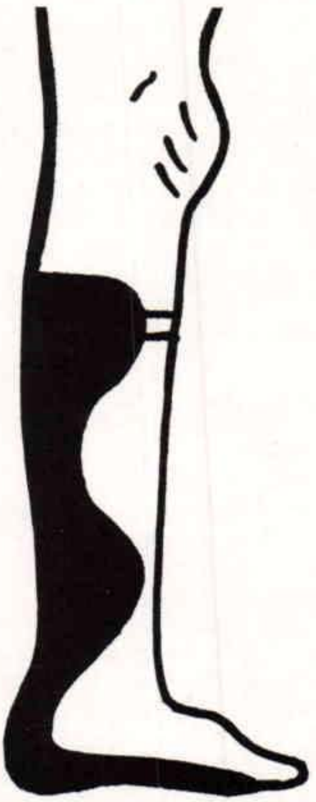

Fig. 2. Outline of the lateral wall of the modified posterior leaf-spring ankle-foot orthosis.

[1]Sabolich Inc., 1017 N.W. 10th Street, Oklahoma City, Oklahoma 73106

Fig. 3. The modified leaf-spring ankle-foot orthosis to control varus.

As we progressed with this technique we began to lower the extension closer to the malleolus. Finally we began to lower the extension to a point where the trim line fell over the apex of the malleolus or even encompassing the entire area. However it is not necessary to do this in order to control the usual varus or valgus condition. The closer the medially directed force is to the anatomical ankle joint, the more effective is the control of lateral instability.

These same principles can be applied to solution to the opposite problem, valgus of the ankle complex, by simply reversing the location of the extension. We have also some cases where both a lateral and medial extension has been indicated, and were fitted with good success.

I have used the terms ankle joint and ankle joint complex loosely when referring to varus and valgus. Perhaps we, as orthotists, should be more specific and refer to the subtalar joint as the primary joint in varus and valgus of the foot, since the ankle joint actually has very little movement in the frontal plane.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 37-38, September 1976

# A NON-STANDARD ABOVE-ELBOW PROSTHESIS FOR A MULTIPLY HANDICAPPED PATIENT

Jonathan Batzdorff[1]

The multiply handicapped above-elbow amputee may find the standard prosthesis for that level too elaborate and cumbersome. When this is so a non-standard prosthesis which accommodates the limited potential of the wearer is indicated.

A 71-year-old female, who was referred to the UCLA amputee clinic by an occupational therapist, had a left elbow disarticulation at the age of six due to trauma. Although she was fitted with a prosthesis at that time, she discarded it later, and for many years performed daily activities satisfactorily with the sound arm and the amputation stump.

At age 70, the patient had a cerebrovascular accident that resulted in paralysis of the right arm and leg. As a consequence, she required assistance in all self-help skills, but learned to ambulate to a limited degree in a wheelchair using her unaffected left leg.

Upon examination, the patient had normal range of gleno-humeral motion on the amputated side. The presence of the wheelchair interfered with scapular motion and humeral extension. Her muscular strength was good, although she tired quickly.

During the initial evaluation the patient explained that her main priorities were independence in feeding and in operating the wheelchair brakes. She assured us that cosmesis was of minimal concern. To these we added several other requirements of the prosthesis. It would have to be light, require minimal training, and be versatile enough to allow an increasing variety of tasks.

A standard above-elbow prosthesis was considered, but several disadvantages precluded its use. The patient possessed neither the energy nor the agility required to operate and position the prosthesis in space, such as is necessary for eating.

Owing to the patient's physical limitations, a non-standard prosthesis was designed. The functional objectives were achieved with a double-walled socket, a passive adjustable elbow joint, and a short forearm section. A Dorrance 99X hook was used (Fig. 1).

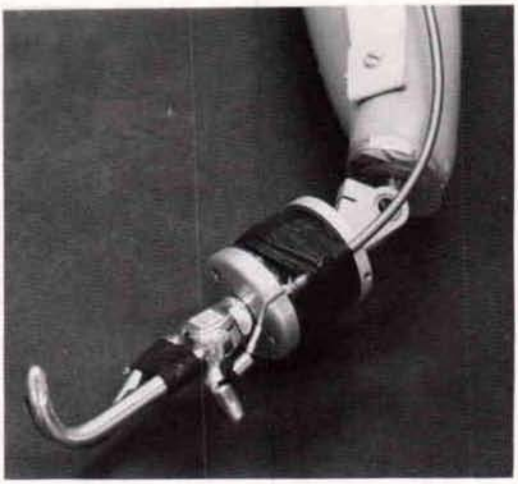

Fig. 1. Special purpose elbow-disarticulation prosthesis provided a hemiplegic. A Hosmer "Universal" shoulder joint is used for the elbow joint.

The socket was trimmed proximally to facilitate maximum range of motion. Distally, it was flattened enough to stabilize the epicondylar area sufficiently to make humeral rotation useful.

A Hosmer "Universal" shoulder joint was used as an elbow joint (Fig. 1) because it permits a wide range of adjustment about three axes; 360° of elbow rotation, 360° of forearm rotation, and 200° of elbow flexion and extension. Specific prepositioning allows for each task to be performed with maximum efficiency. For example, the prosthesis can be adjusted to a position where the terminal device can hold a spoon and maintain it level while food is picked up and placed in the mouth (Fig. 2). It can be oriented also to

[1]Department of Orthopaedic Surgery, University of California at Los Angeles, 1000 Veteran Ave., Los Angeles, Calif. 90024.

Fig. 2. Patient using the special purpose prosthesis shown in Figure 1 to eat.

facilitate other activities such as operation of the wheelchair brakes (Fig. 3). The patient prepositions the hook and forearm by pushing the prosthesis against the sound leg or the wheelchair armrest. The short forearm accommodates the length of a spoon or fork. Terminal device control was omitted during the initial training and evaluation period to avoid reaction to complexity.

After three sessions with the occupational therapist, the patient was able to feed herself after the eating utensil had been adjusted in the hook. She was given two more weeks of practice, then fitted with a standard single-control Bowden cable for active hook operation. The distal cable housing retainer was placed on the forearm to assure that the elbow position would not change during hook operation.

At the present time, eight weeks after beginning her training with the prosthesis, the patient is able to feed herself and operate the brakes of her wheelchair. She is improving her dexterity with the prosthesis by manipulating dominoes, tracing printed designs on paper, and watercoloring.

Fig. 3. Use of special purpose prosthesis shown in Figure 1 to operate wheelchair brake.

Orthotics and Prosthetics, Vol. 30, No. 3, pp. 39–40, September 1976

# NEW PUBLICATIONS

ATLAS OF ANATOMY OF THE HAND. Johan M. F. Landsmeer. Churchill Livingstone; Edinburgh, London, and New York 1976. 349 pp. $55.

The author of this beautifully prepared book is Professor of Anatomy and Embryology at the University of Leiden, and is well known for his work in biomechanics of the hand, some of which was carried out in collaboration with investigators in the United States.

Dr. Lansmeer's work led him to believe that there was a need for an analysis of the structure of the hand, and he has done this using serial sections. Although this book is intended primarily as a reference source of old and new information on the hand for surgeons and anatomists and therefore contains more detail than is usually needed by orthotists, the last chapter, "Functional Considerations," should prove useful to orthotists who are called upon to provide devices for patients with the more severe types of hand dysfunction.

ATLAS OF ORTHOTICS—BIOMECHANICAL PRINCIPLES AND APPLICATION. AMERICAN ACADEMY OF ORTHOPAEDIC SURGEONS. C. V. Mosby Co., St. Louis, 1975, 507 pp., 836 illus., $42.50.

The Committee on Prosthetics and Orthotics of the American Academy of Orthopaedic Surgeons has compiled an outstanding volume on current orthotic appliances and devices. Thirty-six experts have contributed articles describing the latest advancements in the art with particular attention focused on biomechanics and new materials and techniques.

Included also is the current orthotic terminology. The new nomenclature is based on the enumeration of joints encompassed and their control of motion.

One of the most helpful adjuncts is that in several chapters the author describes the Technical Analysis Form for the segment involved and sets forth instructions for its proper usage.

Thanks go to the many persons involved in compiling this most comprehensive and informative book. It certainly belongs in the library of all orthopedic surgeons, orthotists, physiatrists, mechanical engineers, and other individuals interested in assisting handicapped persons.

Dr. Newton C. McCollough III, summarized the necessity for this volume beautifully: "The basis for orthotic prescription should be an accurate biomechanical analysis of the patient, followed by selection of appropriate components, and finally the creation of an orthotic system from the components selected."

H.C. Hittenberger, C.P.O.

EQUIPMENT FOR THE DISABLED. A series of illustrated booklets compiled and edited at Mary Marlborough Lodge, Nuffield Orthopaedic Centre, Oxford; published by the Oxford Regional Health Authority on behalf of the Department of Health & Social Security.

"Clothing & Dressing for Adults", the second of the new *Equipment for the Disabled* booklets to be published in the 4th edition, is now available. It has been completely revised and expanded and provides guidance on selecting ready-made clothes in practical styles which are both fashionable and flattering and draw attention away from disability. Specially designed clothes, particularly for those with incontinence, are also included. Suggested styles and various adaptations, including adapted patterns, are given for the home dressmaker. Dressing aids are also shown.

The booklet, which is fully illustrated with photographs and diagrams, gives brief details of the items shown, an indication of prices, garment washing instructions, manufacturers' and suppliers' addresses. It is intended for use by professional workers caring for disabled people, but disabled people themselves will find much useful advice in the booklet to help them with their clothing problems.

Booklets available from:
Equipment for the Disabled
2 Foredown Drive
Portslade, Sussex, England
BN4 2BB

U.K. prices. Booklets £1.50 each; binders (for 10 booklets) £1.50; postage extra. Overseas prices quoted on request.

MANUAL ON DYNAMIC HAND SPLINTING WITH THERMOPLASTIC MATERIALS. By Maude H. Malick, OTR, Harmarville Rehabilitation Center, Box 11460, Guys Run Road, Pittsburgh, Pa. 15238.

The title of this publication reflects precisely the contents. After a review of the medical principles, the anatomy and the physiology of the hand, and the mechanics of splinting, the author provides step-by-step procedures for designing, fabricating, and fitting various types of hand orthoses. Properties of those sheet plastics suitable for forming over the hand are included as well as the general principles in the use of these new materials. In addition, reasons for design and application are set forth.

This manual is intended for physicians, therapists, and orthotists. It should be readily available to every person in those professions who are responsible for the care of patients with impairments of the hand.

JOINT PRESERVATION TECHNIQUES FOR PATIENTS WITH RHEUMATOID ARTHRITIS. Ruth Ann Watkins, OTR and Dianne Robinson, OTR. Medical Research and Training Center #20, Northwestern University — Rehabilitation Institute of Chicago, 345 E. Superior Street, Chicago, Illinois 60611, 1974, 51 pp.

This booklet has been prepared expressly to help the patient with rheumatoid arthritis to understand a method of treatment devised at the Rehabilitation Institute of Chicago. An explanation of the disease process is followed by instruction in exercise programs and in carrying out certain activities of daily living designed to preserve the affected joints. A supplement addressed to the therapist is included.

This text and illustrations are well done. This booklet should prove to be useful in the management of rheumatoid arthritis.

STEDMAN'S MEDICAL DICTIONARY, 23rd Edition. The Williams and Wilkins Co., 1976, 1678 pp.

The twenty-two editions of Stedman's Medical Dictionary published since the appearance in 1911 of "A Practical Medical Dictionary" by Thomas Lathrop Stedman have consistently provided the medical profession with an authoritative source of definitions. This latest edition is no exception. According to the preface in the twenty-third edition, 15,313 new entries have been included, of which 10,322 are new definitions and the remainder are new cross-references for synonyms. It is gratifying to note that it contains quite acceptable definitions for *orthotics, orthosis, orthotist, prosthetics, prosthesis, and prosthetist.*

# RESOLUTION CONCERNING THE METRIC SYSTEM

The following resolution was adopted by the Board of Directors of the American Orthotic and Prosthetic Association at its meeting in San Diego October 3, 1973:

WHEREAS by Act of Congress it has been determined that the United States should proceed towards adoption of the metric system as used almost universally throughout the rest of the world, and

WHEREAS the technological professions and many segments of the health professions have commonly used the metric system over an extended period of time, and

WHEREAS it is important for members of the orthotic/prosthetic professions to interact with their colleagues in the medical and technological communities for optimum patient service be it hereby

RESOLVED that the American Orthotic and Prosthetic Association endorses the use of the metric system by its members and other orthotic and prosthetic practitioners in the United States, and in witness of this endorsement and Association urges the editors of its journal *Orthotics and Prosthetics* to commence the dual reporting of weights and measurements in both the English and metric systems at the earliest possible date with the objective of employing the metric system solely by the time of the 29th Volume in 1975.

# METRIC SYSTEM
## Conversion Factors

### LENGTH

Equivalencies

angstrom = $1 \times 10^{-10}$ meter (0.0 000 000 001 m)
millimicron* = $1 \times 10^{-9}$ meter (0.000 000 001 m)
micron (micrometer) = $1 \times 10^{-6}$ meter (0.000 001 m)

| To Convert from | To | Multiply by |
|---|---|---|
| inches | meters | 0.0254† |
| feet | meters | 0.30480† |
| yards | meters | 0.91440† |
| miles | kilometers | 1.6093 |

### AREA

| To convert from | | |
|---|---|---|
| square inches | square meters | 0.00063616† |
| square feet | square meters | .092903 |

### VOLUME

Definition

1 liter = 0.001† cubic meter or one cubic decimeter ($dm^3$)
(1 milliliter = 1† cubic centimeter)

| To convert from | To | Multiply by |
|---|---|---|
| cubic inches | cubic centimeters | 16.387 |
| ounces (U.S. fluid) | cubic centimeters | 29.574 |
| ounces (Brit. fluid) | cubic centimeters | 28.413 |
| pints (U.S. fluid) | cubic centimeters | 473.18 |
| pints (Brit. fluid) | cubic centimeters | 568.26 |
| cubic feet | cubic meters | 0.028317 |

### MASS

| To convert from | To | Multiply by |
|---|---|---|
| **pounds (avdp.)** | **kilograms** | **0.45359** |
| **slugs‡** | kilograms | 14.594 |

### FORCE

| To convert from | To | Multiply by |
|---|---|---|
| ounces-force (ozf) | newtons | 0.27802 |
| ounces-force (ozf) | kilogram-force | 0.028350 |
| pounds-force (lbf) | newtons | 4.4732 |
| pounds-force (lbf) | kilogram-force | 0.45359 |

***This double-prefix usage is not desirable. This unit is actually a nanometer ($10^{-9}$ meter = $10^{-7}$ centimeter).**
**† For practical purposes all subsequent digits are zeros.**

**STRESS (OR PRESSURE)**

| To convert from | To | Multiply by |
|---|---|---|
| pounds-force/square inch (psi) | newton/square meter | 6894.8 |
| pounds-force/square inch (psi) | newton/square centimeter | 0.68948 |
| pounds-force/square inch (psi) | kilogram-force/square centimeter | 0.070307 |

**TORQUE (OR MOMENT)**

| To convert from | To | Multiply by |
|---|---|---|
| pound-force-feet | newton meter | 1.3559 |
| pound-force-feet | kilogram-force meters | 0.13826 |

**ENERGY (OR WORK)**

Definition

One joule (J) is the work done by a one-newton force moving through a displacement of one meter in the direction of the force.

1 cal (gm) = 4.1840 joules

| To convert from | To | Multiply by |
|---|---|---|
| foot-pounds-force | joules | 1.3559 |
| foot-pounds-force | meter-kilogram-force | 0.13826 |
| ergs | joules | $1 \times 10^{-7}$† |
| b.t.u. | cal (gm) | 252.00 |
| foot-pounds-force | cal (gm) | 0.32405 |

**TEMPERATURE CONVERSION TABLE**

**To convert °F to °C** $$°C = \frac{°F - 32}{1.8}$$

| °F | °C |
|---|---|
| 98.6 | 37 |
| 99 | 37.2 |
| 99.5 | 37.5 |
| 100 | 37.8 |
| 100.5 | 38.1 |
| 101 | 38.3 |
| 101.5 | 38.6 |
| 102 | 38.9 |
| 102.5 | 39.2 |
| 103 | 39.4 |
| 103.5 | 39.7 |
| 104 | 40.0 |

*A slug is a unit of mass which if acted on by a force of one pound will have an acceleration of one foot per second per second.

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4. Write out numbers less than ten.
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*R. R. Linton, "Post Thrombotic Ulceration of the Lower Extremity: Etiology and Surgical Treatment," *ANNALS OF SURGERY*, Vol. 138:415.

*John J. Cranley, *Vascular Surgery, Vol. II, Peripheral Venous Diseases.* New York, N.Y.: Harper & Row, 1975.

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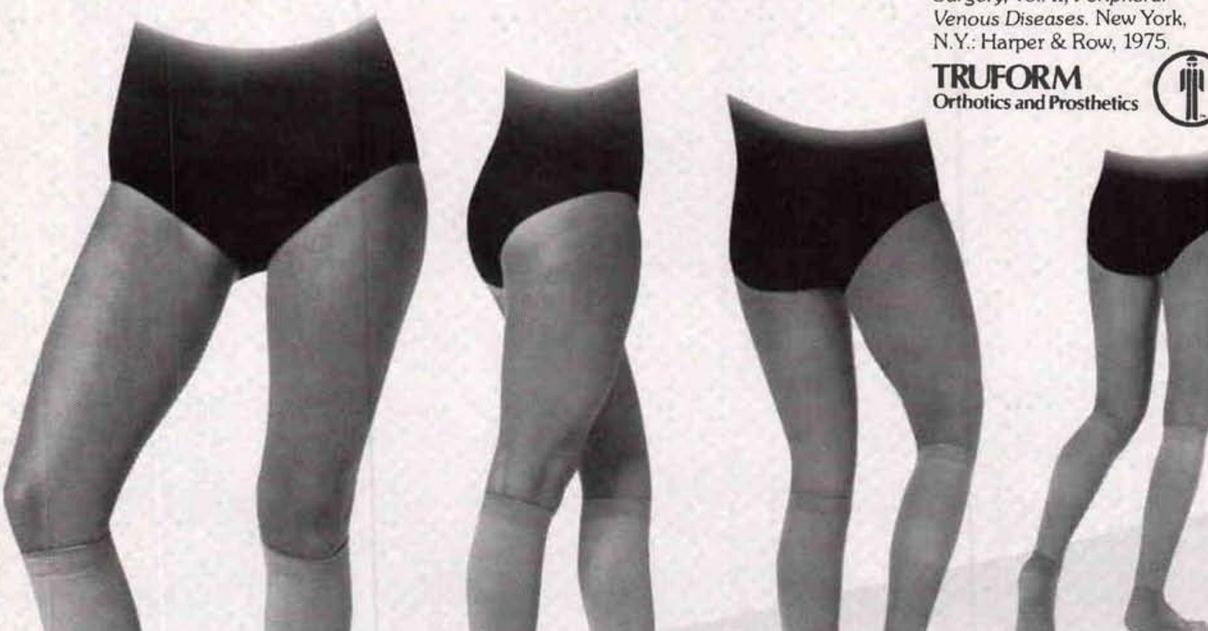



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*References: Vert Mooney, M.D.; Milton Ashby, M.D.; Benjamin E. Lesin, M.D.; Western Orthopedic Conference, October 29, 1975

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